Pre-bolus quantification of arterial input functions by non-steady-state analysis of gradient-echo dynamic imaging

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Introduction

Reproducible perfusion estimates from DCE-MRI data are heavily contingent on reproducible estimation of the arterial input function (AIF), and it is well known that patient-specific AIFs are necessary to achieve optimal estimates[1]. Measuring the AIF is one solution, and this is often achieved by finding a vessel within the dynamic imaging field-of-view (FOV), but this presents several difficulties. The tumour location often means it is impossible to capture a sufficiently large vessel within the FOV, and optimising the dynamic imaging sequence to give good performance for both the tumour and a nearby vessel is difficult as these have contradictory temporal resolution, spatial resolution and signal linearity requirements.

The idea of using a pre-bolus[2] before the main dynamic acquisition has been suggested as a solution to this problem, since the imaging parameters can be specifically optimised, without compromising the tumour imaging. Here we explore the use of a single slice gradient-echo sequence to capture the concentration changes in the descending aorta during the administration of a fraction (here $1/10^{th}$) of a standard dose of contrast agent. Each image is acquired in less than 1 second, so the usual assumption that the magnetisation has reached a steady state may not be appropriate. In this abstract we explore non-steady and steady state modelling, and demonstrate the effect of these models on the estimated AIF.

Methods

Data Acquisition Protocol

Data were acquired on a 1.5T Siemens Avanto using a 2D FFE sequence under free-breathing, and a single 20mm thick coronal-oblique slice was positioned to capture the descending aorta. The imaging parameters were TR/TE = 5.5/1.21 ms, FOV = 440 mm, 128×128 matrix, giving an acquisition time of 0.704 sec per image. To enable T_1 estimation and improve SNR, 60 pre-contrast images were acquired with a flip-angle of 3°, while the dynamic acquisition consisted of 170 images (2 minutes in total) with a flip angle of 20°. At the start of the dynamic acquisition 0.01 ml/kg of Magnevist contrast agent was delivered by power injector at 3 ml/sec, followed by 20 ml of saline. ROIs were drawn in regions at the lower end of the aorta to minimise the effect of pulsatile flow and breathing motion.

Signal Model

We consider two signal models. The first assumes that the magnetization of the blood reaches a steady state, and in this case the pre-contrast T_1 is estimated from the data. The second assumes that the image contrast of the blood is generated by protons receiving N excitations. In this case N is estimated, but the pre-contrast T_1 cannot be simultaneously determined, so a value of 1300 ms is used for a field strength of 1.5T[3]. The steady state model equation is $S = S_0 \sin(\alpha)(1 - E_1)(1 - \cos(\alpha)E_1)^{-1}$, where $E_1 = \exp(-TR/T_1)$, and the non-steady state equation is $S = S_0 \sin(\alpha)(1 - E_1)(1 - (\cos(\alpha)E_1)^N)(1 - \cos(\alpha)E_1)^{-1} + S_0 \sin(\alpha)(\cos(\alpha)E_1)^N$, which tends towards the steady state equation as $N \to \infty$. The change in T_1 is related to the contrast concentration C via $1/T_1 = 1/T_{10} + r_1 C$, where the relaxivity is taken to be $r_1 = 4.3$ s^{-1}mM^{-1}. The concentration time curve is modelled using a functional form based on a raised-cosine that is defined by four parameters[4]; a_b and μ_b describe the first-pass component, a_e and μ_e describe the equilibration phase. Parameter estimation is via least-squares fitting using the residuals between the acquired signals and the signals predicted by the relevant model; a_b, μ_b, a_e, μ_e, S_0 and τ_0 (onset time of contrast enhancement) are estimated for both models, while T_{10} is estimated for the steady state model, and N is estimated for the non-steady state model. Thirteen cases were modelled and results are presented below.

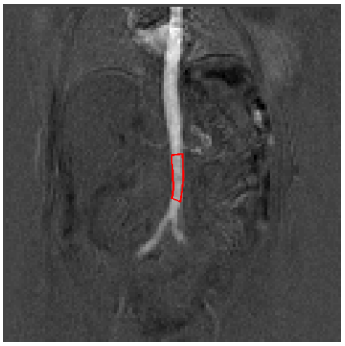

Figure 1 : subtraction image at peak aortic enhancement with ROI shown.

Results and Discussion

Figure 1 shows a typical example of a subtraction image at peak aortic enhancement, including the ROI used for the analysis. Figure 2 shows the corresponding data and fit (top panel) and the inferred contrast agent concentration curves for the two methods (bottom panel). In this instance the estimated number of excitations was 26, while the pre-contrast T_1 estimate for the steady-state analysis was 327 ms. For the 13 cases studied, the average number of excitations estimated was 27.5, range 25-33, while the average pre-contrast T_1 was 359ms, range 308-481 ms. Using the steady-state gradient-echo equation to analyse the data in this way is clearly inappropriate since the native T_1 of blood is around 1300ms. The non-steady state model uses a fixed native T_1 for blood that will be much closer to the truth than the steady state estimates. Simple calculations based on typical blood flow velocities and the TR of the sequence indicate that the estimated number of excitations is plausible, but this needs further testing as the pulsatile nature of the flow makes this calculation quite complex. In all cases the steady state analysis under-estimated the AIF relative to the non-steady state analysis, with average peak value reductions of 25.6%, range 24-28%, and reductions in the value estimated at 2 min of 26.6%, range 23-36%. The consistency between the 13 cases of these statistics, and the estimates of T_1 and N, is of note.

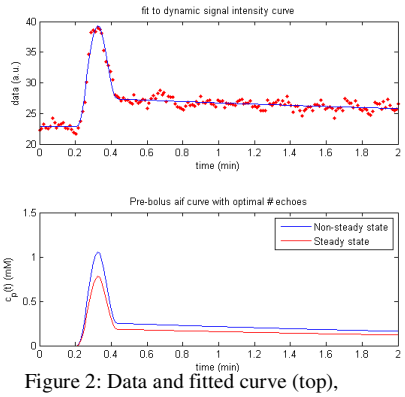

Figure 2: Data and fitted curve (top), inferred AIF curves (bottom).

The pre-bolus measurements will be used to provide a proxy for the full-dose AIF. Assuming the contrast injection flow rates for the pre-bolus and the main DCE-MRI acquisition are the same, this can be achieved by convolving the pre-bolus curve with a rectangular function of height $1/\Delta$ and duration 10Δ, where Δ is the duration of the pre-bolus injection. This has been shown by other workers to give an accurate estimate of the full-dose AIF when the pre-bolus injection duration is much shorter than the duration of the first-pass bolus[2]. In our case, assuming a maximum patient weight of 150kg, the corresponding pre-bolus volume is 3ml, and at 3ml/sec the injection lasts 1 sec, whereas the first-pass bolus lasts around 12 sec, see figure 2, so the constraint is clearly satisfied. With a model-based AIF description the convolution can be implemented either directly on the model-fitted curve, or by a suitable manipulation of the resulting AIF parameters; the best approach is the subject of further work.

Conclusions

Non-steady state analysis of high temporal resolution gradient-echo pre-bolus data has the potential to provide accurate patient-specific descriptions of the arterial input function. Further work is needed to validate this approach, and to assess the impact of using AIFs derived in this way on the tumour perfusion parameter estimates.

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References

[1]Port RE, Knopp MV, Brix G. *Magn Reson Med.* 2001; **45**(6):1030–1038.
[2]Köstler H, Ritter C, Lipp M, Beer M, Hahn D, Sandstede J. *Magn Reson Med.* 2004; **52**(2):296-9.
[3]Stanisz GJ, Odrobina EE, Pun J, Escaravage M, Graham SJ, Bronskill MJ, Henkelman RM. *Magn Reson Med.* 2005; **54**(3):507-12.
[4]Orton MR, d'Arcy JA, Walker-Samuel S, Hawkes DJ, Atkinson D, Collins DJ, Leach MO. *Phys Med Biol.* 2008; **53**(5):1225-39.