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Is aortic wall shear stress affected by aging? An image-based numerical study with two age groups

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ABSTRACT

The size of the larger arteries increases during the entire life, but not much is known about how the change in size affects the blood flow. This study compares the flow field in a group of young males ($N = 10$, age $= 23.5 \pm 1.4$), with a group of older males ($N = 8$, age $= 58.0 \pm 2.8$). Aortic geometries were obtained by magnetic resonance imaging, and the aortic blood flow field was computed using computational fluid dynamics. The aortic wall shear stress was obtained from the computations, and it was concluded that time-averaged wall shear stress decreased with increased age, probably as a consequence of increased aortic diameter and decreased stroke volume, which in turn reduces the shear rates in the aorta. However, the oscillatory shear index, which is a measure of the oscillatory nature of the wall shear stress vector, seemed to be unaffected by aging.

1. Introduction

Blood vessels are continuously affected by the pulsating blood flow and are normally subjected to two types of forces: the blood pressure (BP) and the wall shear stress (WSS). The pressure is directed normal to the wall and the wall shear stress tangential to the wall, and in healthy humans the blood pressure is about 1000 times larger than the shear forces, but both have been found to affect the vessel wall in different ways. The stress in the wall is the result of the residual stresses (present even in a non-pressurized vessel) and the blood pressure, and the stress level determines the amount of stretching in radial, tangential, and axial directions and, thus, affects all cell types in the complex structure of the vessel wall. The wall shear stresses on the other hand, only affect the endothelial cell layer, which is located on the innermost layer of the vessel wall [1].

It is well known that the size of arteries increases during the entire life [2,3], and both arterial diameter and stiffness increase with age [3,4]. As a result of arterial stiffening, aortic pulse wave velocity increases, and is most noticeable in the aorta [4]. In a review article by Shabaan and Duerinckx [5] it was stated that age, blood pressure and BMI, which are all factors linked to the pathogenesis of atherosclerosis, are inversely related to wall shear stress. It should be pointed out that the WSS is not directly dependent on age or BMI, but instead dependent on the local flow and geometry. As such, age-related changes of the vessel have the possibility to affect the WSS.

It has also been found that the endothelial cell production of vasoconstricting growth factors such as angiotensin II and endothelin increases with age, while production of vasodilatory factors such as NO and prostacyclin decreases [6]. Studies have also shown that vasoactive substances released by the endothelial cells are strongly affected by the local shear stresses [5]. Also, the morphology and orientation of the endothelial cells are sensitive to the applied shear forces; under steady shear forces they elongate in the direction of the flow while under oscillatory shear there is no preferable alignment pattern [1,7,8].

This raises the question whether a change in endothelial cell function is caused by age-dependent changes of WSS? It is well known how physiological parameters such as cardiac output, aortic geometry, aortic distensibility and blood pressure change with age [2,4,9–11], but there have only been a few studies on how the WSS is affected. Ultrasound has been used to estimate WSS *in vivo* (see e.g. [12]) and geometrically simplified models have been studied [13]. Image-based computational fluid dynamics (CFD) can complement standard measurement techniques as the flow field and forces can be computed numerically. The simulations make use of MRI measurements for geometry and boundary conditions, and the result will be highly resolved flow fields, from where WSS is extracted with high spatial and

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Table 1
Data on the subjects, expressed as mean value ± standard deviation. Also, statistical significance is indicated with * ($p < 0.05$) or N.S. (not significant).

Parameter	Young group	Old group	
Subjects	10	8	–
Age [years]	23.5 ± 1.4	58.0 ± 2.8	*
Weight [kg]	72.8 ± 3.5	78.8 ± 9.3	N.S.
Height [m]	1.83 ± 0.03	1.81 ± 0.07	N.S.
Systolic BP [mmHg]	125 ± 7.5	130 ± 7.3	N.S
Diastolic BP [mmHg]	68 ± 5.4	79 ± 4.6	*
BMI [kg/m^2]	21.8 ± 1.4	24.0 ± 2.3	N.S.

Table 2
Scale factors used in outflow boundary conditions for the blood vessels leaving the aortic arch. Data expressed as mean value ± standard deviation. No statistical difference between the two groups was found.

Location	Young group	Old group
Brachiocephalic artery	0.469 ± 0.103	0.468 ± 0.102
Left common carotid artery	0.194 ± 0.052	0.129 ± 0.039
Left subclavian artery	0.331 ± 0.065	0.397 ± 0.101

temporal resolution, see e.g. [14]. Normally the aortic wall is considered rigid, but simulations can also take into account the wall motion by either computing the wall displacement due to blood pressure [15] or by prescribing it based on MRI measurements [16].

The goal of this study is to use image-based CFD together with MRI measurements to investigate how aortic WSS magnitude and distribution differ between two age groups — early and late adulthood.

2. Method

2.1. MRI measurements

In total, 18 healthy men were used in the study, 10 who were between 21 and 26 years old (from here on named the “Young” group), and 8 who were between 55 and 64 years of age (from here on named the “Old” group). All individuals were scanned in a 1.5 T MRI scanner (Philips Achieva, Philips Medical Systems, Best, The Netherlands) in order to acquire both geometrical image material of the aortas as well as flow information at the ascending and descending aorta. The MRI method has been described extensively in [17]. Basic information and data on the subjects in each of the two groups are found in Table 1.

2.2. Segmentation

Segmentation was performed in order to create 3D geometrical models from the angiography images obtained by the MRI measurement. The segmentation method used a fast level-set approach implemented in the freely available cardiac analysis software Segment [18]. The method was mainly automatic with some manual interaction; seed points were placed in positions of the blood volume in the aortas and then expanded outwards until it reached the boundary between blood and vessel wall. Since the MR-images are taken at a specific resolution and the segmentation is based on a level-set algorithm, the surface can become slightly rough. To obtain a physiological relevant surface, smoothing with a volume-conserving Gaussian filter was performed. Finally the segmented aorta (which now describes the blood volume) was exported as a STL surface. Details about the segmentation approach can be found in [17].

2.3. Fluid model

The STL-surfaces set the boundary for the flow domain, and were cut in the ascending aorta and in the thoracic aorta. High quality hexahedral meshes were created in ANSYS ICEM 14.0, with 35 boundary layer cells normal to the wall to resolve near-wall flow features. The thickness of the first boundary layer-cell closest to the wall was 10 μm and then grew exponentially until the last cell matched the bulk flow mesh elements. In total, 5.5–6 million hexahedral elements where used in each of the 18 models. Mesh independency checks were performed on meshes with 2.6, 5.5 and 8.8 million cells and it was found that WSS distributions differed only by 3% for the two larger meshes, while the difference was about 5–7% when comparing the smallest and medium sized mesh. It was therefore decided to use meshes with at least 5.5 million cells, which was considered well-resolved compared to other studies where aortic flows have been computed, e.g. [19–21].

The flow was treated as laminar with a density 1060 kg/m^3 and a constant viscosity of 3.5e−3 Pa s, i.e. Newtonian fluid. Subject-specific velocity profiles measured by MRI were prescribed at the inlet in the ascending aorta and mass flow rates were prescribed at the outlets on the branches leaving the arch. The mass flow rates were calculated based on the measured difference between the flow in the ascending and descending aorta multiplied with a scale factor. The scale factor was based on cross-sectional area and was different for all subjects. The average values with standard deviation are shown in Table 2. At the outlet in the thoracic aorta a static pressure was set, while all walls were considered rigid. The brachiocephalic, left common carotid, and left subclavian arteries (the branching vessels in the aorta) were extended 30 mm to minimize numerical problems at the outlets, but these extensions were not included when post-processing the results.

The CFD-simulations were performed in ANSYS CFX 14.0. A sensitivity analysis showed that a time step size of 0.001 s was needed to properly resolve the flow features. The convergence criteria were 1e−5 and domain imbalances were converged well below 0.01%. Both spatial and temporal discretization schemes were second-order accurate. Four cardiac cycles were simulated and data were evaluated from the last cycle to ensure that the results were not affected by transient startup effects. Data were saved every 0.01 s during the last cardiac cycle.

2.4. Post-processing

From the measured and simulated data a number of parameters could be derived. Area strain is a parameter that describes the elastic properties of a vessel. It is computed in a cross-section in the descending aorta and is defined as:

$$A_{\text{strain}} = \frac{(A_{\max} - A_{\min})}{A_{\min}}. \tag{1}$$

To further investigate the flow, the Womersley number was calculated in the ascending aorta. It describes the relation between transient inertial effects and viscous effects and is defined as:

$$\alpha = \frac{D}{2}\sqrt{\frac{2\pi f \rho}{\mu}}, \tag{2}$$

where D is the aortic diameter, f the heart rate in Hz, ρ the density of the blood, and μ the kinematic viscosity of the blood.

From the CFD-simulations, wall shear stress was computed in all subjects. Both the time-averaged WSS (TAWSS) and the Oscillatory Shear Index (OSI) during a cardiac cycle was computed for each subject. The OSI parameter describes the cyclic departure of the WSS vector from its predominant direction [22]. The values range from 0 to 0.5, where 0 means that the instantaneous vector is always aligned with the time-averaged vector, while 0.5 indicates that the instantaneous vector is never aligned with the time-averaged vector. OSI is used to quantify flow reversal, but it is insensitive to shear magnitude and must therefore be used with caution; a large OSI value can indicate a disturbed flow region with high *or* low WSS magnitudes. It is