Modelling of Air Flow in the Human Airways Using Computational Fluid Dynamics and Dynamic Hyperpolarized 3He MRI

B. Brook[1], A. Swift[1], M. N. Paley[1], L. Kasuboski[2], S. Fichele[1], N. Woodhouse[1], K. McCormack[1], R. Hose[1], E. van Beek[1], J. M. Wild[1]

[1]Academic Radiology, University of Sheffield, Sheffield, Yorkshire, United Kingdom, [2]Philips Medical Systems, Cleveland, OH, United States

Introduction Ultrafast imaging of ^{3}He gas has been shown to provide insight into ventilation dynamics in human lungs [1]. In this work we use quantitative time resolved radial projection imaging of ^{3}He [2] to make preliminary validations of computational fluid dynamics (CFD) models of gas flow in the human lungs. The results show good agreement with CFD flow models in the major airways of healthy normals in the inspiratory phase where depolarization due to oxygen can be discounted as a source of signal loss.

Materials All experiments were done on a 1.5T whole body system equiped with a parallel ^{3}He T-R circuit. ^{3}He gas was polarized to 30% by optical pumping with rubidium spin exchange apparatus (Amersham Health). All in-vivo imaging was performed during inhalation of a 300 ml ^{3}He/700 ml N_2 mixture from a Tedlar bag. In-vivo studies were performed with ethics committee approval and informed consent on a healthy subject.

Methods: MRI A dynamic radial projection sequence with sliding window reconstruction was used to image the inspiratory phase. The pulse sequence and data processing is described in detail in [1]. The sequence had the following parameters: 200 mm thick coronal slice, TR 5.4 ms, TE 2.26 ms, 128 views. A flip angle of 10^o was used to bias the delineation of the major airways as opposed to signal in the peripheral lung. Data processing was done with Matlab® code and incorporated a sliding window to generate a continuous time series of data.

Spirometry The flow rate was measured at the mouth while the subject breathed spontaneously from a bag. Several breaths for the same volunteer are plotted in Fig. 1 showing the range of variation for a single subject.

CFD A multi-component flow simulation was carried out using CFX (www.cophit.co.uk) whereby the Navier-Stokes equations are solved for a mixture of ^{3}He and N_2. The appropriate densities are used for each gas, but the viscosity for both gases was assumed to be similar to that of air. The volume fractions of ^{3}He (V_{He}) and N_2 (V_{N2}) are 0.3 and 0.7 respectively. The tracheo-bronchial tree was assumed to be filled with N_2 initially and that at the beginning of the breath there is no flow and pressure is zero everywhere.

Results

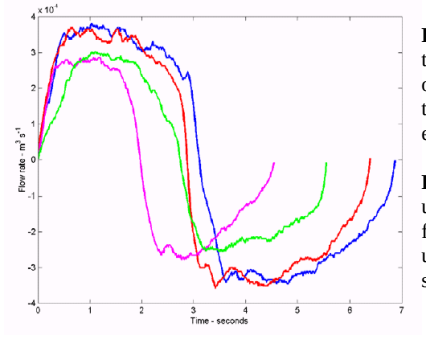

Fig. 1 Flow rate measured at the mouth on 4 separate occasions for a period of time the same as the MRI exam.

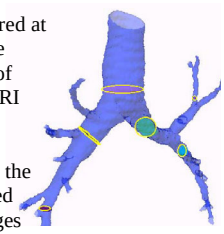

Fig. 2 Volume mesh of the upper airways calculated from high-res. CT images used to estimate the cross sectional areas *A*.

Boundary Conditions The measured flow rate is applied as a transient boundary condition in the form of a velocity normal to the plane at the inlet. Thus plug flow was assumed at this boundary and the normal velocity is calculated via $Q = uA$ where Q is the volume flow rate, u is the normal velocity and A is the cross-sectional area of the inlet at the start of each time step. For these simulations it was assumed that the outlets were open to atmosphere and thus a boundary condition of atmospheric pressure is assumed at the outlets. This is clearly an approximation and coupling of the outlets to the pressures determined from a Windkessel model of the lung will be employed in future simulations. The cross sectional area (A) at a given point was taken from 3D mesh models made from high resolution CT images (Fig.2), in further work we propose to use anatomical proton MRI data from the same volunteers.

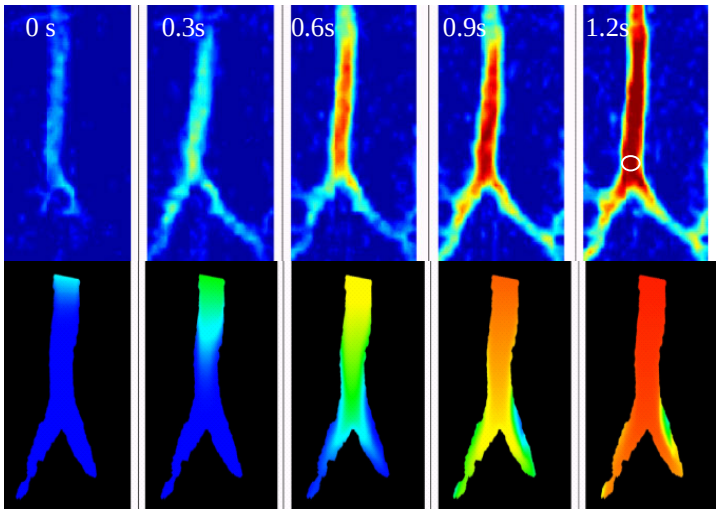

Fig. 3 Dynamic images at 300 ms intervals from the early stage of inspiration.

Fig. 4 Helium concentration calculated from CFD in a central plane through the tracheo-bronchial tree at the same times. A definite resemblance in the data exists despite the fact that the imaging data represents a projection whilst the CFD data represents a central plane through the major airways

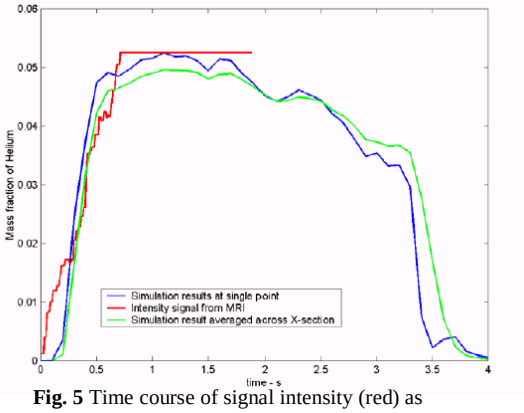

Fig. 5 Time course of signal intensity (red) as measured from the circular ROI circle shown in the 1.2 s image of Fig.3. The blue plot is the gas density as simulated at a point at the center of the circle. The green plot is the simulated data averaged over the average cross section A as measured from the mesh of Fig.2 above the apex.

Discussion We have shown that time resolved imaging of hyperpolarized ^{3}He with a radial sliding window sequence [1] can be used to track the passage of airflow in the major airways in the early inspiratory phase. Furthermore the experimental results in healthy normals are in agreement with CFD models based on the solution of the Navier-Stokes equations for an inhaled mixture of ^{3}He and N_2. Results observed further down the bronchial tree (not shown for sake of space) show less of an agreement which we attribute to two factors; firstly the need for more precise estimates of pressures within the bronchial tree to be derived from an appropriate Windkessel model. Secondly the MR signal in the peripheral lung is influenced by the depolarizing effect of O_2, which can be discounted in the early inspiratory phase in the trachea since the inhaled gas washes out residual O_2. In further work we will be looking to build up a more specific set of boundary conditions in terms of airway geometry and lung pressures in an attempt to deconvolve physiological flow related changes in signal intensity from RF depolarization, the effects of O_2 and diffusion all of which significantly influence the dynamic time course of the ^{3}He signal.

References **[1]** Salerno M, et al MRM 2001; 46:667-77 **[2]** Wild JM et al Magn. Reson. Med 2003 49(6):991-7.