Quantification of Wall Shear Stress using Finite-Element Interpolations in Multidimensional Phase Contrast MR data of the Thoracic Aorta

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PURPOSE:

The wall shear stress (WSS) is an important parameter that allows for evaluating the loss of vascular function in several cardiovascular diseases, including atherosclerosis, aneurysms, stenosis and restenosis among others. In the last years, a B-Spline (BS) based method proposed by Stalder et al. (2008)[1] has been used to quantify the WSS in the aorta from 3D CINE PC-MRI data acquired by cardiovascular magnetic resonance. As noted, the computation of WSS from discrete 2D and 3D CINE PC-MRI velocity data introduces important approximation errors due to limited spatial resolution (1.5x1.5 mm for 2D and 2.5x2.5x2.5 mm for 3D) and numerical differentiation of the velocity field. In order to solve this problem we propose and validate a new method for calculating the WSS distribution in the aorta based on Finite-Element (FE) interpolations[2].

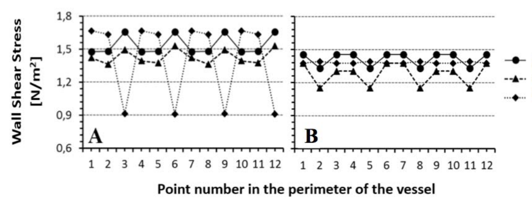

Figure 1. Wall shear stress magnitude evaluated at 12 points on the vessel contour for zero noise level and different voxel sizes: A) 0.48 x 0.48 mm. B) 1 x 1 mm. The methods analyzed were the modified Poiseuille flow (P), the proposed method (FE), and the B-Spline based method (BS).

METHODS:

The velocity field was obtained at discrete locations from 2D and 3D CINE PC-MRI. To this end, the domain of interest is discretized using triangular elements, and the velocities at the center of each voxel obtained from 2D or 3D CINE PC-MRI are interpolated using a finite-element basis, thus generating a continuous approximation for the velocity field. The shear stress tensor over the entire section cut was obtained from a global least-squares stress-projection method, from which the axial WSS vector was obtained. The proposed approach was benchmarked against a modified Poiseuille flow profile, and the robustness of the method was assessed in 12 contour points, by changing the level of resolution and noise. We have also compared our framework with a BS based method previously reported in the literature[1]. To validate the method, we computed the WSS distribution in different aortic sections from a pulsatile aortic phantom, and from 5 healthy volunteers. In the aortic phantom and volunteers, 3D CINE PC-MRI flow data was acquired in 2D cutting planes in 5 sections (AO1=ascending aorta, AO2 & AO3 = aortic arch, and AO4 & AO5 = descending aorta), with a spatial resolution 0.8 mm^2 and temporal resolution 37 ms.

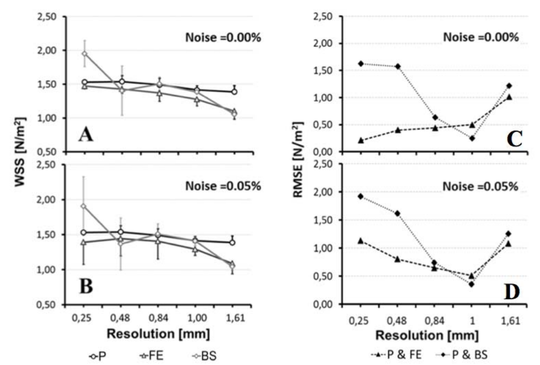

Figure 2. WSS contour mean and standard deviation (left column), and root mean square error (RMSE) (right column) for a noise level of 0% of maximum velocity peak (A - D), 0.05% (B-D). The methods analyzed were the modified Poiseuille flow (P), the proposed method (FE), and the B-Spline based method (BS).

RESULTS:

Local analysis of the WSS in the 12 points showed that our method followed the same distribution shape as the Poiseulle theorical values for any resolution. In contrast the BS method delivered a WSS distribution shape that did not match the theoretical values with large differences for small voxel sizes (Figure 1). The WSS contour mean and standard deviation, and RMS error between the values obtained from Poiseuille and the FE and BS methods are shown in Figure 2. The WSS contour mean values obtained with our and the B-spline based method were in general close to the theoretical values, but both methods worsen in accuracy toward smaller voxel sizes. Left column in Figure 1 shows that, as a general trend for both methods, axial WSS contour mean values decreased as the pixel size was increased. The right column in Figure 2 shows that the RMSE of our method was in general smaller than the RMSE obtained by the B-spline based method, with the exception of a voxel size of 1.00x1.00mm. For the case of zero noise level, our method showed a clear convergence toward the theoretical value as the pixel size is decreased. This trend is consistent with the well-known convergence properties of finite-element approximations[2]. In contrast, for the same case, the B-spline method showed an increase in RMSE as the pixel size decreases.

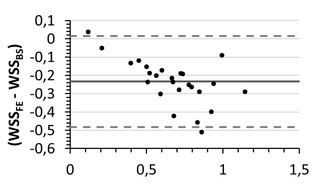

Figure 3. Bland-Altman plot of cardiac phase averaged wall shear stress contour mean comparing the proposed method (FE) and the B-spline method (BS) from volunteer data.

DISCUSSION:

Our results showed that the local WSS values were in good agreement with the theoretical values obtained from the modified Poiseuille flow problem as can be see in the Figure 1. The averaged WSS over the vessel contour showed a systematic, but negligible bias compared to the Poiseuille averaged WSS when subjected to different levels of noise and resolution. In contrast, the BS based method led to greater local differences, and the averaged WSS values were largely affected by the level of noise and resolution. In volunteers, the cardiac cycle average value of the average WSS value was 0.21 ± 0.06 N/m^2, whereas the BS based method yielded 0.45 ± 0.13 N/m^2. The Bland-Altman Figure 3 plot showed a systematic bias between both methods with a mean WSS difference of -0.2338 ± 0.2491 N/m^2.

CONCLUSION:

In conclusion, we have developed a novel methodology to calculate WSS based on FE interpolations, which provides a good approximation of local WSS values, stability when subjected to noise and remarkable convergence properties as the pixel size is decreased.

REFERENCES:

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2. Zienkiewicz OC, et al. The finite element method: Its basis and fundamentals, sixth ed. London: Butterworth-Heinemann. 2005.

FUNDING:

VRI # 44/2011 (Pontificia Universidad Católica de Chile), Anillo ACT 079 and FONDECYT #11100427 and #11121224. JS thanks CONICYT for scholarship for doctoral studies.