Continuous Arterial Spin Labeling with Separate Labelling and Imaging Coils: Implementation Using a Single RF Channel and Amplifier

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Introduction

Arterial spin labelling (ASL) MRI enables non-invasive measurement of cerebral blood flow (CBF) utilizing arterial water as an endogenous contrast agent [1,2]. Among various ASL techniques, continuous ASL (CASL) [3], which uses separate RF coils for imaging and labelling [4], provides high SNR perfusion measurements in the absence magnetisation transfer (MT) artifacts. Conventionally, two independent sets of Proton RF channel and amplifier are required, while the the majority of MRI systems have only one single set, thus demanding the installation of costly RF hardware. This work describes an inexpensive implementation of two-coil CASL using a single RF channel and amplifier.

Methods and Materials

We first consider CASL with alternate labelling and imaging RF pulses (eg, single slice gradient echo). The TTL waveform of the single RF channel (**TTL in**, Fig 1), which gates the RF amplifier, is ON during the labelling and imaging pulses (**lab**, **im**, respectively**)**. However, the RF amplifier output must be switched between the labelling and imaging coils during the respective parts of the sequence. This requires logical separation of the two events (**lab** and **im**). Since **lab** and **im** pulses appear strictly alternating, **TTL in** is input to a positive-edge-triggered flip-flop (74LS74, National Semiconductor) in a divide-by-2 circuit [5]. A reset button initialises the flip-flop, avoiding unpredictable output on power-up. Next, **TTL out** enters a booster circuit comprising fast op-amps (LM318, National Semiconductors) and fast RF push-pull transistors (2N2219A, 2N2905A); two bipolar waveforms (**bias im** and **bias lab**, ON: 5V, OFF: -15V) with high output current (>1mA) are output. These waveforms are used: (i) to actively decouple the labelling and imaging coils and (ii) to actively switch the RF amplifier output between the labelling and imaging coils. The RF switch is of single pole double throw (SPDT) design with PIN diodes (50823077, Agilent Technologies), RF chokes (44μH, EPCOS); isolation between the output ports of the switch was improved by electromagnetically shielding the signal pathways. Total cost of all the components was less than 10 US dollars. Note that the bias waveforms (**bias lab**, **bias im**) are ON from the start of their respective RF pulses (**lab**, **im**) until the start of the following pulse (**im**, **lab**); therefore, the RF coils are connected to the output of the RF amplifier for a period exceeding the duration of the respective RF pulses. The increased period of connection between the RF amplifier and coils has no impact, since the amplifier is gated by the **TTL in** waveform, and is thus ON for only the duration of the RF pulses. When applying multiple imaging pulses between successive labelling pulses (eg multi-slice or spin echo), a dummy TTL ON pulse must be used between imaging pulses, by applying a short (~μs), zero amplitude RF pulse in the pulse sequence.

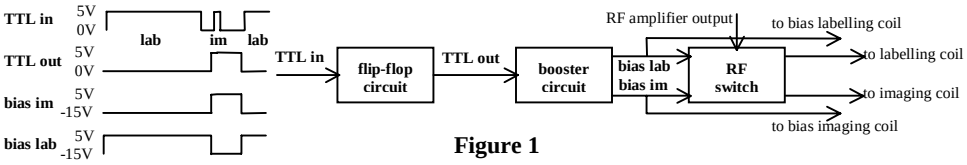

Figure 1

The circuits were evaluated both at the bench and when connected to the MR system (3T, 16cm horizontal bore Magnex magnet, Magnex 10cm-id self-shielded gradient (10kHz/mm max per axis), and an MRRS console). CASL was used to measure CBF changes in the rat (Lister Hooded) induced by 10% hypercapnia. Actively decoupled surface imaging and labelling coils were used (diameters 1.5 and 1.0 cm, respectively), positioned 2.5cm apart. Single-shot, gradient echo echo-planar imaging was used with 64x64 matrix, FOV=3x3cm, TE=15ms, TR=3s, labelling period of 2.850s. Paired images (ASL and control) were acquired alternately by varying the sign of the offset frequency for the labelling RF pulse. CBF values were calculated as in [4], using λ=0.9, T_1=1.4s α=0.8.

Resultss and Discussion

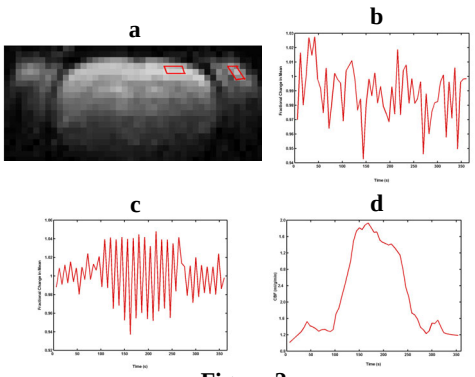

Figure 2

The following measurements were made with the circuit connected to the scanner: lag between input waveform **TTL in** and output waveform **bias lab** (**bias im)** < 50 ns, insertion loss of the RF switch < 0.6 dB, isolation between output ports of the RF switch > 53dB. Measurements on phantoms confirmed that the circuitry had no effect on RF power level and image SNR.

Figure 2 shows results from the hypercapnia experiments. In (a) an EPI image is shown with two ROIs: one on the muscle around the scull and the other on the cortex; the time course of the raw EPIs for these ROIs is shown on (b) and (c), respectively. The ROI from the cortex (c) shows the oscillatory pattern characteristic of the alternating labelled and unlabelled EPI acquisitions. The increase in the magnitude of these oscillations denotes CBF increase due to hypercapnia. The absence of oscillations in the muscle ROI (b) confirms the lack of MT effects using the two-coil CASL. (d) shows the CBF time course for the cortex ROI. The baseline CBF of 1.2ml/g/min [2] is increased by 77% under hypercapnia.

These results demonstrate that the described circuitry enables good quality two-coil CASL using a single RF channel and amplifier. Currently, it is used in studies of concurrent CASL and laser doppler flowmetry measurements of CBF changes in activation (submitted to ISMRM 05).

Conclusions

A method for performing two-coil CASL is described using a single RF channel and amplifier. It uses simple and inexpensive electronics to manipulate the TTL pulses of the RF channel and to switch the output of the RF amplifier selectively to the imaging and labelling coils. The method was implemented in a small bore MRI scanner, and achieved good quality CASL for measurement of cerebral blood flow changes under brain activation.

References [1] Silva AC et al, *J Cereb Blood Flow Metab* 19: 871-879 1999 [2] Sicard K et al, *J Cereb Blood Flow Metab* 23: 472-481 2003 [3] Detre JA et al, *Magn Reson Med* 23: 23-45 1992 1989 [4] Silva AC et al, *Magn Reson Med* 33: 209-214 1995 [5] Horowitz P and Hill W, The Art of Electronics, 2nd Ed, CUP, 1989

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