# A Quantitative Simulation Approach for Optimization of Spiral K-Space-Sampled Time-Resolved Contrast-Enhanced MRA

**T. Song[1], A. F. Laine[1], Q. Chen[2], H. Rusinek[2], G. Laub[3], R. Kroeker[4], L. Bokachev[2], and V. S. Lee[2]**

[1]Biomedical Engineering, Columbia University, New York, NY, United States, [2]Radiology, NYU School of Medicine, New York, NY, [3]Siemens Medical Solutions USA, Los Angeles, CA, United States, [4]Siemens Medical Solutions USA, Winnepeg, ME, Canada

**Introduction**

Time-resolved k-space undersampling techniques for contrast-enhanced 3D MRI have many clinical applications including organ perfusion and MRA studies. However, there is a trade-off between high temporal resolution and high spatial image quality with fewer artifacts. The extent of k-space undersampling theoretically should be guided by the spatial resolution desired for key enhancing structures, bounded by temporal resolution needs and image artifact minimization. Our purpose was to develop a multi-vessel MRA model to determine the relationship between imaging parameters and image quality.

**Methods**

(1) A computer simulated 3D multi-vessel phantom (60x60x60 matrix) was made to simulate vessels of different diameters from D=1 3, 7, 11, 15, to 18 voxels (Fig.1). A gamma-variate function was to generate various aortic enhancement curves [1]. For example, if 10 ml bolus of Gd-DTPA was injected intravenously over 5 sec, an aortic enhancement curve could be simulated by gamma-variate function with alpha = 20, tmax = 10sec, maximum amplitude=300, and minimum amplitude = 100 [2]. (2) Time-resolved contrast-enhanced MR technology (TWIST, Siemens [3]) splits k-space into two separate regions: *A* (lower spatial frequency) and *B* (higher frequency). K-space points are reordered by their radial distances and angles shown in Fig. 2(a). *A* and *B* are sampled in a fashion shown in Fig. 2(b).With *A* acquired more frequently, images can be updated more rapidly. Region *B* is undersampled with respect to *A* and represented by *N* trajectories: *Bi* (*i = 1, 2, 3, …*), each trajectory spanning the whole B region. According to the acquisition definition, *pA (percentage of A)* that is sampled with each acquisition can be defined as *kc/kmax,* and can, in turn, be related to the spatial resolution of the frequently sampled data (alternatively, *pA* can be expressed as the percentage area of k-space). *pB (percentage of B)* can be defined as *1/N* called sampling density in region *B*. (3) We hypothesize that the inverse relation of k-space voxel size and image voxel size should help determine *pA(=kc/kmax)*. *pA* should be larger than *Dvoxelsize/Dvesselsize=1/D (D is diameter)* in order to reconstruct signal intensity (SI) precisely. For example, for a vascular diameter *D=3* voxels, pA should be at least 0.33, while for *D = 11*, pA should be at least 0.09, *pB* can then be used to adjust the desired temporal resolution, balanced against increasing artifacts with k-space undersampling. Temporal accuracy was evaluated by calculating errors from averaged signal intensity aortic enhancement curve estimation. In order to assess image ringing artifacts caused by k-space undersampling, differences between reconstructed and true images, normalized by true image SI, were calculated. Total error index was calculated as the sum of these two errors.

**Results**

In our simulation, the acquisition time (TA) without TWIST is 13 sec. We initially fixed *pB = 1/3* and investigated *pA=0.5, 0.33, 0.25, 0.20, 0.10* (*=40%, 17%, 10%, 6%, and 2%* of the k-space area), resulting *TA = 7.7, 5.8, 5.2, 4.9, and 4.5sec*, respectively. For *pB = 1/7,* the same group of *pA* values results in *TA = 6.2, 3.8, 3.0, 2.6, and 2.1sec.* Averaged signal intensities in two regions of interest (ROI) over time were calculated at two positions (Fig 1) in the phantom: *D=3* and *D=11* and plotted in Fig.3. More accurate representations of enhancement curves are obtained with larger vessels. *pA = 0.33* provides the highest fidelity with truth by balancing spatial resolution with sufficient temporal resolution to capture the dynamic changes of the enhancement profile. To assess for errors, differences between reconstructed image and ground truth image were normalized by ground truth SI value for each voxel (example shown in Fig 5). Root mean square of normalized differences was used to calculate errors caused by artifacts. Errors from differences of SI versus time curves and from artifacts were integrated and summed as a criterion for quantitative optimization shown in Fig.4. For *D=3*, optimal *pA=0.33* with smallest error (peak aortic enhancement underestimated by 14.1%). Considering both image artifacts and temporal resolution needed to accurately represent the enhancement curve used for this phantom, optimal *pB=1/3*. For D=11, optimal *pA <0.25,* and *pB=1/7*, which gave the best estimation of SIs (peak enhancement underestimated by 7.3%) as well as smallest integrated errors (rms = 0.10).

**Conclusions**

A multi-vessel MRA phantom provides a useful method for analyzing k-space undersampling parameters for time-resolved MRA, both in terms of providing the best representation of enhancement curves and quantitatively evaluating ringing artifacts caused by undersampling.

**References**

[1] K. T. Bae, Radiology, 2003, Vol. 227, No. 3, 809-816. [2] M. T. Madsen, Phys. Med. Boil., 1992, Vol. 37, No 7, 1597-1600. [3] R. Kroeker et al. Siemens Application Document on Time-resolved Contrast-enhanced MRA: TWIST, 2006.

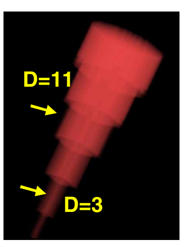


Fig.1 . 3D vessel phantom.

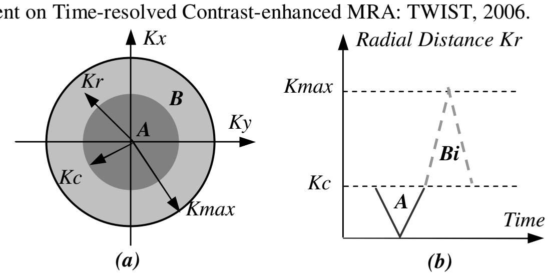


Fig. 2. (a) K-space is divided into A and B regions in phase encoding plane. (b) Acquisition sampling order.

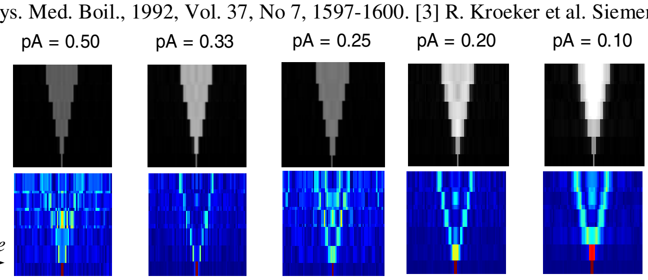


Fig. 5. Top: A snap shot of one of reconstructed slices with pB=1/7, and bottom: normalized difference between reconstructed image and ground truth.

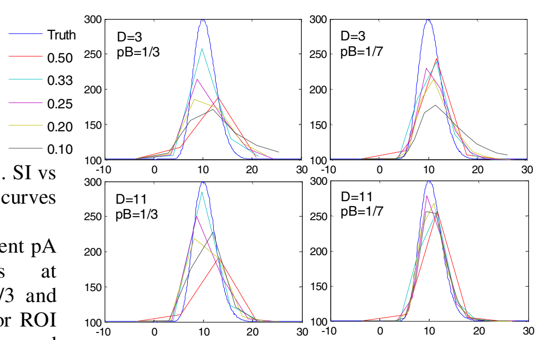


Fig. 3. SI vs time curves with different pA values at pB=1/3 and 1/7 for ROI D=3 and D=11.

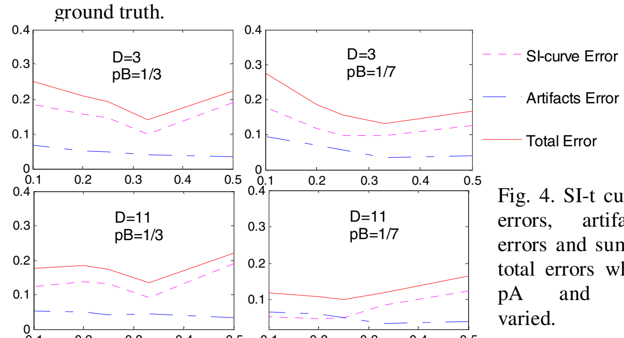


Fig. 4. SI-t curve errors, artifacts errors and sumed total errors when pA and pB varied.