

Is a Flat Inlet Profile Sufficient for WSS Estimation in the Aortic Arch?

J. Renner¹, D. Loyd¹, T. Länne², M. Karlsson¹

¹Department of Management and Engineering

²Department of Medical and Health Sciences/Physiology

Linköping University

SWEDEN

johan.renner@liu.se

Abstract: - Atherosclerosis is one of the main reasons for cardiovascular disease which cause many deaths every year especially in the Western world. The development of atherosclerosis is strongly believed to be influenced by hemodynamic forces in the arteries e.g. wall shear stress (WSS). Estimations of WSS are therefore very important. By combining magnetic resonance imaging (MRI), image processing and computational fluid dynamic (CFD) simulations, it is possible to estimate subject specific WSS in the human arteries. The framework for performing such work includes i.e. using inlet boundary conditions which, however, will influence the final result i.e. the WSS distribution. This paper aims to investigate the influence of the inflow boundary condition in the human aorta with comparing two settings for the inflow: 1) subject specific inlet profile measured with MRI and 2) uniform profile with the subject specific mass flow rate. The analysis of WSS will be performed both on spatial location along the artery as well as on the temporal location in the cardiac cycle.

Subject specific data have been used for geometry, inflow velocity profile and blood viscosity. The recommendation due to our findings from nine healthy subjects, is that a measured subject specific inlet boundary condition must be used in order to get a subject specific WSS distribution; the difference in WSS is 8-34% compared to using a mass-flow correct uniform profile. Temporal variations were clearly seen in the WSS differences due to the different inflow velocity profiles used. The lowest influence of the inlet boundary condition was found at peak velocity in the cardiac cycle. The aortic geometry does not form the flow in such extent (compared to the influence by inlet boundary condition) to obtain a more correct WSS distribution further away from the inlet at the systolic parts of the cardiac cycle. The shape of the vessel has only a significant influence at low velocities i.e. the diastolic phase of the cardiac cycle.

Key-Words: Wall Shear Stress – Aorta - CFD - Subject Specific - Inlet Boundary Condition - Uniform Velocity Profile

1 Introduction

The number one killer in the Western world is cardiovascular diseases originating from development of atherosclerosis. Risk factors such as e.g. smoking and hypertension are well known but hemodynamic factors are also highly important for atherosclerosis studies today. The rapid development of in-vivo image acquisition, image analysis, computational fluid dynamics (CFD), as well as computer power has enabled the possibility to perform non-invasive time resolved subject specific simulations of the blood flow in the human body. This makes it possible to perform highly resolved (both spatially and temporally) in-vivo estimations of hemodynamic parameters using computer simulations.

The aim of the computer simulations in this paper is to describe the blood velocity distribution and the

flow-induced friction force on the arterial wall, i.e. the wall shear stress (WSS). Hemodynamic factors derived from WSS are of great interest due to its coupling to atherosclerosis genesis and development, see (1).

The influence of hemodynamic forces on atherosclerosis location was initially divergently postulated by (2) and (3). They found that high and low WSS indicate atherosclerosis risk regions. To this day the coupling between WSS and atherosclerosis is still not fully understood, (4). However, the most recent studies show a coupling of hemodynamic parameters such as WSS with indicators of atherosclerosis; e.g. (5).

Our proposed method uses CFD simulation techniques combined with magnetic resonance imaging (MRI) information for arterial geometry as well as measured blood velocity information. With this approach the velocity distribution near the

vessel wall is calculated with very high resolution thus enabling a high resolution estimate of the WSS distribution.

The design of CFD simulation models contains three main parts: defining the geometry, defining the governing equations and material properties, and defining the boundary conditions, (6). A correct geometry is essential in performing blood flow simulations and an in-vivo arterial geometry is therefore crucial in making subject specific estimates of WSS, (7-9). The governing equations for the flow are the Navier-Stokes equations. Typical boundary conditions for human cardiovascular system handles: the wall, the inflow and the outflow(s).

Many factors will influence the final result of the WSS calculation: the geometry interpretation (called segmentation), the boundary conditions as well as the chosen flow equations. The influence of the inflow boundary condition has only been examined in a few studies, where the most prominent ones are the studies by (8) and (10). In the work by (11) they found that the variation of the inlet boundary condition caused differences on the velocity contours but no changes in outflow flow rates. Analyzes by (10) concluded that the geometrical effect on WSS distribution was much larger than the effect of inlet boundary condition variation.

The fact that the inlet profile has an influence on the WSS distribution is known. However, to what extent the influence is relevant for different locations in the human aorta and different times in the cardiac cycle will be addressed in this paper. We present and discuss the temporally influence on WSS of the inflow boundary condition using a measured subject specific velocity profile (from MRI) and comparing it to a uniform velocity profile (with the correct subject specific mass flow) on a group (N=9) of normal volunteers, see table 1. Our hypothesis is that the arterial geometry will shape the WSS distribution more then the inflow boundary condition the further downstream the WSS distribution is compared.

2 Method

The in-vivo measurement technique of MRI was used in order to achieve both geometrical as well as flow information. Nine healthy normal male volunteers (age 26.6 ± 1.3 , BMI 21.9 ± 1.3) see table 1, were scanned in a 1.5T MRI scanner (Philips Achieva, Philips Medical Systems, Best, The Netherlands). For geometrical data of the whole aorta a 3D gadolinium-enhanced gradient-echo sequence was used within a breath hold (echo time 1.6 ms, repetition time 5.3 ms, and flip angle 40° ,

field of view $400 \times 360 \times 80$ mm, acquisition matrix $400 \times 230 \times 80$, SENSE factor 1.5). A 30 ml (0.5 mmol/ml) contrast bolus (Omniscan, Amersham Health, Oslo, Norway) was injected at 2.0 ml/s. Randomly segmented central k-space ordering (CENTRA) was used. The final geometrical data were transferred into a resolution of $0.78 \times 0.78 \times 1.00$ mm.

Velocities in the aorta were measured in a 2D plane perpendicular to the main flow direction in the ascending aorta distal to the aortic root (echo time 2.3 or 2.4 ms, repetition time 3.9 or 4.0 ms, flip angle 15, velocity encoding range 1.50 or 2.00 m/s). Acquisitions were made in a breath hold with SENSE factor 2.0 and retrospective cardiac gating to a vectorcardiogram. Velocity information was gained in a 10 mm thick slice (field of view of 350×297 mm, acquisition matrix 144×122) with spatial resolution of 1.37×1.37 mm and temporal resolution of 40 time-frames per cardiac cycle. Corrections for of concomitant gradient fields and eddy currents effects were made directly on the scanner. A level set approach was used for the segmentation, where a region was expanded out from one or several seed points, \cite{Sethian_99}. The seed points were placed by the operator and expand automatically toward the aortic contours. The expansion is influenced by a speed image and the local curvature. In order to significantly increase the speed of the level set segmentation was a rough curvature estimate used, (12).

The segmentation data in form of binary information was then smoothed with a 2 mm Gaussian smoothing filter with spatial size of 19 pixels. The description of the 3D aortic geometry was converted into a surface description using marching cubes, (13). Finally it was converted to a stereo-lithography file format (.stl). Approximate time for the segmentation process was 20 minutes per subject. The segmentation algorithm was implemented into a cardiac image analysis software package, (14), which is freely available at (<http://segment.heiberg.se>).

The CFD models consisted of 1.5-2.5 million cells with four prism layers at the wall boundary. This mesh sizes and construction were determined by performing mesh independence analyzes for the WSS in the aortic arch and descending aorta. The time dependent Navier-Stokes equations were solved using the commercially available software ANSYS Fluent 6.1 (Ansys Inc., Pittsburgh, Pennsylvania, USA). The flow was considered non stationary and laminar, (15), with the blood

considered as an incompressible fluid with a density of 1060 kg/m^3 . Subject specific fluid behavior regarding viscosity was chosen to Newtonian fluid model with measured subject specific viscosity ($0.0046 \pm 0.0007 \text{ Ns/m}^2$), measurement performed using the ReoRox® Jr. device (MediRox AB, Nyköping, Sweden). The inlet boundary condition at the ascending aorta was of two different types for each subject: 1) the measured subject specific velocity profile (shown in Figure 1 for all subjects at max acceleration time t_1) and 2) a uniform profile with subject specific (measured) mass flow rate. The choice of a uniform profile as the simplified case instead of a fully developed profile was made due to the fact that the flow profile in the ascending aorta is far from a fully developed and that a uniform profile is a more relevant assumption, Figure 1 and (16).

Outflow boundary conditions (specified fractions of the inflow) were prescribed unified in all subjects. The outlets mass-flow fractions used were 10%, 5%, 5% and 80% in brachiocephalic artery, left common carotid artery, left subclavian artery, and descending aorta. The aortic wall in the computational models was considered rigid with a no slip boundary condition.

Simulations were performed, for two cardiac cycles in order to ensure independence of initial conditions, on the Linux cluster monolith at the Swedish National Supercomputer Center, NSC, (<http://www.nsc.liu.se>). Typically 8 CPUs were used for 100 h to simulate two cardiac cycles.

WSS estimations from the nine individuals were gained for a whole cardiac cycle. Comparison data were extracted at the three cross-sections of the aorta placed at the ascending part, mid arch, and descending section; named CS1, CS2 and CS3, see Figure 2. WSS were extracted from these three cross-sections at four times. The times were chosen according to maximum acceleration, maximum velocity, maximum retardation, and minimum velocity at the inlet; denoted t_1 - t_4 , see Figure 2.

The final set of WSS data used for the analysis were the WSS values for the 9 subjects at three spatially different cross-sections and at these four time positions resulting in 108 possible comparisons between the two inlet profile conditions. Each of the three cross-sections consisted of 360 spatial evenly distributed points on the wall. For evaluations, RMS values were calculated for each cross-section and time according to:

$$RMS = \sqrt{\frac{\sum \Delta_i^2}{i}} \cdot 100 \quad (1)$$

where

$$\Delta_i = \left(\frac{WSS_{measured,i}}{\max_{i=1}^{360} (WSS_{measured,i}, WSS_{uniform,i})} - \frac{WSS_{uniform,i}}{\max_{i=1}^{360} (WSS_{measured,i}, WSS_{uniform,i})} \right) \quad (2)$$

Where *measured* denotes measured profile and *uniform* denotes uniform profile with subject specific mass flow rate.

3 Results and Discussion

Two questions were of special interest for this study. First, is the use of a measured inflow velocity profile important in order to get reliable WSS result or is it sufficient enough to use a uniform profile with subject specific mass flow rate? Secondly, is the geometry influence on the WSS distribution so dominating that a subject specific inlet velocity profile is of limited importance when moving downstream the aorta?

3.1 Results

The differences in WSS values between the simulations with measured subject specific inlet velocity profile and uniform inlet velocity profile with subject specific mass flow rate for subject #4 is shown in Figure 3. There are significant differences in the WSS distribution and the differences vary by cardiac cycle time location (t_1 - t_4). Circumferential mean WSS values for the two inlet velocity profile settings were calculated for the different individuals, spatial positions, and time positions. These data are presented in Figure 4, where the WSS values are plotted against each other and the solid line marks the no difference positions. All points should coincide with this line if the two inflow boundary conditions resulted in the same WSS result. The subject mean differences in percent between measured and uniform inlet velocity profile are presented in Figure 5 by using RMS values described in equation (1) and (2).

3.2 Main Findings

Our focus was to determine the influence on the WSS distribution using a uniform inlet velocity profile (with measured subject specific mass flow) and a measured subject specific inlet velocity profile. Prominent differences in WSS distribution were found when using a mass-flow correct uniform profile compared with using a subject specific

measured inlet profile, see Figure 5. In Figure 4 it can be noted that the difference was smallest at time positions t_2 and t_3 which are at maximum velocity and maximum retardation. The over/under estimation structure, between measured profile and uniform velocity profile, shifts between time t_1 and t_4 (maximum acceleration and minimum velocity). This behavior is also indicated in the one subject example in Figure 3. Differences between the three cross-sections were generally very small, with the exception of time t_4 (minimum velocity) there the difference between the different inlet boundary condition is much larger in CS1 then in CS2 and CS3. However, the WSS distribution became significantly different when using the simplified (uniform) inflow boundary condition compared to using a subject specific profile, see Figure 4 and 5. The use of a measured inlet velocity profile is therefore of crucial importance when attempting to estimate WSS. At peak systole i.e. time t_2 the differences in WSS were significantly smaller then at the other times.

The influence on WSS distribution due to arterial geometry was not so prominent that the WSS distribution from the uniform profile becomes more similar to the WSS distribution from the measured velocity profile, when comparing the WSS distribution more and more downstream the aorta. However, in the first three time positions t_1 - t_3 (maximum acceleration, maximum velocity, and maximum retardation) the individual order between the three cross-sections is the same, but it is dramatically shifted at minimum velocity time t_4 , where the geometrical influences become more prominent, Figure 5. This implies that the temporal behavior is important to study. The main difference on WSS influence of inflow profile between time t_4 and the other time positions is that there exists a retrograde flow with low velocity. At low velocities there is probably a larger influence from the geometry, see (17), which may be explained by that the flow becomes more friction dependent at low velocities.

3.3 Comparison and other findings

The influence of inlet boundary condition studied by (11) were evaluated by using uniform, parabolic and a realistic (but not subject specific) profile applied on the femoralis bifurcation. Evaluations were performed by using computed velocity profiles and flow rates in the two outlets. Their main findings were that a geometrically idealized model is behaving clearly different in the flow rate results when varying the inflow boundary condition which

enhances the importance of a subject specific geometry. In (10) the focus is the question if a fully developed velocity profile is sufficient to get reliable WSS parameters and compare it with a realistic inlet profile. The realistic velocity profiles were gained by applying different (curved) inlet sections to the simulation inlet in the carotid artery. The approach by (10) was evaluated using cardiac cycle average WSS values and oscillating shear index (OSI). They reported that geometrical variation (realized by performing 3 different scans on each subject) is clearly influencing the WSS more than the inlet boundary condition.

Our results show that, in the aorta, the geometry importance does not influence the flow in such an extent that the influence from inflow boundary condition decreases when moving downstream from the inlet in the model of the aorta. The reasons of the different conclusions in our work compared to (10) on geometrical importance compared to the inlet velocity profile are probably that the evaluation was performed in a different temporal scope. Improvement in the knowledge on WSS distribution variation due to inlet boundary condition was that in our study the temporal variations were shown and evaluated. But in (10) the temporal variations were being included in the cardiac cycle average WSS and OSI. Our results show that importance of geometry is more prominent at low velocities which correspond to the diastolic part. The diastolic part is more than 50% of the cardiac cycle at rest, which indicates that temporal average parameters such as cardiac cycle average WSS and OSI consist of a large part of diastolic values. Both (11) and (10) use relevant velocity profiles, but our study uses measured subject specific velocity profiles as well as subject specific viscosity, which is a further step into making subject specific aortic WSS estimations.

3.4 Conclusions

In conclusion subject specific models have been used for geometry, inflow velocity profile as well as viscosity in nine subjects. The recommendation due to our findings is that a subject specific inlet boundary condition should be used in order to get a subject specific WSS distribution for the human aorta. The influence on WSS distribution by aortic geometry compared to the inlet velocity profile influence is only significant at diastolic times in the cardiac cycle. Importance in both scope and analysis is the temporal variations which must be taken into account if temporal resolved WSS distribution is of interest.

Acknowledgement

This work has been conducted within the Center for Medical Image and Science Visualization (CMIV) at Linköping University, Sweden. CMIV is acknowledged for provision of financial support and access to leading edge research infrastructure for medical imaging.

Computer resources for making this studie possible were supplied by National Supercomputer Center (NSC) at Linköping University, Sweden.

The authors are grateful to Dr Tino Ebbers for MRI acquisition and to Dr Einar Heiberg for segmentation tool development.

Subject	Age [years]	Wiegth [kg]	Length [m]	BMI [kg/m ²]	Viscosity [kg/ms]
1	24	68	1.79	21.2	0.0043
2	24	77	1.84	22.7	0.0062
3	23	69	1.81	21.1	0.0039
4	21	67	1.82	20.2	0.0045
5	23	73	1.86	21.1	0.0046
6	24	72	1.87	20.6	0.0046
7	26	71	1.78	22.4	0.0055
8	23	83	1.84	24.5	0.0037
9	24	75	1.79	23.4	0.0041
	23.6±1.3	73±5	1.82±0.03	21.9±1.3	0.0046±0.0007

Table 1: Age, weight, length, BMI and blood viscosity for the nine subjects.

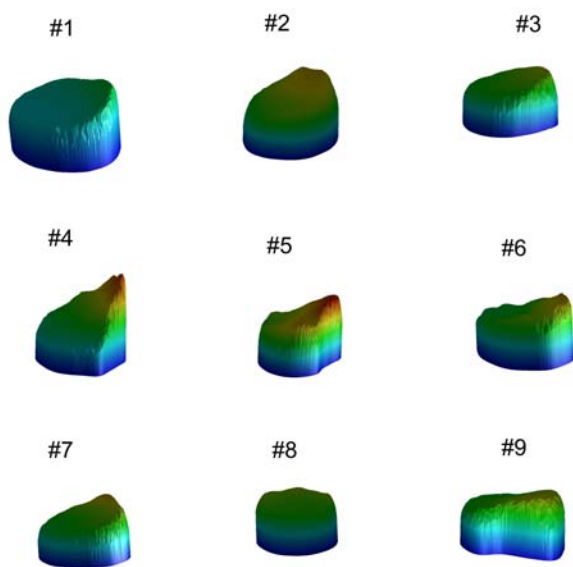


Figure 1: Subject specific inlet velocity profiles for all nine subjects at time t_1 (maximum acceleration).

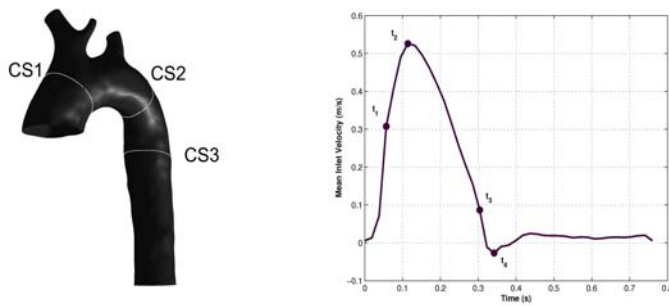


Figure 2: Left: Subject specific aortic geometry with the three analyzed cross-sections CS1-CS3 marked. Right: Mean inlet velocity in one subject, with time positions $t_1 - t_4$ marked. The positions correspond to maximum acceleration, maximum velocity, maximum retardation, and minimum velocity.

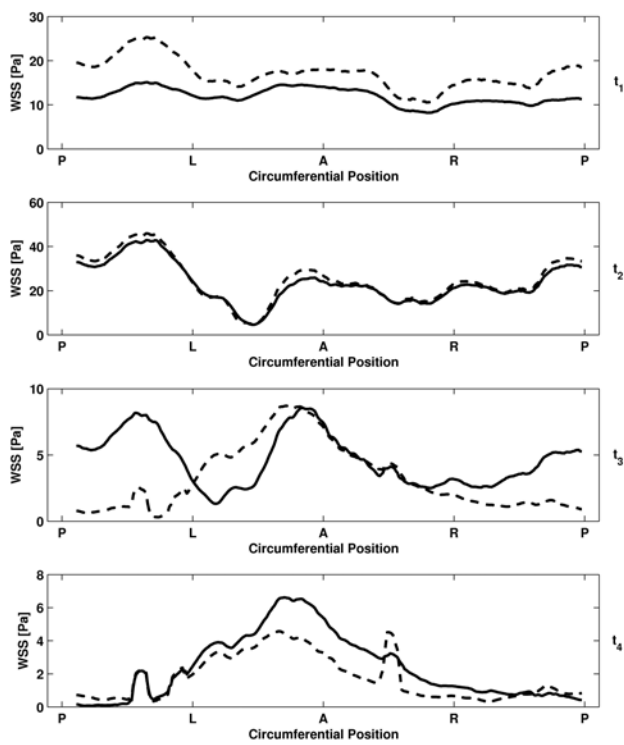


Figure 3: WSS values around CS3 for subject #4 (descending aorta) where P, L, A and R denotes posterior, left, anterior, and right. Solid line is WSS from the measured inlet profile and the dashed line comes from the uniform inlet velocity profile (note the different y-axis scales).

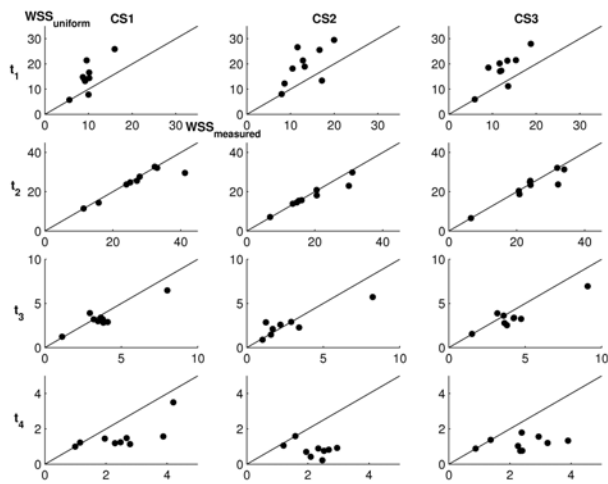


Figure 4: Cross-section mean WSS values, for subject specific measured inlet profile WSS_{mp} versus mass-flow correct uniform inlet profile WSS_{fp} ($N=9$). Displayed at the three cross-sections CS1-CS3 and four time positions $t_1 - t_4$.

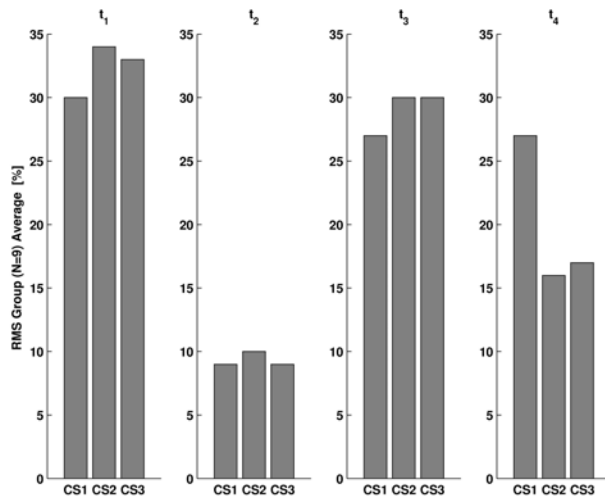


Figure 5: Subject average of RMS of the percentage of differences in WSS between subject specific inlet profile and mass flow correct uniform inflow profile. The average RMS values are shown at four different time positions $t_1 - t_4$ and three cross-sections CS1-CS3.

References

1. Chatzizisis YS, Coskun AU, Jonas M, Edelman ER, Feldman CL, Stone PH. Role of endothelial shear stress in the natural history of coronary atherosclerosis and vascular remodeling: Molecular, cellular, and vascular behavior. *J Am Coll Cardiol.* 2007 Jun 26;49(25):2379-93.
2. Fry DL. Acute vascular endothelial changes associated with increased blood velocity gradients. *Circ Res.* 1968 Feb;22(2):165-97.
3. Caro CG, Fitz-Gerald JM, Schroter RC. Atheroma and arterial wall shear. observation, correlation and proposal of a shear dependent mass transfer mechanism for atherogenesis. *Proc R Soc Lond B Biol Sci.* 1971 Feb 16;177(46):109-59.
4. Steinman DA, Taylor CA. Flow imaging and computing: Large artery hemodynamics. *Ann Biomed Eng.* 2005 Dec;33(12):1704-9.
5. Augst AD, Ariff B, McG Thom SA, Xu XY, Hughes AD. Analysis of complex flow and the relationship between blood pressure, wall shear stress, and intima-media thickness in the human carotid artery. *Am J Physiol Heart Circ Physiol.* 2007 Aug;293(2):H1031-7.
6. Svensson J, Gårdhagen R, Heiberg E, Ebbers T, Loyd D, Länne T, et al. Feasibility of patient specific aortic blood flow CFD simulation. *Med Image Comput Comput Assist Interv Int Conf Med Image Comput Comput Assist Interv.* 2006;9(Pt 1):257-63.
7. Svensson J, Gårdhagen R, Karlsson M. Assesment of geometrical influence on WSS estimation in the human aorta. *WSEAS Transactions on Fluid Mechanics.* 2006;1:318-26.
8. Morris L, Delassus P, Callanan A, Walsh M, Wallis F, Grace P, et al. 3-D numerical simulation of blood flow through models of the human aorta. *J Biomech Eng.* 2005 Oct;127(5):767-75.
9. Myers JG, Moore JA, Ojha M, Johnston KW, Ethier CR. Factors influencing blood flow patterns in the human right coronary artery. *Ann Biomed Eng.* 2001 Feb;29(2):109-20.
10. Moyle KR, Antiga L, Steinman DA. Inlet conditions for image-based CFD models of the

carotid bifurcation: Is it reasonable to assume fully developed flow? J Biomech Eng. 2006 Jun;128(3):371-9.

11. Morris L, Delassus P, Grace P, Wallis F, Walsh M, McGloughlin T. Effects of flat, parabolic and realistic steady flow inlet profiles on idealised and realistic stent graft fits through abdominal aortic aneurysms (AAA). Med Eng Phys. 2006 Jan;28(1):19-26.

12. Nilsson B, Heyden A. A fast algorithm for level set-like active contours. Pattern Recog Lett. 2003;24(9-10):1331-7.

13. Lorensen WE, Cline HE. In: Marching cubes: A high resolution 3D surface construction algorithm. SIGGRAPH '87: Proceedings of the 14th annual conference on computer graphics and interactive techniques; July; New York, NY, USA: ACM Press; 1987. p. 163-9.

14. Heiberg E, Wigstrom L, Carlsson M, Bolger AF, Karlsson M. In: Time resolved three dimensional automated segmentation of the left ventricle. Lyon, France. ; 2005. p. 599-602.

15. Fung Y. Biomechanics : Motion, flow, stress, and growth. New York ; Berlin: Springer; 1990.

16. Kvitting JP, Ebbers T, Wigström L, Engvall J, Olin CL, Bolger AF. Flow patterns in the aortic root and the aorta studied with time-resolved, 3-dimensional, phase-contrast magnetic resonance imaging: Implications for aortic valve-sparing surgery. J Thorac Cardiovasc Surg. 2004 Jun;127(6):1602-7.

17. Renner J, Gårdhagen R, Ebbers T, Heiberg E, Lanne T, Karlsson M. A method for subject specific estimatin of aortic wall shear stress. accepted for publication in WSEAS Transactions on Biology and Biomedicine.