AIF Correction for Non-Uniform k-space Data Weighting Effects in First-Pass Cardiac Perfusion MRI

Sohae Chung[1], and Leon Axel[1]
[1]*Center for Biomedical Imaging, Radiology Department, NYU Langone Medical Center, New York, NY, United States*

Introduction: To obtain first-pass cardiac perfusion images, a saturation-recovery (SR) preparation with TurboFLASH readout can be used to capture the contrast enhancement dynamics. However, when using a SR preparation, non-uniform k-space data weighting in the phase-encoding direction, due to data acquisition during the transition toward the steady state, can lead to a filtering effect. This may lead to systematic overestimation of the image-derived arterial input function (AIF), with a resulting bias in the perfusion calculations. In this work, numerical simulations were used to correct for non-uniform k-space data weighting effects on the AIF.

Method: A SR TurboFLASH imaging sequence with centric k-space reordering was implemented on a 3T MR scanner (Tim Trio; Siemens). Image parameters included: slice thickness = 8mm, TE/TR = 1.2/2.4ms, TD = 50ms, flip angle = 10°, and receiver bandwidth = 1008Hz/pixel. A proton density-weighted (PDw) image was also acquired for normalization [1]. Three water phantoms with different concentrations of Gd-DTPA and six healthy volunteers (29±12 years old) were tested; in vivo, the AIF measurements were performed at the aortic root. The T_1 values were calculated using a single-point T_1 method [1,2]. For the phantom, a reference T_1 value was calculated from a multi-point spin-echo inversion recovery method [3]. For signal correction, a numerical model of each object geometry was generated and the corresponding k-space data-weighting filters for the T_1-weighted (T_1w) and PDw images were calculated using the Bloch equations with the same imaging parameters and k-space trajectory used in MR imaging (Fig. 1). Using the numerical simulations, for the phantoms, a fitting algorithm was used to correct the T_1 values and for the in vivo data, the MR signals were corrected using a look-up table, including a pair of signals simulated with and without non-uniform k-space data weighting filters.

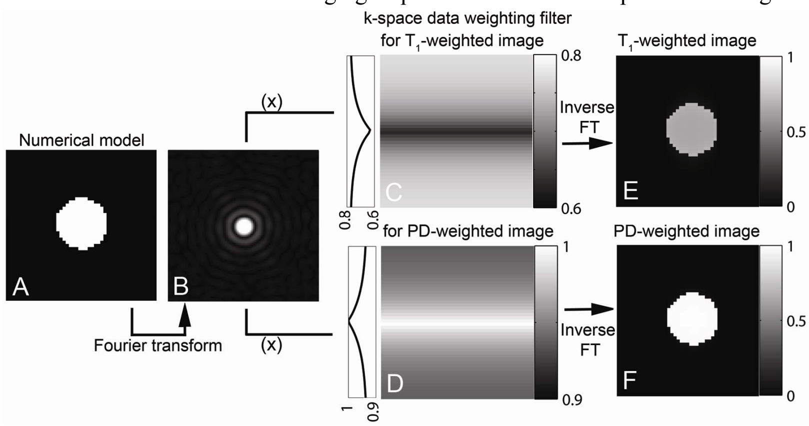

Figure 1. (A) Numerical model of the object geometry. (B) The corresponding k-space representation. The k-space data weighting filters for (C) T_1w and (D) PDw images. Simulated (E) T_1w and (F) PDw images after the inverse Fourier transform.

Results: As seen in Fig. 2, the profiles of the normalized MR signals and the simulated signals of the phantoms showed excellent agreement. For the phantoms 1-3, the T_1 values calculated directly from the single-point method were 26.5ms, 83.5ms and 434.8ms, and the corresponding % errors with the reference T_1 values (33ms, 100ms, 507ms) were 19.5%, 16.5% and 14.7%, respectively. The T_1 values from the fitting procedure, using the simulated model, were 33.3ms, 99.2ms and 507ms and the corresponding % errors were 0.9%, 0.8% and 0.5%, respectively. For in vivo data, a look-up table was calculated as shown in Fig. 3A and was used to correct the observed MR signals (Fig. 3B). The average absolute percent differences of the observed and corrected signals for all subjects was 18.4% ± 5.7%.

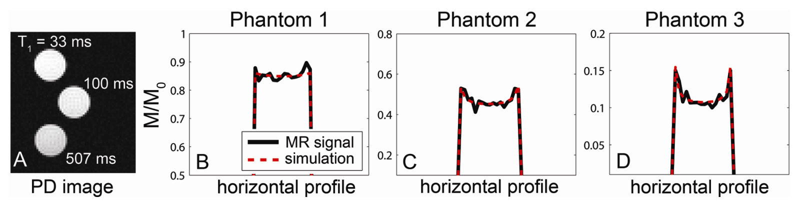

Figure 2. (A) PDw images of three phantoms. (B-D) The horizontal profiles of the normalized MR signals (solid line) and the corresponding simulated signals (dashed lines).

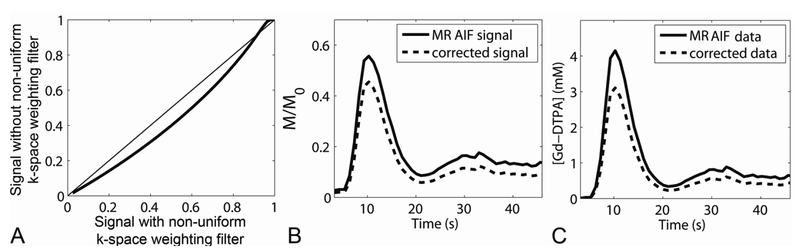

Figure 3. (A) A look-up table showing the signals with and without the non-uniform k-space data weighting filters. The thin line represents the line of identity. Measured and corrected (B) MR signals and (C) the corresponding Gd-DTPA concentrations.

Discussion: This new method allows numerical correction for the effects of non-uniform k-space data weighting on T_1w signals acquired using a SR TurboFLASH imaging sequence with centric k-space ordering. Using a look-up table approach, the observed signals can be corrected easily and rapidly, and the corrected signals can be used as more accurate inputs for tracer kinetics modeling.

References: [1] A. Cernicanu and L. Axel. Acad Radiol. 2006; [2] E. Breton, et al., JMRI. 2011; [3] H. Carr and E. Purcell. Phys Rev. 1954;