Multi-shot Diffusion-Weighted Split-Echo PROPELLER MRI of the Abdomen

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Introduction

Diffusion-weighted MRI (DWI) provides a non-invasive method for in vivo evaluation of tissue water mobility. Most DWI studies employ single-shot DW-EPI techniques which can suffer severe artifacts and image distortion in abdominal imaging applications. Multi-shot TSE-based DW-PROPELLER [1] may overcome these limitations. DW-PROPELLER for brain [2] and abdominal [3] imaging has demonstrated improved image quality and the potential for high resolution DWI. Despite of these improvements, multi-shot DW TSE acquisition in the abdomen is particularly challenging due to artifacts resulting from violation of Carr-Purcell-Meiboom-Gill (CPMG) conditions. Incoherent signal phase due to motion during DW preparation can lead to destructive interference between spin echo and stimulated echo signals and consequent rapid signal loss. The SPLICE (split-echo acquisition of fast spin echo signals) technique may avoid this problem by separating the two signal components [4,5]. The purpose of our study was to investigate the feasibility of combining SPLICE and DW-PROPELLER techniques (DW-SP-PROPELLER) for abdominal DWI. In phantom and normal volunteer studies we demonstrate that DW-SP-PROPELLER can mitigate the non-CPMG artifacts sometimes present when using conventional DW-PROPELLER.

Methods

PROPELLER These techniques use a multi-shot TSE acquisition strategy with each segment of data acquired as a single rectilinear blade along a propeller-shaped *k*-space trajectory (**Fig. 1**). The concentric *k*-space region of each blade permits phase correction before segment combination and image reconstruction. Our implemented pulse sequence was based upon the BLADE sequence (Siemens implementation of PROPELLER TSE).
SPLICE Two families of echoes (E1 and E2) with destructive interaction can be separated with the first spin echo refocused asymmetrically with regard to symmetric TSE convention. The SPLICE technique uses an unbalanced readout gradient to acquire the two signal components separately by extending the original readout gradient. For our DW-SP-PROPELLER sequence (**Fig. 1**) two separate PROPELLER k-space datasets were phase corrected and reconstructed. The two magnitude images were summed for final image reconstruction.
MRI All imaging experiments were performed using a 1.5 T clinical scanner (Magnetom Sonata, Siemens Medical Solutions). DW-PROPELLER and DW-SP-PROPELLER images were acquired in phantom models and three normal volunteers. For phantom studies two cylindrical vials (distilled water and ethanol) were imaged using a single-channel head coil. We compared both relative SNR and ADC accuracy for both sequences. Volunteer studies were performed at a single axial abdominal slice position using a flexible anterior 6-channel phased-array abdominal coil and a posterior spinal array coil. Common imaging parameters for DW-PROPELLER and DW-SP-PROPELLER: TR/TE = 3000/90ms, 5mm slice thickness, 400 Hz/pixel BW, 400x400 mm^2 FOV, 128 matrix (3.0x3.0 mm^2), 3 signal averages, ETL = 17, 59 segments, free-breathing data acquisition with respiratory bellow triggering. Motion-probing gradients separated by a slice-selective 180° refocusing pulse provided the requisite diffusion-weighting. DW-PROPELLER: echo spacing = 6.9ms. DW-SP-PROPELLER: echo-spacing = 9.4ms and τ (interval between each E1 and E2 pair) = 2.6 ms.

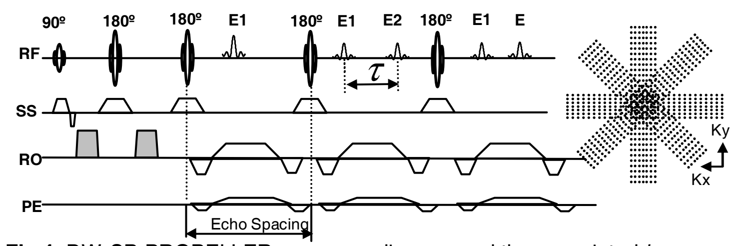

Fig 1. DW-SP-PROPELLER sequence diagram and the associated *k*-space pattern. Two families of echoes (E1 and E2) are separately acquired by extending the readout gradients. Diffusion gradients (shaded area) are applied along the readout encoding direction.

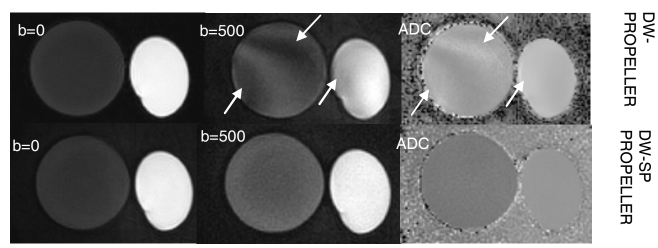

Fig 2. DWIs and ADC maps of Water (W) and Ethanol (E) phantom using DW-PROPELLER (upper) and DW-SP-PROPELLER (lower). Notice non-CPMG artifacts (arrows) in DW-PROPELLER images not present in DW-SP-PROPELLER images.

Results

As shown in **Table 1**, for each phantom specimen, SNR for DW-SP-PROPELLER decreased by approximately $\sqrt{2}$ as expected compared to DW-PROPELLER at b=0 s/mm^2, while SNR for DW-SP-PROPELLER increased compared to DW-PROPELLER at b=500 s/mm^2, mostly likely due to the suppression of non-CPMG artifacts on DW-SP-PROPELLER. Mean ADC values measured by DW-SP-PROPELLER were more consistent with literature values ($ADC_{water} \approx 2.5\times10^{-3} mm^2/s$ and $ADC_{ethanol} \approx 1.0\times10^{-3} mm^2/s$). DW-SP-PROPELLER images and ADC maps were more homogenous (i.e. lower SD of ADC) than those of DW-PROPELLER (**Fig. 2**). DW images and reconstructed ADC maps acquired using DW-PROPELLER and DW-SP-PROPELLER sequences are shown for a representative normal volunteer in **Fig. 3**. The unstable breathing pattern of human subjects can lead to severe artifacts obscuring abdominal organs. These signal-drop banding artifacts due to non-CPMG conditions were well mitigated by using DW-SP-PROPELLER.

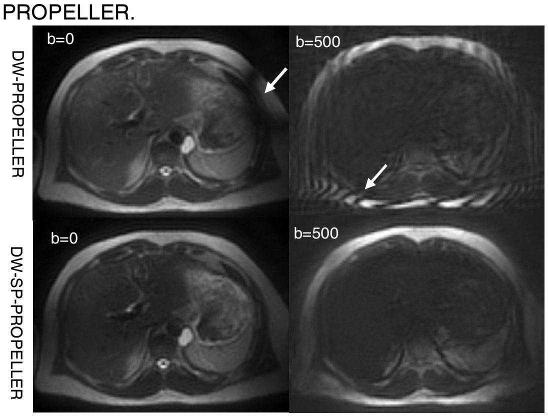

Fig 3. DW-PROPELLER (upper) and DW-SP-PROPELLER (lower) images of a normal volunteer. Notice non-CPMG artifacts (arrows) in DW-PROPELLER images not present in DW-SP-PROPELLER images.

Conclusion

The SPLICE technique was highly effective for overcoming non-CPMG artifacts for abdominal DW-PROPELLER MRI. DW-SP-PROPELLER offers the potential for robust high-resolution DWI particularly beneficial for abdominal oncologic imaging applications.

Table 1.		Water	Ethanol
SNR DWI (b = 0 s/mm^2)	PROPELLER	183	26
	SP_PROPELLER	128	20
SNR DWI (b = 500 s/mm^2)	PROPELLER	50	12
	SP_PROPELLER	53	16
ADC ($\times10^{-3} mm^2/s$) Mean ± SD	PROPELLER	2.8±0.32	1.7±0.61
	SP_PROPELLER	2.4±0.05	1.0±0.16

Reference: [1] Pipe et al. MRM. 2002;47(1):42-52. [2] Forbes et al. Radiol. 2002;225 :551-5 [3] Deng et al. Invest Radiol.2006;41(10):769-75 [4] Norris et al. MRM.1992;27:142-64 [5] Schick et al. MRM. 1997:38:638-44