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Errors in the estimation of wall shear stress by maximum Doppler velocity

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ABSTRACT

Objective: Wall shear stress (WSS) is an important parameter with links to vascular (dys)function. Difficult to measure directly, WSS is often inferred from maximum spectral Doppler velocity (V_{max}) by assuming fully-developed flow, which is valid only if the vessel is long and straight. Motivated by evidence that even slight/local curvatures in the nominally straight common carotid artery (CCA) prevent flow from fully developing, we investigated the effects of velocity profile skewing on V_{max}-derived WSS.
Methods: Velocity profiles, representing different degrees of skewing, were extracted from the CCA of image-based computational fluid dynamics (CFD) simulations carried out as part of the VALIDATE study. Maximum velocities were calculated from idealised sample volumes and used to estimate WSS via fully-developed (Poiseuille or Womersley) velocity profiles, for comparison with the actual (i.e. CFD-derived) WSS.
Results: For cycle-averaged WSS, mild velocity profile skewing caused ±25% errors by assuming Poiseuille or Womersley profiles, while severe skewing caused a median error of 30% (maximum 55%). Peak systolic WSS was underestimated by ~50% irrespective of skewing with Poiseuille; using a Womersley profile removed this bias, but ±30% errors remained. Errors were greatest in late systole, when skewing was most pronounced. Skewing also introduced large circumferential WSS variations: ±60%, and up to ±100%, of the circumferentially averaged value.
Conclusion: V_{max}-derived WSS may be prone to substantial variable errors related to velocity profile skewing, and cannot detect possibly large circumferential WSS variations. Caution should be exercised when making assumptions about velocity profile shape to calculate WSS, even in vessels usually considered long and straight.

1. Introduction

Wall shear stress (WSS) represents the shearing force exerted by blood flow on vascular endothelial cells, and is widely thought to play a key role in vascular reactivity [1], wall thickening [2] and atherosclerosis [3,4]. Although direct and precise measurement of WSS metrics (such as mean, maximum, and circumferential variation of WSS) in the clinic would therefore be highly desirable, this goal has not yet been achieved, primarily due to resolution limits of current imaging modalities. For example, WSS is non-linearly underestimated by magnetic resonance imaging (MRI) [5], while accurate measurement of velocities near the moving arterial wall via Doppler ultrasound is extremely difficult [6].

A popular alternative to direct measurement of near-wall velocities is to estimate WSS from an assumed velocity profile. A common approach in Doppler studies has been to assume a parabolic (i.e. Poiseuille) profile and calculate WSS via the resulting formula $4\mu V_{max}/D$; where μ is blood viscosity, D is the vessel diameter and V_{max} is the maximum instantaneous velocity [1,2,7–10]. This neglects the possible blunting of the systolic velocity profile caused by flow pulsatility, so a number of investigators have instead used a Womersley profile to estimate WSS [11,12].

Underpinning the calculation of WSS from parabolic or Womersley profiles is the assumption that the velocity profile is axisymmetric and fully-developed (with V_{max} lying on the centreline), thus limiting the measurement to vessels that are long and straight. However, vessels typically assumed to be long and straight

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(e.g. brachial, femoral and common carotid arteries (CCA)) may often harbour undeveloped or skewed velocity profiles owing to mild curvature [13–15]. Velocity profile skewing has three potential consequences for estimating WSS via Doppler ultrasound. First, a small, centrally-located sample volume may not detect true V_{max}. Second, the departure from ideal Poiseuille or Womersley profiles may lead to errors in calculated WSS. Finally, velocity profile skewing is likely to be associated with circumferential WSS variations, which cannot be captured by V_{max}-based calculations.

The aim of this study was to determine the likely errors in Doppler WSS measurement arising from these three issues, using anatomically realistic, image-based computational fluid dynamics (CFD) models of the CCA. This study extends our recent work investigating the accuracy of volume flow variables derived from Doppler-V_{max} [16].

2. Methods

2.1. Study participants and image acquisition

Eighteen subjects (age range, 37–84 years; mean ± SD, 58 ± 15) without carotid stenosis were selected from a subset of the VALIDATE study (Vascular Aging – The Link That Bridges Age To Atherosclerosis) representing ‘normal vascular aging’, consisting of subjects recruited from the Baltimore Longitudinal Study of Aging [17]. The study was approved by institutional review boards and subjects provided written informed consent. Contrast-enhanced magnetic resonance angiograms (CEMRA) and phase contrast magnetic resonance imaging (PCMRI) sequences were acquired as described by Hoi et al. [18].

2.2. Computational fluid dynamics (CFD)

Model geometry was obtained by segmenting the right carotid bifurcation from CEMRA images using the Vascular Modelling ToolKit (www.vmtk.org), with the CCA segmented to its thoracic origin, and CFD was performed with a validated solver, as in Ref. [16]. Second-order tetrahedral volume meshes were generated with a nominal edge length of 0.25 mm, previously demonstrated by our group to adequately resolve carotid WSS distributions [19]. Inlet and outlet pulsatile flow rates were obtained from PCMRI images and prescribed at the CCA and internal carotid artery (ICA) boundaries, assuming fully-developed axial Womersley velocity profiles, having adjusted ICA flow by the factor CCA/(ICA + ECA) at each time point to ensure instantaneous flow conservation. A traction-free boundary condition was used for the external carotid artery (ECA), vessel walls were assumed to be rigid and blood viscosity and blood density were 0.035 cm^2/s and 1.06 g/cm^3 respectively.

2.3. Data analysis

The data analysis procedure was similar to that described in Ref. [16]. Briefly, two-dimensional axial velocity profiles were extracted from CFD data at 3, 7 and 11 maximally-inscribed sphere radii proximal to the bifurcation, i.e. 1.2 ± 0.2, 2.2 ± 0.4 and 3.3 ± 0.5 cm from the bifurcation apex respectively, consistent with the reported range of 1–3 cm (see Ref. [16]). The degree of velocity profile skewing in the common carotid artery was classified by analysis of the high velocity region (HVR) [13], a contiguous area of pixels containing the highest velocities and covering 25% of the lumen area. Using an automated algorithm [13], the shape of the HVR boundary was used to classify the cycle-averaged velocity profile as Type I (axisymmetric), Type II (skewed) or Type III (highly skewed, or crescent-shaped) (Fig. 1).

From the total data set, six Type I, six Type II and twelve Type III profiles were selected, with a greater number of Type III profiles selected because these were expected to exhibit the greatest departure from fully-developed axisymmetric flow. As in Ref. [16], if multiple profiles from the three slice locations in a given subject had the same type, only one of these was included in the analysis (selected randomly). Note that Ford et al.’s analysis of PCMRI data of the undiseased CCA [13] revealed a prevalence of 36%, 24% and 40% for Types I, II and III respectively.

2.4. Idealised virtual Doppler ultrasound

Most studies measuring WSS from V_{max} have employed a small Doppler sample volume placed in the centre of the vessel [1,7–9]. Alternatively, the sample volume may be positioned at the perceived location of highest velocity (‘max-line’ velocity), then for the purposes of calculation, assumed to lie on the centreline [11]; the latter approach may be preferable in the presence of significant velocity profile skewing [16]. To perform an idealised virtual Doppler ultrasound on the 2D velocity profile data, we therefore placed 1.5 × 1.5 mm square sample volumes 1) in the centre of the lumen or 2) centred at the location of peak maximal velocity (Fig. 1). The virtual Doppler acquisition involved extracting the highest velocity from within the sample volume.

2.5. Wall shear stress calculation

WSS was calculated from V_{max} taken from the respective sample volumes by assuming Poiseuille or Womersley velocity profiles (see Appendix). This approach assumes axisymmetry, and hence that WSS is uniform circumferentially. However, this is not true for the CFD-derived WSS, and therefore reference WSS values were obtained by first extracting values from the CCA wall in 1.5 mm thick slices centered at the chosen slice location (Fig. 2). Since the surface nodes were randomly (albeit evenly) distributed with

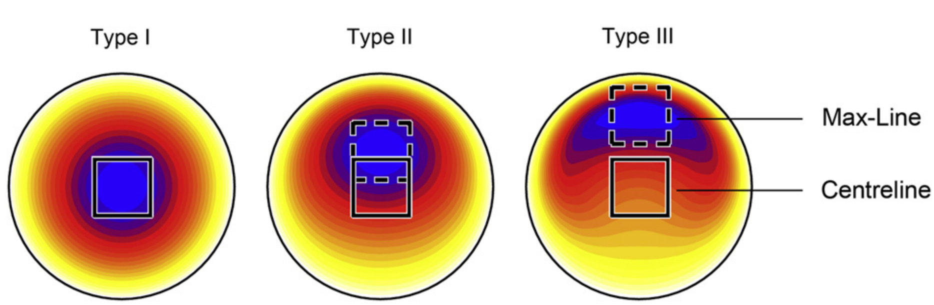

Fig. 1. Illustration of the idealised sample volumes (centreline, max-line) in the virtual Doppler ultrasound, shown in relation to the three velocity profile classifications: axisymmetric (Type I), skewed (Type II) and crescent (Type III).