An inexpensive and programmable separated coil CASL system

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Introduction

Compared with other multi-slice arterial spin labeling (ASL) methods, continuous ASL (CASL) with a separate labeling coil provides higher SNR and lower RF power deposition (SAR) and is not confounded by magnetization transfer (MT) effects. Previous implementations [1, 2] required two independent sets of proton RF channel and amplifier, which are not available on most clinical MRI scanners, and used commercial RF instruments, such as PTS RF source (Programmed Test Sources, Inc, MA, USA) and NMR power amplifier, amounting to substantial RF hardware cost. One alternative implementation of separate coil CASL is to utilize a single RF channel and amplifier on the scanner by inserting a high power RF switch between the scanner RF power amplifier and the coil system [3]. This approach splits the RF power into a low power labeling period and a high power excitation period in the time domain. However, this method is limited by the maximum RF duration permitted on clinical scanners and requires scanner hardware modification. The present work describes an inexpensive alternative for two-coil CASL system with minimal scanner hardware modifications, using a system on chip (SOC) direct digital frequency synthesizer (DDS) [4]. Taking advantage of the DDS, this design also provides capability for remote programmable control for easy setup and debugging of the ASL sequence. In addition, this design may have applications in parallel RF excitation [5].

Methods and Materials

Fig.1 shows the system diagram for the generation of the RF waveform during the labeling period and the detuning of the labeling coil controlled by pulse sequence. The RF signal is directly generated by a single chip DDS (AD9954, Analog Devices, MA, USA) driven by a reference from scanner. Its 400MHz high speed core provides 0.01 Hz resolution, 95dB 1 MHz spur free dynamic range and -110 dBc/1KHz phase noise performance. Compared with costly commercial RF instruments, this single chip design significantly reduces the cost of the labeling system without sacrificing the RF signal performance. In addition, the DDS provides a programmable user interface. Therefore, it can be connected to a PC via parallel/serial port and remotely controlled by the scanner's host computer through a network. Instead of manually adjusting the RF in the MR equipment room from time to time, we can directly set the labeling RF parameter using the scanner console by remotely controlling the DDS RF amplitude and frequency output. To gate the RF signal, an RF switch with 80dB at 120MHz isolation capability was used (AD8180, Analog Devices, MA, USA). Subsequently, the RF signal was filtered and amplified by a 120MHz center frequency band pass filter and a mini 500MHz 10-Walt power amplifier with the TTL disable input which improves the RF isolation further (HD17070, HD Comm. Corp, NY, USA). The RF power was monitored with a power meter. The TTL gating signal came from the scanner and was converted by a home-built pin diode driver and sent to the coil to activate a pin diode in the labeling coil. As shown in Fig.1, this design needs minimal scanner hardware modification as it only uses the RF reference and TTL trigger signal from the scanner. A 65mm x 30mm de-tunable butterfly surface coil was constructed and used for labeling (Fig.2). The coil was placed on the neck at ~18 cm from the center of imaging volume. All studies were performed on a 3T Siemens Trio Scanner (Siemens Medical Solutions, Malvern, PA) using the Siemens birdcage head coil for excitation and reception of MR signal. Labeled (applying RF before spin echo EPI acquisition) and control images (no applied RF) were acquired in an interleaved fashion using EPI (TR/TE = 5s/27ms, FOV = 25cm, Resolution = 64X64, slice thickness = 5mm) with 96 averages, taking approximately 16 min. Subtracting the labeled images from the control images yielded perfusion weighted images. Labeling was performed with a 4 second pulse along with a 2.0mT/m labeling gradient and was followed by a post labeling delay of 0.7s.

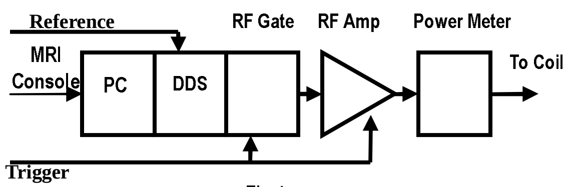

Fig.1

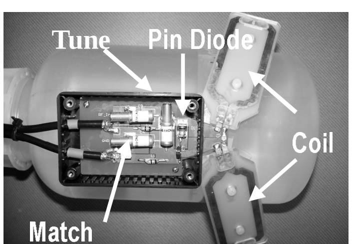

Fig.2

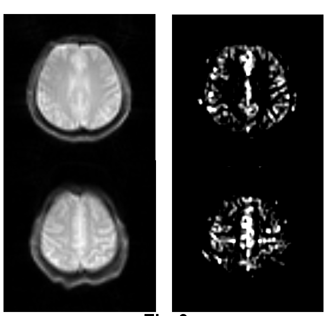

Fig.3

Results

The circuit was evaluated on the bench before it was connected to the scanner. The isolation between RF on and off was over 110dB. The insertion loss of the coaxial cable from power meter to coil was 3dB. The actual RF power delivered to the labeling coil was 500mW.

Fig.3 presents the perfusion weighted imaging results. Two axial slices for control images are shown in the left column and perfusion images created using the labeling coil and sequence are showed in the right column. The results show that the system works properly.

To further validate the efficacy of this system, we also quantified CBF by measuring a 35-year old healthy male volunteer using our two-coil setup as well a traditional CASL acquisition [6] assuming T_{1b} of 1.49 sec, blood–brain partition $\lambda = 0.9$ and inversion efficiency (α) of 1 and .68 for the separate coil and traditional acquisitions, respectively. For gray matter, the ratio of the subtracted image to the control image ($\Delta S/S_{CONTROL}$) is 1.9%, demonstrating improvement relative to 1.1% achieved in the same session with traditional CASL. CBF derived from acquisition with the two-coil setup was 69.5 ml/100 g/min (using equations from ref. [2]) while that derived from the traditional CASL (using equations from ref. [6]) was 74.7 ml/100 g/min, in agreement with the expected resting value for gray matter of 60 ml/100 g/min [7].

Conclusion

In this work, a new design for two-coil CASL labeling system is presented. The present design utilizes a low-cost SOC DDS to directly generate the labeling RF waveform without sacrificing RF performance, reducing the total cost to ~$2000. Compared with a single RF channel approach, this design requires minimal scanner hardware modifications. Moreover, this design offers programmable interface for easy sequence setup and debugging. Performance of this system for CASL is demonstrated.

Reference

1. S. L. Talagal et al. Magn Reson Med 52:131-140 (2004). 2. G. Zaharachuk et al. Magn Reson Med 41:1093-1098(1999). 3. N. G. Papadakis et al. Concepts Magn Reson Part B 29:145-152(2006). 4. Y. Jiang et al. Rev Sci Instrum. 73:3329-3331(2002). 5. AD9959 Datasheet. Analog Devices (2005). 6. J. Wang et al. Magn Reson Med. 50:599-607(2003). 7. N. A. Lassen et al. Physiol Rev 39:183-238 (1959).

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