In vivo Cardiac Diffusion MRI via Second Order Motion Compensated Diffusion Weighted Driven Equilibrium Balanced Steady State Free Precession (SOMOCO-DW-DE bSSFP)

Christopher Nguyen[1,2], Zhaoyang Fan[2], Rohan Dharmakumar[2], Troy LaBounty[3], James Min[3], Daniel Berman[3], and Debiao Li[2,4]

[1]*Biomedical Engineering Interdepartmental Program, University of California, Los Angeles, Los Angeles, CA, United States,* [2]*Departments of Biomedical Sciences and Imaging, Biomedical Imaging Research Institute, Cedars-Sinai Medical Center, Los Angeles, CA, United States,* [3]*Cedars-Sinai Medical Center, Los Angeles, CA, United States,* [4]*Department of Bioengineering, University of California, Los Angeles, Los Angeles, CA, United States*

Introduction: Cardiac diffusion MRI (CDMRI) has the potential to identify acute myocardium ischemia and assess the chronic change of myofiber orientation after a myocardial infraction [1, 2]. Serious technical challenges such as cardiac motion and low SNR limit the application of the technique *in vivo*. Only a few previous *in vivo* studies demonstrate the feasibility of CDMRI in humans but make use of a DW-EPI sequence that either inherently has poor SNR efficiency in the case of STEAM DW encoding [3] or requires the use of a reduced FOV [4]. Both previous techniques employ a first order motion compensation (MOCO) to account for cardiac motion but must limit their diffusion encoding duration (T_{diff}) to less than 30 ms to avoid non-constant velocity motion. This severely hinders the ability to achieve an acceptable b-value (~300s^1mm^{-2}) for CDMRI with the clinical max gradient strength limited to 40 mT/m. For *in vivo* CDMRI, we propose a novel application of DW driven equilibrium balanced steady-state free precession (DW-DE bSSFP) [5] to include second order MOCO (SOMOCO) that not only allows for a longer T_{diff} and acceptable b-values, but also takes advantage of the high SNR of bSSFP to ensure a large FOV.

Materials and Methods: A twice refocused spin echo (TRSE) diffusion weighted preparation (DW-PREP) was used to reduce eddy current artifacts [6] with small crusher gradients straddling the refocusing pulses to suppress stimulated echoes that may form from imperfect refocusing [5] (Fig. 1). Two additional gradients (δ_2) beyond the conventional four gradients used in TRSE were placed between the two middle gradients (δ_1) to allow for SOMOCO. The SOMOCO DW-PREP was designed via simulation coded in Matlab (The Mathworks, Natick, MA) to adjust δ_1 and δ_2 such that m_1 and m_2 are approximately zero while maximizing b-value and minimizing both T_{diff} and TE.

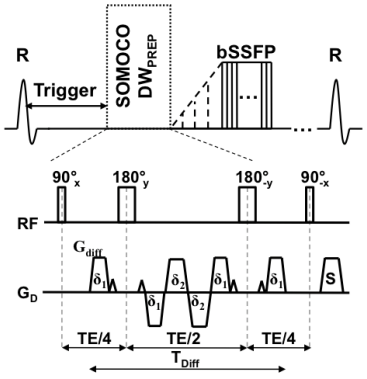
Phantom and *in vivo* volunteer experiments were performed at 1.5T (MAGNETOM Avanto, Siemens) using a 12 channel body matrix coil (TR/TE=233.19ms/1.3ms, FOV=256x256mm^2, 128^2 matrix, 7mm short axis (SA) slice, δ_1=8.5 ms, δ_2=13.75ms, G_{diff}=40mT/m, T_{diff}=73.1ms, b=0, 100, 200, 385 s^1mm^{-2}). Diffusion encoding was along readout direction for all experiments. Phantom experiment of a 10 cm diameter sphere of water compared SOMOCO-DW-DE bSSFP and DW-EPI (TR/TE=1000ms/62ms, FOV=256x256mm^2, 128^2 matrix, 7mm slice, 16 NEX, GRAPPA R = 2) to test the accuracy of derived apparent diffusion coefficient (ADC) value. Four healthy volunteers were imaged using the parameters above during the quiescent cardiac phase of diastole for a single breath-hold with parallel imaging (GRAPPA R=2) and only acquiring b=0, 100, 385 s^1mm^{-2} for a total scan time of 9 seconds. ADC maps were calculated offline assuming a monoexponential fit in Matlab segmenting out lungs and areas outside of the patient with the b=0 s^1mm^{-2} DW image for clarity. Manual segmentation of the left ventricle (LV) was used to calculate the mean and standard deviation (SD) of the ADC values for each volunteer. SNR was calculated from the b=385 s^1mm^{-2} DW image. Noise levels were determined by averaging SD's of 6 ROIs in the air region outside the chest.

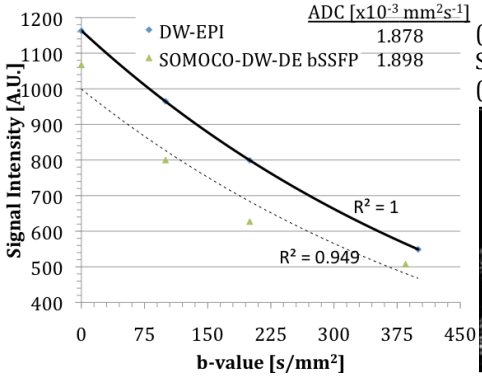
Results: The phantom experiment showed great agreement (~1% difference) of calculated ADC values between SOMOCO-DW-DE bSSFP ($1.898x10^3 mm^2s^{-1}$) and DW-EPI ($1.877x10^3$ mm^2s^{-1}) (Fig. 2). Table 1 shows the measured ADC values in the LV myocardium for the four volunteers with values ranging 1 to 3 $x10^3$ mm^2s^{-1} and the SNR of the b=385 DW image with values ranging 5 to 7. Fig. 3 shows the b=0 s^1mm^{-2} image and ADC map for volunteer 2.

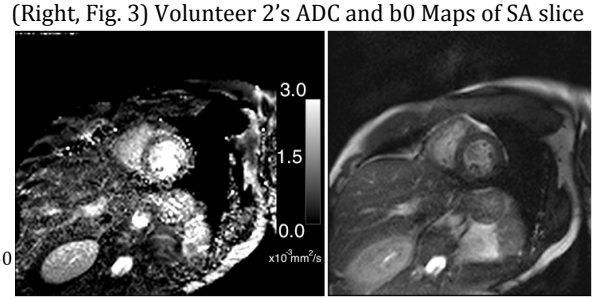
Conclusion: We have shown the feasibility of DW-DE bSSFP in the application of *in vivo* cardiac diffusion within clinically human limited MR parameters (i.e. maximum gradient strength) by employing a SOMOCO DW-PREP. ADC values acquired from the four volunteers are consistent with prior *in vivo* human cardiac diffusion studies [1,3,4]. Further optimization of SOMOCO-DW-DE bSSFP is needed to increase SNR, which include shortening the TE of the DW-PREP and using a more robust chest array coil.

Table 1 – ADC values and SNR of LV Myocardium of 4 Volunteers

LV Myocardium	Volunteer 1	Volunteer 2	Volunteer 3	Volunteer 4
ADC [$10^{-3}mm^2s^{-1}$]	2.12 ± 0.979	2.01 ± 0.958	2.09 ± 1.05	1.92 ± 0.913
SNR b = 385 DWI	5.73	6.33	5.46	7.63







(Left, Fig. 1) Pulse sequence diagram. (Mid, Fig. 2) Phantom Signal vs b-value curve of SOMOCO-DW-DE bSSFP vs DW-EPI (Right, Fig. 3) Volunteer 2's ADC and b0 Maps of SA slice

[1] Wu, et al. Circ 114:10 (2006). [2] Hsu, et al. Am J Heart Physio 275:697 (1998). [3] Dou, et al. MRM 48:105 (2002). [4] Gamper, et al. MRM 57:331. (2007). [5] Jeong, et al. MRM 50:821 (2003). [6] Reese, et al. MRM 49:177 (2002).