Quantitative Contrast-Enhanced Myocardial Perfusion MRI: Simulation of Bolus Dispersion in Constricted Vessels

D. Graafen[1], K. Münnemann[1], S. Weber[1], and W. G. Schreiber[1]

[1]Section of Medical Physics, Department of Radiology, Mainz University Medical School, Mainz, Germany

Introduction

Quantification of myocardial blood flow by means of dynamic T1-weigthed MRI requires the knowledge of the arterial input function (AIF). In first pass myocardial perfusion imaging the AIF is usually estimated from the left ventricle (LV) [1-3]. Dispersion of the contrast agent bolus may occur between the LV and the tissue of interest, e.g. as a result of the vascular flow profile, and/or at a stenosis. This effect can be described mathematically by a convolution of the LV AIF with a vascular transport function (VTF): $AIF_{out}(t) = AIF_{in}(t) \otimes VTF(t)$ [4,5]. A quantitative parameter of the VTF which reflects the dispersion is the variance σ^2_{VTF} which can be calculated from AIF_{in} and AIF_{out}: $\sigma^2_{VTF} = AIF_{out}^{(2)} / AIF_{out}^{(0)} - AIF_{in}^{(2)} / AIF_{in}^{(0)} + (AIF_{in}^{(1)} / AIF_{in}^{(0)})^2 - (AIF_{out}^{(1)} / AIF_{out}^{(0)})^2$, with $AIF^{(n)}$ being the n^{th} integral moment of the AIF [5]. If dispersion occurs it will lead to systematic underestimation of blood flow, which was demonstrated in a simulation study based on the assumption of an exponential VTF [4]. The aim of this study was to simulate the dispersion along a simplified coronary artery with different stenosis by using the computational fluid dynamics (CFD) approach.

Material and Methods

Simulations were performed on straight vessels with a length of 100 mm and a diameter of 3 mm having typical dimensions of coronary vessels. Stenoses with different degrees of area reduction (60%, 70%, 80% and 90%), a length of 5 mm and two different cosine shapes - axialsymmetric and asymmetric – were integrated in the vessels 20 mm behind the inlet. The computational meshes consisted of approximately 250,000 quadrilateral cells mapped throughout the domain. The flow equations were solved with the laminar model of a commercial CFD software package (FLUENT, Fluent GmbH, Darmstadt, Germany). Two different boundary conditions at the inlet were simulated – to simplify matters both were steady conditions. For the myocardial perfusion at rest a flow of 0.1 m/s was chosen whereas for perfusion at stress a total gauge pressure of 1010 Pa was applied to realize the decrease of the velocity with increasing degree of area reduction of the stenosis. The pressure for the stress state was calculated from the results of the vessel without a stenosis with an inlet velocity of 0.5 m/s. A mixture of two species (blood and contrast agent) was simulated in order to derive the transport of the contrast agent. Due to the small mass fraction of the contrast agent (MFCA) of less than 0.003 the rheological properties of the contrast agent were neglected. For simplification blood was assumed to be a Newtonian fluid. Therefore, the mixture had a constant density of 1050 kg/m^3 and a dynamic viscosity of 0.04 Poise. The injection of the contrast agent was described by a gamma-variate function obtained by fitting the AIF of a volunteer measured in the LV. The typical duration of a myocardial perfusion measurement of 40 s was simulated with a constant time step size of 0.02 s. The area weighted average of the MFCA was calculated on several cross sections between the inlet and the outlet perpendicular to the axial vessel direction. To quantify the effect of dispersion, the variance σ^2_{VTF} was calculated for the inlet and each simulated cross sectional curve.

Results

The simulations show that the variance in resting condition is larger than under stress conditions. The main influence of the stenosis was observed immediately behind it (Fig.1). After the initial increase of the variance behind the stenosis a reduction of the variance occurred approximately until the end of the recirculation zone of the velocity field remaining only a small offset $\Delta\sigma^2$ in comparison to the variance curve in the straight unconstricted vessel with the same inlet velocity. At resting condition the effect of the stenosis on the variance at the end of the vessel was only small (Fig.2a). At stress condition with a constant inlet pressure, the decrease of the inlet velocity yielded a much stronger dependence of the variance as a function of degree of the stenosis. The variance for the stress state was clearly less than that for the resting state except for the cases with a stenosis of 90% due to the loss of the myocardial perfusion reserve (MPR). Both simulated shapes of the stenosis increase the absolute values of $\Delta\sigma^2$ with increasing area reduction, though the axial-symmetric yields a positive whereas the asymmetric shape produces a negative offset $\Delta\sigma^2$ (Fig.2b).

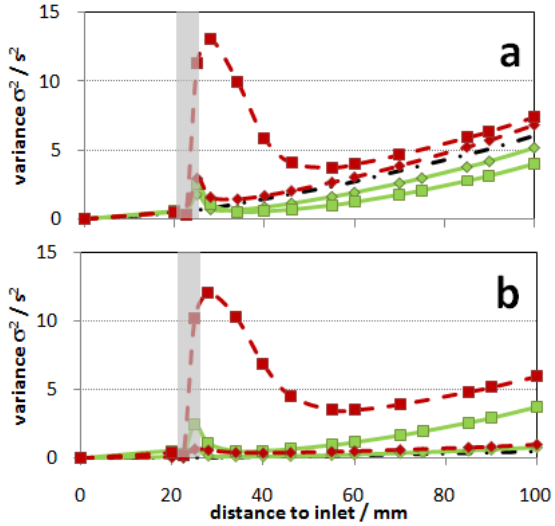

Fig. 1: Development of the variance σ^2 along the vessels for (a) the resting state and (b) the elevated state. The symbols represent the different degree of area reduction (▲:70%; ■:90%). The black curves depict for the development in the unconstructed vessel, the red for the axialsymmetric and the green for the asymmetric stenosis. The position of the stenosis is highlighted.

Discussion

Under steady boundary conditions dispersion is higher in the resting state than under stress conditions. Dispersion is mainly influenced by the amount of the inlet velocity. By comparison the effect of the stenosis under a constant inlet velocity causes only small change of the variance of the VTF, which depends both on the degree of area reduction and on the shape of the stenosis.

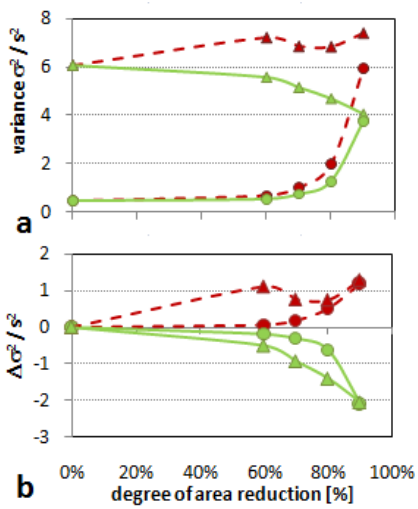

Fig. 2: (a) Variance σ^2 and (b) offset $\Delta\sigma^2$ at the outlet of the vessel versus the stenosis degree of area reduction for two different shapes of stenosis (axialsym., asymmetric). (▲: Resting condition; ■: Stress condition).

The reason for the strong inlet velocity dependence might be the smaller mean vascular transit time. However, this implicates a larger error at quantification of myocardial blood flow (MBF) in the resting state [4]. In the absence of a stenosis we calculated $\sigma^2_{VTF} = 6\ s^2$ for the resting condition and $\sigma^2_{VTF} = 0.5\ s^2$ under stress conditions. Using an exponential residue function for the VTF and the mathematical model MMID4 Schmitt et al. [4] found for these variance values an error of mean fitted MBF of approximately -10% for $\sigma^2_{VTF} = 0.5\ s^2$ and an error of approximately -25% for $\sigma^2_{VTF} = 6\ s^2$. The more severe underestimation of resting MBF leads to an overestimation of the MPR of about 20% for a healthy volunteer with a MPR of 5. For patients the overestimation would decrease with decreasing MPR.

An implicit assumption of our study is that of steady state conditions, because in this case dispersion is mainly affected by the mean velocity. This is a simplification which needs to be overcome in future simulations under conditions of pulsatile blood flow and additionally the complex geometry of coronary vessels should be considered.

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References

[1] Jerosch-Herold, M., et al., Med Phys, 1998. 25: p.73-84; [2] Wilke, N., et al., Cardiol Clin, 1998. 16: p.227-46; [3] Weber, S. et al., J Magn Reson Imaging, 2008. 28: p.51-9; [4] Schmitt, M. et al., Magn Reson Med, 2002. 47: p.787-93; [5] Calamante, F., et al., Magn Reson Med, 2006. 55: p.1180-5