Motion Correction of Multiple b-values (MCMB) Diffusion-Weighted Imaging

Y. Mazaheri[1], R. K. Do[2], J. Zhang[2], E. Morris[2], O. Akin[2], and H. Hricak[2]
[1]Medical Physics, Memorial Sloan Kettering Cancer Center, New York, New York, United States, [2]Radiology, Memorial Sloan Kettering Cancer Center

INTRODUCTION

Displacements caused by motion and image distortions caused by eddy currents are major sources of artifacts in diffusion-weighted (DW) MR body imaging. In studies that require collecting images at multiple b-values, such as for incoherent intravoxel motion (IVIM) the problem is exacerbated. We propose an acquisition scheme designed for collecting series of single-shot spin-echo EPI (SS SE-EPI) DW MR images with subsequent co-registration and referred to as motion correction of multiple b-values (MCMB). Non-rigid registration was then performed using a free form deformation (FFD) algorithm based on B-splines using a normalized mutual information-based similarity measure to correct for spatial misalignment during post-processing of DW imaging (DWI) volumes due to rigid and non-rigid bulk motion. The method was used to obtain multi b-value data for modeling IVIM signal from the prostate, breast, and liver of human subjects.

MATERIALS AND METHODS

All studies were performed on a 3.0-T unit (Signa; GE Healthcare, Milwaukee, WI). The MCMB-DWI technique collects multiple acquisitions of 2D slices at multiple b-values using a conventional SS SE-EPI acquisition. The acquisition scheme is illustrated in Fig. 1. The data are acquired during two phases.

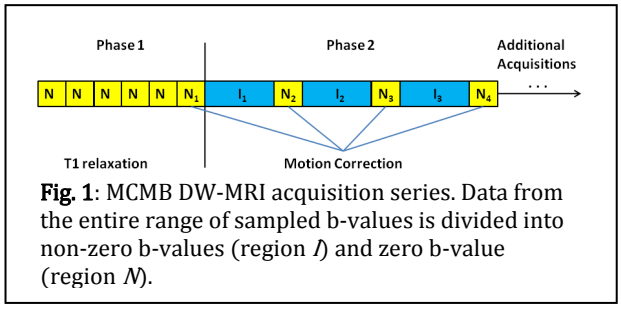

Fig. 1: MCMB DW-MRI acquisition series. Data from the entire range of sampled b-values is divided into non-zero b-values (region *I*) and zero b-value (region *N*).

Phase 1: A series of images are acquired without diffusion weighted gradients (b = 0) to minimize T1-weighting due to incomplete recovery of longitudinal magnetization at shorter TRs. Excitation of data during this period will allow the longitudinal magnetization to reach equilibrium even with short repetition times (TR) minimizing T1-weighting due to incomplete longitudinal magnetization recovery at shorter TRs (1).

Phase 2: Phase 1 is followed by a series of images with non-zero b-values. The distribution of b-values is designed so that high (H) and low (L_1, L_2, L_3, …) b-values are interspersed, to allow nearest-neighbor interpolation of transformation parameters estimated at lower b-values to be used for correction of motion when image signal is significantly attentuated at higher b-value images and the SNR is not sufficient for accurate image registration. Data were acquired assuming isotropic diffusion with the modified SS SE EPI diffusion imaging sequence (GE) with the following parameters: 9 b-values: 0, 29, 44, 66, 99, 148, 222, 333, 500 s/mm^2; TR/TE = 4000/75.7 ms; [FOV]=24×24cm^2; 7-mm thickness; 4 slices; acceleration factor = 2; acquisition time = 2 min.

Displacement Correction Scheme: Deformable co-registration of a two-dimensional (2D) floating image, F, to a reference image, R. was based on the FFD cubic B-splines method described by Rueckert et al (2). Each *N* region from phase 2 was first co-registered to the initial region (N_1) from phase 1. For the non-zero b-value images *I*, especially higher b-value images with reduced SNR, where the contrast differs strongly from that in lower b-value images, the spatial transformations between adjacent *N* regions were estimated by linear interpolation and then applied to the images of interest.

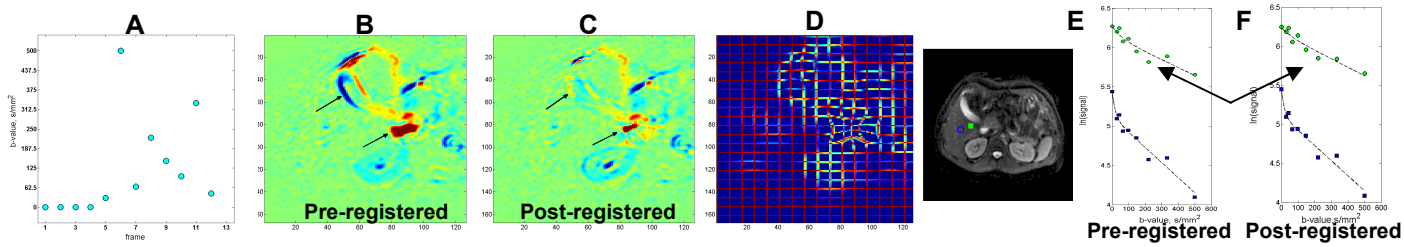

Fig. 2. Liver example image registrations performed using mutual information similarity measures. (**A**) MCMB data acquisition series; b-value vs. frame distribution; (**B**) Pre-registration: frame 4 - reference frame (frame 1) (**C**) Post-registered: frame 4 - reference frame (frame 1) Arrows show residual has decreased substantially due to registration (**D**) Deformed grid (8×8)of control points. In areas where the transformation requires significant change, the deformation is large, whereas in other areas where a lot of deformation is not required the deformation is smaller. Plots of ROI signal intensity from a series of increasingly DW-MR images and bi-exponential fits (parameters: D*, D, and f and χ^2) (**E**) ROI placed on tumor (green) and benign liver (green) prior to registration Green ROI (tumor) (D*= 19.8×10^{-3} mm^2/s, D= 1.1×10^{-3} mm^2/s, f=0.07 , and χ^2=1473.0), Blue ROI (benign liver) ((D*= 59.0×10^{-3} mm^2/s, D= 1.9×10^{-3} mm^2/s, f= 0.28 , and χ^2=756.3)and (**F**) post registration Green ROI (D*= 19.8×10^{-3} mm^2/s, D= 19.8×10^{-3} mm^2/s, f= 0.11 , and χ^2=738.8), Blue ROI ((D*= 19.8×10^{-3} mm^2/s, D= 2.0×10^{-3} mm^2/s, f= 0.28 , and χ^2=1059.3).

RESULTS AND DISCUSSION

Fig. 2 shows the performance of the displacement correction scheme on a human liver. Even small patient-motion during imaging can produce large artifacts and inaccuracy in estimated parameters in multi-b-value DWI signals. Detection and correction of displacement improves data robustness and allows high quality DW images to be obtained on standard clinical scanner hardware. The method presented here extends the work by Rohde et al (3). The method is flexible as it allows for adjusting the spacing of region *N*, motion correction during IVIM imaging, and designing sampling schemes to maximize precision of ADC measurements through acquisition of the optimum number of averages at each b-value.

REFERENCES: [1] Kim DH, et al. Magn Reson Med. 2008;59(1):216-20. [2] Rueckert D,et al. IEEE transactions on medical imaging. 1999;18(8):712-21. [3] Rohde GK, et al. Magn Reson Med. 2004;51(1):103-14.