Imaging of localized inert gas washout rates with ^{3}He MRI

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Introduction: Measurement of inert gas washout is an established method in pulmonary physiology, which is very sensitive to ventilation heterogeneity and disease processes at the level of the small airways [1,2]. Conventionally, washout is measured globally at the mouth, providing a single diagnostic index for the whole lung. Imaging of inert gas washout with ^{13}N PET has been proposed as a means of measuring pulmonary mechanics [3]. ^{3}He MRI washout measurements have been used previously to quantify ventilation in rat lungs [4]. Recently it was shown that the ^{3}He MRI signal can be used to obtain ^{3}He human ventilation images and washout data using a single dose of ^{3}He, thus obtaining complementary information on static ventilation distribution and the global washout rate [5,6]. The present work takes the ^{3}He MRI method further, with the aim of measuring washout rates directly on a regional or pixel-by-pixel basis.

Methods: Inert gas washout was tracked by imaging the ^{3}He ventilation distribution after each breathing cycle (exhalation-inhalation) using a 2D SPGR sequence on a GE HDx 1.5T scanner with a quadrature coil wrapped around the chest of a healthy volunteer (m, 29yr). The experiment was performed under informed consent and ethical approval. Pulse sequence parameters were: three axial 10 cm slices, acquired sequentially and covering the whole lung, FOV (35cm)^2, 64^2 matrix, TE/TR 1.1/3.1 ms, flip angle 2.9°. Image sets were acquired at twelve time points, with a delay of 3 s between scans. Scan duration for a single time point was ~0.6 s. The subject inhaled a mixture of 200ml ^{3}He (polarized to ~20% by Rb spin exchange equipment (GE Healthcare)) and 800ml N_2 from a Tedlar bag, and held his breath until the end of the second slice set. Then the subject performed a single breathing cycle at tidal volume between each of the subsequent time points.

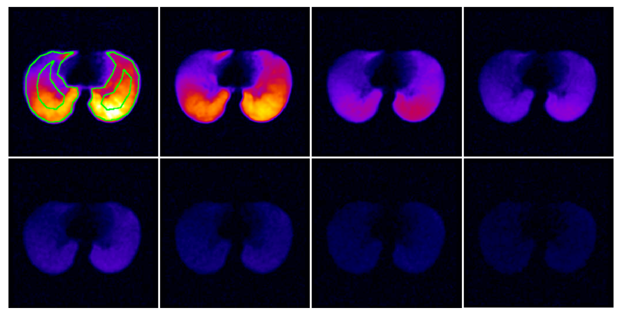

Figure 1: The first 8 images of the central slice. Subject held his breath until after the second image, then a single ventilation cycle (exhalation-inhalation) between images.

Results: Figure 1 shows the first eight images from the central slice in identical windowing. Pixel intensity decay between the first and second images results from RF depolarization and T_1 decay, while subsequent images contain an additional washout contribution. SNR of the first eight images was in the range from 133 to 8, allowing measurement of signal decay in different lung regions. Acquisition of the first two images at breath-hold permits compensation for RF and T_1 decay, as RF amplitudes and delays between images remain constant. A compensation factor map is obtained by division of the first two images, which can then be applied to subsequent images, leaving only washout of gas as source of signal loss. A "core and peel" ROI is depicted in the first image of Fig. 1, and Fig. 2 shows the evolution of the mean signal in the corresponding regions on a logarithmic scale after compensation for RF and T_1 decay for each ROI. Mono-exponential washout is observed in all ROIs, as expected for healthy lungs [7]. The apparent residual fraction of ^{3}He remaining in the lungs after each breathing cycle is 0.78 ± 0.02 (mean ± standard deviation over all ROIs), which is consistent with the global result 0.80 ± 0.04, observed with a global washout measurement on the same volunteer [6]. A linear pixel-by-pixel fit in MATLAB to the logarithm of the corrected intensities of images 2-8 yields the map of washout slopes depicted in Fig. 3, which shows the efficiency of inert gas washout directly on a regional basis.

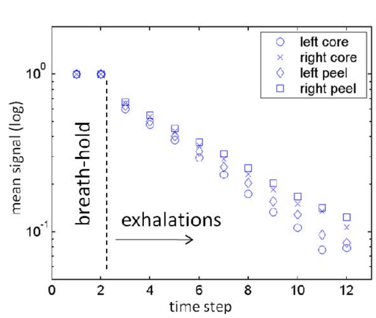

Figure 2: Mean signal evolution in the core/peel ROIs visible in Fig. 1.

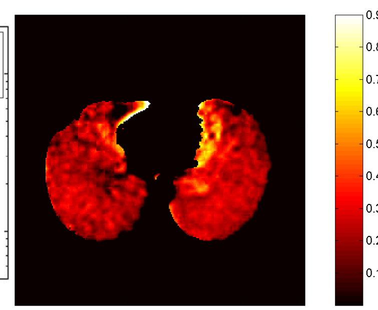

Figure 3: Map of the slope of the logarithmic signal evolution after compensation for RF and T_1 decay.

Discussion: Direct regional imaging of inert gas washout rates is feasible using ^{3}He MRI, as demonstrated in this work. Successful washout measurement depends on the ability of observing signal for a sufficient number of cycles. A flip angle of 2.9° was used to this effect, which results in reduced RF depolarization compared to the optimal flip angle of ~10° for a single image of the used matrix size. A constant delay between acquisitions permits a simple compensation for RF and T_1 decay, but substantial subject cooperation is required to ensure that only a single breathing cycle takes place during the delay. Alternatively respiratory gating could be used, precluding the decay compensation approach used here. The resulting map of washout rates in Fig. 3 is mostly homogeneous, as expected for healthy lungs. Inhomogeneous regions in the anterior lung and around the heart are likely to result from motion artefacts and mis-registration between images. Image registration algorithms could be used to improve the presented method further. Some additional imperfections arise from interaction between 2D slices, although care was taken to cover the whole lung with excitations. As such, this method is a promising approach to direct imaging of inert gas washout rates.

References: [1] Aurora et al., Resp. Physiol. Neurobiol. 148:125:139 (2005); [2] Horsley et al., Thorax 63:135-140 (2008); [3] Wellman et al., J. Nucl. Med. 51:646-653 (2010); [4] Deninger et al., Magn. Reson. Med. 48:223-232 (2002); [5] Deppe et al, ISMRM 17, 2186 (2009); [6] Deppe et al., Magn. Reson. Med., in press (2010); [7] West, Respiratory Physiology: The Essentials, Lippincott Williams & Wilkins 2003 **Acknowledgements:** Funding by UK EPSRC grant no. EP/D070252/1, polarizer support by GE