Feasibility of regional renal blood flow and vascular volume fraction measurement with cardiac-output corrected MR renography

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INTRODUCTION

Regional blood perfusion, as an important parameter of renal function, reflects the efficiency of oxygen delivery to renal tissue and of glomerular filtration. A perfusion map is needed to document hypo-perfusion in a local region of a kidney. One important clinical application is assessing the significance of a segmental branch renal artery stenosis. MRI has provided multiple options for quantifying perfusion. MR renography using ultra-low gadolinium dose has been validated by many groups, mainly for estimating glomerular filtration rate (GFR) (1, 2). Images of high temporal resolution (1-3 sec per frame) are usually acquired for the initial 30-60 secs to record the vascular phase. Artifacts in arterial input function (AIF) are very common and lead to errors in perfusion quantification. These artifacts can be corrected given the value of patient's cardiac output (CO) (3), resulting in markedly improved precision of GFR.

In this study we performed a voxel-by-voxel analysis for the vascular phase of low-dose Gd-enhanced MR renography data and assessed the impact of CO-corrected AIF on the precision of renal perfusion measurements.

MATERIALS AND METHODS

After written informed consent, five human subjects (2 males and 3 females, age 45 ± 13 yrs) were enrolled in this study. At a 1.5 T system (Avanto, Siemens), 2D prospective ECG-gated phase-contrast velocity images were acquired for CO measurement: perpendicular to the ascending aorta 20 mm above the aortic valve; TR/TE/FA 48.9 ms/3.09 ms/30°, slice thickness 6 mm, field of view 270 × 320 mm, matrix 216 × 256, 20 frames per cardiac cycle, acquisition time ~20 sec, one breath hold. Heart rate was recorded continuously. CO was obtained by multiplying the stroke volume (the area under flow vs time curve) by the heart rate. For MR renography, abdominal images covering aorta and both kidneys was performed using coronal 3D FLASH (TR/TE/flip angle=2.84ms/1.05ms/12°, FOV 425×425 mm^2, voxel 1.7×1.7×2.5 mm^3, acquisition time 3s). Before contrast injection, five acquisitions were done in a single breath hold, to establish baseline signal. A 4 ml bolus of Gd-DTPA was injected, followed by 20 ml saline flush both at 2 ml/s. Eight seconds following the start of Gd-DTPA injection, 10 3D acquisitions were repeated continuously for 30 s, during which the subject suspended respiration as long as possible. After a break of 18 sec, 12 additional volumes were acquired during separate 3 s breath-holds spaced over 5 min.

AIF was sampled at 3 locations along abdominal aorta (above renal artery (A1), at the level of renal artery (A2), and below renal artery (A3)), representing different degrees of inflow artifact. AIF correction was applied to each level yielding cA1, cA2, and cA3 (3). For each voxel signal intensity curves were converted to gadolinium concentration (4), and concentration vs. time curve were modeled as the convolution of AIF with the impulse retention function:

$$R(t) = \begin{cases} 0 & t < t_0 \\ F & t_0 \le t < t_0 + MTT, \\ E \cdot e^{-k \cdot t} & t \ge t_0 + MTT \end{cases}$$

where F is perfusion, t_0 is bolus arrival time and MTT is the mean transit time of tracer in the voxel. The exponential term k accounts for the tracer in tubular and interstitial space. The product F·MTT represents the vascular fraction (V) of the voxel. Fitting was based on Levenberg-Marquardt method and restricted to data acquired in the first 30 sec. Separate mappings were performed with six AIF variants: A1, A2, A3, cA1, cA2, cA3. Cortical and medullary regions of interest (ROIs) were drawn on the MR image with maximal cortico-medullary contrast and all parametric maps were averaged for each ROI. For each kidney region and each parameter, standard deviation (STD) was first calculated across different AIF (A1, A2, A3) in order to assess the effect of AIF artifact. We then calculated the mean value and the standard deviation of STD across all kidneys. The analysis was repeated for the parameter estimated using the CO-corrected AIFs (cA1, cA2, cA3).

RESULTS AND DISCUSSION

Figure 1 shows representative parameter maps and the residual error. The fitting error was 15% - 20% in the cortex. Error was larger in the medulla, due to much lower vascularity (thus lower SNR). The large noise in MTT is most likely caused by inadequate temporal resolution (3-sec). Parameter values across all subjects are shown in Table 1. Cortical perfusion averaged 2.3 ml/min/ml, medullary perfusion was 0.4 ml/s/ml; vascular volume fraction for cortex was 22%, and for medulla 4% (Table 1, row 1), in agreement with literature volumes.

Table 1: Renal vascular parameter values, and their variances due to AIF artifacts.

	F_{Cx} (ml/min/ml)	V_{Cx}	MTT_{Cx} (s)	F_{Med} (ml/min/ml)	V_{Med}	MTT_{Med} (s)
Value from cA2	2.3±0.5	0.22±0.05	5.9±0.5	0.4±0.2	0.04±0.02	6.0±0.4
STD w/o AIF correction	4.9±1.3	0.56±0.16	0.6±0.2	1.0±0.5	0.12±0.06	0.7±0.2
STD with AIF correction	0.3±0.2	0.02±0.03	0.4±0.2	0.07±0.04	0.01±0.01	0.2±0.1

The variability of perfusion was remarkably reduced with the use of CO-based AIF correction (Table 1, rows 2 and 3). After CO-correction, the variability of cortical perfusion and vascular volume fraction was ~10%-15%, and for medulla, ~20%. In summary, the proposed model combined with AIF correction allows a clinically useful estimate of regional renal perfusion and vascular volume fraction.

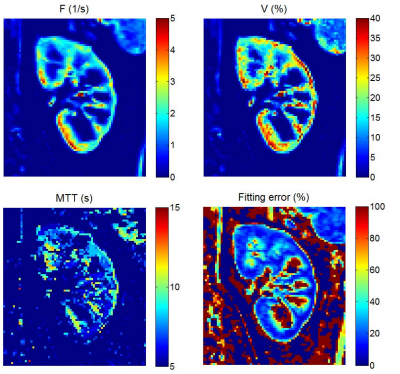

Figure 1: Example of parameter maps: perfusion (F), vascular fraction (V), MTT, and fitting error

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