Diffusion-weighted MR Imaging of Kidneys Using Targeted-SPLICE

N. Jin[1], J. Deng[2], and A. C. Larson[1,3]
[1]Departments of Radiology and Biomedical Engineering, Northwestern University, Chicago, IL, United States, [2]Children's Memorial Hospital, Chicago, IL, United States, [3]Robert H. Lurie Comprehensive Cancer Center, Chicago, IL, United States

INTRODUCTION

Diffusion-weighted (DW) MRI can provide information about in vivo water diffusion rates and microcirculation; these measurements can differentiate normal tissues from aberrant tissues at various disease states. DWI may be particularly useful for functional interrogation of the kidney given the high blood flow rates through this organ (1,2). Changes in measured diffusion rates should reflect alterations in water mobility expected in accompany various renal disease states. Single-shot DW spin-echo echo-planner imaging (DW-SE-EPI) is commonly used for DW acquisition; however, this technique can experience image distortion and chemical shift artifacts (3). Alternative HASTE split acquisition of fast spin-echo signals for diffusion imaging (SPLICE) technique (4) can effectively reduce image distortion; but SPLICE still suffers from image blurring caused by T2-filtering effects due to the long echo-train length. We recently developed targeted-SPLICE technique by combining the inner volume imaging (IVI) technique (5,6) with SPLICE for DWI without image distortion and blurring. The purpose of our study is to apply targeted-SPLICE technique for DWI in the kidneys and compare these targeted-SPLICE diffusion measurements to conventional DW-SE-EPI measurements.

METHOD

Targeted-SPLICE Sequence A DW-HASTE sequence was modified to produce targeted-SPLICE sequence depicted in **Fig. 1**. Diffusion sensitivity was obtained by utilizing velocity-compensated diffusion-encoding gradients around the first two non slice-selective 180° RF pulses. Diffusion-encoding gradients were applied along one of the principal gradient directions. A HASTE technique is used for rapid acquisition. To solve the CPMG problem, the SPLICE technique was applied by prolonging the duration of the original readout gradients to completely separate the echoes E1 and E2 (4, **Fig. 1**). IVI technique is used for the targeted reduced FOV acquisition. Slice-selective gradients for spatially selective RF excitation were applied along the phase encoding (PE) direction for the 90° RF pulse and along slice-selection direction for the 180° refocusing RF pulses. The signal is only generated within the overlapping region of the excitation and refocusing volumes, this allows for a reduction of the FOV along PE direction, hence reducing the PE steps and shortening the echo train length.

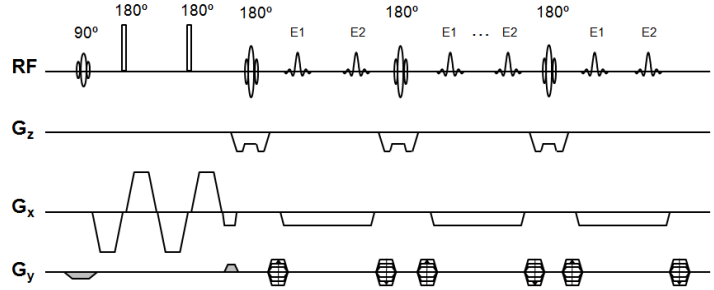

Fig. 1. Pulse sequence diagram of targeted-SPLICE sequence. Readout gradients are prolonged to completely separate echoes **E1** and **E2**. A slice-selective 90° RF excitation is performed in the PE direction

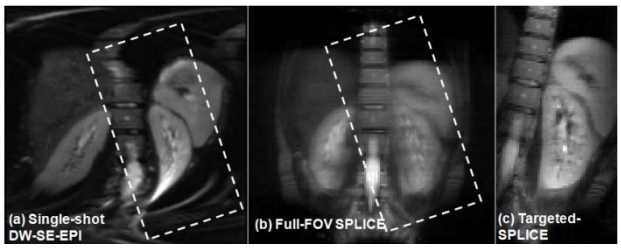

Fig. 2. DWI at b = 0sec/mm^2, acquired using single-shot DW-SE-EPI (**a**), full-FOV SPLICE (**b**) and targeted-SPLICE (**c**). Targeted-SPLICE effectively removed the geometric distortion in (**a**) and image blurring in (**b**).

MRI Experiments were performed in seven healthy volunteers using 1.5 T Magnetom Espree scanner (Siemens Medical Solutions) with a body matrix coil and spinal array for signal reception. Axial and coronal TSE images were acquired for localization. First, a coronal single-shot DW-SE-EPI acquisition with slice encompassing both kidneys was performed with the following parameters: TR/TE, 3500/71 ms; slice thickness, 5 mm; spatial resolution, 2.34^2~3.14^2 mm^2; nine signal averages; bandwidth, 1400 Hz/pixel and partial Fourier factor, 6/8. 6 b-values were used: 0, 50, 100, 200, 300, and 500 sec/mm^2. Next, reduced-FOV DWI of the left and right kidneys were acquired separately in seven and four volunteers using targeted-SPLICE sequence with the same spatial resolution as single-shot DW-SE-EPI. The following parameters are used: TR/TE, 3500/61 ms; slice thickness, 5 mm; nine signal averages; bandwidth 640 Hz/pixel; 46.9% PE-FOV; 11 b-values: 0, 20, 40, 60, 100, 150, 200, 250, 300, 400 and 500 sec/mm^2. Full-FOV SPLICE images were also acquired at the same position with the similar parameters for comparison. The diffusion gradients were applied in three orthogonal directions and subsequently averaged to minimize the effects of diffusion anisotropy. Respiratory triggering was used for all acquisitions.

Data Analysis For both single-shot DW-SE-EPI and targeted-SPLICE methods, diffusion parameters were calculated on a voxel-by-voxel basis using Matlab software (The Math Works Inc., Natick, MA). First, an apparent diffusion coefficient (ADC) map was calculated by employing the non-linear Levenberg-Marquardt algorithm to fit the mono-exponential function $S(b_i) = S_0 \cdot \exp(-ADC \times b_i)$. Next, in order to separate the contributions from pure diffusion and microcirculation, a bi-exponential fitting was performed according to the equation: $S(b_i) = S_0 \cdot [(1-f) \cdot \exp(-D \cdot b_i) + f \cdot \exp(-D^* \cdot b_i)]$ where f is the perfusion fraction, D is the diffusion parameter representing the pure diffusion and D* is the diffusion parameter representing microcirculation within the voxel. Mean ADC, f and D were measured in the cortex and medulla.

Statistics Analysis All statistics were performed using SPSS (SPSS, Chicago, IL, USA). Pair-wised student t-tests were used to compare the mean diffusion parameters (ADC, f and D) of the kidney cortex and medulla calculated using DW-SE-EPI and targeted-SPLICE methods.

RESULTS

Fig. 2 shows DWI (b = 0sec/mm^2) with significant field inhomogeneity (due to poor shimming) acquired using single-shot DW-SE-EPI (**a**), full-FOV SPLICE (**b**) and targeted-SPLICE (**c**) respectively. Single-shot DW-SE-EPI images were severely distorted. Although full-FOV SPLICE method reduced distortion artifacts, images were blurred in the phase-encoding direction due to the T2 decay. Overall, no image distortion or blurring artifacts were observed in the targeted-SPLICE images, which provided better anatomical details. **Table 2** shows the diffusion parameters of the kidney cortex and medulla measured using single-shot DW-SE-EPI and targeted-SPLCE. There were no significant differences in the mean ADC, f and D of the kidney cortex and medulla calculated by two methods. ADC, f and D were significantly higher in the cortex than in the medulla (**Table 2, Fig. 3**).

CONCLUSION

Targeted-SPLICE effectively reduced geometric distortion and image blurring and produced accurate diffusion parameter measurements in the kidney; target-SPLICE is a promising method for abdominal DWI.

Table 2. Diffusion Parameters

	ADC (×10^{-3}mm^2/s)		f		D (×10^{-3}mm^2/s)	
	Cortex	Medulla	Cortex	Medulla	Cortex	Medulla
Single-shot DW-SE-EPI	2.88 ± 0.25	2.72 ± 0.17	0.18 ± 0.04	0.17 ± 0.03	2.30 ± 0.15	2.25 ± 0.10
Targeted-SPLICE	2.90 ± 0.24	2.68 ± 0.17	0.20 ± 0.04	0.18 ± 0.03	2.23 ± 0.22	2.16 ± 0.23

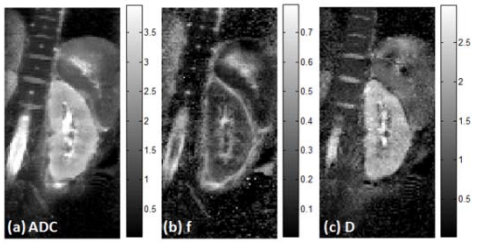

Fig. 3. Diffusion parameter maps generated from targeted-SPLICE images: (a) ADC, (b) perfusion fraction f, (c) pure diffusion D.

References: (1) Thoeny et al. *Radiology*,2005;235:911-917. (2) Thoeny et al. *Radiology*,*2006;*241:812-821.(3) Farzaneh et al. *MRM*. 1990;14:123–139. (4) Schick et al. *MRM*, 1997;38:638-644. (5) Feinberg et al. *Radiology*, 1985;156:743–747. (6) Saritas et al. *MRM*, 2008;60:468-473