Pulmonary Oxygen Mapping with ^{3}He MRI at Very-Low-Field

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Introduction

A key measure of the effectiveness of pulmonary ventilation and perfusion is the alveolar partial pressure of oxygen, p_AO_2. MRI using hyperpolarized ^{3}He has, in recent times, provided the first regionally-selective measure of p_AO_2 [1-3], which has been correlated with the traditional ventilation-to-perfusion ratio, V/Q [2]. p_AO_2 is derived from the attenuation in the ^{3}He MRI signal over time, and for accuracy, requires precise knowledge of the RF excitation flip-angle. However, in clinical MRI systems operating at high field and frequency, this measurement has proven non-trivial, due to coil-loading effects of different human subjects and the general B_1 inhomogeneity of the RF coils used at 1.5 and 3.0 T. As a result, RF flip-angle calibrations must be incorporated into every measurement, resulting in numerous variations of this technique, and requiring the use of non-renewable magnetization in the flip-angle determination in every experiment, reducing the sensitivity of the technique to p_AO_2 [3]. We employ an open-access, very-low-field human MRI system to study pulmonary function with subjects in a variety of postures [4-6]. However, operation of this system at 210 kHz has resulted in a simplification of the p_AO_2 measurement technique in comparison to high-field methods.

Methods

Our open-access human MRI system was optimized to operate at B_0 = 6.5 mT (65 G) applied field, allowing ^{3}He MRI at 210 kHz. The magnet design allows us to use a highly-homogenous solenoidal RF coil, while the very-low Larmor frequency makes coil-loading effects negligible. For phantom measurements, a Tedlar bag was filled with ~ 500 cm^3 of polarized ^{3}He gas directly from a home-built spin-exchange polarizer, or the ~ 80 cm^3 optical pumping cell from the polarizer was placed in the imager. We acquired 2D gradient-recalled-echo images, without slice selection, over a 50 cm FOV, data size 128 × 32, TR/TE = 64/10 ms, NEX = 1, FA = 3º, in ~ 2 seconds. Multiple 2D images were acquired with 10 ms inter-image delays to calculate excitation flip-angle maps from the ^{3}He MRI signal decay. Spectroscopic flip-angle calibrations were performed at different locations in the RF coil using the same method but without spatial localization. pO_2 maps were obtained from the same imaging protocol but with 5 s inter-image delays.

Results

Spectroscopic and spatially-localized flip-angle measurements on ^{3}He phantoms are shown in Figure 1.

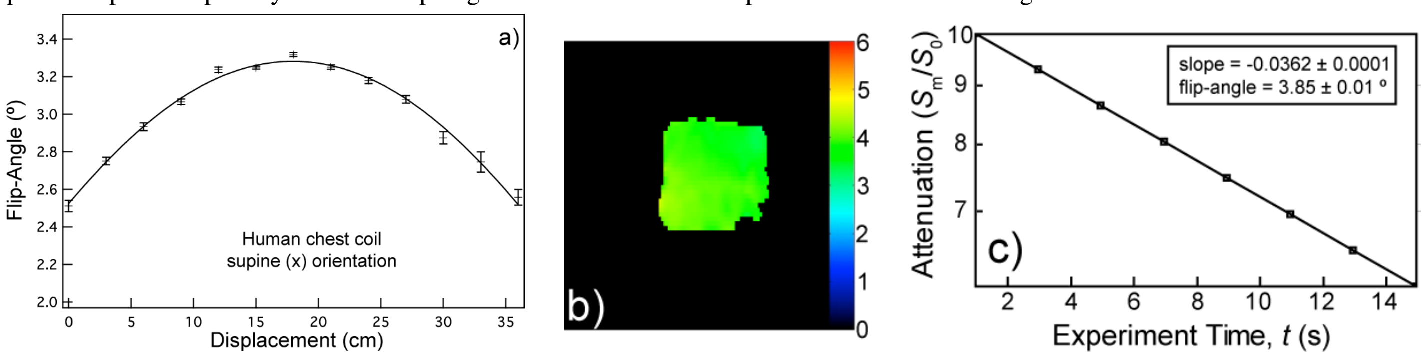

Figure 1: a) ^{3}He flip-angle derived from spectroscopic measurements as a glass cell of ^{3}He was placed at different positions along the x axis of the human chest coil. b) Calculated flip-angle map obtained from ^{3}He signal decay in 8 successive images, obtained from ^{3}He gas in a Tedlar bag. The scale bar values have units of degrees. c) ^{3}He signal decay from one of the pixels in the image in b).

Discussion

The solenoid RF coil design is provides a considerably more homogeneous B_1 environment than the coil designs used in most clinical MRI scanners. Our novel magnet design allows subjects to be imaged while horizontal or upright, using a solenoid RF coil. Fig. 1a) shows the measured and calculated variation in flip-angle along the length of our chest coil. Variation is less than 10%, and well characterized. Therefore, the flip-angle map shown in Fig. 1b) demonstrates minimal B_1 variation in the central 20 × 20 cm region of the coil, exhibiting a flip angle of 3.9 ± 0.2º. A subsequent MRI-based pO_2 measurement on a phantom filled with pO_2 = 68 ± 5 torr yielded an average value throughout the bag of 60.5 ± 7 torr, without the need to perform an in-situ flip-angle calibration. As we have shown previously that RF coil loading effects are minimal and reproducible at a Larmor frequency of 210 kHz [6], operation at such a low frequency permits a simplified p_AO_2 measurement procedure, possibly with increased accuracy in the p_AO_2 measurement.

Acknowledgements

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