Improved Venous Output Function using MR Signal Phase for Quantitative 2D DCE-MRI in Human Brain

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Introduction: Quantitative dynamic contrast-enhanced (DCE) MRI in the human brain provides valuable diagnostic information (1). For quantitative DCE-MRI, the contrast agent concentration-vs-time ([C](t)) in the superior sagittal sinus (SSS) gives the venous output function (VOF). The VOF can be used to correct partial volume artifacts in the arterial input function (AIF), which is crucial for accurate estimation of perfusion parameters (2,3). Unfortunately, measuring the VOF with MR signal magnitude (|S|) can be difficult due to inflow, especially for multislice (2D) sequences, and saturation of |S| at high [C] (4-6). Some researchers have been investigating the use of MR signal phase (φ) for measuring the VOF and/or AIF for quantitative DCE-MRI (5-8). φ is linear and non-saturating with [C] (9-11); it is relatively insensitive to blood flow (12), partial volume effects (5), and flip angle variations (5-13); and it typically has a greater SNR than |S| (13). It is therefore hypothesized that investigating the use of MR signal phase (φ) for measuring the VOF and/or AIF for quantitative DCE-MRI (5-8).

Methods: Raw data were saved from twenty-eight 2D DCE-MRI studies performed during routine, clinical, Gd-enhanced brain exams (1.5T Siemens Symphony). A spoiled gradient echo sequence was used with the following parameters: TR=45 ms, double TE = 2.06 and 5.48 ms, flip = 90°, four 5.5 mm-thick transverse slices (2.75 mm gap), temporal resolution = 2.2 s, Gd dose = 0.07-0.1 mmol/kg. An ROI was drawn inside the SSS of each slice, providing |S|(t) and φ(t). VOF was computed from |S|(t) using standard signal magnitude, extrapolating to TE = 0 ms and assuming T1=1250 ms (5-8,14). VOF was computed from φ(t) (TE=5.48 ms), accounting for the angle of the segment of SSS with respect to the main magnetic field (5-13). The peak amplitude, area-under-the-curve up to 30 seconds (AUC30), and washout amplitude (mean from 80 to 100 seconds) were computed for each VOF.

Results and Discussion: Figs 1a and 1b show, for one study, whole-blood VOF and VOF for a function of slice (inferior-superior). The peak amplitude of VOF varied significantly as a function of slice location (1-way ANOVA, p<0.001) whereas that of VOF did not (p=0.9). This likely reflects the insensitivity of φ to inflow and partial volumes, compared to |S|. Therefore, only the slice with max VOF was used for the final VOF calculation, whereas the average of all slices was used for the final VOF calculation. Fig 1c and Table 1 show average and study-to-study variation of VOF and VOF as well as comparison with a recently published population-based AIF (14), which should have characteristics similar to a VOF. VOF had a smaller coefficient of variation in peak, AUC, and washout than VOF (f-test, p<0.03) and also resembled the pop. AIF much more closely.

Conclusion: For 2D DCE-MRI in human brain, phase-derived VOFs are more precise and more accurate than magnitude-derived VOFs.

Table 1

<table>
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<tr>
<th>VOF</th>
<th>AUC30 (mM s)</th>
<th>Washout (mM)</th>
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<tbody>
<tr>
<td>1.5 ± 0.9</td>
<td>300 ± 180</td>
<td>0.4 ± 0.2</td>
</tr>
<tr>
<td>6 ± 2</td>
<td>900 ± 300</td>
<td>0.9 ± 0.3</td>
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References:
1. Espig et al. Topics MRI ‘06; 17: 89.
2. Østergaard. JMRI ‘05; 22: 710.
10. Akbudak et al. MRM ‘96; 36: 809.

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