Original Research

Rapid Measurement of Renal Artery Blood Flow With Ungated Spiral Phase-Contrast MRI

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Purpose: To verify the potential of ungated spiral phase-contrast (USPC), which has been shown to provide accurate and reproducible time-averaged measurements of pulsatile flow, for rapid measurement of renal artery blood flow (RABF) in vivo.

Materials and Methods: The RABF rates of 11 normal human subjects and one patient with renal failure were measured with USPC within six seconds.

Results: Rapid USPC scans produced reproducible RABF measurements (SD ≤ 9%) that agreed with the normal RABF rates known from the literature. The RABF rates of the patient with renal failure were substantially less (<50–65%) than the normal RABF rates.

Conclusion: The results demonstrate that it is now possible to obtain rapid and consistent RABF measurements within six seconds with USPC.

Key Words: renal; vascular; flow; fast imaging; phase-contrast; spiral
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ACCURATE MEASUREMENT of renal artery blood flow (RABF) are clinically important for evaluating acute and chronic renal failure (1–5), and RABF has also been implicated in acute allograft rejection (3). However, it is difficult to assess RABF in vivo because of the small calibers and frequently variable anatomy of the renal arteries, and currently there is no gold standard available for RABF measurement. While para-aminohippuric acid clearance is frequently used as a surrogate for RABF in clinical settings, it requires the use of difficult procedures (4) and is well known to be inaccurate. Positron emission tomography (PET) has been demonstrated to be a tool for measuring RABF (5), but PET indirectly measures RABF and is not readily available. MR phase-contrast (PC) techniques have shown a potential for measuring RABF directly and noninvasively. Because of the small vessel sizes (3–10 mm diameter) and variable flow velocities (10–30 cm/second) involved, the ideal MR sequence must achieve submillimeter spatial resolution along with sensitivity to relatively low blood-flow rates. The sequence also has to cover the flow in the main renal arteries and any accessory renal arteries quickly. Current techniques such as cardiac-synchronized (gated) methods require relatively long scan times, and therefore they are adversely affected by respiratory motion in the upper abdomen (2,6,7). Faster gated methods may provide scan times within a breath-holding interval, but their low spatial resolution or low temporal sampling degrades the accuracy and repeatability for measuring RABF (7).

Recently, ungated PC imaging with spiral trajectories, or ungated spiral PC (USPC), was shown to rapidly measure time-averaged flow rates accurately and reproducibly even in relatively small arteries and in the presence of strong pulsatility (8). In this work we investigated the potential of USPC specifically for obtaining rapid RABF measurements in vivo. Our results demonstrate that consistent RABF measurements within a very brief scan time can be obtained with USPC.

MATERIALS AND METHODS

Pulse Sequence and Postprocessing

USPC is a non-cardiac-synchronized PC technique that uses interleaved spiral k-space trajectories (8). The two flow encodings (FEs) in the through-plane direction are alternated every excitation, and the spiral readouts are rotated every odd excitation. Table 1 shows the imaging

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Table 1.
USPC Imaging Parameters for RABF Measurement

	N	FOV	Δx	θ	S-thick	TR	TE	Readout	Vmax
Protocol-1	40	30 cm	1 mm	30°	10 mm	15 ms	3.5 ms	7.4 ms	100 cm/s
Protocol-2	40	24 cm	0.8 mm	30°	10 mm	15 ms	3.5 ms	7.5 ms	100 cm/s

N = Number of interleaved spiral trajectories, Δx = spatial resolution, θ = flip angle, S-thick = slice thickness, Vmax = maximum encoded velocity.

parameters for USPC. For 40-interleaved spiral trajectories, we manually order the trajectories instead of using a bit-reversal method (9,10), so that the consecutive acquisitions will be maximally separated in k-space. If the spiral trajectories in k-space are numbered from 1 to 40 in terms of consecutive rotation angles, the trajectories can be shown in temporal order as follows: {1, 21, 11, 31, 5, 25, 15, 35, 8, 28, 18, 38, 3, 23, 13, 33, 40, 20, 10, 30, 6, 26, 16, 36, 7, 27, 17, 37, 2, 22, 12, 32, 9, 29, 19, 39, 4, 24, 14, 34}.

This USPC pulse sequence is played out continuously with a relatively short TR over a few cardiac cycles (i.e., over six seconds for renal application). USPC achieves high spatial resolution, and the scan time for one complete image is relatively long (1.2 seconds, i.e., 80 TRs for 40 interleaves and two FEs), which may increase flow artifacts depending on the k-space trajectories (8). In USPC, interleaved spiral k-space trajectories provide significant flow-artifact suppression, and a pseudo-randomized interleave ordering and minimum-first-moment FE scheme further reduce flow artifacts (8). Note that protocol 2 uses a different FOV and spatial resolution compared to the original USPC protocol (protocol 1; see Table 1). However, protocol 2 keeps the same imaging matrix size (300×300) and scan time for one complete image as those used in protocol 1, which suggests that protocol 2 will perform as well as protocol 1 based on the analyses presented in the original paper on USPC (8).

To acquire the appropriate time-averaged flow measurement of a periodic flow waveform, one must average over an integer multiple of cardiac cycles. In USPC, we can observe the periodicity from the same raw-data sets acquired to measure the time-averaged flow rate by constructing a graph called a cumulative-average velocity plot (CAVP) (8). Each point of the plot represents the velocity average based on using all the data collected up to that point in time. Figure 1 shows the CAVP of RABF in a normal human subject. To speed up the image reconstruction process for constructing a CAVP, we utilize the linearity of the image-reconstruction process (11), i.e., we first reconstruct single-spiral-interleave images and then perform the necessary temporal-averaging operation corresponding to each point in the plot. Finally, to construct the plot we perform a simple phase-difference (PD) operation in PC imaging (12) on the temporally averaged images corresponding to each point in the plot.

From the constructed CAVP, we choose the excitations that correspond to the low-to-low dips over more than one cardiac cycle (Fig. 1). We then temporally average those selected excitations to produce two PC images (8). Finally, we perform the conventional PC postprocessing (PD operation) on those two PC images to produce the RABF measurement. To produce a region of interest (ROI) with a smooth, circular vessel-boundary line, we employ a conventional thresholding technique that uses the peak magnitude pixel inside the lumen as a reference (12).

Experimental Methods

Studies were performed on a 1.5 T GE Signa whole-body scanner (GE Healthcare, Milwaukee, WI) capable of 40 mT/m gradient amplitude and 150 mT/m/msec slew rate. We used a body coil for RF transmission and a 5-inch surface coil for signal reception. The subjects were positioned supine, and the surface coil was placed posteriorly to facilitate the slice prescription procedure using an interactive real-time scan. Eleven normal volunteers (nine males and two females, age range = 24–30 years) and one patient with renal failure (male, age = 85 years) were scanned. The institutional review board approved all imaging protocols, and informed consent was obtained before each subject was scanned.

We prescribed an oblique slice perpendicular to the renal artery to obtain through-plane flow using a real-

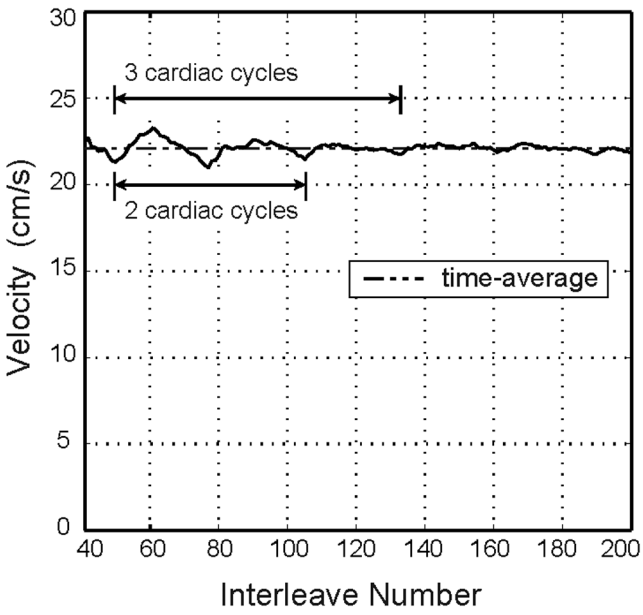

Figure 1. CAVP of RABF to determine the period of a cardiac cycle. This plot was constructed from the same raw-data sets for RABF measurement with USPC. One "interleave number" in the plot is equivalent to 2TR = 30 msec. Two examples of selecting the final data set that covers integer multiples of cardiac cycles are illustrated.

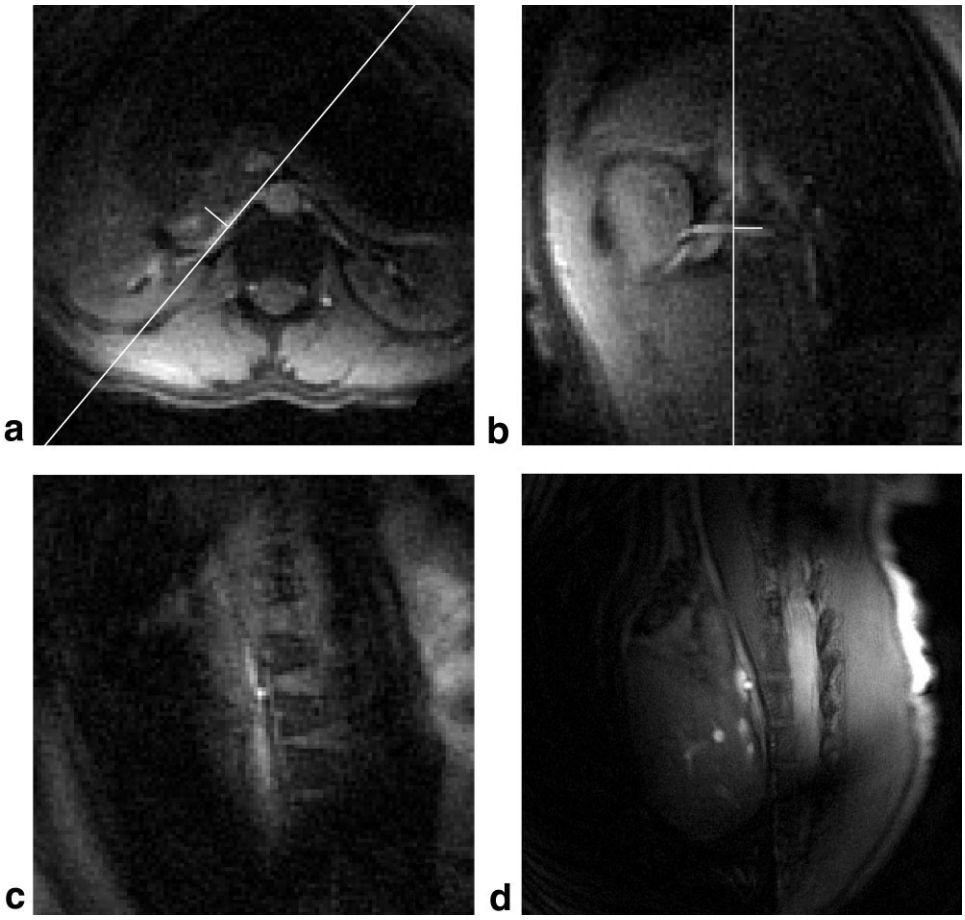

Figure 2. Prescription of an oblique slice, perpendicular to the renal artery, to obtain through-plane flow using a real-time interactive imaging system. **a:** On this axial image through both kidneys, the oblique plane parallel to the right renal artery is prescribed. **b:** On this oblique plane parallel to the right renal artery, the oblique slice (**c**) that will be used by USPC is prescribed perpendicular to the right renal artery. **d:** USPC (protocol 1) image at the prescribed plane. The whole prescription time was less than 30 seconds.

time imaging system (10), which provides interactive control over scan planes and imaging parameters through an external workstation (the workstation also reconstructs and displays images in real time). The real-time imaging sequence used four-interleave spiral trajectories with a 12-msec readout to achieve FOV/resolution = 20 cm/1.78 × 1.78 mm^2 at 104 msec per image. Other imaging parameters were TE/flip angle/slice width = 4.2 msec/30°/5 mm. Figure 2 shows the procedure used. After we acquired the axial image through both kidneys, we first prescribed the oblique plane parallel to the renal artery and then prescribed the oblique slice perpendicular to the renal artery. The whole prescription time was generally less than 30 seconds. The USPC pulse sequence was then played out over a brief six-second breath-holding interval. The breath-hold was done in a neutral lung position, which is approximately at early expiration phase. For the patient, the six-second USPC scan was performed without breath-holding. Protocol 1 was used for the normal subjects, and protocol 2 was used for the patient (Table 1).

RESULTS

Figure 3 shows the RABF rates in the right renal arteries of the 11 normal volunteers and the patient, corrected for their body-surface area (m^2) (13). (The results from the first five subjects were also presented in our previous work (8).) The error bars for each measurement represent the standard deviations (SDs) of three to five repeated measurements with USPC for each subject in a single session, and the time between the re-

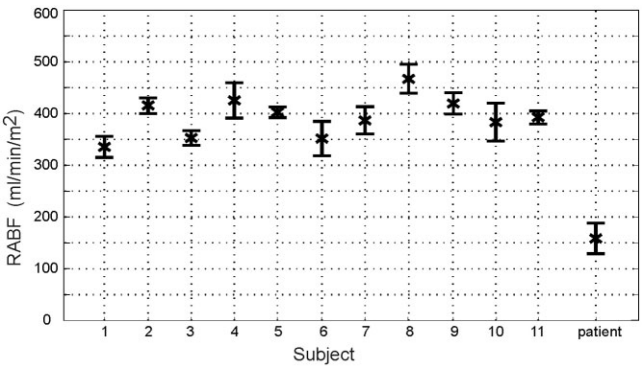

Figure 3. RABF measurement in vivo with USPC in six seconds. Right RABF rates in 11 normal human subjects and one patient with renal failure are measured and corrected for their corresponding body-surface area (m^2). With the normal subjects, a rapid USPC (protocol 1) scan produced consistent RABF measurements that agreed with the normal RABF rates (380 mL/min/m^2). With the patient, the RABF rates measured with USPC (protocol 2) were substantially less (<50–65%) than the normal RABF rates, which agreed with the other measurement that predicted the RABF rate to be half of the normal rates.

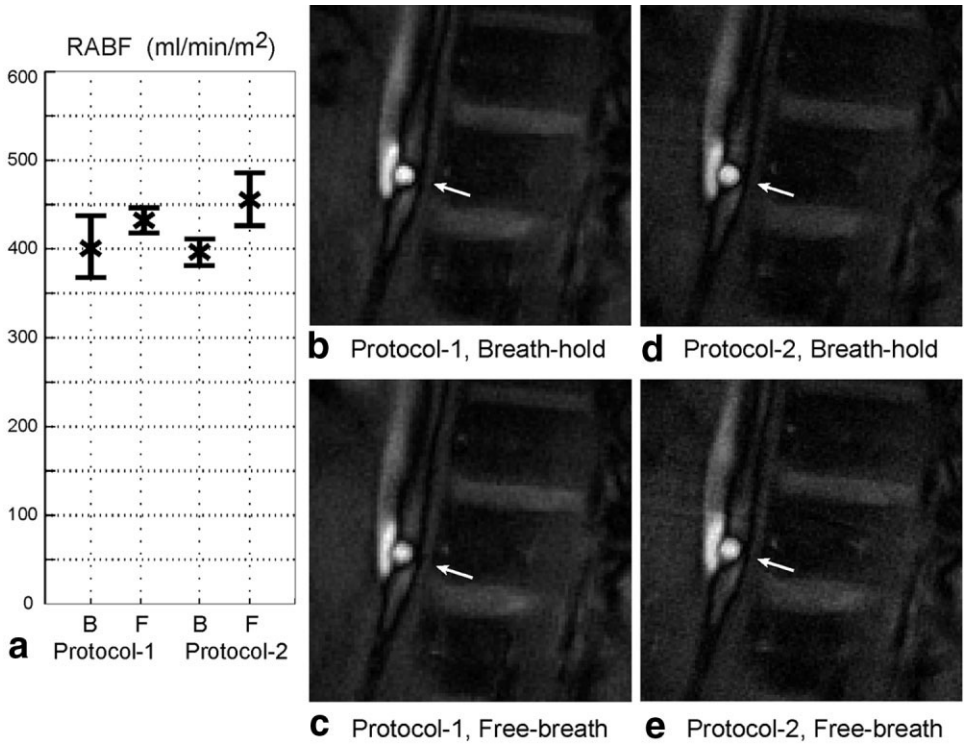

Figure 4. RABF of subject 5 in Fig. 3 measured again with protocols 1 and 2 in six seconds. The RABF measurements of subject 5 in this figure were obtained 31 months after the RABF measurements in Fig. 3. Right RABF rates are shown in **a**, and the corresponding magnitude images are shown in **b–e**. Protocols 1 and 2 yield minimal differences in RABF measurements and magnitude images, which show minimal vessel movement and minimal flow artifacts. A slight overestimation is observed with free-breathing, possibly due to an increase in the ROIs measured.

peated measurements was approximately 10 seconds. RABF measurements of normal subjects (SD $\leq$ 9%; average of mean = 394.8 $\pm$ 38.4 mL/min/m^2) agreed with the normal RABF rates (approximately 380 mL/min/m^2 per kidney) known from the literature (14). The RABF rates of the patient were substantially less (<50–65%) than the normal RABF rates, which corresponded with the elevated serum creatinine level of 2.0 mg/dL.

Figure 4 shows the effects of protocol 2 and free-breathing on the RABF measurements. As shown in this figure, and from the analyses presented in the original USPC paper (8), the differences in the protocols between the normal subjects (protocol 1 with breath-hold) and the patient (protocol 2 with free-breathing) apparently had a minimal effect on the measurement. Note that even after 31 months, USPC produced reproducible results.

The magnitude images showed minimal flow artifacts and sharply defined margins of arteries, indicating minimal vessel movement (Fig. 4b–e and 5). Note that for each measurement, the actual amount of data used for final flow quantification covered only two to four cardiac cycles, as illustrated in Fig. 1. This suggests that the scan time can be further shortened. However, in our studies we chose a scan time of six seconds to facilitate the scan prescription and ensure a sufficient signal-to-noise ratio (SNR) even in large subjects.

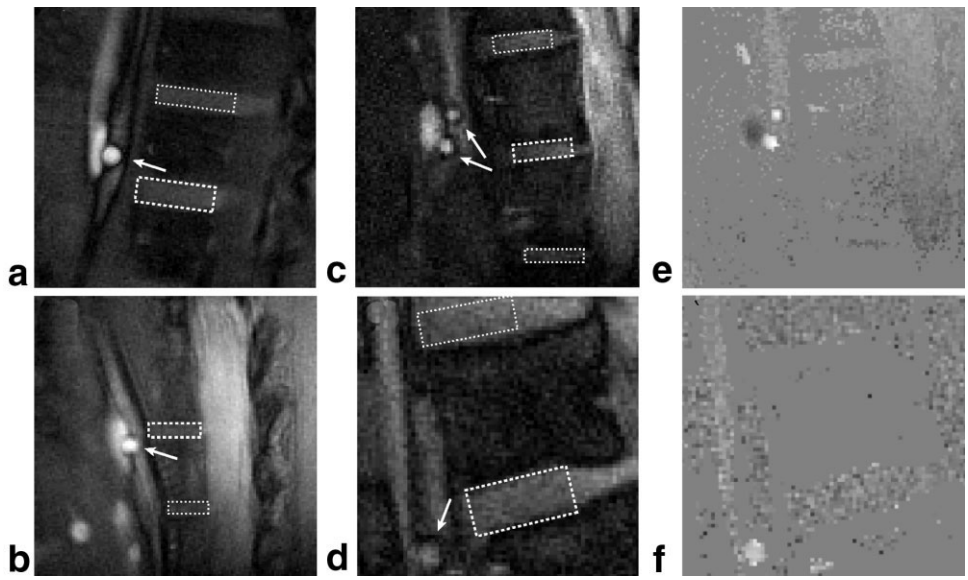

Figure 5. USPC magnitude and phase images. **a–d:** The arrows show renal arteries, and the dashed lines show static regions used for local phase-offset correction. Image a is the same as Fig. 4d, b is the same as Fig. 2d but zoomed in, c is from subject 10, and d is from the patient. **e** and **f:** The corresponding phase-offset-corrected phase images of c and d, respectively.

DISCUSSION

The results demonstrate that rapid and consistent RABF measurements can be obtained with USPC within a very brief scan time. Rapid USPC scans of the normal subjects produced consistent RABF measurements that agreed with the normal RABF rates known. The RABF rates of the patient, which were substantially less than the normal rates, agreed with his known renal dysfunction, which suggests that USPC may be useful even when RABF flow is compromised by renovascular disease.

Potential sources of errors in RABF measurements with USPC include 1) errors in local phase-offset correction (12); 2) errors unique to USPC, as described in Park et al (8) (i.e., errors from the asymmetric temporal velocity distribution and the inflow magnitude-modulation effects, errors from selected excitation sets when the final flow quantification does not cover exact integer multiples of cardiac cycles, and partial averaging because the number of selected excitation sets is not exactly equal to the multiples of *N*); 3) partial-volume effects (PVEs) in renal arteries with small diameters (12,15); 4) artifacts caused by misalignment between a prescribed oblique slice and the sensitivity region of a receiving surface coil; 5) blurring of the vessel in the magnitude image due to vessel compliance or displacement during the scan; and 6) physiological variation of RABF during the scan. These six sources of errors were addressed or analyzed as described below.

The errors in local phase-offset correction can be minimized by choosing more than one static region necessary for correcting the phase-offsets, especially when static regions are far from the vessel in the prescribed slices (Fig. 5). However, since the phase-offsets in the ROI were extrapolated from the static regions in our studies, the nearest static region was weighted to minimize the error in estimating the phase-offset. The observed variability (<2%) resulting from this phase-offset correction was small compared to the RABF rates in all of our studies.

The errors that are unique to USPC were minimal compared to the RABF rates in all our studies. The errors from the asymmetric velocity distributions and inflow effects were minimal because the imaging objects and protocols (e.g., Vmax, flip angle, etc.) were similar to those described in Ref. 8. We were able to minimize the errors resulting from the fact that selected excitations sets did not cover exact integer multiples of cardiac cycles, as well as errors from the partial-averaging process (8), by choosing the final data that corresponded to the low-to-low dips of the CAVP (see Fig. 1), and using more than one cardiac cycle for the final flow quantification. Note that a CAVP eventually approached the desired time-averaged velocity, which suggests that there is a trade-off between measurement errors and scan times (Fig. 1). One can further minimize these USPC errors by using scan times that are longer but still on the order of only a few seconds.

In all of our studies, including the multiple-renal-artery case shown in Fig. 5a, the diameters of the arteries were sufficiently large, compared to the spatial resolution, resulting in approximately <10% error from PVEs (15).

If objects exist outside of the FOV of the prescribed oblique slice, but inside the sensitivity region of a receiving coil, they create swirling artifacts that are observed in a typical spiral-readout image with insufficient FOV. Therefore, before prescribing an oblique acquisition slice from the real-time scout scans, one should take care in positioning the subject and surface coil. When a relatively large FOV compared to the sensitivity region of the receiving coil is used, and the coils are carefully positioned with the use of scout scans, there will be no or only weak swirling artifacts near the edge of the FOV, as seen in our experiments.

Vessel displacement during the scan can cause blurring of a vessel boundary in the magnitude image. The diameter or the position of renal arteries may vary during the scan due to the pulsatility in the renal arteries. In terms of measured velocity in a voxel near the vessel boundary, PVEs in the temporal direction can occur, i.e., the voxel contains velocities of surrounding static materials and those of blood at the vessel edge (12). However, because the observed variations in the shape and size of the vessel boundaries were visibly and quantitatively minimal, blurring and its accompanying error were also minimal in all of our studies, including the patient scan without breath-holding.

The final potential source of errors is the physiological variation of RABF during and between the scans due to inconsistent breath-holding. In future work, we can retrospectively gate USPC to the respiratory cycle in order to further analyze the effect of breath-holding on RABF.

One may be able to further minimize the variability or improve the accuracy of USPC for RABF measurement by optimizing sequence parameters that depend on or are independent of the vessel and the flow waveform of interest. Further experiments or analytical analysis of USPC using (k, t)-space (16) may reveal the optimal *N* for specific imaging conditions. Other centric sampling k-space trajectories (e.g., projection reconstruction (PR), rosette (17), variable-density spiral (9), etc.) may be used with ungated PC imaging. Real-time PC techniques (ungated PC with high temporal resolution) (18,19) may also provide alternative ways to measure RABF with short scan times; however, the low spatial resolution of the real-time methods may degrade accuracy.

In conclusion, we have shown that with USPC it is possible to obtain consistent RABF measurements (which are difficult to obtain with gated methods because of respiratory motion in the upper abdomen) with a brief six-second scan. In the future, a larger clinical study will help to establish the variability in normal subjects and patients with renal failure. Furthermore, USPC may be able to produce reproducible and clinically useful measurements of flow in other parts of the body, such as the mesenteric artery (20) and cerebral blood flow.

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