

Shear stress depends on vascular territory: comparison between common carotid and brachial artery

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Dammers, Ruben, Frank Stiff, Jan H. M. Tordoir, Jeroen M. M. Hameleers, Arnold P. G. Hoeks, and Peter J. E. H. M. Kitslaar. Shear stress depends on vascular territory: comparison between common carotid and brachial artery. *J Appl Physiol* 94: 485–489, 2003. First published October 4, 2002; 10.1152/jappphysiol.00823.2002.—Shear stress (SS) is thought to be constant throughout the vascular system. Evidence for this supposition is scarce, however. To verify this hypothesis *in vivo*, we assessed common carotid (CCA) and brachial artery (BA) peak and mean wall shear rate (SR) noninvasively in 10 healthy volunteers (23.7 ± 3.4 yr) with an ultrasound SR estimation system. SS was estimated from SR and calculated whole blood viscosity. SR was higher ($P < 0.05$) in the CCA (mean: 359 ± 111 s⁻¹; peak: 1,047 ± 345 s⁻¹) than in the BA (mean: 95 ± 24 s⁻¹; peak: 770 ± 170 s⁻¹). Whole blood viscosity was higher in the BA than in the CCA (5.1 ± 0.7 vs. 3.3 ± 0.6 mPa·s; $P < 0.001$). Peak SS did not differ between the CCA and the BA, whereas mean SS was significantly higher in the CCA (1.15 ± 0.21 Pa) than in the BA (0.48 ± 0.15 Pa; $P < 0.001$). These results demonstrate that BA SS strongly deviates from CCA SS *in vivo*.

Poiseuille; shear rate; whole blood viscosity

IT IS COMMONLY ASSUMED THAT wall shear stress, the viscous drag exerted by the flowing blood on the vessel wall, is rather constant throughout the vascular system (14, 15, 21, 23, 27, 32, 33). This assumption is anchored in Murray's law on the basis of the principle of minimum work, which states that the cube of the radius of a stem vessel equals the sum of the cubes of the radii of the vessels of the subtree distal to a stem. More recent research has extended Murray's law in that the radii of both stem vessel and subtree are dependent on flow, i.e., shear stress, blood pressure, and vascular smooth muscle tone throughout the lifetime of an organism (21, 27, 28).

The above assumption regarding a constant shear stress, however, is mostly based on theoretical and *in vivo* studies referring to Poiseuille's law (9, 12, 13, 27, 28, 34), which supposes a Newtonian fluid under non-pulsatile conditions in a straight vessel. This is at least

an inaccurate supposition, because human large artery blood flow is pulsatile, and the vascular system is a distensible instead of a rigid tube (18, 20). Moreover, because of the viscoelastic properties of human blood, whole blood viscosity (WBV) might vary over the cardiovascular system, thus influencing wall shear stress or vice versa (6, 7).

Therefore, we hypothesize that wall shear stress varies with anatomical location. Indeed, in separate studies, our laboratory has shown that mean wall shear stress is on the order of 1.2 Pa in the common carotid artery (CCA) (24), 0.4 Pa in the common femoral artery, and ~0.5 Pa in the superficial femoral (17) and brachial arteries (BA) (3). In the present study, we investigated the individual differences, if any, in wall shear rate and stress in the elastic CCA and the muscular BA of healthy volunteers. Furthermore, according to the literature, shear stress would not only be the same in various vascular segments, but it should also be related to each other, i.e., when shear stress changes in one vascular segment, this would have its effect on another vascular segment as well. Therefore, we were also interested in a possible intra-individual correlation between the shear stresses of both the CCA and BA.

MATERIALS AND METHODS

Subjects and subject characteristics. The study was approved by the joint Medical Ethical Committee of the University of Maastricht and the University Hospital of Maastricht. Ten assumed healthy volunteers (3 women) were included in the study. They were aware of the investigational nature of the study and gave written, informed consent. All volunteers were nonsmokers, and none reported medication intake or a history of cardiovascular, cerebrovascular, or peripheral vascular disease. Also, hemoglobin, glucose, total cholesterol, high- and low-density lipoprotein, and triglyceride values were within the normal ranges. Additional subject characteristics are summarized in Table 1.

The measurements were always performed in the morning between 8:00 AM and noon after an overnight fasting of at least 10 h. Subjects were examined after an acclimatization

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Table 1. *Subject characteristics*

Parameter	Mean \pm SD
Age, yr	23.7 \pm 3.4
Body mass index, kg/m ²	22.2 \pm 2.2
Weight, kg	61.8 \pm 14.9
Height, m	1.80 \pm 0.13
Systolic blood pressure, mmHg	121.4 \pm 9.2
Diastolic blood pressure, mmHg	64.5 \pm 5.5
Mean arterial pressure, mmHg	86.5 \pm 5.2
Pulse pressure, mmHg	56.7 \pm 11.4
Hematocrit, %	43.0 \pm 4.2
Plasma viscosity, mPa·s	1.13 \pm 0.05

Characteristics of 7 male and 3 female healthy volunteers are shown.

period of 15 min in supine position in a room with a temperature of 22–24°C. Artery diameter and wall shear rate measurements were performed on the left BA and the left CCA. Right arm BA blood pressure was measured every 5 min by a semiautomated oscillometric device (Dinamap 1846P, Critikon, Tampa, FL). The measurement procedure for both diameter and wall shear rate, each with a length of 5 s and covering about six cardiac cycles, was repeated five times with a time interval of 1 min between recordings.

Shear rate estimation system. To assess the instantaneous blood velocity and shear rate distribution along a selected line of observation with high axial resolution, an ultrasound system (Ultramark 9 plus, Advanced Technology Laboratories, Bellevue, WA) with a broadband (5–9 MHz) curved array transducer (1, 24) was switched to a wide-band M-mode with a high pulse repetition frequency (10 kHz). From a depth of \sim 2 cm, the returned acoustic signals had an effective bandwidth on the order of 2.5 MHz, which was equivalent to a spatial resolution of \sim 300 μ m. Each velocity estimate was based on half-overlapping data segments of 300 μ m in depth and 10 ms in time and was calculated by means of a modeled cross-correlation function for the radio frequency signals. This results in an instantaneous velocity distribution, providing peak systolic velocity (in mm/s) and mean center stream velocity (in mm/s) averaged over a heart cycle, in the center of the lumen. The instantaneous shear rate distribution follows from the radial derivative of the velocity profile at each site and each time instant. The personal computer memory presently available allows a recording of 6 s for shear stress measurements and up to 20 min for diameter measurements, which is sufficient to capture data over five complete heartbeats. Reproducibility has been quantified in previous studies for both the CCA and BA and proved to be moderate to good (3, 25).

The same system can also be used in a perpendicular plane of observation to measure diameter (in mm) and distension (in mm) of the vessel of interest. A more detailed description of this system is provided by Brands et al. (1).

Measurement procedure in the BA. The BA was visualized in B-mode with the arm in \sim 45° abduction and the palm of the hand facing upward. The ultrasound system was switched to M-mode, and diameter and distension measurements were taken perpendicular to the wall. Subsequently, a suitable line of sight was selected that crossed the artery at an angle of 70° with the longitudinal axis of the artery, allowing the measurement of velocity and wall shear rate.

Measurement procedure in the CCA. The CCA was visualized in B-mode with the head of the subject slightly tilted in the contralateral direction. In each subject, the left carotid artery was investigated because, in a previous study, no

differences were found between the left and right carotid arteries (25). The common, internal, and external carotid arteries were evaluated for the presence of plaques. The tip of the bifurcation was used as a landmark. Subsequently, in M-mode, a line of observation (perpendicular to the wall) was positioned 2–3 cm proximal to the bifurcation to assess diameter and distension. Thereafter, at an angle of 70° with the longitudinal axis of the artery, velocity and wall shear rate were determined.

Wall shear rate assessment via Poiseuille's law. According to Poiseuille's law, the velocity profile of a Newtonian fluid under nonpulsatile conditions in a straight vessel exhibits a parabolic shape. To assess differences between wall shear stress as measured with the shear rate estimation system and as calculated according to Poiseuille's law, if any, mean wall shear rate was approximated (γ_P ; s⁻¹) with

$$\gamma_P = \frac{2n \cdot MV}{D} \quad (1)$$

where MV is the average center stream velocity, n is the bluntness factor of the instantaneous velocity profile, and D is the lumen diameter. For $n = 2$, Eq. 1 pertains to a parabolic velocity profile that may develop under nonpulsatile conditions in a straight vessel. Furthermore, the peak bluntness factor of the velocity profiles of both the CCA and BA was obtained by substituting the parameters measured with the wall track and shear rate estimation system in Eq. 1.

Shear stress calculation. With the use of the approximation proposed by Weaver et al. (30) for healthy subjects, WBV (in mPa·s) was estimated from plasma viscosity (η_0 ; mPa·s) that was determined with a glass capillary viscometer (Mikro-Ostwald Viskosimeter, Schott-Geräte, Hofheim, Germany), hematocrit (Hct; %), and mean wall shear rate (γ ; s⁻¹)

$$\log \text{WBV} = \log \eta_0 + (0.030 - 0.0076 \log \bar{\gamma}) \cdot \text{Hct} \quad (2)$$

To obtain wall shear stress (τ ; Pa), the measured wall shear rate was multiplied by the calculated WBV

$$\tau = \text{WBV} \cdot \gamma \quad (3)$$

Statistical analysis. Statistical management and analysis was performed by using the SPSS 10.0 package (SPSS, Chicago, IL). Values are presented as means \pm SD and are compared by using ANOVA. To relate vessel wall characteristics to hemorheologic factors and BA to CCA parameters, simple linear regression analysis was performed. Statistical significance was considered at a P value of <0.05 .

RESULTS

In Table 1, standard volunteer characteristics are shown. No subject was excluded from the study because of deviant blood parameter values and/or the presence of plaque. CCA and BA vessel wall param-

Table 2. *Common carotid and brachial artery vessel wall parameters*

	Common Carotid Artery	Brachial Artery
Heart rate, min ⁻¹	57 \pm 6	59 \pm 7
Distension, mm	0.77 \pm 0.10	0.058 \pm 0.035*
Diameter, mm	6.7 \pm 0.5	3.7 \pm 0.7*
Relative distension, %	11.6 \pm 2.0	1.5 \pm 0.8*

Values are means \pm SD. *Significantly different from common carotid artery ($P < 0.001$).

Table 3. Effect of changes in hematocrit and/or plasma viscosity on whole blood viscosity at different shear rates calculated with Weaver's approximation

MWSS, s ⁻¹	Change in WBV for 1% Change in Hct, %	Change in WBV for 1% Change in η_0 , %
100	2.13	0.74
250	1.86	0.74
500	1.73	0.74
750	1.43	0.74
1,000	1.31	0.74

MWSS, mean wall shear stress; WBV, whole blood viscosity; η_0 , plasma viscosity.

ters are depicted in Table 2, where obvious differences can be observed between the elastic CCA and the muscular BA in healthy subjects. Relative distension, defined by the quotient of distension and diameter, is significantly higher in the elastic artery (1.5 ± 0.8 vs. $11.6 \pm 2.0\%$; $P < 0.001$).

CCA WBV, calculated with Eq. 2, was significantly lower than in BA WBV (3.3 ± 0.6 vs. 5.1 ± 0.7 mPa·s; $P < 0.001$). Hematocrit and plasma viscosity were assumed to be equal (Table 1), and under physiological shear rate conditions ($100 \text{ s}^{-1} < \gamma < 1,000 \text{ s}^{-1}$) the effect of changes in hematocrit and plasma viscosity on WBV, calculated with Eq. 2, is negligible, as illustrated in Table 3, which leaves shear rate as relevant parameter.

Although peak systolic velocity tended to be less in the BA, no significant deviation was found (Table 4). It should be noted, however, that during the diastolic phase there was zero flow in the BA, which can be explained by the high-resistance vascular bed in the lower arm and hand. Mean velocity, however, was almost 400% higher in the CCA ($P < 0.001$; Table 4) as expected, due to the diastolic flow in the latter artery. The same holds for measured mean shear rate, which is higher in the CCA ($P < 0.001$; Table 4). Calculated mean shear rate, according to Poiseuille, was also less in the BA than in the CCA. When measured and calculated mean shear rate in the CCA were compared,

Table 4. Hemodynamic parameters in the common carotid and brachial artery

	Common Carotid Artery	Brachial Artery
SV, mm/s	869 ± 256	680 ± 160
MV, mm/s	295 ± 81	$77 \pm 27^*$
Peak wall SR, s ⁻¹	$1,047 \pm 345$	$770 \pm 170^\dagger$
Mean wall SR, s ⁻¹	359 ± 111	$95 \pm 24^*$
Peak-to-peak wall SR, s ⁻¹	$1,015 \pm 389$	971 ± 234
Peak wall SS, Pa	3.4 ± 0.8	3.9 ± 0.8
Mean wall SS, Pa	1.15 ± 0.21	$0.48 \pm 0.15^*$
Peak-to-peak wall SS, Pa	3.28 ± 1.05	$4.86 \pm 1.07^\ddagger$
γ_P , s ⁻¹	179 ± 55	$83 \pm 25^*$
τ_P , Pa	0.57 ± 0.10	$0.42 \pm 0.14^\dagger$

Values are means \pm SD. SV, peak systolic velocity; MV, mean velocity; SR, shear rate; SS, shear stress; γ_P , mean wall SR according to Poiseuille's law; τ_P , mean wall SS according to Poiseuille's law. * $P < 0.001$ compared with common carotid artery; $^\dagger P < 0.05$ compared with common carotid artery; $^\ddagger P < 0.01$ compared with common carotid artery.

however, a significant difference was found. More precisely, Poiseuille underestimated the mean shear rate considerably (179 ± 55 vs. $359 \pm 111 \text{ s}^{-1}$; $P < 0.001$). For the BA, no difference between calculated and measured mean shear rate was observed. Linear regression showed no correlation between BA and CCA mean wall shear rate ($r = -0.04$, $P = 0.92$). Peak shear rate was slightly higher in the CCA (Table 4), whereas peak-to-peak (maximum - minimum) shear rate was not different (Table 4). No correlation between peak shear rate in the CCA and BA was found ($r = 0.23$, $P = 0.52$).

The peak bluntness factor n of Eq. 1 was calculated for both the CCA and the BA to determine whether a parabolic flow profile, as assumed by Poiseuille, was realistic in these vascular segments. It showed that BA blood flow showed a rather parabolic profile with $n = 2.1$, whereas in the CCA the flow profile is somewhat flattened with $n = 4.0$. The difference in n was significant with $P < 0.001$, which can be explained by the greater relative distension of the CCA (Table 2) (20, 29).

Mean wall shear stress appeared to be significantly higher in the CCA (1.15 ± 0.21 vs. 0.48 ± 0.15 Pa, $P < 0.001$), whereas peak wall shear stress was equal in both arterial segments ($P = 0.19$; Table 4). Peak-to-peak wall shear stress, however, was significantly lower in the CCA (3.28 ± 1.05 vs. 4.86 ± 1.07 Pa; $P = 0.004$). Via Poiseuille's equation, CCA mean wall shear stress was lower than when approximated with Weaver's (30) expression (Eq. 2) (0.57 ± 0.10 vs. 1.15 ± 0.21 Pa; $P < 0.001$), whereas BA mean wall shear stress was similar with the use of both expressions ($P = 1.00$; Table 4). No correlation was observed between BA and CCA mean wall shear stress ($r = 0.31$, $P = 0.39$; Fig. 1). Also, no relation was found between both arteries regarding peak wall shear stress ($r = 0.23$, $P = 0.52$).

BA mean wall shear stress correlated to BA diameter ($r = 0.77$, $P = 0.009$), whereas in the CCA no relation

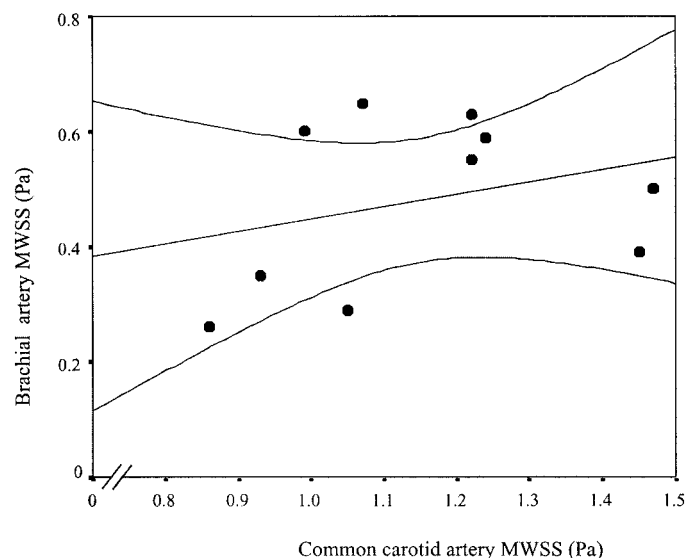


Fig. 1. Brachial artery mean wall shear stress (MWSS) is not correlated to common carotid artery MWSS.

was found between diameter and shear stress. When correlating distension or relative distension to mean wall shear stress, no correlation was found in either the BA or the CCA.

DISCUSSION

In the present study, we observed a significant difference in mean wall shear stress in two anatomical and physiological distinct arteries, namely the CCA (~1.2 Pa) and the BA (~0.5 Pa) of healthy individuals. Common opinion, however, is that wall shear stress is rather constant throughout the vascular tree (2, 14, 15, 21, 23, 27, 32, 33).

Our observations were based on shear rate measurements with the use of a custom-built shear rate estimation system that has proven good reproducibility for shear rate measurements at both anatomical locations (3, 16, 19, 25). Shear stress was approximated by the product of shear rate and WBV, calculated with *Eq. 3* as proposed by Weaver et al. (30). The dynamic WBV is considerably lower in the CCA than in the BA (3.3 and 5.1 mPa·s, respectively). Table 3 shows that this difference can substantially be attributed to the mean wall shear rate. This is a valid conclusion, because blood is indeed a shear-thinning non-Newtonian fluid, thus its viscosity decreases as the shear rate increases. Furthermore, variations in shear rate are mainly caused by changes in the viscous properties of blood (2, 5).

Although irrelevant for intrasubject evaluation, the influence of both hematocrit and plasma viscosity on viscosity must not be underestimated. For instance, an increase in hematocrit due to higher hemoglobin content, as observed in racing horses (7), would show an increase in viscosity. Also, at different shear rates (0.1–208 s⁻¹), hematocrit levels account for 67–84% of variability in WBV in 128 normotensive adults, as shown by de Simone et al. (4). The plasma protein content, especially fibrinogen and albumin, mainly accounts for the effect of plasma viscosity on WBV (6, 30).

Although it is generally accepted that WBV is affected by shear rate, many research reports (9–11) assume a fixed viscosity for all studied subjects. This is not justified, however, because wall shear rate is site dependent, and mean shear rate, hematocrit, and plasma viscosity vary among subjects.

Besides the effect of WBV, vessel wall characteristics are of great influence on shear rate distribution during a heart cycle. Due to the greater compliance of the CCA, for instance, the velocity profile will be more blunted, compared with the stiffer BA velocity profile (Table 2) (5, 20), resulting in a peak bluntness factor (*Eq. 1*) of 4.0 for the CCA and 2.1 for the BA. Measured BA shear stress was more or less equal to calculated mean wall shear stress with Poiseuille, which supports the impression that BA flow is parabolic in shape. In the present study, CCA mean wall shear stress, however, was largely underestimated by using Poiseuille's equation for a parabolic flow profile. The assumption of a parabolic velocity profile throughout the vascular tree is therefore erroneous.

Interestingly, when the mean flow profile is known, e.g., through blood velocity measurements at different radial positions within an arterial lumen (8), WBV can be approximated with the use of Womersley's equation of velocity profile (26, 31). This accentuates our findings regarding the importance of taking into account the flow profile in relation to shear rate and WBV.

Blood pressure is also known to affect shear stress, i.e., the higher the blood pressure, the higher the shear stress. Thus the lower shear stress in veins is in agreement with their relatively lower blood pressure (21, 22, 27). This implies a complicated interplay between blood pressure, blood flow, shear stress, WBV, and vessel wall smooth muscle tone. It is postulated that the interaction can be explained by an optimization principle that minimizes the total energy required to 1) drive the blood flow, 2) maintain the blood supply, and 3) maintain a regulatory smooth muscle tone in the vessel walls (27). In other words, the vessel attempts to maintain a constant wall shear stress and circumferential wall stress. In our study, the influence of mean transmural blood pressure can be ignored because measurements were done within each subject in supine position, with both arteries at heart level.

Another intriguing finding was that no relation could be observed between the shear rate and stress of the CCA and BA, suggesting that shear stress is regulated on a local level. Indeed, we believe that mainly the characteristics of the peripheral vascular bed are of great influence on the shear stress in either artery. The low-resistance vascular system of the cerebrum accounts for an increase in diastolic flow in the CCA. In turn, this particularly increases mean shear stress, which is in accordance with our findings of a higher mean shear stress in the carotid artery and an equal peak shear stress value (Table 4). As we have discussed before (3), it is of interest for the BA, connected to a vascular bed that is subjected to great changes in metabolic demands, to maintain a low baseline shear stress. This would namely allow a large acute rise in flow and thus shear stress. The maximal tolerable shear stress for the vessel wall would, therefore, never be reached despite the minor flow-dependent diameter increase. In other words, minimal work is needed for the vessel to keep shear stress within physiological ranges.

In summary, we have illustrated that mean shear stress differs between vascular beds, in particular between the BA and CCA. Also, no direct relation could be observed between the shear stresses of these two anatomical sites. It is important to realize that viscosity and shear rate are complexly interrelated, and that viscosity is not the same throughout the arterial tree. Furthermore, the assumption of a Poiseuille blood flow in vivo should be questioned because of the distensible vascular system and the viscoelastic properties of blood.

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