Single Scan 3D pO2 Mapping with Hyperpolarized 3He MRI

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Introduction 3D mapping of lung pO2 with HP ^{3}He has previously been shown to be less sensitive to errors caused by inter-slice gas diffusion that are inherent in 2D methods [1]. In this work, a single scan method of 3D mapping of lung pO2 with ^{3}He was developed to increase scan speed and reduce errors associated with breath-hold repeatability and mis-registration in multiple breath-hold 3D p02 mapping. The accuracy of the sequence is demonstrated with gas phantoms and in vivo experiments on a healthy normal.

Theory If a time series of *n* 3D images is acquired during the same breath-hold, then the n^{th} image has an intensity given by:

$$\ln(A_n/A_0(\cos\alpha)^{nN}) = \left(-\frac{1}{\xi}\int_0^{t_n} pO_2(t)dt\right) \quad \textbf{[1]}$$

Where α is the flip angle, *N* is the number of RF pulses applied per image acquisition ($N=N_Y N_Z$) and the term in brackets represents the T1 decay due to mixing with O_2, where ξ = 2.61 bar × s at 310 K [2,3].

In previous work [1], two sets of experiments were performed with different inter-image delay times, with the same α, allowing elimination of α from **[1]** to give the pO_2 as a function of the mean inter-image time. In the new single scan method, two images are acquired immediately after one another, allowing a B1 reference scan of α to be made, as proposed previously for a single scan 2D method [4]. The contribution from T1 decay between the 1st and 2nd images was minimized with use of a short TR sequence but was nevertheless accounted for in the α calculation by normalizing the second time point with $\exp(-\Delta t/T1_0)$, where a starting estimate of $T1_0$ =20 s was assumed (p_0 =0.13 bar in vivo). A set of T1 weighted 3D images (i= 3:n) were then acquired later at longer intervals when the T1 shortening of the signal due to mixing with oxygen has taken place –see Figure 1. With prior knowledge of α from the 3D B1 reference scan, the pO_2(t) can then be calculated from the later images.

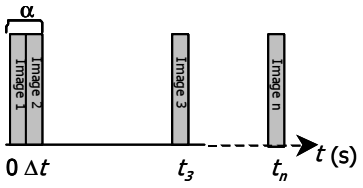

Fig. 1. Sequence timing diagram. α is first calculated from the first two images. T1 is then calculated from the remaining time course following elimination of α from Eq. 1

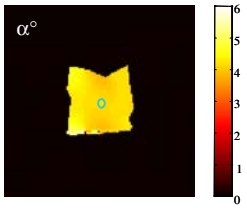

Fig. 2. α map of the bag phantom calculated from Image 1 & Image 2. The birdcage coil has uniform B1.

Materials & Methods All work was performed on a 1.5T whole body MRI system (Philips Eclipse) with a homebuilt low pass birdcage T-R resonator for the phantom experiments and a flex saddle T-R coil for the in vivo work (Medical Advances). The ^{3}He gas was polarized to 26% with rubidium spin exchange apparatus (GE Health). Phantom studies were performed using one litre volume Tedlar bags containing 800 ml N_2, 100 ml ^{3}He and 50 ml O_2 at 1 bar. The in vivo imaging was performed at breath-hold with a dose of 400 ml ^{3}He/600 ml N_2 following tidal breathing of room air.

MR Methods An optimised 3D gradient echo sequence [1] was used with a 300μs hard pulse. The sequence parameters were:

Phantom experiments - N*z* =8 pe views, N*y*=28 pe views, *Nx* =128 samples, BW 50 kHz, TR 4 ms, TE 2.2, flip angle of $\alpha \approx 4°$. n= 7: (t_1=0 s, t_2=1, t_3=15.5, t_4=21.5, t_5=27.5,t_6=33.5,t_7=39.5 s) The timing scheme of the phantom image time series is shown in Figure 1.

In vivo experiments - N*z* =12 pe views, N*y*=48 pe views, *Nx* =128 samples, BW 50 kHz, TR 4 ms, TE 2.2, flip angle of $\alpha \approx 2°$. n= 4: (t_1=0, t_2=2.3, t_3=15.5, t_4=21.5 s) –total breath-hold 22 seconds.

The pO_2 was then calculated on a pixel-pixel basis according to Eq.[**1**] using Matlab code which involved; median filtering of the magnitude images and correction for the initial T1 decay in the interval t_2 as described above.

Results

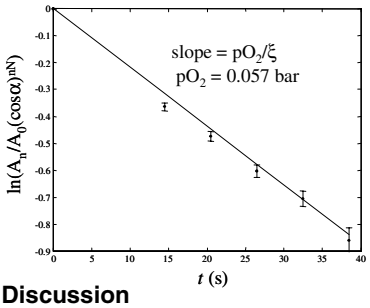

Fig. 3. Linear least squares fit to Eq. 1 from the circular ROI indicated in Fig.2. The closed phantom has a constant pO2 determined from the volume fraction (50/950) = 0.053 bar. The fitted result of 0.057 bar is in good agreement.

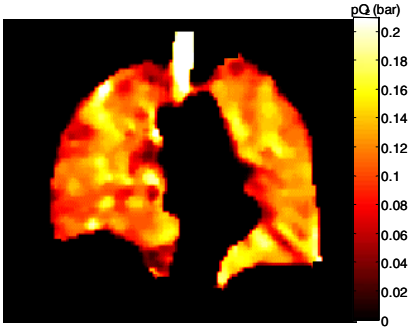

Fig. 4. in-vivo pO2 map derived from one of the 3D image slices in a healthy subject. The sequence was performed at breath-hold with a dose of 400 ml 3He/600 ml N2 following tidal breathing of room air.

Discussion

A single scan method of mapping pO2 in 3D has been shown to give accuracy to within 7% in phantom experiments. There is a margin of error in gas dispensing of at least ± 5 ml so this is as likely to be due to inaccurate knowledge of the composition of our gas phantom as it is to errors incurred by estimating the T1 decay between the first two images. The in vivo values for the initial pO_2 (p_0) are in agreement with those published previously with a 2 scan 3D method [4] and whole lung projection methods [3]. The 3D acquisition minimizes errors in the calculation of pO2 incurred by inter-slice diffusion mixing of gas polarization [1] that are inherent in 2D methods [4]. The ability to perform the examination in a single scan has obvious advantages in vivo where motion between breaths is inevitable, furthermore the method saves on a dose of HP ^{3}He. The initial 3D B1 mapping phase of the acquisition could be used as a reference scan for the subsequent oxygen sensitized images acquired with reduced phase encoded parallel imaging techniques or indeed for parallel phase encoded ventilation images acquired on subsequent breath-holds.

References [1] Wild JM et al. Magn Reson Med. 2005;53(5):1055-64. [2] Saam B et al. Phys. Rev. A. 1995 52(1):862-865. [3] Deninger AJ et al. J Magn Reson. 1999; 141:207-2161. [4] Fischer MC et al. Magn Reson Med. 2004;52:766-73.

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