Methodology for Optimal Quantitative Flow Analysis by planar analysis of CINE Phase Contrast 2D or 3D MR data

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Introduction: Time-resolved (CINE) *phase contrast* (PC) MRI allows measurement of time resolved three-directional velocities within entire 3D vascular structures [1, 2]. The information contained in the PC data is not limited to volumetric flow, but also includes derived quantities such as *wall shear stress* (WSS), the viscous drag of blood on the arterial wall, which is believed to play a key role in atherogenesis [3-6]. However, the lack of suitable analysis tools for the evaluation of complex 2D and 3D CINE PC MRI datasets still prevents the routine clinical use of such predictors. High dimensional 3D visualizations methods do exist but remain complex and permit only subjective and qualitative evaluation of blood flow characteristics. The purpose of this study was the development of quantitative analysis tools in order to derive flow parameters from CINE PC MR data. The method presented here combines accurate segmentation and robust data analysis methodologies into a planar analysis tool providing blood flow and vessel wall parameters [2]. Particular attention was given to methodological challenges such as arterial motion and limited resolution of the 2D or 3D CINE PC MRI data. Initial results in volunteer and patient studies of the thoracic aorta illustrate the potential of such parameters to characterize the link between disturbed blood flow and mechanical properties of the vessel wall. Furthermore, synthetic flow data was used to test the accuracy of the proposed quantification method.

Materials and Methods: All experiments were performed at 3T (Trio, Siemens, Germany) using a respiration controlled and ECG gated rf-spoiled gradient echo sequence with 3-directional velocity encoding in 2D (*2D-CINE-3dir.PC*: spatial resolution: 1.24-1.82 x 1.25-1.82 x 5 mm^3, temporal resolution: 24.4 ms, venc=150 cm/s) and 3D (*3D-CINE-3dir.PC*: spatial resolution: 2.71-2.93 x 1.58-1.69 x 2.60-3.0 mm^3, temporal resolution: 48.8 ms, venc=150 cm/s) [1]. Data preprocessing included eddy current correction and optionally velocity aliasing correction. Planar analysis planes transversal to the arteries were either directly acquired (*2D-CINE-3dir.PC*) or retrospectively interactively extracted from *3D-CINE-3dir.PC* data using 3D visualization software (Ensight, CEI, USA) and then imported into an in-house analysis tool based on Matlab (MathWorks, USA) [2]. Frame-wise segmentation was interactively performed using *cubic B-spline smooth contours* (Fig.1). In order to compensate for small local fluctuations due to noise, data was first filtered with a Gaussian low-pass filter of fixed radius (2mm) to prevent introduction of scaling dependencies. To compensate for the limited spatial resolution of the PC MR data, *cubic B-spline interpolation* was used based on its excellent compromise of interpolation quality and computational burden [7]. Moreover, *B-spline finite difference property* reduces the derivation to the use of another basis function [7], i.e. *analytical derivatives are obtained simultaneously with the interpolation process at almost no extra computational cost*. Those B-spline properties were used to obtain interpolation and derivatives for the segmentation spline (interpolated coordinates, derivatives and normal vectors) and velocities (interpolated velocities and derivatives on the contour) as required for flow parameter calculations. Based on the segmented vessel lumen contour and "*Green's theorem*", area and flow are efficiently computed from single integrals (eq.1-2). The numerical integration of (eq.3) is carried out at the original resolution (size N) while (eq.1-2) are realized using interpolation on ∂D (size N_i). The computational complexity for the area and flow calculations are thus $O(N_i)$ and $O(N_i \cdot N)$, respectively (compared to $O(N_i^2)$ with the same oversampling without using "Green's theorem"). WSS is derived from the slope of the measured velocities at the arterial wall and can be described using a deformation tensor [8] (eq.4-5). Calculation of the deformation tensor (eq.5) requires 3-dimensionnal derivation of the velocity vector field. Assuming that the 2D analysis plane is transversal to the vessel surface and enforcing a no flow condition at the vessel surface, the derivatives along the axial direction disappear and eq.4 can be used on 2D planes with 3D velocity fields to provide 3D WSS vectors. All data processing tasks were integrated into a graphical user interface which allowed for interactive segmentation and display of more than 20 physiological and flow parameters. The flow quantification tool has been evaluated on Synthetic data, a MR flow phantom (not presented here), 11 volunteers and 5 patients.

$$WSS = \eta \cdot \vec{n} \times \left((2\dot{\varepsilon} \cdot \vec{n}) \times \vec{n} \right) \quad (4)$$

$$\text{with } \dot{\varepsilon}_{ij} = 1/2 \left(\partial V_i / \partial x_j + \partial V_j / \partial x_i \right) \quad (5)$$

$\vec{n}$: inward normal; η : viscosity

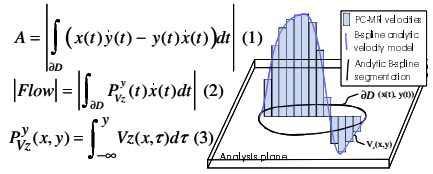

Fig. 1 Schematic illustration of B-spline based segmentation and velocity modeling

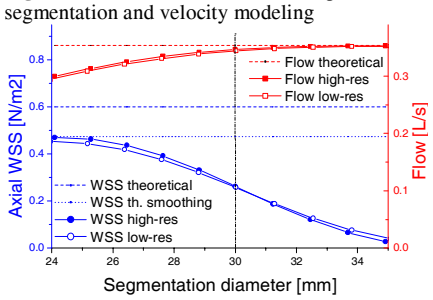

Fig. 2 Flow and WSS dependence on the segmentation for 2 synthetic parabolic datasets (mean velocity 0.5m/s, diameter 30mm, flow 353 L/s, WSS 0.6 N/m2). The dotted line corresponds to the WSS derived from the maximum velocity gradient after smoothing. High-res and low-res are 0.15 and 1.5 mm/voxel, respectively.

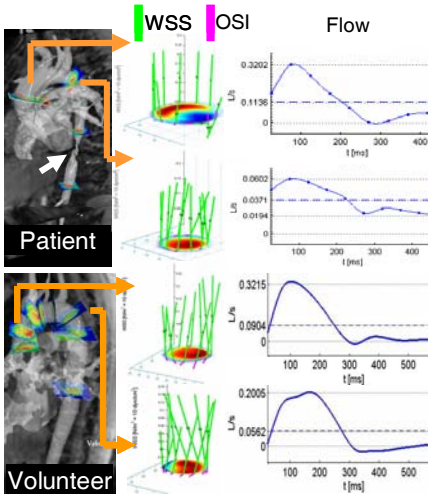

Fig. 3 WSS analysis and flow quantification: The placement of retrospectively defined 2D analysis planes transecting the thoracic aorta (ascending aorta (AAo) and proximal descending aorta (DAo)) allowed for the quantitative evaluation of mean WSS, normalized OSI and flow in a volunteer (bottom) and a patient (top) with coarctation in the DAo (white arrow). Altered hemodynamics in the patient DAo led to vortex formation and locally retrograde flow in the AAo that eventually induced low and oscillating WSS along the inner wall compared to findings in the normal volunteer.

Results: Optimized planar flow and vessel wall quantification was successfully applied to synthetic data, a MR flow phantom and in human studies. For each dataset, several flow parameters of interest were derived: geometrical information (time-resolved: area, perimeter, equivalent diameter), flow (time-resolved flow, flow acceleration, flow per cardiac cycle, regurgitant flow ratio, resistance index, pulsatility index), Reynold number, kinetic energy and wall shear stress (vectorial WSS and *oscillatory shear index* (OSI)). From Fig. 2, it is evident that the required smoothing step is introducing an intrinsic limitation regarding WSS, but conserves blood flow. Unlike flow, WSS is relatively dependant on the accuracy of the segmentation while both parameters (flow and WSS) exhibit only minor resolution dependency. Fig. 3 illustrates the application to a volunteer and a patient with a coarctation and the resulting altered hemodynamics distal to the coarctation.

Discussion: Quantification of CINE PC-MRI data is a challenging task because of the limited spatio-temporal resolution, SNR and the difficulty to accurately segment vessel lumen. Most approaches until now have consisted in using either a restrictive flow model (e.g. paraboloid [9]) or numerical flow simulations [10]. However, due to the limitations of those methods (restrictive model, time consuming flow simulations and difficulty to simulate flexible walls), clinical applications have been limited. In contrast, the method presented here aims at a direct quantification of flow parameters by using Green's theorem and cubic B-spline interpolation with their essential finite difference property to provide optimal quantification. Simple parameters such as the flow volume can be accurately quantified even for low resolution data while local flow parameters such as WSS are more limited by the spatio-temporal resolution and noisiness of MR data (Fig. 2). Nevertheless, the spatial repartition of velocity that MR offers is translated into the derived WSS and initial in vivo results on volunteers and patients are promising. Although the estimated WSS is limited in its absolute accuracy, WSS patterns may have the potential to analyze the impact of altered hemodynamics on the vessel wall. Our WSS measurements are in good agreement with published results in the abdominal aorta derived from phase-contrast MRI [11-13] which delivered similar average WSS values over the course of the cardiac cycle (0.18 to 0.95 N/m2). Further similar WSS values were also reported in a recent study of WSS in different segments of the descending thoracic aorta [5].

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