Investigating the influence of phased arrays and parallel imaging for input function definition in DCE-MRI

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Introduction Physiologically-meaningful dynamic contrast-enhanced MRI (DCE-MRI) measurements of microvascular characteristics are not possible without an accurately-measured arterial input function (AIF). Various competing factors limit our ability to directly measure the AIF, including constraints on our temporal resolution, which leads to serious undersampling errors (1). The use of phased array coils allows improved SNR and/or spatial resolution over quadrature coils. If used in conjunction with parallel imaging methods such as SENSE it is also possible to reduce image acquisition time and therefore improve temporal resolution but often with an SNR penalty. This study used a flow phantom to take MR measurements of rapid changes in contrast agent concentration typical of an AIF using a quadrature (Q) body coil, phased array coil and the same phased array coil with SENSE for comparison against independent high-temporal resolution optical measurements to investigate whether any cost in terms of accuracy or precision could be established.

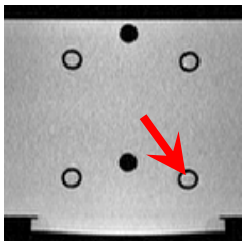

Figure 1: Axial slice of phantom using phased array coil showing measurement location.

Materials and Methods A closed circuit flow phantom as described in (2) was used with water doped with Omniscan 0.5 mmol/ml (gadodiamide, GE Healthcare) to reduce the T_1 to 1.5 s (approximately that of blood). A manual injection of a 1 ml well-mixed bolus containing equal volumes of Omniscan and visible dye (Supercook Black Food Colouring, Supercook, Leeds, UK) was made at a rate of ~2 ml/s. DCE-MRI measurements of concentration were taken at the core of the phantom for comparison with optical measurements taken outside the magnet bore. The DCE-MRI data were acquired using a 1.5 T Philips Intera system (Philips, Best, NL) and a 3D volume of 25 slices, 4 mm thick with FOV 165 mm square, a 128 × 128 matrix, TR of 4.0 ms and TE of 1.0 ms. Baseline variable flip angle measurements of T_1 were made using a Fast Field Echo (spoiled gradient echo) acquisition with flip angles 2°, 10° and 20°. These were followed by dynamic series using the same sequence with a 20° flip angle. The Q body coil and body phased array coil scans had temporal resolution 5.0 s and 45 time point acquisitions. The final acquisition used the phased array coil with SENSE factor 2 giving a temporal resolution of 2.6 s over 90 time point acquisitions. MR-derived concentrations were measured at the 20th slice in the direction of flow for the section of tubing indicated in Figure 1 using automatically extracted regions of interest (following a procedure similar to that described in (3)) then converted into mean concentrations for each location using the procedure outlined in (4) with the relaxivity as measured during the experiment as 5.4 $s^{-1}mM^{-1}$. SNR measurements were taken from uniform regions of interest within the static portion of the phantom in the central slice of each volume at a pre-bolus time point for each respective scan. Optical measurements were taken using a light dependent resistor onboard a DrDAQ data logger (Pico Technology, Cambridge, UK) and converted into concentrations using a model of the pre-scan calibration as shown in Figure 2 (2).

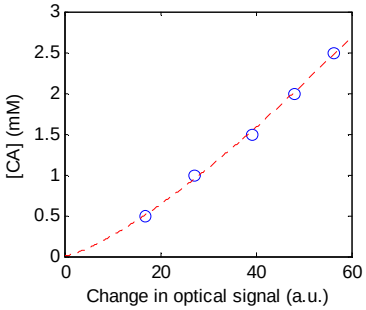

Figure 2: Pre-scan calibration relating change in optical signal to contrast agent concentration (circles), with fitted model (dashes) $y = 0.012x^{1.3218}$.

Results and Discussion Figure 3 shows optical measurements of contrast agent concentration which are aligned to overlay the MR measurements using a least squares fit. The four peaks correspond to four passes of the bolus through the entire flow phantom. Later recirculation peaks are more likely to illustrate inflow effects than the first peak for each of the protocols, which tends to demonstrate undersampling as a result of the more rapid change in contrast agent concentration. The sum of squares errors for the three alignments were 0.12 mM^2 (Q body coil, 45 points), 0.11 mM^2 (phased array, 45 points) and 0.23 mM^2 (phased array plus SENSE, 90 points). This suggests there is not much variation in error between the techniques but the SENSE acquisition appears to more precisely follow the changes in concentration as a result of the improved temporal resolution. Dynamic image SNR was measured as 2.9 (Q body coil), 5.5 (phased array) and 3.0 (phased array plus SENSE), indicating a large SNR penalty in using SENSE. However the resultant SNR is comparable to that obtained when using the Q body coil. The ability to compare the MR measurements against independent optical measurements is a useful application of this system.

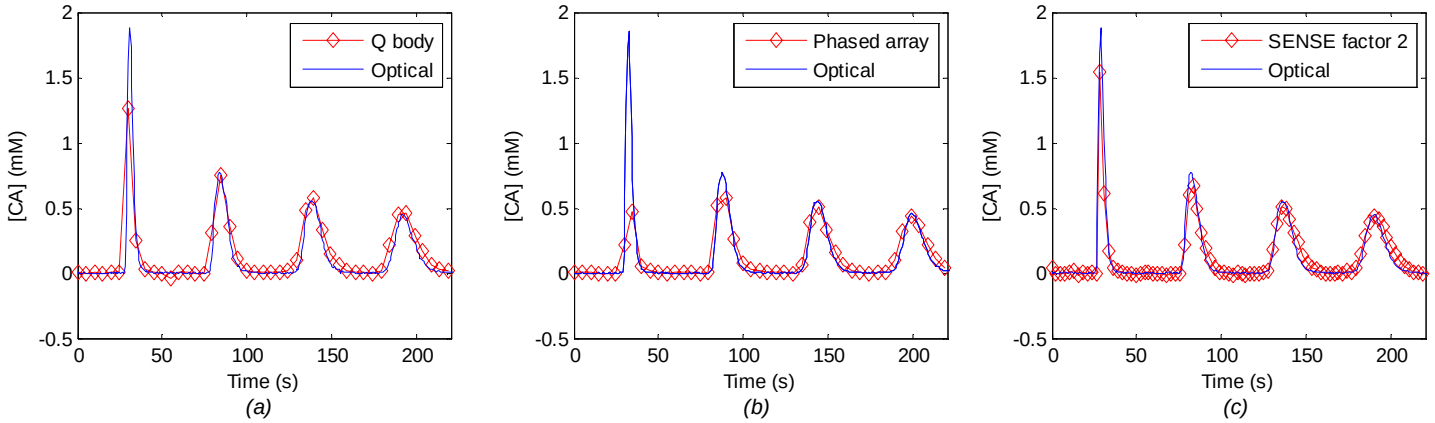

Figure 3: Results of optically-measured concentrations of contrast agent plotted with MR measurements taken from the phantom using (a) a Q body coil, (b) a phased array body coil and (c) the phased array body coil with SENSE factor 2.

Conclusion The use of SENSE with our flow phantom allows comparable accuracy in contrast agent concentration measurement to the use of the Q body coil, as demonstrated using independent optical measurements, but at higher temporal resolution. This allows more accurate sampling of the rapid changes in concentration associated with AIFs. While SNR is degraded when using parallel imaging, a SENSE factor of 2 provides similar SNR to the use of the Q body coil but at twice the temporal resolution.

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References (1) Henderson, E., *et al.*, *Magn. Reson. Imag.*, 16:1057-1073, 1998. (2) Little, R.A., *et al.*, *Proc of 15th ISMRM*, 2985, 2007. (3) Parker, G.J.M., *et al.*, *Magn. Reson. Med.*, 56:993-1000, 2006. (4) Li K., *et al.*, *J. Magn. Reson. Imag.*, 12:347-357, 2000.