A Low-Flip-Angle, Phase-Based Method for Accurately Calibrating the Transmitter in Hyperpolarized-Gas MRI

J. P. Mugler, III[1], G. W. Miller[1], T. A. Altes[1], E. E. de Lange[1], J. R. Brookeman[1]
[1]Radiology, University of Virginia School of Medicine, Charlottesville, VA, United States

Introduction: Calibrating the transmitter voltage for each subject is a standard procedure that is performed for every clinical MRI exam. This calibration is critical for obtaining consistent and optimum image quality, and is thus integrated into the software of commercial scanners. However, established transmitter-calibration procedures, which typically use high-flip-angle (90° or greater) RF pulses and are often iterative, are not suited to the non-equilibrium nature of the hyperpolarized-gas magnetization. Thus, these standard procedures cannot be used for hyperpolarized-gas MRI. The goal of this work was to develop a phase-based method for accurately calibrating the transmitter voltage that does not require high flip angles and can therefore be integrated into a breath-hold image acquisition, thus eliminating the need for a separate dose of hyperpolarized gas to calibrate the transmitter voltage for each subject.

Theory: Consider the application of a composite RF pulse with flip angles α_x, α_y, α_{-x} and α_{-y}, where x and y denote axes in the rotating frame. If we start with a z-component only and α is very small, this pulse will move the magnetization in a square pattern in the transverse plane and return it essentially to the origin. However, as α increases and non-linear behavior becomes important, the pattern begins to deviate from a perfect square and the magnetization no longer returns to the origin. For example, Fig. 1 illustrates this concept for a pulse with α equal to 20°. The points marked 2, 3, ... in Fig. 1 show the position of the magnetization after the pulse is repeated that number of times. We find (solid lines in Fig. 2): (1) the phase of the transverse magnetization has a unique correspondence to the flip angle α; and (2) the amount of phase shift per degree of flip angle can be increased by applying a "multi-cycle" (where a "cycle" is defined as the sequence α_x, α_y, α_{-x}, α_{-y}) pulse. Therefore, by measuring the phase angle of the transverse magnetization after applying such a multi-cycle RF pulse, we can derive the value of α, providing a means to calibrate the transmitter voltage. To deal with the classic problem of measuring the phase in the presence of background sources of phase shift, we simply perform measurements at two angles, such as $\alpha/3$ and α, and subtract the corresponding phase measurements to eliminate phase shifts common to both. For typical values of 5° and 15° for $\alpha/3$ and α, only 7% of the magnetization is consumed. The sensitivity of this technique to off-resonance signals is overcome by applying a half cycle (i.e., α_x, α_y) as the last part of the RF pulse.

Methods: Experimental confirmation of the theoretical predictions was obtained by performing studies on a commercial 1.5-T scanner (Sonata, Siemens Medical Solutions). Water phantoms were used to systematically evaluate the performance of the phase-based calibration method for 10.5 and 20.5-cycle pulses. The method was also tested in two healthy volunteers by using the 10.5-cycle RF pulse as an excitation pulse in a gradient-echo pulse sequence and acquiring ^{3}He lung images. The flip-angle values predicted from these images were compared to those from the transmitter calibration method that we currently use for routine ^{3}He studies. ^{3}He gas was polarized by collisional spin exchange with an optically-pumped rubidium vapor by using a commercial system (Model 9600 Helium Polarizer, Magnetic Imaging Technologies, Inc.). All experiments were performed under a Physician's IND (# 57866) for imaging with hyperpolarized ^{3}He following a protocol approved by our institutional review board. Informed consent was obtained in all cases.

Results and Discussion: Figure 2 shows experimental verification of the proposed method for $\alpha/3$ and α RF pulses with 10.5 (4 ms) or 20.5 (8 ms) cycles. For these measurements in a water phantom, the solid line is the theoretical behavior assuming that the manufacturer's transmitter-calibration procedure, on which the flip angle values in the plot are based, is exactly correct. It is reasonable to expect that even established calibration methods may be off by a few percent; the dotted lines denote the theoretical behavior that results for an error of ±5% in the manufacturer's calibration. Within these error bounds, there is excellent agreement between theoretical and experimental values. Figure 3 shows ^{3}He GRE lung images generated by using 10.5-cycle $\alpha/3$ and α transmitter calibration pulses for excitation, and the resulting phase-difference map. The phase-difference map varies smoothly except near the bottom of the lungs, in the vicinity of the susceptibility interface at the diaphragm. For transmitter calibration in the lung, we are interested in the flip angle integrated over a thin axial slab that is centered along the superior/inferior direction, and thus the poor behavior near the diaphragm is not a problem. The phase differences calculated along the length of the lung by using the central phase-encoding step for each image, which is the only data that would be collected for an actual transmitter calibration, were, except near the diaphragm, within 5% of the values obtained by averaging the pixel values in the phase-difference map in the left-right direction. This result supports the validity of using data that is spatially encoded only by using a readout gradient to estimate the flip angle averaged over an axial slab through the lung. The same measurements in a second subject produced equivalent results. The flip angles derived from the phase measurements were consistent with the crude transmitter calibration that we currently use for routine ^{3}He studies. Additional studies are needed to find the optimum trade-off between sensitivity, which favors calibration pulses having a large number of cycles, and resistance to off-resonance effects, which favors calibration pulses having a small number of cycles.

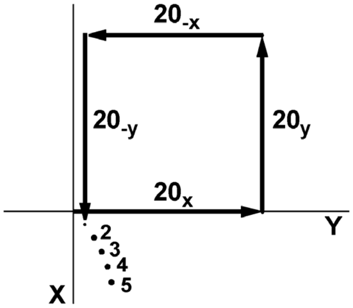

Fig. 1. Path of the magnetization in the transverse plane for a 20°$_x$, 20°$_y$, 20°$_{-x}$, 20°$_{-y}$ RF pulse.

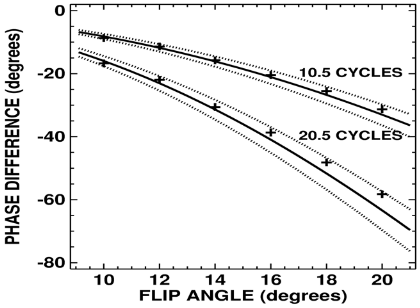

Fig. 2. Theoretical (lines) and experimental (+) phase differences versus flip angle for 10.5-cycle (4 ms) and 20.5-cycle (8 ms) RF pulses.

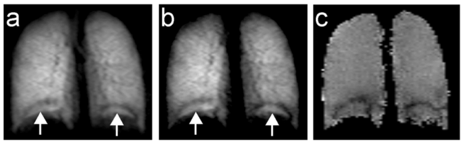

Fig. 3. (**a**, **b**) Coronal ^{3}He lung MR images obtained from a volunteer by using 10.5-cycle $\alpha/3$ (a) and α (b) transmitter calibration pulses for excitation. The only off-resonance-induced artifacts that are apparent occur in the vicinity of the diaphragmatic border (arrows). (**c**) The phase-difference map calculated from the images in (a) and (b). TR/TE, 120/3.8 ms; matrix, 56x128; FOV, 315x420 mm; thickness, projection.

Conclusions: We have demonstrated a phase-based method that can accurately determine the transmitter calibration using only low-flip-angle (< 20°) RF pulses. Since the procedure consumes only a few percent of the non-equilibrium magnetization, it can be integrated into any pulse sequence, obviating the need for a separate calibration dose of gas. This transmitter-calibration procedure provides an efficient means to achieve the desired flip angles for each subject without administering additional hyperpolarized gas, thus permitting optimum image quality, including maximum SNR, to be obtained consistently, which is critical if hyperpolarized-gas MRI of the lung is to become a technology that is practical for widespread clinical use.

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