High Resolution Mouse Kidneys Perfusion Imaging using Pseudocontinuous ASL (pCASL) at Very High Field (11.75T)

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Introduction: A method that would allow *in vivo* and non-invasive reliable assessment of the renal microvascular perfusion would be very valuable for many diseases such as hypertension, ischemia or acute renal failure which have shown to be linked to damage or loss of renal microvessels. For this purpose, Arterial Spin Labeling (ASL) had shown great potential for measuring renal blood flow (RBF) [1-3], by offering a simple quantification model and a blood contrast specificity without any contrast agent injection. Unfortunately, ASL signal is weak, in the order of few percents of the tissue signal. Therefore, to obtain reliable quantitative blood flow maps, several data averages are required, making the technique prone to motion artifacts, especially in the abdomen area. So far, most of the reported human and small animal ASL kidneys studies have been performed with the Flow alternated Inversion Recovery (FAIR) technique[1-4]. However, the recent pseudo-continuous (pCASL) technique[5], which has been shown to provide higher efficiency than pulsed techniques (e.g. FAIR) and which is now largely applied for brain studies, could be more adapted for kidneys perfusion studies[6]. This work presents the application of the pCASL technique at very high field (11.75T) for high resolution mouse renal perfusion measurement. The first part of the study was dedicated to the comparison of pCASL and FAIR in terms of sensitivity and quantitative blood flow values. In a second part, pCASL was used for multislice and coronal orientation acquisitions.

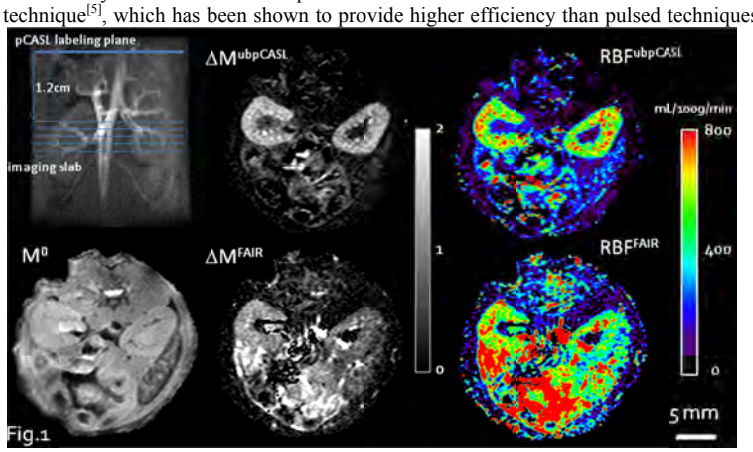

Fig.1

Methods: Experiments were performed at 11.75T on a vertical MR system (Bruker, AV 500WB, transmitter/receiver volume coil:∅ 2cm, length 3cm) on anaesthetized mice (C57BL/6j, 10 weeks, weight 25±1g, N=5).
presat-FAIR experiments were performed with a single inversion time (TI=1.3s) and a recovery time TR^{FAIR}=3.4s after global presaturation[7]. **pCASL experiments** were performed with the unbalanced (ubpCASL) scheme[8] (b_{1ave}= 4.7 μT (fa 33°), G_{max}/G_{ave}=90/10 mT/m, Hanning pulse duration δ=200μs and repetition rate Δt=450 μs, labeling duration τ=3s and post-labeling delay, *w*=0.3s). The transverse labeling plane was located ~1.2cm above the imaging slab perpendicularly to the descending aorta (Fig1). Magnetization differences, $\Delta M^{ubpCASL}$ and ΔM^{FAIR} were averaged 5 minutes. Quantitative RBF values were obtained by derivation of the classical CASL and PASL equations[9], with T_{1a}=2.1s and assuming blood transit time delays values of δ^{FAIR}=10ms and δ^{pCASL} = 100 ms: $RBF^{pCASL} = \Delta M^{pCASL}/(2M_b^0.\beta.T_{1app}.e^{-\delta_{pCASL}/T1a}.e^{-(w-\delta pCASL)/T1app}.(1-e^{-\tau/T1app}))$ and $RBF^{FAIR} = \Delta M^{FAIR}/(2M_b^0.\alpha.e^{-\delta_{FAIR}/T1a}.(e^{-(TI-\delta FAIR)/T1a}-e^{-(TI-\delta FAIR)/T1app})/(1/T_{1a}-1/T_{1app}).(1-e^{-TR_{FAIR}/T1a}))$. M_b^0 (equilibrium magnetization), α (FAIR inversion efficiency) and T_{1app} (apparent relaxation time) were determined with a slice-selective inversion recovery prescan[7] whereas the pCASL inversion efficiency (β) was measured in the renal arteries, ~1cm below the labeling plane[10]. For imaging in the axial plane, a 2-shot SE-EPI sequence (FOV 2.5x2.5cm^2, slice thickness 0.75mm, matrix 128x128, fat sat) was used for both the pCASL/FAIR comparison and multislice acquisition (5 sequential slices, total read-out time <200ms). For coronal images acquisition, a single-shot RARE sequence (TE=2.4ms, TE_{eff}=28.8ms, fat sat) was used. All the acquisitions were synchronized to the respiratory rate (80±10 bpm).

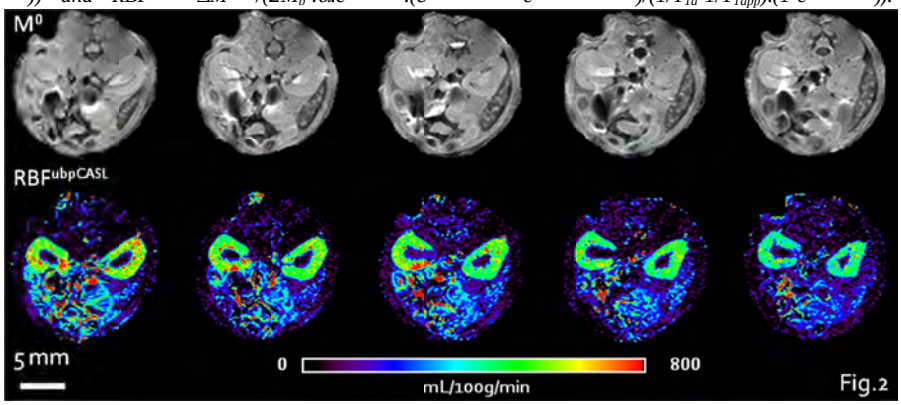

Fig.2

Results: Figure 1 shows a M^0 map along with typical ΔM and quantitative RBF maps obtained with presat-FAIR and ubpCASL techniques. This latter provided a significantly higher sensitivity ($\Delta M^{upCASL} >> \Delta M^{FAIR}$). This was confirmed by the quantitative analyses performed in the cortex ($\Delta M^{upCASL} \sim 1.34 \times \Delta M^{FAIR}$, N=5). Quantitative RBF values (*mean (ROI standard deviation) ± group standard deviation*) were measured in the cortex (RBF^{FAIR}=517(171)±111 mL/100g/min, $RBF^{ubpCASL}$=535(101)±92 mL/100g/min with β=0.68) and in the medulla (RBF^{FAIR}=112(78)±66 mL/100g/min, $RBF^{ubpCASL}$=135(70) ±56 mL/100g/min with β=0.68). Figure 2 shows multislice axial kidneys M^0 images and corresponding RBF maps obtained in a single imaging session with ubpCASL 2shot SE-EPI. Perfusion signal was homogenous across the 3 first slices with no apparent loss of sensitivity. For the distal slices, a slight loss of sensitivity appeared. Figure 3 shows single-shot M^0 RARE coronal image of the kidneys along with the corresponding ΔM and RBF maps, free of motion artifacts.

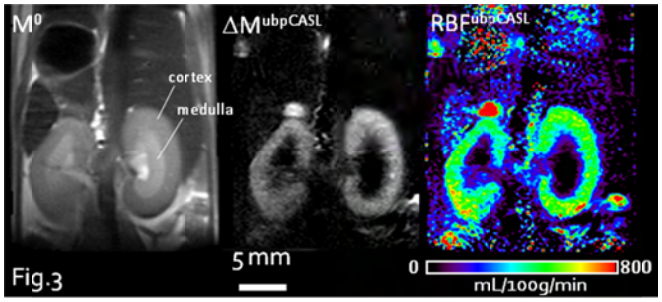

Fig.3

Discussion: This work presents the application of pCASL at very high field for mouse renal perfusion measurement. The pCASL technique combined with fast imaging provided highly resolved (200x200μm^2/pixel) RBF maps within 5 minutes in both axial and coronal planes. The respiratory gating combined with single-shot RARE and 2-shot EPI helped producing images of good quality, free of susceptibility and motion artifacts. With an inversion efficiency of ~0.68, pCASL offered a sensitivity gain of ~+34% compared to the FAIR technique and the RBF values measured in cortex and medulla were in the range of the literature data. Further optimization could be performed. First, background suppression should help in reducing motion-related subtraction errors in ASL images[6], which would lead to more accurate RBF values. Then, at very high field and in the abdomen area, B_1 and B_0 inhomogeneities could lead to pCASL inversion efficiency loss. Several strategies[11] could be envisaged in order to achieve β values higher than 0.68. On the overall, pCASL combined with fast imaging appeared to be a high sensitive and robust method well suitable for studying mouse models of renal vascular diseases.

References*: [1] Karger N. et al., MRI (2000) [2] Winter J.D. et al., JMRI (2011) [3] Martirosian P. et al MRM (2004) [4] Rajendran R. et al, Proc. ISMRM2010 [5] Dai et al., MRM (2008). [6] Robson, P. et al., MRM (2009). [7] Duhamel et al., MRM (2008) [8] Duhamel et al., MRM (2011) [9] Buxton et al., MRM (1998) [10] Zhang et al, MRM (1993) [11]Jahanian et al, NMR Biomed (2011)