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ORIGINAL PAPER

Wall shear stress calculation in ascending aorta using phase contrast magnetic resonance imaging. Investigating effective ways to calculate it in clinical practice

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Abstract *Introduction:* There is growing evidence that atherosclerosis, as well as endothelial biology, depend on arterial wall shear stress (WSS). Several methods of WSS calculation with varying degrees of complexity have been proposed. This study aimed at investigating whether the most straightforward and easier to apply of these methods give comparable results in clinical practice.
Methods: Complete velocity encoding measurements using phase contrast magnetic resonance imaging were performed in 20 patients at a level perpendicular to the long axis of the ascending aorta approximately 2 cm above the aortic valve. WSS was calculated at this location on maximum systole. MR imaging was accomplished on a 1.5 T scanner. Four methods were applied to calculate WSS; three of them are based on the predictions of Poiseuille's theory of flow, while the last one is based on calculations resulting by the application of the definition of WSS.
Results: WSS calculated with the above mentioned methods was found to be in the range 4.2 ± 1.8 to 3.5 ± 1.7 dynes/m^2. The velocity profile at the site of measurements can be described with a parabolic equation of the form $u = ar^2 + br + c$ with an average $r^2 = 0.83$, which is in good agreement with Poiseuille's theory of flow. Comparison of the results shows no statistically significant differences between WSS measurements calculated with these methods.
Discussion: The four methods are equivalent in calculating WSS at the ascending aorta when blood flow velocities have a good parabolic distribution.

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Introduction

MR imaging is considered one of the most reliable, non-invasive methods for the determination of various arterial flow characteristics such as blood flow velocity, shear rate, and shear stress. Flow characteristics are of great clinical importance, since there is growing evidence that wall shear stress plays a significant role in endothelial homeostasis and focal distribution of atherosclerotic lesions [1,2]. Several studies have shown that atherosclerosis occurs predominantly at certain locations of the vascular tree where arteries have relative complex geometry such as bifurcations and curvature that result in ''disturbed'' blood flow behavior, whereas arterial sections that exhibit ''undisturbed'' blood flow are generally spared by atherosclerosis [3,4].

The determination of WSS at the arterial tree is possible by non-invasive methods such as ultrasound [5–7] and pulsed Doppler ultrasound [8] and invasive methods such as intravascular ultrasound [9] and intravascular Doppler ultrasound [10]. The disadvantages of the non-invasive ultrasound methods are their applicability only to superficial arteries such as the carotid arteries, and their limited spatial resolution which makes the determination of flow velocity near the vessel wall difficult [6,7,11]. The main disadvantages of the invasive ultrasound methods is their invasive nature since they are performed at the catheterization laboratory and the presence of the measuring probe into the flow field which unavoidably affects the original intravascular flow conditions and thus the measured values of WSS [11].

MR imaging is a non-invasive technique providing velocity data that can be precisely matched with anatomic pictures. Various flow data can be extracted from the acquired images, including velocity profiles and flow rates, and subsequently fluid dynamic quantities such as shear rate and shear stress can be calculated.

The aim of the current study is to apply four straightforward and easy to apply methods of WSS calculation at the ascending aorta on maximum systole using phase contrast MR imaging and to examine whether these methods give comparable results in order to be used in clinical practice.

Theoretical background

Wall shear stress (WSS) expresses the force per unit area which is exerted by a flowing fluid on the surface of the conduit tube in the direction of the local tangent. Following the Newtonian, incompressible fluid approximation, WSS depends on the dynamic viscosity of the fluid μ and the velocity gradient near the tube wall, namely the wall shear rate (WSR) and is defined as [12]:

$$\mathrm{WSS} = \mu \mathrm{WSR} = \mu \frac{du}{dr} \tag{1}$$

where WSS is the wall shear stress, measured in pressure units (N/m^2 or dynes/cm^2, with a relation N/m^2 = 10 dynes/cm^2); μ is the dynamic viscosity of blood, measured in poise (1 poise = 1(dyne·s)/cm^2); du/dr is the velocity gradient of the blood which is called Shear Rate (SR), measured in 1/s. When this gradient is considered at the vessel wall, we have the Wall Shear Rate (WSR).

The dynamic viscosity of a fluid expresses its resistance to flow and it is low if a trivial force on a fluid layer produces a high velocity of that layer relatively to an adjacent layer and vice versa [13,14]. WSR expresses the rate of flow velocity increase with distance when moving away from the vessel wall where according to the no slip condition the velocity is zero.

Eq. (1) implies that dynamic viscosity is an inherent physical property of the fluid showing how easily the fluid is sliding [12]. It also implies that the velocity gradient is a linear function of the applied stress. Water and many other liquids exhibit such behavior and are for that reason called Newtonian or ideal fluids [12,15]. A Newtonian fluid is, by definition, one in which the coefficient of viscosity is constant at all rates of shear [15]. On the other hand, whole blood which is essentially a suspension of erythrocytes, leukocytes and platelets in blood plasma can be considered as a Newtonian fluid only under special conditions [15].

Materials and methods

MR imaging was performed on a 1.5 T scanner (Intera 1.5 T, Philips Medical Systems, Best, The Netherlands) using a 5-element phased array cardiac coil for signal reception. As part of our cardiac MRI protocol, a breath-hold phase contrast gradient echo sequence with retrospective cardiac gating was obtained at a level perpendicular to the long axis of the aorta approximately 2 cm above the aortic valve. Sequence parameters were as follows: TR = 4.2 ms, TE = 2.6 ms, FA = 15°, image acquisition matrix = 144 × 142 (frequency × phase encoding steps), reconstruction matrix = 256 × 256, FOV = 39 cm × 39 cm, percentage FOV = 0.86, ST = 10 mm, 1 NSA, VENC = 150 cm/s.

Methodologies that have been previously used for the estimation of WSS at the arterial tree using blood flow velocity measurements by phase contrast MRI were employed. All methodologies require the value of dynamic viscosity for the estimation of WSS. We have used the constant value of 0.035 (dyne·s)/cm^2, considering blood as a Newtonian fluid.

Average flow method (AFM)

This method of WSS estimation is based on Poiseuille's theory of flow and the resulting Hagen–Poiseuille formula. Poiseuille flow describes the type of laminar flow under steady (non-pulsatile) conditions of a Newtonian fluid in a cylindrical tube with rigid walls. Under these conditions flow is fully developed into a parabolic velocity profile where the velocity is highest at the center of the tube and zero at the tube walls. It can be derived that in Poiseuille flow, WSS can be mathematically derived by the Hagen–Poiseuille formula [12,15]:

$$\mathrm{WSS}_{\mathrm{systolic}} = \frac{4\mu Q}{\pi R^3} \tag{2}$$

In Eq. (2), μ is the dynamic viscosity of the fluid, Q is the volume flow and R is the inner radius of the conduit tube at the site of flow measurement. Using phase contrast MRI