Renal Pseudo-continuous Arterial Spin Labelling (pCASL) MRI: A Repeatability Study.

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Authors: M. Sokolska¹, D. Thomas¹, A. Bainbridge¹, X. Golay¹, S. Taylor¹, S. Punwani¹, D. Pendse²
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Aims and objectives

The kidneys are highly vascularised organs, receiving approximately 20% of cardiac output under normal circumstances. Renal perfusion may be abnormal in focal renal disease (such as carcinomas), global renal disease (such as renal artery stenosis, acute renal injury, chronic nephritides) and complications of renal transplant.

Imaging of renal perfusion is therefore of huge clinical interest but has traditionally created a real dilemma for the radiologist. Cross-sectional perfusion imaging with CT or MRI relies on the administration of contrast media which may indeed worsen renal function (as in iodine-based contrast agents) or may be contraindicated (such as gadolinium based agents in renal failure).

Arterial Spin Labelling (ASL) is an advanced MRI technique, developed for neuroimaging. It allows for the assessment renal perfusion without contrast agent injection. ASL uses the water-spins in blood as an endogenous contrast agent, obviating the need for injection of contrast media. Furthermore ASL offers quantification of renal blood flow (RBF) which may be a useful imaging biomarker.

The work presented in this poster aims to investigate the potential for renal perfusion imaging with pseudo-continuous ASL\(^1\) (pCASL) in healthy volunteers, assessing both feasibility and within-patient repeatability.

Methods and materials

**Ethical Considerations:**

Institutional and Research Ethics Committee approval was obtained for this study.

**MRI Protocol:**

Seven healthy adult volunteers were recruited to this study (3 females, age 25-35). All subjects underwent renal ASL perfusion measurements in a 3.0T Phillips Ingenia wide-bore MR scanner. Volunteers were positioned with their arms extended on dedicated armrest above their heads. Respiration and cardiac cycles were monitored with respiratory bellows and pulse oximetry. The main protocol consisted of localiser scans, pseudo-continuous Arterial Spin Labelling (pCASL) and phase contrast MRA. A coronal-oblique ASL imaging volume was positioned along the long axis of both kidneys.
to reduce in and out of plane motion. The labelling plane was carefully positioned above the kidneys, perpendicular to the descending aorta. Labelling duration (t) was 1.65s, post labelling delay 0.9s. pCASL labelling train consisted of 0.5ms Hanning RF pulses with FA=18°, maximum gradient 0.6 G/cm and Gmax/Gmean equal to 10. Gradient-echo EPI was used for the readout module with 108x99 acquisition matrix, 320x320 FOV, 10 slices (2mm gap), SENSE factor of 3 and TE/TR = 7/4000ms. 20 control-label pairs were acquired. To reduce motion between acquisitions, two pairs were acquired within a 16s breath-hold. To estimate labeling efficiency, quantitative, a cardiac-gated phase contrast scan was also acquired at the exact position of labeling plane (2x2x5mm^3 FOV, VENC = 90-100, TR/TE 5.6 / 3.3, 12 phases). 5 volunteers underwent a repeat pCASL MRI during the same scanning session. Perfusion data was acquired from both kidneys in each of 12 separate pCASL acquisitions with a total of 24 data sets.

**Fig. 1:** Saggital (left) and Coronal (right) views of the abdomen, showing ASL labelling plane (White Lines) with respect to the aorta and the kidneys.

**References:** Imaging, UCLH - 2PG/UK

*Fig. 1 on page 6*
**Fig. 2**: Sagittal (left) and Coronal (right) views of the abdomen, showing ASL imaging field of view (FOV) and shim boxes.

**References**: Imaging, UCLH - 2PG/UK

**Motion Correction**:

To further reduce misalignment between consecutive pairs, post-processing motion correction was employed. First, the images were cropped to restrict FOV to the left and right kidney separately. Next, a mask was drawn manually on each kidney and used to constrain the rigid registration to the region of interest. Control and labelled images were registered to a template created by refined registration of control images using DTITK\(^2\) software.

Pairwise subtraction was then performed, and these were averaged to create a perfusion-weighted image (PWI). These processing steps improved the quality of PWIs, especially in volunteers that performed breath holds in an inconsistent manner.
**Labelling Efficiency estimation:**

Maximum velocity in each cardiac cycle phase was used to estimate labelling efficiency by numerical simulation of the Bloch Equations, assuming laminar flow. Then, all efficiencies were integrated over one full cardiac cycle. The resulting estimated efficiency fraction was used for RBF quantification.

**Fig. 3:** LEFT: Sagittal image showing position of phase-contrast imaging plane
UPPER RIGHT: Axial phase-contrast imaging of the aorta
LOWER RIGHT: Estimated blood velocity in aorta, during average cardiac cycle

**References:** Imaging, UCLH - 2PG/UK
Fig. 3 on page 7

**Quantification:**

The general kinetic model was used to quantify RBF using FSL\(^3,4\), with blood T1 set to 1.65s and partition coefficient \(l = 0.9\). Mean value of M0 within whole kidney mask was used for calibration. Cortical RBF was calculated by averaging signal within a manually segmented cortex mask created for each kidney.
Images for this section:

**Fig. 1:** Saggital (left) and Coronal (right) views of the abdomen, showing ASL labelling plane (White Lines) with respect to the aorta and the kidneys.
Fig. 2: Sagittal (left) and Coronal (right) views of the abdomen, showing ASL imaging field of view (FOV) and shim boxes.
Measurement of aortic blood-velocity with Phase-contrast imaging

**Fig. 3:** LEFT: Sagittal image showing position of phase-contrast imaging plane
UPPER RIGHT: Axial phase-contrast imaging of the aorta
LOWER RIGHT: Estimated blood velocity in aorta, during average cardiac cycle
Results

**Imaging Findings:**

Maximum and mean velocity in the descending aorta are shown in Figure 1. Perfusion maps generated were of sufficient tissue contrast and spatial resolution to differentiate between cortex, medulla and renal columns. Mean labelling efficiency, calculated from phase-contrast imaging was 0.72 (range 0.71-0.74) and was used in the quantitative model of RBF. A typical renal perfusion map is shown in Figure 2. Tissue contrast and spatial resolution of the imaging allowed for differentiation between cortex, medulla and renal columns. Mean perfusion values for the whole kidney were 209±39 mL/100g/min (range 104-271 mL/min/100g) and for the renal cortex 243±55 mL/100g/min. Mean perfusion difference between the left kidney (241mL/100g/min) and right kidney (245mL/100g/min) was not significantly different from zero.

![Right kidney blood-flow maps](image)

**Fig. 4:** Quantitative renal blood flow maps: Multiple sagittal-oblique images of the right kidney showing calculated renal blood flow per voxel.

**References:** Imaging, UCLH - 2PG/UK
**Repeatability:**

Mean perfusion difference was not significantly different from zero for both whole kidney and cortex. Bland-Altman plot showing inter-scan variability is shown in Figure 3. The mean within-session difference for RBF in the whole kidney was -1.6 ml/100g/min with 95% CI of [-77, 74].

![Bland-Altman plot showing within-patient repeatability of pCASL renal blood-flow measurement. Red dotted lines represent 95% confidence intervals.](image)

**Fig. 5:** Bland-Altman plot showing within-patient repeatability of pCASL renal blood-flow measurement. Red dotted lines represent 95% confidence intervals.

**References:** Imaging, UCLH - 2PG/UK
Images for this section:

**Fig. 4:** Quantitative renal blood flow maps: Multiple sagittal-oblique images of the right kidney showing calculated renal blood flow per voxel.
**Fig. 5:** Bland-Altman plot showing within-patient repeatability of pCASL renal blood-flow measurement. Red dotted lines represent 95% confidence intervals.
Conclusion

In this preliminary study we have demonstrated the feasibility of using pCASL-GEPI to measure renal perfusion in healthy volunteers. The technique is quick to acquire and both tissue contrast and spatial resolution suggest likely utility in the clinical setting.

In one volunteer, inconsistency between test-retest was observed. We hypothesise that this was caused by a reduction of labelling efficiency due to variable B0 homogeneity in the labelling plane.

In all other volunteers, the same session repeatability was acceptable, although the authors note that normalisation for blood-pressure or cardiac output was not performed.

Further work is ongoing to evaluate the technique in pathological disease states and to compare the pCASL data with contrast-enhanced MRI.

Personal information

References

3. http://fsl.fmrib.ox.ac.uk/fsl