ARTERIAL SPIN LABELING PERFUSION STUDIES OF THE PROSTATE WITH AN ERC

X. Li1, C. Kalavagunta1, M. T. Nelson2, and G. J. Metzger1

1Center for Magnetic Resonance Research, University of Minnesota, Minneapolis, MN, United States; 2Diagnostic Radiology, University of Minnesota, Minneapolis, MN, United States

Introduction: Prostate perfusion using arterial spin labeling (ASL) can potentially benefit patients with renal dysfunction or MR contrast-agent allergies in the diagnosis, prognosis and therapeutic monitoring of prostate cancer. However, practical challenges facing the clinical applicability of this technique in the prostate include: (a) low signal noise ratio (SNR), (b) intrinsically low blood flow (PBF) and (c) the signal loss or distortion due to the susceptibility around air-tissue interface between the prostate and rectum. The later is particularly troublesome as echo planar imaging (EPI) is still the fastest/typical method for reading out the multi-slice ASL data. Some of these challenges can be overcome by using an endorectal coil (ERC) which has significantly higher SNR than surface arrays, decreases motion of the prostate and, when inflated with susceptibility matching fluids, reduces B0 induced distortions (1,2). This paper, reports on initial results obtaining quantitative ASL in the prostate at 3T with an ERC.

Materials and Methods: All subjects provided written informed consent to participate in an IRB approved protocol. A 3T Siemens TIM Trio whole body scanner was used for all imaging with a whole body transmit coil. On receive, a standard surface array configuration was used with a balloon-type endorectal coil (BPX-30, Medrad, Pittsburgh, PA) inflated with a perfluorocarbon was used. Quantitative ASL was performed using FAIR (3) with Q2TIPS (4). Q2TIPS was performed 10 mm above the superior side of the imaging section, as the readout. Imaging slices were positioned to ensure that the most superior slice of the prostate slice began at the base of the prostate.

Perfusion studies using multiple post-bolus delay times were performed to evaluate how perfusion signal changes with delay time: TR/TE = 3000/9 ms, FOV = 220 x 200 mm2, matrix size = 64 x 64, in-plane resolution = 3.44 x 3.44 mm2, slice thickness/gap = 5/1 mm, number of slices = 5, phase oversampling = 10%, acquisition order = descending, GRAPPA iPAT factor = 2 with 24 reference lines, partial Fourier = 7/8, shimming mode = advanced 3D shimming over whole slab, number of measurements = 260, selective/spatially-confined inversion slab = 50/230 mm, labeling time (TI1)/post-bolus delays (TI2) = 0.7/(0.4, 0.8 1.2, 1.6, 2.0) s. The order of post-bolus delay times were randomized for each subject. For scans with delay time equal to 0.8 s, 300 EPI volumes were acquired, but 260 were used in the analysis of perfusion signal change with respect to delay time. High-resolution perfusion imaging used the same parameters with the exception of the following: TE = 16 ms, FOV = 256 x 256 mm2, matrix size = 128 x 128, in-plane resolution = 2 x 2 mm2, partial Fourier = 6/8, number of measurements = 300, post-bolus delay = 0.8 s. For both low- and high-resolution perfusion scans, the superior saturation pulse train lasted during the entire delay period by using 20 mm saturation slabs with a 25 ms pulse interval, and proton density (PD) images (M0) were acquired with the same EPI imaging parameters but a long TR (8 s). T2-weighted turbo spin echo images were obtained as an anatomic reference. Motion correction and co-registration were performed with SPM (Functional Imaging Laboratory, University College London).

Perfusion-weighted (PW) images and signals were all expressed as a percentage change with respect to M0. A single blood compartment model (5) was used for prostate blood flow (PBF) quantification (equation 1). Measurement errors due to either random noise (or white noise), estimated as the standard deviation of the background signals of EPI images, or temporal noise (or physiological noise), estimated as the standard deviation of time series of measured mean perfusion signals in the prostate (equations 2 and 3, respectively), were evaluated.

\[ \text{PBF} = \frac{\Delta M}{12} \times \frac{\text{TI1} \times \text{EXP}^{-\text{TI1}/T_{1b}}}{2} \]

where \( N_{\text{voxel}} \) represents the total number of averaged voxels, \( N_{\text{vol}} \) the number of averaged perfusion weighted images, \( T_{1b} \), 1 of the blood, is assumed 1660 ms.

\[ E_{\text{random}} = \frac{\sqrt{N_{\text{vol}}}}{\sqrt{N_{\text{vol}}}} \times \frac{\Sigma_{\text{random}}}{\sqrt{2/\Sigma_{\text{random}}}} \]

Conclusion: High quality high-resolution ASL perfusion imaging maps can be obtained by using an ERC. The inhomogeneity on receive across the prostate (especially along A-P direction) as a result of the sensitivity profile of the ERC are automatically corrected in the calculated PW maps (Figures 1 and 2). Temporal error is the major source of error for prostate perfusion imaging (Figure 3). Measurement error due to random noise can be easily reduced to a very low level (less than 2% of PBF) for both low- and high-resolution perfusion imaging. The temporal error tends to increase with imaging resolution, but can be reduced to less than 10% for high-resolution prostate perfusion imaging by using more than 120 averages for a typical subject (Figure 3). Future studies will provide a more comprehensive and quantitative analysis of perfusion parameters measured in both healthy individuals and prostate cancer patients to investigate if these methods could augment or replace contrast based vascular measurements.

Acknowledgements: This study was supported by BTRC P41-RRO080879.