A dynamic contrast-enhanced MRI flow phantom to validate arterial input function measurements

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Introduction It is generally assumed that individual measurement of the arterial input function (AIF), when compared with the use of assumed or population averaged AIFs (1), improves the precision and accuracy of measurements of the dynamic contrast-enhanced MRI (DCE-MRI) kinetic modelling parameters. However, it is difficult to sample the first pass peak of the AIF to a sufficient temporal resolution in the presence of competing constraints for acquisition parameters such as the need for a high signal to noise ratio (SNR), good spatial resolution, and volume coverage (2, 3). Flow phantoms have been developed previously for the investigation of AIFs, such as that by Ivancevic *et al* (4). Our method has the potential to compare and contrast the efficacy of different AIF acquisition protocols by providing a 'gold standard' AIF measurement via the independent high temporal resolution optical measurement of a first pass peak of contrast agent.

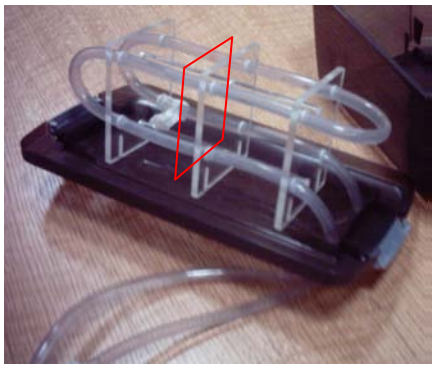

Fig 1: Internal configuration of flow phantom with slice of interest highlighted

Materials and Methods PVC tubing (length 20 m, external/internal diameter 14/10 mm) was arranged within a rigid phantom (Fig. 1) and attached to an impeller pump (Grant Instruments, Cambridge UK) which delivered a flow rate of 56.6-57.8 cm/s. The phantom was bathed in water doped so that T_1 = 0.83 s and water doped to the same T_1 was circulated around the system. Omniscan 0.5 mmol/ml (gadodiamide, GE Healthcare) was manually injected at a rate of approximately 2 ml/s as a well-mixed 1 ml bolus containing 0.5 ml contrast agent and 0.5 ml visible dye (Supercook Black Food Colouring, Supercook, Leeds, UK). The resultant changes of concentration of bolus were measured simultaneously using DCE-MRI at the phantom and a light dependent resistor (LDR) in the scanner control room. The MR scans were conducted on a 1.5 T Philips Intera system (Philips, Best, Netherlands) using a 3D volume of 25 slices, FOV 165 mm × 165 mm, matrix size 128 × 128, slice thickness 4 mm, TR = 4 ms, TE = 1.02 ms for all acquisitions and a Synergy body coil. Baseline T_1 measurement was performed using a variable flip angle Fast Field Echo (FFE; spoiled gradient echo) acquisition with flip angles 2°, 10° and 20°. The dynamic series used the same sequence with flip angle 20° and temporal resolution 4.97 s with a total of 42 volume acquisitions. MR-derived concentration measurements were taken from the slice shown in Fig. 1 and were evaluated using the method described by Li *et al* (5), assuming a relaxivity of 4.5 $s^{-1}mM^{-1}$. The LDR-derived concentration measurements were made at an average temporal resolution of 2.08 ms in the control room using a 3 cd LED with a dominant wavelength of 625 nm (Part AGI-3N4TR, AgiLight Inc, Texas US). The LDR signal was recorded using a DrDAQ data logger (Pico Technology, Cambridge, UK) over a period of 210 s. A calibration related the LDR signal output to concentration levels of the black dye (and therefore of the contrast agent in the mixed bolus) using measurements of known concentrations.

Results and Discussion Figure 2 shows a T_1-weighted acquisition from the slice of interest within the FFE volume. Dynamic MR-derived concentration measurements were made in each of the tubes shown in this slice and Fig. 3 illustrates the data from acquisition site (b) overlaid with the values taken from the LDR readings. The MR-derived concentrations appear to be overestimated marginally, possibly due to inflow effects or due to an inaccurate estimate of relaxivity. The LDR measurements have the same form as the MR data and the higher temporal resolution LDR-derived concentrations better define the first pass peak. Figure 4 shows a scatterplot of MR-derived concentration readings against LDR-derived concentration readings for 39 samples made at the same time points. A regression line (R^2 = 0.975) emphasises the good correspondence between the MR and optical data, although the measurements do not follow the line of identity due to the systematic differences noted in Fig. 3. The clustering of the points near the origin demonstrates the paucity of samples representing measurements of relatively high concentration.

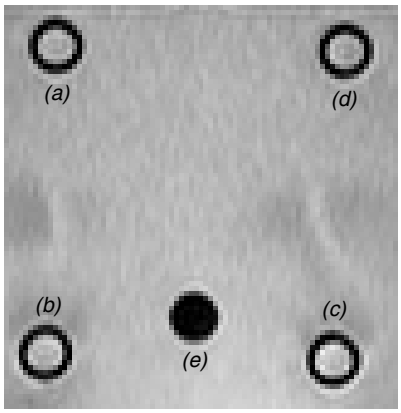

Fig 2: Axial slice of phantom showing potential acquisition sites (a)-(d) (labelled in order of flow) and internal phantom support (e)

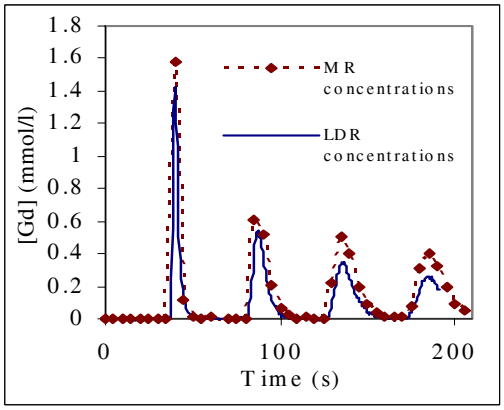

Fig 3: Time series of MR-derived concentration measurements vs independent LDR-derived concentration measurements

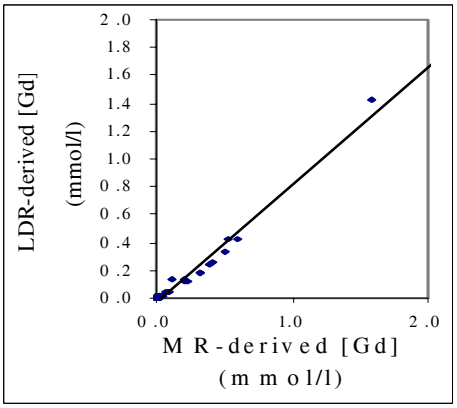

Fig 4: Scatterplot of LDR concentration measurements against each MR measurement

Conclusion The measurement technique introduced here provides a way to evaluate independently the accuracy of MR-based AIF acquisition techniques against independent optical measurements that approach a 'gold standard'. It may be used to compare alternative approaches to acquisition or analysis and to avoid, or assess the impact of, confounds specific to MR such as inflow effects. This will allow AIF acquisition strategies to be developed appropriate to the constraints present in varying clinical settings.

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References (1) Parker, G.J., *et al.*, *Magn. Reson. Med.*, 56:993-1000, 2006. (2) Henderson, E., *et al.*, *Magn. Reson. Imag.*, 16:1057-1073, 1998. (3) Roberts, C., *et al.*, *Magn. Reson. Med.*, 56:611-619, 2006. (4) Ivancevic, M.K., *et al.*, *Magn. Reson. Med.*, 50:885-891, 2003. (5) Li K.-L., *et al.*, *J. Magn. Reson. Imag.*, 12:347-357, 2000.