Determination of the vascular input function using magnitude or phase-based MRI: influence on Dynamic Contrast-Enhanced MRI model parameters in carotid plaques

R.H.M. van Hoof[1,2], M.T.B. Truijman[1,3], E. Hermeling[1,2], R.J. van Oostenbrugge[2,4], R.J. van der Geest[5], A.H. Schreuder[6], A.G.G.C. Korten[7], N.P. van Orshoven[8], B. Meens[9], M.J.A.P. Daemen[2,10], J.E. Wildberger[1,2], W.H. Backes[1], and M.E. Kooi[1,2]

[1]Radiology, Maastricht University Medical Center, Maastricht, Netherlands, [2]Cardiovascular Research Institute Maastricht (CARIM), Maastricht University, Maastricht, Netherlands, [3]Clinical Neurophysiology, Maastricht University Medical Center, Maastricht, Netherlands, [4]Neurology, Maastricht University Medical Center, Maastricht, Netherlands, [5]Radiology, Leiden University Medical Center, Leiden, Netherlands, [6]Neurology, Atrium Medical Centre, Heerlen, Netherlands, [7]Neurology, Laurentius Medical Centre, Roermond, Netherlands, [8]Neurology, Orbis Medical Centre, Sittard, Netherlands, [9]Neurology, VieCuri Medical Centre, Venlo, Netherlands, [10]Pathology, Academic Medical Centre, Amsterdam, Netherlands

Target Audience: Researchers involved in dynamic contrast-enhanced perfusion studies of the microvasculature.

Purpose: A reliable vascular input function (VIF) is important for quantitative analysis of atherosclerotic carotid plaque microvasculature using dynamic contrast-enhanced (DCE) MRI. In tumor imaging and brain perfusion studies, it has been demonstrated that a magnitude-based VIF (m-VIF) is rather sensitive to flow artefacts[1]. The purpose is 1) to compare the m-VIF and phase-based VIF (ph-VIF) from the internal jugular vein and 2) to investigate the influence of different VIFs from the carotid artery and internal jugular vein on DCE MRI model parameters in carotid plaques.

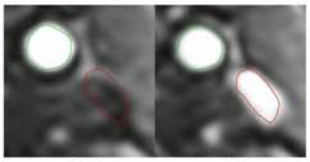

Figure 1: Magnitude MR images showing the carotid artery (green) and jugular vein (red) before (panel A) and after (panel B) contrast arrival.

Methods: *MRI Acquisition.* 21 patients with 30-99% carotid stenosis underwent an ECG gated 3T DCE MRI (T_1W 3D FFE with a flip angle of 35°) on a 3T Achieva TX whole body MRI system (Philips Achieva TX, Philips Healthcare, Best, The Netherlands) using a dedicated 8-channel carotid RF coil (Shanghai Chenguan Medical Technologies Co., Shanghai, China). Four patients were scanned with a high temporal resolution (approx. 4 seconds) (TR/TE = 23.15/3.17 ms) but lower spatial resolution (in-plane acquired/reconstructed resolution 1.56/0.45 mm^2). A saturation slab was placed 15 mm cranial to the imaging slice in order to suppress magnitude MR signal of blood in the internal jugular vein before contrast injection. The other 17 patients were scanned using a lower temporal resolution (approx. 20 seconds) (TR/TE = 11.61/5.65 ms) and higher spatial resolution (in-plane acquired/reconstructed resolution: 0.63/0.25 mm^2).

Data Analysis. The four dynamic acquisitions with a high temporal resolution scans were used to construct group-averaged m-VIF and ph-VIF[2], while the other 17 dynamic acquisitions with a high spatial resolution were used for calculation of K^{trans} (measure of plaque microvasculature) using the determined group averaged m-VIF and ph-VIF. Pearson's correlation between K^{trans} using the various VIFs was calculated.

Calculations. To investigate flow influence on m-VIF, first, the m-VIF was calculated from the signal intensity-time curves by neglecting flow and by assuming a steady state as is common in literature[2]. Second, the effect of Poiseuille flow on the m-VIF was estimated by calculating the average relative signal enhancement of the jugular vein based on the concentrations as derived from the dynamic phase images using the Bloch Equations, incorporating the (local) blood velocity and a cranially positioned spatial saturation slab at a distance of 15 mm from the imaging slice.

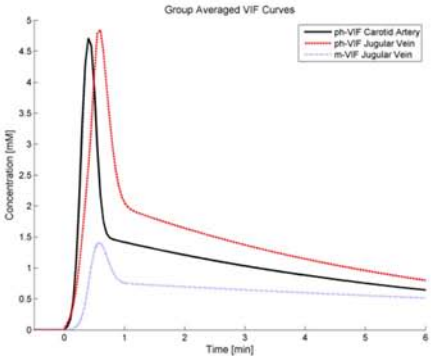

Figure 2 : Group averaged Vascular Input Functions determined from the jugular vein (phase method, red dotted line; magnitude method, blue dashed line) and carotid artery (phase method, black solid line).

Results: Determination of the m-VIF in the carotid artery was not possible due to MR inflow effects, which resulted in absence of a first pass peak. For one subject, the m-VIF could also not be determined in both the jugular veins due to the absence of a first pass peak while the first pass peak was clearly visible in the phase images. Determined group averaged VIFs are shown in figure 2. When comparing the phase-based VIFs of the internal jugular vein and the carotid artery, the peak concentration in the internal jugular vein is achieved approximately 8. seconds later and the first pass peak is broader in the internal jugular vein. Calculated peak concentrations of jugular vein m-VIFs were on average 4-fold lower than for the jugular vein ph-VIFs ($p<0.001$) when neglecting flow and assuming a steady state, resulting in different K^{trans} values determined using group averaged m-VIF and ph-VIF. Despite these differences, strong and significant correlation between K^{trans} was found (Figure 3, Pearson's correlation ranging from 0.88 to 0.99, $p<0.001$). Taking into account a theoretical Poiseuille flow with a maximum velocity of 10 cm/sec, it is shown that the difference in peak signal enhancement between the measured magnitude signal and the averaged simulated signal is diminished (figure 4).

Discussion: Our results show a strong influence of flow on the magnitude-based method for determination of the Gd concentration-time curve, even with the use of a saturation slab, leading to a large underestimation of the peak Gd concentration compared to phase-based methods. The various VIFs leaded to different K^{trans} values. Despite this, a strong significant correlation between the K^{trans} parameters using the various VIFs was found. Calculations showed 1) a large influence of the Poiseuille flow on the (average) relative signal enhancement and 2) that the discrepancy between the m-VIF and ph-VIF can be explained by flow artefacts.

Conclusion: m-VIFs are strongly influenced by flow, even with the use of saturation slabs. Despite this, a strong correlation between K^{trans} parameters determined using m-VIF and ph-VIF was found. It is expected that a ph-VIF results in a quantitative more realistic value of K^{trans}. Therefore, we advise to use a ph-VIF for quantitative DCE MRI analysis.

References: 1. Garpebring et al. Magn Reson Mater Phy. 2011;24:233-245, 2. Parker et al. MRM. 2006;56:993-1000.

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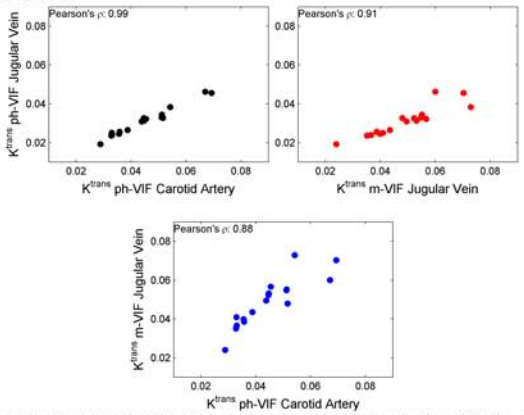

Figure 3: Scatter plots of median Ktrans values determined using the three VIFs shown in Figure 1. Pearson's correlation coefficients range from 0.88 (ph-VIF Carotid Artery vs m-VIF Jugular Vein) to 0.99 (ph-VIF Carotid Artery vs ph-VIF Jugular Vein).

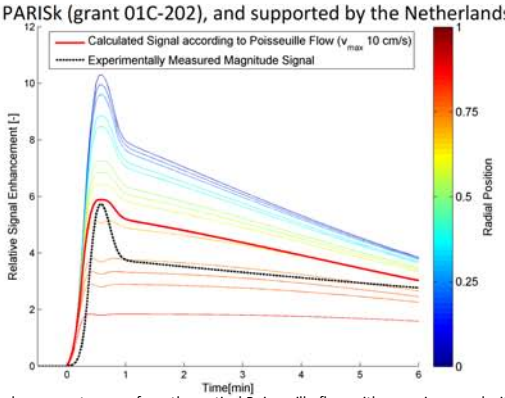

Figure 4: Simulated signal enhancement curves for a theoretical Poisseuille flow with a maximum velocity of 10 cm/second. Signal enhancement curves were calculated based on concentrations derived from the measured ph-VIF curve (figure 2, black) and a calculated curve based on 2D Poiseuille flow profile. Individual calculated pixel intensities are plotted by thin lines (various colours, colorbar indicating radial position, ranging from centre, 0, to near wall, 1) . Near the wall, signal saturation occurs due to a low local blood velocity resulting in increased T_1 relaxation between the saturation slab and imaging slice. Average simulated signal enhancement is shown in red (thick solid line), while the experimentally measured average magnitude signal enhancement is shown in black (thick dotted line).