Investigating Absolute Quantification in MR Perfusion Studies

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INTRODUCTION

Cerebral blood flow (CBF) variations are encountered in a variety of pathological conditions. The strong relationship between flow values and the cellular condition in ischemic stroke, for example, makes CBF a key parameter for indicating whether or not a tissue is likely to become compromised. Relative (compared to normal tissue) CBF values can be obtained using magnetic resonance (MR) protocols; however absolute flow quantification is not possible. A major limitation to achieving this goal is the presence of partial-volume effects (PVE) [1, 2] where the signal from a voxel contains information from more that one tissue type. PVE lead to low accuracy and reproducibility of perfusion estimates. Chen *et al*. [3] used simulations to establish the nature of the dependency of CBF estimates on PVE. In addition, the partial-volume (PV) level cannot be estimated directly from the current MR perfusion protocol. [3] Studies have been performed trying to reduce or avoid PVE errors. [4, 5, 6] van Osch *et al*. [5, 6] have evaluated a method for correcting PVE based on signal analysis that is only applicable to arteries that run parallel to the main magnetic field. Unfortunately this is not the case for the middle cerebral artery (MCA), a vessel typically used in MR perfusion studies. A new method is therefore needed for arbitrarily-oriented vessels. The work presented here suggests a means to estimate PV levels in perfusion imaging. By estimating and then correcting for PVE, absolute flow quantification may be possible without the need for cross-calibration with flow values obtained from other imaging techniques.

METHOD

The principles of phase contrast (PC) imaging were used to estimate the PV level in a vessel that runs parallel to the image plane. Voxels in the image were assumed to have a signal intensity $s(t)$, $s(t) = \alpha \cdot s_b(t) + (1 - \alpha) \cdot s_t(t)$, where α is the fraction of volume occupied by the artery (from 0 to 1) and $s_b(t)$ and $s_t(t)$ are the signals from blood and brain tissue, respectively. The voxel size was set to 1.7mm x 1.7 mm x 5 mm. A symmetric bipolar gradient was used to encode blood velocities into the signal phase [7]. The velocity encoding (v_{ENC}) ranged from almost zero to 150 cm/s [8]. Blood flow was considered to have either a parabolic flow (expected in big vessels) or a plug flow (small vessels) profile. Seven healthy volunteers were scanned using a 4-point PC encoding scheme (1 slice at the MCA level, 5 mm thickness, NEX =1 or 4) with varying v_{ENC} values. α values estimated from these images using the right-left velocity encoding direction were compared to those generated from a high-resolution 3D-TOF (time-of-flight) protocol (64 slices, 1mm slice thickness, FOV=26 cm). Linear regression was used for the comparison. All the images were collected using a 3.0T MR scanner (Signa; GE Healthcare, Milwaukee, WI).

RESULTS

Simulation results indicate (Fig. 1) that as the v_{ENC} value is decreased the signal intensity is decreasingly sensitive to blood flow. For various PV levels, the signal strength tends (at small v_{ENC}) towards the values associated with the tissue content of the voxel. This enables the calculation of the α parameter, that is, the indicator of the arterial fractional volume in the voxel. Results from the clinical data (Fig. 2) show that there is a good correlation between the PC-based α values and those estimated from the high resolution TOF images. The estimated α values ranged from 0.3 to 0.7 and can be translated to vessel diameters of 1.5 mm and 3.5 mm, respectively, for a 5 mm slice thickness. These values are in the expected range for healthy subjects. It is also shown in Fig. 2 that the overall accuracy of the PC technique to estimate α is good (slope = 0.915, (R^2 = 0.4)) with high precision (tight confidence interval (CI)). α values obtained using different number of averages in the PC protocol (NEX=4 and NEX=1) were similar (not shown), suggesting that these estimates are not overly sensitive to noise. In practice α could be calculated at a few pixels and then averaged to improve precision.

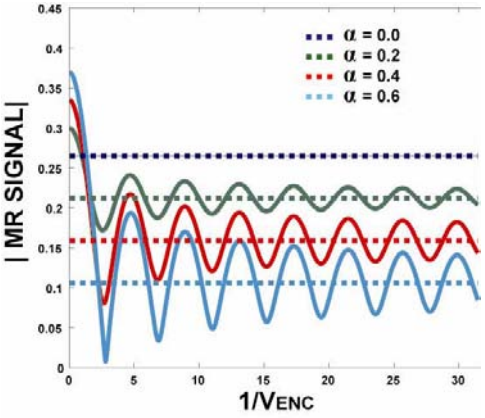

Fig. 1. MR imaging in voxels with various PV levels. The MR signal decays with decreasing v_{ENC} values to a signal level representative of only the tissue component in the voxel.

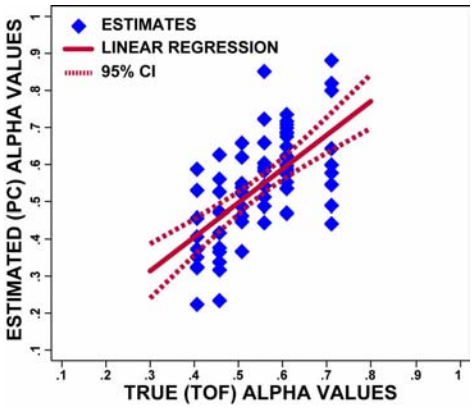

Fig. 2. Estimated vs true α values from 7 clinical sets (NEX = 1). PC-based α values are in the normal range and in agreement with those obtained from TOF images.

CONCLUSIONS

Simulations and clinical data support the idea that phase-contrast principles can be added to existing MR perfusion imaging protocols to estimate and compensate for the partial volume effects. Extensions of the work presented here will combine the PC protocol to the perfusion MR protocol and will also address the effects of the PV correction on the CBF estimates.

1 Conturo TE *et al. JMRI* 2005;22:697-703.
2 Smith AM *et al. MRM* 2000;43:559-64.
3 Chen JJ *et al. JMRI* 2005;22:390-99.
4 Vonken EJ *et al. JMRI* 1999;10:109-17.
5 van Osch MJ *et al. JMRI* 2005;22:704-09.
6 van Osch MJ *et al. MRM* 2001;45:477-85.
7 Moran PR *et al. MRI* 1982;1(4):197-203.
8 Pelc NJ *et al. JMRI* 1991;1(4):405-413.