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An Inexpensive and Programmable RF Transmitter Setup for Two-coil CASL

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Abstract

An inexpensive and programmable RF transmitter for two-coil continuous arterial spin labeling (CASL) system is presented in this paper. While previous implementations of two-coil CASL require the expensive MR RF instruments, the present design utilizes a low-cost system on chip direct digital synthesizer and a mini-size communication RF power amplifier to generate the labeling RF waveform without sacrificing RF performance. Compared with a single RF channel two-coil CASL approach, this design requires minimal scanner hardware modifications. Moreover, this design offers a programmable interface for easy sequence setup and debugging. Performance and ease of use are validated experimentally.

Keywords

perfusion; continuous arterial spin labeling; separate labeling coil; RF transmitter

INTRODUCTION

Pulsed (PASL) and continuous arterial spin labeling (CASL) are established tools for imaging perfusion with MRI, allowing for noninvasive assessment of brain function and pathology (1). Compared with single coil CASL (2), CASL with a separate labeling coil (two-coil) has the advantages of higher signal-to-noise ratio (SNR), the ability to acquire oblique slices, reduced RF power deposition and absence of magnetization transfer (MT) effects (3–5).

Most implementations of two-coil CASL (3–5) require two independent sets of proton RF channel and amplifier, a hardware requirement that is not currently available on most clinical MRI scanners. Consequently, implementation of two-coil CASL often requires the availability of costly commercial RF instruments such as Programmed Test Sources (PTS) RF source and MR RF power amplifier. This substantial RF hardware cost poses an obstacle for the routine availability and application of two-coil CASL.

One inexpensive alternative implementation of two-coil CASL is to utilize the RF channel and amplifier on the scanner by inserting a high power RF switch between the scanner RF power amplifier and the coil system (6). This approach splits the RF power into a low power

labeling period and a high power excitation period in the time domain. Although this implementation significantly reduces cost, it is limited by the maximum RF-on time permitted on clinical scanners causing sub-optimal labeling efficiency. More importantly, it requires scanner RF chain hardware modifications which vary by scanner and are quite complicated.

The present paper describes an alternative inexpensive solution for two-coil CASL with minimal scanner hardware modifications, using a system on chip (SOC) direct digital synthesizer (DDS) (7) and mini-size communication RF power amplifier. The programmable DDS technique facilitates setup and debugging of the CASL sequence. This design is also simple and fully integrated, requiring minimal cost. Additionally, utilization of a communication RF power amplifier instead of an expensive MR RF power amplifier further reduces cost. A preliminary account of this work was presented at the 16th annual meeting for the International Society for Magnetic Resonance in Medicine (8).

METHODS

Design

Fig. 1 shows the schematics and the photograph of the proposed setup for gating a continuous wave RF output during the labeling period and to detune the coil after labeling. The setup requires only two inputs from the scanner. First, a 10 MHz reference signal ensures that the labeling RF is coherent with the scanner center frequency. Second, the scanner Transistor-Transistor Logic (TTL) triggering signal is used to control the labeling timing. In this paper, we refer to this TTL as “triggering signal” and the RF triggered by this TTL as “labeling signal.”

A newly introduced integrated circuit (IC), AD9954 (Analog Devices, Norwood, MA), is used to generate the labeling RF signal. The DDS core in the AD9954 provides 32-bit frequency resolution (0.01 Hz tuning steps) and 14-bit phase resolution (0.02° tuning steps) for a sinusoidal output in the digital domain. A 400 mega samples per second (MSPS) digital to analog converter (DAC) is used to convert the digital signal to an analog RF waveform. The output frequency can range from DC to 180 MHz when referenced to a 400 MHz system clock, covering Larmor frequencies for fields up to 7 T. An amplitude modulation block is inserted between the DDS core and the DAC enabling control of RF output amplitude. A 14-bit digital amplitude scaling register and a digital multiplier are integrated in this amplitude modulation block. The value of this register that is used to scale the RF signal amplitude can be provided by control software according to the arterial labeling power. The DDS chip is driven by the 10 MHz external reference clock from the scanner console. Because the frequency of this reference is lower than 20 MHz, the minimum allowable reference input for AD9954, a passive coaxial frequency multiplier (Model: MK-2, Mini-Circuits Inc, Brooklyn, NY) is used to convert the scanner 10 MHz signal to meet the AD9954 reference input specification. An on-board phase lock loop (PLL) in AD9954 is used to permit the multiplication of the reference clock 20 times to 400 MHz.

This SOC chip also provides a flexible 4-line serial configuration protocol allowing programming with a computer parallel/serial port or micro-controller unit (MCU). In the present implementation, this DDS is connected to a Pentium III control personal computer (PC) via PCI bus. A PCI9052 (PLX tech Inc, Sunnyvale, CA) and a Spartan II (Xilinx Inc, San Jose, CA) Field Programmable Gate Array (FPGA) are used for data transfer and logic control between the DDS and the PC. This computer is remotely controlled by the scanner's host computer through a network (Microsoft remote desktop connection).

On the control PC, a program written in C++ automatically calculates the desired labeling frequency according to

$$f_{\text{labeling}} = f_0 + \gamma G \Delta r, \quad [1]$$

where f_0 is the center resonance frequency of the scanner, γ is the gyromagnetic ratio of proton, G is the labeling gradient, and Δr is the distance between centers of the labeling coil and scanner field. The program subsequently configures the DDS by storing the frequency tuning word (FTW) in the DDS frequency register using

$$FTW = (f_{\text{labeling}} \times 2^N) / \text{SYSCLK}, \quad [2]$$

where N is the frequency resolution (32 bits in present design) and SYSCLK is the system clock. The power of labeling RF is also controlled by this program as the scaled RF signal,

$$S(t) = A \cdot \cos(2\pi f_{\text{labeling}} t + \theta), \quad [3]$$

where A is the amplitude controlled by a 14-bit digital register in AD9954 and θ is the initial phase offset of the DDS. Instead of manually adjusting the RF in the MR equipment room as is done in previous implementations (3–5), the program allows remote control of the labeling RF parameters from the scanner console.

The SOC chip contains all the necessary functional blocks including DDS Core, high speed DAC, amplitude modulation block, PLL and serial configuration port integrated into a single chip. In addition, it can provide 95 dB spur free dynamic range at 1 MHz span and -110 dBc/KHz 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 quality.

Following the DDS, a single-pole double-throw RF switch with 80 dB isolation capability at 120 MHz (AD8180, Analog Devices, Norwood, MA) is used to gate the RF signal. This switch is controlled by the triggering signal from the scanner. Subsequently, the RF signal is filtered and amplified by a 120 MHz center frequency band pass filter and a mini-size 10 MHz – 500 MHz 10-Watt power amplifier (Model: HD17070, HD Comm. Corp, NY). To reduce RF interference and artifacts during MR signal acquisition, this power amplifier also provides a blanking signal controlled by the triggering signal to shut down the RF output. A two channel oscilloscope (Model: TDS3052B, Tektronix Inc, Richardson, TX) was used to measure the transient response of the RF power amplifier. The first channel was connected to a pickup coil 60 cm away from the labeling coil to visualize the RF power amplifier output signal. The second channel was connected to the triggering signal which synchronized the acquisition of the oscilloscope.

Finally, the labeling RF is delivered to the labeling coil through a power meter. A webcam (Model: VX3000, Logitech Inc, Fremont, CA) remotely monitors the value of the power meter through the network. As a result, the user can readily monitor the RF power to ensure safety throughout the scanning duration.

The lumped element equivalent circuit diagram and the photograph of the labeling coil are shown in Fig. 2. The loops and circuitry are made from a Flame Retardant 4 printed circuit board (PCB) with 0.95 mm thickness. There are two rectangular current loops ($6.6 \text{ cm} \times 3 \text{ cm}$) connected in series on the coil with a 130 degree angle. The left (L1, L2) and right (L3, L4) loops containing series coupling capacitors (C1, C2 and C3) are tuned by a variable capacitor (C4). Variable capacitors C5 and C6 are used for coil matching. The PIN diode actively switches the coil between labeling (tuned) and isolated (detuned) working states according to the triggering signal from the scanner. Inductor L5 and capacitor C4 form a parallel resonance circuit to detune the coil when the PIN diode is short circuited. A single chip PIN diode driver (Model: 9351, Impellimax Inc, Nashua, NH) is used to transfer the triggering pulse from TTL protocol to a +5V/-12V PIN diode switching signal. A 39Ω resistor is put in series at the output PIN diode driver, and the output current is approximately 110 mA when coil is detuned. To prevent RF interference with the PIN diode driver, RF chokes are placed in series with the control signal loops. All of the electronic components and coil loops are placed on one side of the printed circuit board (PCB). To ensure subject safety, all components are shielded by high impedance plastic covers. The Q factor of the labeling coil and the insertion loss of the coaxial cables were measured using a network analyzer (Model: 8753D Hewlett Packard Inc, Palo Alto, CA).

Image Acquisition

All studies were performed using a 3T Siemens TimTM system (Siemens Healthcare, Erlangen, Germany) equipped with gradients capable of 40 mT/m amplitude. A volume TEM head coil was used for excitation and reception of the images. A homebuilt 65 mm $\times$ 30 mm de-tunable butterfly surface labeling coil was placed on the neck, approximately 18 cm away from the isocenter of the scanner to continuously invert the blood flowing bilaterally in the carotid and vertebral arteries.

A diagram of the two-coil CASL sequence used is shown in Fig. 3. Labeled (applying RF before acquisition) and control images (no applied RF) were acquired in an interleaved fashion using spin-echo EPI (TR/TE = 4.4 s/27 ms, FOV = 25 cm, resolution = 64×64 , slice thickness = 5 mm, 11 ascending slices). Adiabatic flow driven RF labeling (9) was achieved with a 3 s pulse applied simultaneously with a 2.5 mT/m labeling gradient and was followed by a post labeling delay (PLD) of 700 ms.

Experiments

To evaluate the two-coil CASL system, two types of scans were acquired in a healthy 29 year old male subject. First, a resting scan was acquired with 60 averages (120 images raw images in 8.8 minutes) in 11 slices covering the sensorimotor cortex while the subject closed his eyes. Second, a functional scan was acquired with 80 perfusion images (160 images raw images in 11.7 minutes) in 11 slices covering the visual cortex during four 88 second fixation blocks interleaved with four 88 second blocks of flashing checkerboard stimulus (covering 8 radial degrees of the visual field and reversing contrast at 8 Hz). Visual paradigms were programmed in PresentationTM (www.neuro-bs.com), while a serial response box (Current DesignsTM, www.curdes.com) was used for synchronization with the MR scanner.

Data Analysis

All functional data were analyzed using SPM2 (Wellcome Department, University College of London, London, UK) and custom MATLAB (MathWorks Inc., Natick, MA) programs. First, all data were preprocessed for motion correction and subtracted (control – label) to generate perfusion-weighted images. Second, CBF was quantified under the assumption that magnetization transfer was absent using a published formula (10)

$$CBF = \frac{\lambda \Delta S}{2\alpha T_{1a} M_0} \cdot \frac{1}{e^{ATT/T_1} e^{-(w-ATT)/T_1} (1 - e^{-\tau/T_1})} \quad [4]$$

where λ is the water brain-blood partition coefficient, ΔS denotes the difference between control and labeled images, α represents labeling efficiency, T_1 is the longitudinal relaxation of gray matter, M_0 is the equilibrium magnetization of the brain, ATT is the arterial transit time, T_{1a} is the T_1 of arterial blood, τ is the labeling duration (3 s) and w is the post-labeling delay (700–1340 ms, calculated by slice). M_0 was approximated by the control images and α of 0.75, T_1 and T_{1a} of 1.33 and 1.49 s, respectively (4), λ of 0.98 ml/g (11) and ATT of 0.98 s (12) were used. Next, images were normalized to MNI space for further analysis. Resting CBF data were then masked with segmented anatomical data to quantify gray matter SNR by normalizing the mean difference image (control – label) dividing by the mean control image. Functional data were spatially smoothed (FWHM kernel of 6 mm) and analyzed with a general linear model with correction for a false discovery rate (FDR) using a threshold of $p < 0.05$ (13).

RESULTS

The circuit and coil were evaluated on the bench before it is connected to the scanner. The insertion loss of the coaxial cable from power meter to the coil was 3 dB. The actual RF power delivered to the labeling coil was approximately 1.4 W. The unloaded and loaded Q factors of the surface labeling coil were found to be ~75 and ~40, respectively. Therefore, approximately 47% of the RF power (~0.66 W) applied to the coil was deposited in the tissue (14) without exceeding the current FDA guidelines. Based on the coil geometry (rectangle 6.6 cm × 3 cm), assuming the coil has no resistive loss and using a 1900 ml Siemens water phantom with 0.375% NiSO₄ and 0.5% NaCl (Model: 8624186), we calculated the B1 at the center of each loop using the Biot-Savart law. The B1 is approximately 19 mG when the RF power delivered to coil is 1.4 W. To our knowledge, the B1 value varies depending on the coil loading. Fortunately, the two-coil CASL application is fairly insensitive to B1 variation owing to the flow driven adiabatic inversion RF pulses (15). Fig. 4 presents the transient response of the RF power amplifier when the labeling RF waveform was turned on and off. The transient time is approximately 50 ms and less than 1 μ s for RF on and off, respectively. Fig. 5 presents the resting state perfusion-weighted imaging results. For gray matter, the ratio of the subtracted image to the control image ($\Delta S/S_{CONTROL}$) is 0.95% reflecting similar SNR to other two-coil systems (0.4–1.3% (5) and 0.8% (3)). Average gray matter CBF was 52.6 ml/100g/min, within previously reported physiological range (80 ml/100g/min (3) and 48–62 ml/100g/min (4)). Fig. 6 shows the visual activation results from the same volunteer. As expected, suprathreshold voxels are located in the occipital regions of both hemispheres, consistent with the bilateral visual stimulus. The absence of spurious activation in unrelated brain regions also confirms the specificity of CBF to brain function, demonstrating the applicability of current system to functional studies.

DISCUSSION

Although two-coil CASL has been successfully used for several years, costly MR hardware and complicated implementation have limited its use as a routine perfusion imaging method. While the cost can be reduced by modifying the scanner's RF channel (6), this modification is complicated by the need to work with the manufacturer's RF system and possible limitation of RF on time. The SOC DDS presented in this paper inexpensively addresses the need for an additional RF channel without requiring scanner hardware modification. The

only connections with the scanner, the RF reference and TTL triggering signals, are available on most clinical scanners for device synchronization and fMRI stimulus triggering, respectively. In this regard, the proposed CASL system provides an inexpensive and general solution, independent of the specific architecture of the scanner RF chain, making it affordable and portable.

Another advantage of using the DDS for the two-coil CASL system is its high level of integration and its relevance to recent MRI hardware development. Several studies have demonstrated the principles and feasibility of applying DDS to MR applications (7,16). Recent increase in demand for high performance communications applications have led to DDS integration of all necessary functional blocks into one SOC chip with higher phase/amplitude resolution and system clock. These developments improve signal quality and enable simplified hardware design of the CASL RF transmitter because many on-board discrete components can be omitted.

In addition, DDS also provides a programmable interface for setting the RF signal frequency, amplitude and phase. For example, the user can easily program the AD9954 via the 4-line serial configuration interface using PC parallel/serial port or MCU. This feature enables the user to remotely control the DDS via network for easy set-up and sequence debugging.

Furthermore, taking advantage of the rapidly growing semiconductor industry, several individual SOC DDS are integrated into a single coin-sized chip in our design. This can dramatically simplify the MR transmitter design. For example, the AD9959 integrates 4 independent 500 MSPS DDS channels. This feature provides a potential application for parallel excitation which requires multiple RF channels, with independent control of RF phase and amplitude for each channel.

In general, most MR applications require strong MR RF power amplifier with very fast RF on/off switching transient response. For example, peak B1 amplitudes between 15 and 30 μT inside a whole-body transmit coil require a peak power of 15 to 20 KW on a typical 1.5 T whole-body MRI system (17). Because the RF power requirement is proportional to the square of the B_0 strength, higher field strength scanners require much stronger RF power. In addition, to minimize TE and TR, the duration of the RF excitation pulse should be as short as one millisecond. As a result, MR RF power amplifiers have a very fast RF on/off switching transient response. However, for CASL, instead of using an expensive MR RF power amplifier, we used an inexpensive 10 W mini-size communication RF power amplifier with RF ON/OFF control option for two reasons. First, for two-coil CASL, due to the localization of labeling by a neck surface coil, the second proton RF channel only requires a relatively low output (several Watts) (3–5). Second, compared with several seconds of RF labeling, the 50 ms RF on and less than 1 μs RF off transient response time of this mini-size RF power amplifier is sufficiently short. In addition, unlike conventional MR RF amplifiers which are operated in a pulse mode, the labeling RF power amplifier described here works in a continuous wave mode, making it unlimited by the duty cycle.

The increased cost efficiency does not reduce data quality, as demonstrated by the resting and functional data. However, gray matter CBF obtained in this study, while within the expected physiological range, is at the low end of values reported by previous studies (4,5). One potential source of underestimation could be the increased number of slices compared to ASL studies with reduced coverage; a recent study demonstrated that increasing the number of slices could excite blood spins in transit and reduce CBF in dorsal slices (18). Future work will investigate the dependence on slice number and optimize the imaging parameters for more accurate CBF measurements.

CONCLUSION

A new setup for performing two-coil CASL has been developed. Relative to previous systems using expensive MR RF instruments, the present design utilizes a low-cost SOC DDS and mini-size communication RF power amplifier to generate the labeling RF waveform without sacrificing RF performance, reducing the total cost to approximately \$2000. Compared with a single RF channel approach, this design requires minimal scanner hardware modifications, making the proposed system portable. Moreover, this design offers a programmable interface for easy sequence setup and debugging. The performance and ease of use has been validated experimentally, demonstrating its potential for widespread applications.

Acknowledgments

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Figure 1. (A) The hardware block diagram of the proposed two coil CASL system. (B) A photograph of homebuilt DDS RF transmitter. (C) A photograph of RF power amplifier and PIN diode driver

Figure 2.

(A) and (B) present the circuit diagram and a photograph of the butterfly surface labeling coil respectively.

Figure 3.

The Two-coil CASL sequence diagram. RF labeling applied during T1 period which followed by a T2 post-labeling delay before a multi-slice EPI acquisition. Labeling Coil is only tuned in T1 period.

Figure 4. (A) and (B) present the RF ON/OFF transient response of RF power amplifier respectively. The triggering signal acts as a gate to the labeling signal. Amplitude scaling is 5V/div for triggering signal and 100 mV/div for labeling signal.

Figure 5.
Mean $\Delta S/S_{\text{CONTROL}}$ for resting state perfusion-weighted imaging.

Figure 6.
Visual activation results show bilateral activation ($p < 0.05$ corrected for FDR).