Calibration of RF Transmitter Voltages for Hyperpolarized Gas MRI

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INTRODUCTION: Magnetic resonance imaging of lungs using hyperpolarized gases, ^{129}Xe and ^{3}He, has the potential to become an important diagnostic technique for clinical evaluation of lung function and pathology. Hyperpolarized gas MRI has the advantage of significantly increased signal to noise ratio but requires non-standard strategies for data acquisition due to the non-reversible loss of polarization after every RF excitation. As there is a fixed amount of gas polarization available for each MR scan, an accurate quantification of flip angle is a must for optimal imaging protocols. Conventional RF calibration methods used in proton imaging are not suitable for hyperpolarized imaging due to the non-reversible loss of magnetization. Several RF calibration procedures have been proposed (1,2). These methods require inhalation of an extra portion of hyperpolarized ^{3}He gas or the presence of bags of ^{3}He gas inside the scanner with the subject and are not ideally suited for clinical settings. Here we describe a simple procedure for flip angle calibration for MRI with hyperpolarized gases that does not require any extra ^{3}He gas or the presence of an additional phantom in the scanner. It also has advantages for clinical scanning as it practically eliminates additional time needed for calibration and requires only a one-time procedure that can be done during a routine QA protocol. Our approach is based on the hypothesis that for a given flip angle there exists a linear relationship between the required RF transmitter voltages for protons and the hyperpolarized gas nuclei. Here we demonstrate a method to determine this linear relationship and use a standard ^{1}H calibration procedure to predict the ^{3}He RF voltage required for a desired flip angle.

METHODS: The RF power required for a given flip angle changes with the coil load for both ^{1}H and ^{3}He transmit coils. For calibration purposes we need to establish a relationship between the transmitter voltages needed for 1ms 180° rectangular pulses for the ^{1}H coil (V^{1H}) and the ^{3}He coil (V^{3He}) for a range of physiologically relevant loads. According to our hypothesis, this relationship is linear and can be presented by the following equation:

$$V^{3He} = \lambda V^{1H} \tag{1}$$

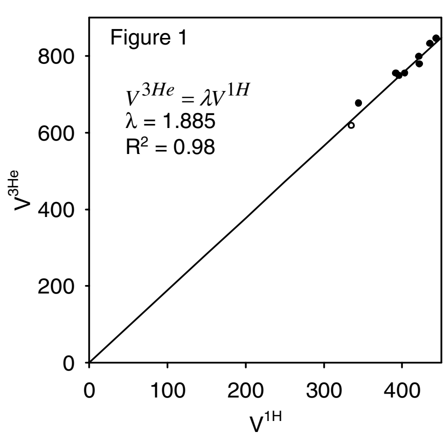
Below we describe a procedure to determine parameter λ. All experiments were done on a Siemens 1.5 T Magnetom Sonata system with a custom built Helium coil (3). Subjects ranging in weight from 120 lbs to 220 lbs were placed inside the magnet and a standard ^{1}H calibration procedure was used to determine V^{1H} . To determine V^{3He} we used a single-turn un-tuned inductive pick-up loop coil, terminated in 50 ohms resistance at an oscilloscope. This pick-up loop coil was fixed at a chosen location in the ^{3}He transmit coil. We recorded the voltages from the pick-up coil (U) with different loads inside the ^{3}He coil and 100V as the transmitter voltage. At the resonance frequency of ^{3}He at 1.5T the spatial distribution of the transmit field is not affected by the presence of the load: only the overall amplitude of the transmit field is affected by the load. Therefore the pickup voltages recorded by the pickup coil are directly proportional to the RF fields in the center of the coil. The pickup voltages U can be converted to V^{3He} using the following proportionality relationship

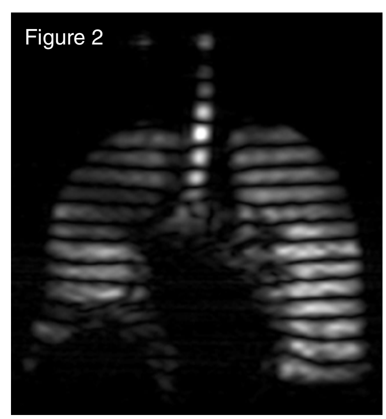
$$V^{3He} / U^{180} = 100/U , \tag{2}$$

where U^{180} is the pickup voltage when the transmitter voltage corresponds to 180° flip angle. U^{180} can be calculated using the pick-up coil geometry and RF frequency or determined experimentally. We used the experimental approach and estimated U^{180} voltage on a phantom by measuring a series of pick-up voltages simultaneously with the intensity of the MR FID signal from the phantom while varying the transmitter voltage. A standard sinusoidal fit determined the flip angle and thus established U^{180} . The measurement parameters used for the FID experiment were: TR/TE = 8000/3.2 ms, averages = 8, vector size = 1024 and spectral width = 2000Hz. Once V^{1H} and V^{3He} are found, the calibration parameter λ can be estimated using linear regression (Eq 1).

In vivo validation was done on a subject with inhalation of 300ml of hyperpolarized ^{3}He gas diluted with 700ml of oxygen. One coronal slice was acquired with spin tagging technique (4) with the following imaging parameters: FOV = 448mm, resolution = 128, slice thickness = 30mm, TR/TE = 7.5/4 ms, and 30mm tagging wavelength. A 45° flip angle for tagging pulses and 3° flip angle for imaging pulses were used with voltages determined from the above relationship.

RESULTS: Fig 1 shows the measured plot between ^{1}H and ^{3}He transmit voltages, confirming our hypothesis of the linear relationship between the RF voltage required for a given flip angle in both ^{1}H and ^{3}He imaging. Closed circles are data from subjects and open circle represents data from phantom. Eq 1 was used to fit the data resulting in a correlation coefficient, r^2 = 0.96 and slope, λ = 1.885, i.e. for our setup the ^{3}He coil needs approximately twice as much voltage as ^{1}H for the same flip angle. Fig 2 shows a representative image from the *in vivo* experiment. It clearly demonstrates the signal is nearly zero between the stripes, confirming the correct calculation of 45° RF pulse.





CONSLUSIONS: We have demonstrated a simple procedure to calibrate the RF pulse voltages in hyperpolarized gas imaging. The approach can be easily incorporated on any scanner and in clinical settings. The desired calibration curve between the proton transmitter voltage and hyperpolarized gas transmitter voltage needs to be determined only once for a particular MRI scanner and hyperpolarized gas imaging coil and can be done as part of QA protocol for the ^{3}He coil.

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