Quantitative perfusion and capillary permeability measurements in lung parenchyma using T_1-weighted DCE-MRI

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Introduction

Measurements of lung perfusion are important for understanding and monitoring conditions such as COPD. Previously we have assessed perfusion parameters in lung tumours[1]. Here we examine the non-tumour-bearing lung using the same analysis techniques to asses the applicability of the adiabatic approximation to the tissue homogeneity model[2] (AATH) in this setting. This model allows estimates of blood flow (F) contrast agent extraction fraction (E), distribution volume (v_e), capillary blood volume (v_b), and transit time (τ), and capillary permeability surface area product PS.

Methods

Data were acquired in 3 patients presenting with lung tumours on a Philips Gyroscan NT Intera system at 1.5 Tesla. The DCE-MRI protocol consisted of 3D gradient echo (FFE) acquisitions with a temporal resolution of 4 s. Native tissue T_1 was determined using 3 separate acquisitions with 4 averages and flip angles of 2^0, 10^0, and 30^0; TR = 3.5 ms; TE = 0.99 ms. The dynamic acquisition consisted of 100 single average volumes (flip angle 30^0, TR = 3.5 ms; TE = 0.99 ms). Image matrix for all scans was 128 x 128 in-plane, with 25 slices acquired using over-contiguous slicing. An elliptical k-space window was utilised to maintain a short total acquisition time. The body resonator was used for RF transmission and reception. 0.1 mmol/kg Omniscan (Amersham Health, Amersham, UK) was administered as a bolus using a power injector at a rate of 2 ml/s.

To reduce the detrimental effects of motion on the time series analysis we utilise an affine registration scheme with 12 degrees of freedom available in FSL version 1.3 (available at www.fmrib.ox.ac.uk/fsl). Significant chest wall and bulk motion was removed with this approach[1].

Custom-built software allowed analysis of the dynamic time series to provide estimates of microvascular parameters. The AATH model was fitted using the simplex method[3,4]. Fitting starting values for flow (F) and distribution volume (v_e) were defined from the output of a simpler kinetic model[5] ("extended Tofts model"[6,7]); E was initialised at 0.3; multiple initial values for τ were defined to reduce fitting instability[5,8]. Haematocrit was assumed to be 0.42. A tissue/air volume ratio of 0.1 was assumed to allow conversion of parameters measured in MR-visible tissue to measurements per unit volume of lung. The arterial input function was determined using an automated algorithm[9]. We defined regions of interest in normal appearing non-tumour-bearing lung in the right upper lobe, as this region is less prone to motion than the lower lung.

Results

Figure 1 shows the concentration time series plus fits to the data using the AATH model; table 1 shows fitted parameter values.

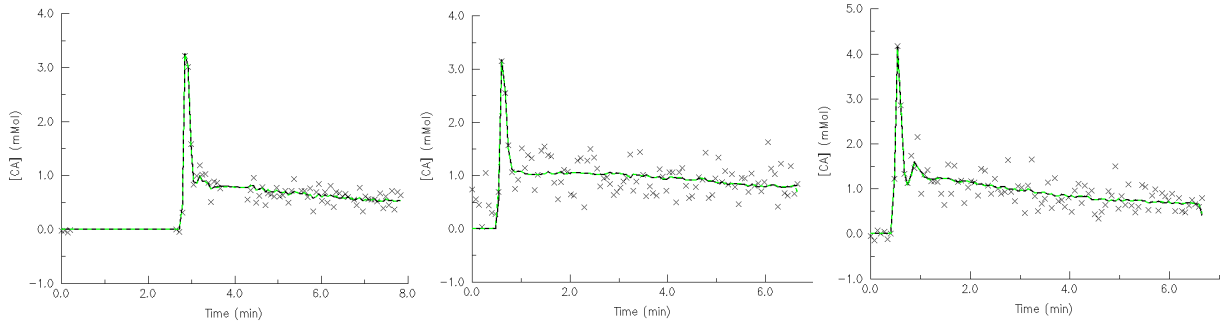

Figure 1. *Fits of AATH model to contrast agent concentration time courses in upper right lobe ROIs in 3 patients. Output from extended Tofts model used for AATH fitting initialisation.*

Patient	v_e †	v_b †	τ (s)	E	F † (ml.ml^{-1}.min^{-1})	PS (ml.g^{-1}.min^{-1})	$K^{trans} = EF\rho$(1-Hct) (min^{-1})
1	1.5 %	5.3 %	1.1	0.0048	1.68	0.081	0.047
2	3.2 %	4.7 %	2.4	0.021	0.67	0.14	0.079
3	2.1 %	6.6 %	1.3	0.012	1.71	0.21	0.122

Table 1. *Parameters derived from AATH fit to concentration time courses.* †*assuming 90 % air space.*

Conclusion

Our acquisition protocol has short TE, thus reducing signal loss due to T_2* effects. The short TR, in combination with the elliptical k-space sampling scheme, enables rapid volumetric data acquisition. Our values of flow are within the range of previously published values in the upper parts of the normal lung[10]. However, the values of blood volume are slightly lower than some previous studies, which may reflect sub-clinical disease in the normal appearing lung in the patients studied. Our estimates of v_p, v_e, and F are affected by the assumed tissue/air volume ratio, which may not be accurate, but which could be measured independently. The AATH model allows more potentially useful parameters to be defined than simpler kinetic models, which may be of benefit in the study of COPD and other conditions.

References

1. Parker, G.J., *et al.*, Proc. Int. Soc. Magn. Reson. Med., Toronto, 2003.
2. St. Lawrence, K.S. and Lee, T.-Y., *J. Cereb. Blood Flow. Metabol.*, 18, 1365, 1998.
3. Press, W.H., *et al.*, *Numerical recipes in C: The art of scientific computing*, 2 ed. Cambridge: Cambridge University Press, 1992.
4. Buckley, D.L., *et al.*, *Magn. Reson. Med.*, 32, 646, 1994.
5. Buckley, D.L., *Magn. Reson. Med.*, 47, 601, 2002.
6. Tofts, P.S., *et al.*, *J. Magn. Reson. Imag.*, 10, 223, 1999.
7. Daldrup, H., E., *et al.*, *Magn. Reson. Med.*, 40, 537, 1998.
8. Henderson, E., *et al.*, *J. Magn. Reson. Imag.*, 12, 991, 2000.
9. Parker, G.J., *et al.*, Proc. Int. Soc. Magn. Reson. Med., Toronto, 2003.
10. Ohno, Y., *et al.*, Proc. Int. Soc. Magn. Reson. Med., Toronto, 2003.

Acknowledgements This work was supported by AstraZeneca.