A Comparative Study of Arterial Spin Labeling and Dynamic Contrast Enhanced Perfusion Magnetic Resonance Imaging in the Kidneys

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Introduction

Perfusion, the blood flow in microvasculature, is an important index of physiology and pathophysiology. In the kidneys, perfusion has been used to evaluate chronic renal allograft nephropathy (1) and renal cell carcinoma (2). Arterial spin labeling (ASL) and dynamic contrast enhanced (DCE) MRI are well-developed techniques for quantitative perfusion measurement. Perfusion MRI provides advantages of minimal or none invasiveness as compared to nuclear medicine, and relaxes dependence on variable anatomy of vasculature than flow measurement based on phase-contrast MRI. While 3T MRI has become more common over the past few years for the gain in signal-to-noise ratio (SNR), most renal perfusion studies had been performed at the field strength of 1.5T (1-3). Specifically, the longer T1 at 3T benefits ASL. To the best of our knowledge, head-to-head cross-validation between ASL and DCE remains absent for 3T. In the present study, we examined the compatibility of ASL and DCE in measuring renal blood flow (RBF) at 3T by performing both techniques on a cohort of healthy volunteers and one patient with renal artery stenosis.

Materials and Methods

This study was IRB-approved and conducted in compliance with the ethical guidelines. Eight healthy volunteers (2 women, 6 men, 25-68 y/o) and one male patient (34 y/o) diagnosed with renal artery stenosis were recruited and all gave informed written consent before partaking in this study. MR imaging was performed on a 3T scanner (Tim Trio, Siemens, Erlangen, Germany) using the spine matrix coil and a flexible torso matrix coil for signal reception, and the body coil for RF transmission. After scout and T2-weighted turbo spin-echo anatomic imaging, ASL images were collected using the pseudocontinuous labeling scheme (4) (labeling duration = 1.5 s, postlabeling delay = 1.2 s) and gradient-echo echo-planar readout (slice thickness = 6mm, 5 coronal slices, field-of-view = 30 cm, matrix = 128x128, GRAPPA acceleration factor = 2, TR = 3 s, TE = 17 ms). The labeling plane was axial and 70-80 mm above the center of the imaging volume such that it was approximately perpendicular to the abdominal aorta. Ten runs of ASL were conducted and each comprised 6 measurements (the first 2 were discarded) and the duration of breath holding was within 20 s. For DCE imaging, 0.0125 mmol/kg of Gd-DTPA was injected at a rate of 4 ml/s using a power injector followed by a 15 ml saline flush while 0.025 mmol/kg was used on two of the subjects to test the effect of T1 saturation. Saturation-recovery TurboFLASH was used for image acquisition (TR = 200 ms, TE = 0.98 ms, matrix = 144x192, slice thickness = 8 mm, 3 coronal slices plus 1 axial slice). A total of 80 measurements were collected while contrast agent was injected after 5 measurements. All images were corrected for bulk motion and coil sensitivity. The tag and control images of ASL scans were pair-wise subtracted and averaged to produce perfusion-weighted images which were then converted to RBF maps using the abdominal aorta as an internal reference. Labeling efficiency was assumed to be 0.75. With DCE data, RBF was computed by $[C_t(t) \text{ deconv. } C_a(t)]_{t=0}$, where $C_t(t)$ and $C_a(t)$ were the concentration-time curves of tissue and the abdominal aorta, respectively, and $C(t) \propto S(t) - S_0(t)$, where S_0 was the signal before the arrival of contrast agent. Block-circulant deconvolution (5) was adopted for its insensitivity to the arrival timing of bolus. Bilateral kidneys in the center slice were segmented into cortex and medulla based on the anatomic images from which average RBF was calculated. Data from two volunteers were excluded due to excessive motion and failure in breath holding.

Results

Fig 1 is the scatter plot of RPF values obtained with ASL and DCE. The values derived from DCE were apparently overestimated on the two subjects injected with 0.025 mmol/kg Gd-DTPA (hollow symbols) due to signal saturation in the aorta, and were excluded from statistical analysis. For five subjects with the dose of 0.0125 mmol/kg Gd-DTPA, a liner correlation was found between ASL and DCE measurements ($r = 0.9$, $p < 0.005$). Both ASL and DCE were able to distinguish the reduced RBF in the patient with renal artery stenosis (see arrows). Mean RBF is summarized in Table 1.

Fig 1.

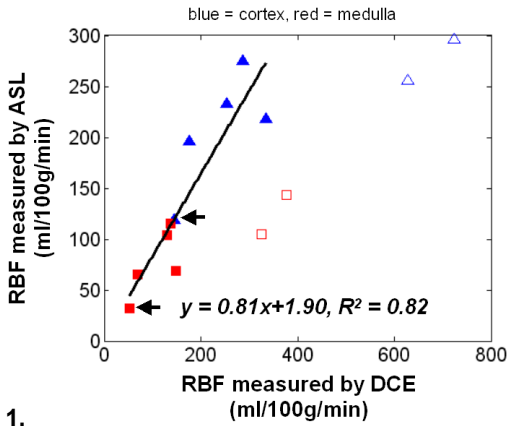

Table 1.

RBF (ml/100g/min) N = 5 Mean±SD	ASL		DCE	
	Cortex	Medulla	Cortex	Medulla
	227±59	91±37	239±78	108±44

Discussion

The concentration of contrast media used in DCE (usually 0.02 mmol/kg at 1.5T) needs to be reduced to avoid signal saturation at 3T. For PCASL, the single-compartment model of tissue was adopted as opposed to the blood pool usually used in the brain because renal capillaries are more permeable to water than cerebral capillaries with blood-brain barrier. The air in abdomen is very likely to perturb local field and thus degrade labeling efficiency even though the subject holds breath during the scan. Calibration of labeling efficiency will be necessary to further improve PCASL quantification. We conclude that ASL and DCE provide compatible measurement of renal perfusion at 3T.

References

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