Improving DSC-MRI by Orientation-corrected Phase-based AIF and VOF

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Purpose: Quantitative DSC-MRI perfusion imaging requires signal values that can be accurately converted to tracer concentration values. Whilst tracer concentration can accurately determined for tissue [1], measuring vascular concentration (e.g. arterial input function AIF and venous output function VOF) remain challenging and probably the leading source for incorrect quantification. Conventional single-shot GRE-EPI suffers from clipping of the first-pass peak. Moreover, typical TRs are <=$T1_{blood}$ and T1-shortening counteracts the expected T_2* effect, which in turn make quantification of vascular concentration values difficult. Yet another problem for accurate quantification is the non-linear relationship [2] between intravascular tracer concentration and ΔT_2*. It is well known, though, that susceptibility agents induce a change in the resonance frequency [2] that is linear with the tracer concentration and does not depend on T_1. This change in frequency can be assessed by change in MR signal phase and could potentially deliver better estimates of tracer concentration. The observable phase effect depends however on the orientation of the vessels relative B_0 (Eq. 1) and information about vessel orientation is thus warranted. To avoid extra cumbersome measurements we propose to estimate the vessel orientation from the magnitude data of the dynamic EPI scan itself.

Methods: Intravascular concentration of paramagnetic tracer can be estimated [2] using Eq. 1, where θ is the tilt angle of the vessel relative to B_0, t_{TE} is the echo time, $\Delta\varphi$ is the tracer-induced change of phase relative to the background phase, ρ is molar susceptibility of gadolinium (ρ = 0.026 ppm/mM in cgs units), and k is a currently unknown scaling factor (empirically determined k = 0.02). To estimate θ, so called tubular-filtering [3] was used on the maximum of R_2*-shortening values derived from magnitude signals in DSC-MRI data. The latter can be seen as a 'poor-man's 3D angiography'. Tubular filtering of such a dataset yields an estimate of the vessels orientation. The tubular filter is inherently robust against noise and suppresses non-cylindrical structures (typically imaging artifacts). It also suppresses objects of sizes much larger or much smaller than the selected kernel size. Experimental verification was performed on MR test data using multi-echo GRE-EPI (PERMEATE) sequence [1], TR=1.225s, matrix 96x96, TE1,2,3 = 15;34;51ms, FOV=24cm, 15 slices, slice thickness 5mm, gap 2mm, B_0= 1.5T, and an in-house written data reconstruction.

$$c_b = k\frac{\Delta\varphi}{\rho t_{TE} \gamma B_0\left(\cos^2\theta - \frac{1}{3}\right)} \quad \text{Eq. 1}$$

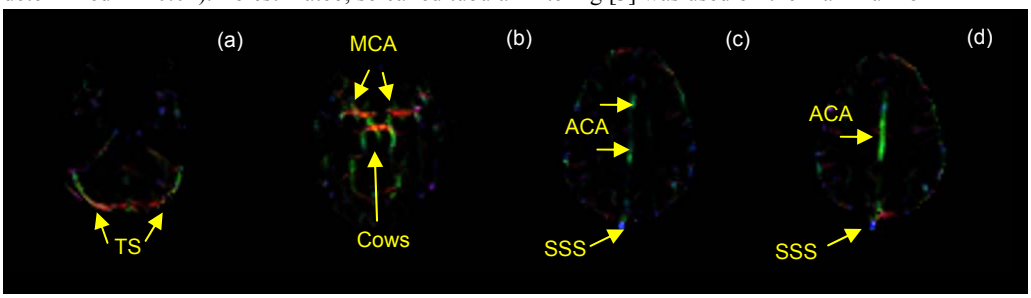

Fig. 1: Vessels and their orientation as detected with the tubular filter. Red indicates orientation in *x* direction, green in *y* direction and blue in *z* direction. (a) transverse sinuses (TS), (b) middle cerebral artery (MCA) and Circle of Willis (CoW). (c),(d) cerebral artery (ACA) and superior sagittal sinus (SSS).

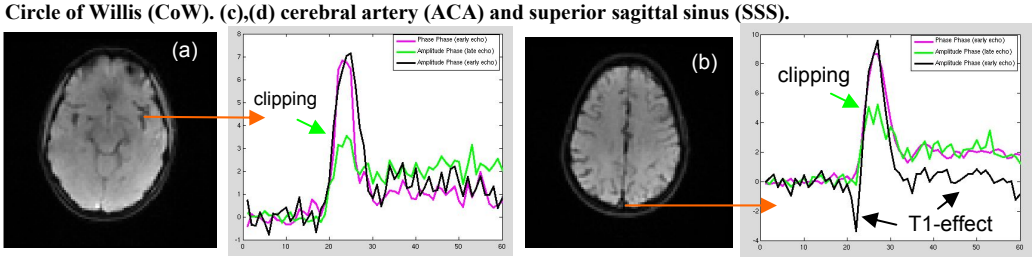

Fig. 2: (a) Vessel oriented perpendicular to B_0 (e.g. AIF in MCA) and (b) parallel to B_0 (e.g. VOF in SSS). Plots of concentration signals obtained at locations with different vessel orientations. The plots show estimated concentration during bolus passage using phase-signal from early echo (purple), and magnitude signals from late echo (green) and early echo (black). Note the clipped peak in the time course derived from the late echo as well as the T_1-induced underestimation of the concentration derived from the early echo.

Results: Fig. 1 shows an example of the major intracranial vessels and their orientation relative to B_0 as detected by the tubular filter. In Fig. 2 vascular gadolinium concentration values are shown that were obtained either from phase (first echo) or magnitude data (first and third echo). Despite that the concentration signals derived from MR magnitude data were corrected for non-linear tracer effect in bulk blood [2], concentration values for vessels running either parallel or perpendicular to B_0 were different when derived from magnitude data. Moreover, the T1-dependency of short-TE magnitude data at early bolus arrival can be clearly seen.

Discussion: A new method for improving the estimation of vascular Gd concentration has been proposed. One prerequisite is however the availability of complex data. Of note is also that partial voluming should be avoided to assure the validity of the phase relationship. Also, the magic angle effect has to be considered, i.e. phase changes will vanish for = 54.7° and thus the method proposed herein cannot be used for vessels oriented at this angle to B_0. Since phase information has a generally a lower contrast-to-noise ratio (CNR) than the magnitude counterpart it is therefore advisable to select vessels oriented either parallel or perpendicular to B_0. Major advantages of the phase-based approach are its immunity to log-Ricean transformed noise (present in perfusion signals when computed in magnitude data) and immunity to T1-artifacts as well as linearity of signal with respect to Gd concentration.

References:

[1] Newbould R.D. et al: MRM 58:70-81, 2007
[2] Kjolby B.F. et al: MRM 56:187-197, 2006
[3] Frangi A. et al: Multiscale Vessel Enhancement Filtering, Proceedings of MICCAI 1998

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