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The mechanical properties of exposed human common carotid arteries in vivo*)

Die mechanischen Eigenschaften der freigelegten menschlichen A. carotis communis in vivo

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With 4 figures and 2 tables

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Summary

In exposed common carotid arteries of 15 patients (36–74 years) undergoing neck surgery, the intra-arterial pressure (P) was recorded by means of a catheter-tip manometer and, at the same site, the external diameter (D) by means of a contact-free photoelectric device. On the average, the pulsatile diameter changes were 5.6% of the end-diastolic diameter at pulse pressures of about 50 mm Hg. Due to viscoelasticity, the P–D diagrams exhibited hysteresis loops. Using the criterion of loop elimination, an iterative procedure was applied which permitted, by the use of an appropriate computer program, the separation of the purely elastic and the purely viscous components of the P–D relationships. In all cases, the purely elastic P–D curves markedly deviated from linearity. The tangential elastic modulus (E_t) and the pulse wave velocity (c) calculated from these curves were normalized by dividing these quantities by the respective end-diastolic values and plotted against the normalized external diameters. During each pulse cycle, E_t increased, with increasing diameter, by a factor between 1.2 and 3.5, while c increased by a factor between 1.1 and 1.9 with reference to the respective end-diastolic values.

In the last two decades, numerous investigations on vessels in vitro have led to a reasonable understanding of the mechanical properties of excised human and animal arteries (for a review see [2]). In contrast, there is not much information on the elastic behaviour of arteries in vivo and what is available is contradictory. The main problem encountered in the determination of the elastic properties of arteries in vivo is undoubtedly the inherent difficulty of recording the circumference or the diameter with adequate resolution and frequency response. Most information on arteries in vivo has been obtained with gauges in physical contact with the arterial wall. In some of these studies interest was focussed on the human carotid artery, part of which is supplied with receptors and nerves that play an

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important role in the regulation of blood pressure. *Greenfield* et al. (4) found the distensibility of the exposed human carotid artery to be relatively small, while a non-invasive ultrasonic echo technique employed by *Arndt* (1, 5) indicated an unexpectedly high distensibility of this vessel. It was argued (1, 2) that surgical exposure of the vessel would result in a marked decrease in distensibility. Obviously a gauge applied directly to an artery represents an unacceptable load and, if attached by suturing or gluing, may cause structural changes in the vessel wall. But there is no conclusive evidence to suggest that the mere exposure of an artery causes significant changes in its elastic behaviour. We are of the opinion that the possible effects of the exposure of an artery on its distensibility have not yet been adequately clarified.

The first aim of the present study was to establish whether and to what extent surgical exposure of an artery influences its distensibility. For this purpose, the pulse wave velocity as a measure of distensibility was determined in canine femoral arteries before and after surgical exposure. The second aim of this study was the experimental determination of the elastic properties of the human common carotid artery on the basis of pressure and diameter the latter being recorded by means of a contact-free device.

Methods

Experiments were performed on six dogs weighing 20–30 kg anaesthesized with sodium pentobarbital (30 mg/kg). A cathetertip manometer bearing two pressure sensors at a distance apart of 6 cm (Millar PC 771) was inserted into the carotid artery and advanced into the femoral artery. The pulse-wave velocity was determined from the time interval between corresponding points located on the lower parts of the ascending limbs of the two pressure pulses. An initial series of pulses was recorded with intact skin and tissue in the femoral region. The skin was then opened and the femoral artery carefully freed from surrounding tissue for a length of about 10 cm. Immediately following this preparation a second series of pulses was recorded. In two of the dogs (Nos. 5 and 6 in table 1), a third recording was carried out after a 20-minute period during which the artery was free and not protected from drying.

The studies of the pressure-diameter relationship of the human carotid artery were performed in a total of 15 patients (aged 36–74) undergoing neck surgery. During the course of the surgical procedure, the common carotid artery was exposed for a length of about 10 cm. A catheter-tip manometer (Millar PC 350 A) was introduced through the superior thyroid artery. In 9 patients, it was possible to place the pressure-sensitive membrane of the tip manometer within the carotid exactly at the level of the diameter recording. In the remaining patients, the tip manometer was kept in the side branch. Recording of the diameter of the common carotid artery was carried out by means of a contact-free photoelectric gauge with two photocells (8, 9). The two cantilevers of this instrument were gently pressed against the tissue so that the carotid lay free over the photocells. The frequency response of this gauge was flat up to 200 Hz and the time lag of the signal about 0.2 msec. The resolving power was about 1 μm and the sensitivities of the photocells were adjusted to be uniform within $\pm$ 1% over their whole surfaces.

The pressure and diameter traces were stored on analog magnetic tape (Bell & Howell VR 3200). Analysis of the data was carried out after analog-digital-conversion (Hewlett Packard 5610 A), by means of a digital computer (HP 2100 A).

Theory

As is shown below, the human carotid artery exhibits a marked nonlinear behaviour. For this reason it is not permissible to calculate the tangential elastic modulus (E_t) and the pulse wave velocity (c) from formulae based on Fourier analysis. E_t and c must, therefore, be represented in the time domain, rather than in the frequency domain. We express pressure, stress, E_t and c as functions of the external diameter which itself is a function of time.

To determine E_t, we need the stress-strain relationship. The tangential stress of an arterial segment is given by

$$\sigma_t = P \cdot r_i/h = Pe/(1 - e) \quad (1)$$

substituting $e^2 = 1 - 4S/(\pi D^2)$
where P = transmural pressure, r_i = internal radius, h = wall thickness, D = external diameter, and S = cross-sectional area of the wall. The expression containing S and D is calculated from the quotient r_i/h assuming incompressibility of the arterial wall and longitudinal constraint of arteries in situ.

Under these assumptions, S is constant for a given arterial segment. σ_t can be expressed as a function of D if it is possible to determine a unique relationship between pressure and diameter from in vivo P–D recordings. Although the orginal pressure-diameter relationship shows hysteresis due to the viscoelasticity of the arterial wall, a unique relationship based solely on elasticity, can be obtained in the following way. The pulsatile blood pressure is imagined to comprise two parts, the elastic part (P_{el}), which is related to the tangential wall strain and the viscous part (P_{vis}) which is related to the rate of change of the tangential wall strain. In eq. (1), only the elastic part of the transmural pressure is considered. We have recently described a procedure for the separation of elastic and viscous parts of the pressure (8). In principle this procedure consists in subtracting a term related to the rate of change of the diameter from the recorded pressure (P). The magnitude of the term is found by trial and error using as a criterion the disappearance of the P–D hysteresis loop. The subtracted term is then equal to the viscous part (P_{vis}) and the difference ($P_{el} = P - P_{vis}$) is equal to the elastic part of the recorded pressure. As is illustrated in Figure 3, the P_{el}–D relationship no longer shows any loop and, therefore, allows us to calculate the purely elastic stress (σ_t) as a unique function of the diameter. In every case we examined, these P_{el}–D curves of the human carotid artery exhibit a marked nonlinearity.

The nonlinear P_{el}-D curves can be approximated by a second degree polynomial

$$P_{el} = A + B\,D + C\,D^2 \quad (2)$$

where A, B, and C are coefficients obtained from the P_{el}-D curve by the method of least squares.

Generally, we define the tangential elastic modulus as:

$$E_t = d\,\sigma_t/(du/u) \quad (3)$$

where u is the circumference related to the mean wall layer:
$u = 2 \cdot \pi \cdot (r_i \cdot r_e)^{1/2}$, with r_i = internal radius and r_e = external radius. The relation between u and D is given by

$$u^2 = 4 \cdot \pi^2 \cdot r_i r_e = \pi^2 D \left(D^2 - \frac{4S}{\pi}\right)^{1/2}. \tag{4}$$

Eqs. (1), (2), and (4) inserted into eq. (3) give

$$E_t = \frac{D^3 e^2}{D^2 - \frac{2S}{\pi}} \left[(B + 2CD) \frac{e}{1-e} + (A + BD + CD^2) \frac{4S}{\pi D^3 e (1-e)^2} \right]. \tag{5}$$

In this way, E_t can be calculated as a function of D.

The pulse wave velocity (c) is obtained from the modified version of *Weber's* formula:

$$c^2 = \frac{1}{2\varrho} \cdot \frac{dP_{el}}{du/u} \tag{6}$$

where ϱ = density of blood.
From eq. (4) we obtain

$$du/u = \frac{D^2 - \frac{2S}{\pi}}{D^2 - \frac{4S}{\pi}} \cdot \frac{dD}{D} \tag{7}$$

Inserting this equation into eq. (6) gives

$$c^2 = \frac{D \cdot \left(D^2 - \frac{4S}{\pi}\right)}{2\varrho \cdot \left(D^2 - \frac{2S}{\pi}\right)} \cdot \frac{dP_{el}}{dD} \tag{8}$$

and regarding eq. (2)

$$c^2 = \frac{D \left(D^2 - \frac{4S}{\pi}\right) \cdot (B + 2CD)}{2\varrho \left(D^2 - \frac{2S}{\pi}\right)}. \tag{9}$$

This equation is used for the determination of c as a function of D.

Results

As can be seen from Table 1, a comparison of the values of the pulse wave velocities determined on unexposed and exposed canine femoral arteries reveals only small differences, all of which fall within the range of the standard deviation.

The mean of the end-diastolic diameters of the common carotid arteries of 15 patients was 8.69 ± 1.05 mm. The mean of the relative pulsatile diameter changes was 5.6%, the mean of the pulse pressures was 52 mm Hg. In Figure 1 typical tracings of pressure and diameter recorded in the carotid artery of a 68-year-old patient are shown. These recordings also demonstrate the stability and accuracy of the contact-free photoelectric diameter recording system. This can be seen in the slow registration of pressure and diameter (right) where the blood pressure changes of the second order and higher also give rise to corresponding diameter changes.

Table 1. Comparison of pulse wave velocities determined in femoral arteries of 6 dogs before, and 1 min. after, exposure. Values labelled (*) of dogs 5 and 6 were obtained 20 min. after exposure. Values are mean ± SD, number of determinations in brackets.

Dog	Pulse wave velocity unexposed (cm/sec)	Pulse wave velocity exposed (cm/sec)		Mean blood pressure (mm Hg)
1	1150 ± 60 (15)	1190 ± 50 (15)		175 ± 9
2	780 ± 40 (15)	700 ± 70 (15)		112 ± 8
3	950 ± 70 (18)	1040 ± 60 (15)		147 ± 6
4	1050 ± 50 (18)	1060 ± 70 (18)		168 ± 10
5	650 ± 70 (8)	590 ± 40 (6)*	570 ± 30 (8)	110 ± 7
6	890 ± 70 (11)	960 ± 60 (6)*	930 ± 40 (8)	123 ± 6

Figure 2 shows three P–D loops taken from recordings in a 64-year-old patient. During the registration extrasystoles caused the blood pressure to vary considerably. It can be seen that with increasing pulse pressure the nonlinearity of the P–D-relation becomes more conspicuous. Such non-linear behaviour was always clearly visible when the pulse pressure exceeded about 40 mm Hg.

As pointed out in the theoretical section, for the calculation of the tangential elastic modulus and the pulse wave velocity we use the purely elastic pressure-diameter (P_{el}–D) relationship which was determined for each evaluated pulse. An example is given in Figure 3. The P_{el}–D curves of the 9 patients were approximated by second degree polynomials (eq. [2]) and inserted into eq. (5) in accordance with the procedure described in the theoretical section. Since it was not possible to measure the wall thickness of the carotids of the patients, we assumed it to be 15% of the external diameter (10). From this value the cross-sectional area (S) of the arterial wall was calculated. The ratio of the elastic moduli E_t/E_{to} as a function of the diameter ratio D/D_o is plotted in Figure 4a. The suffix "o" indicates the respective end-diastolic values. It may be seen, that E_t/E_{to} increases within the course of a pulse cycle by a factor between 1.2 and 3.5.

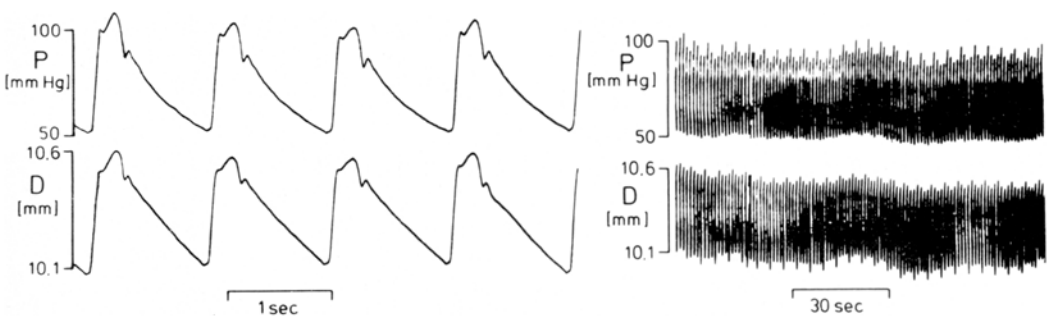

Fig. 1. Simultaneous recordings of pressure (P) and diameter (D) at the same site in the left common carotid artery of a 68-year-old male patient.

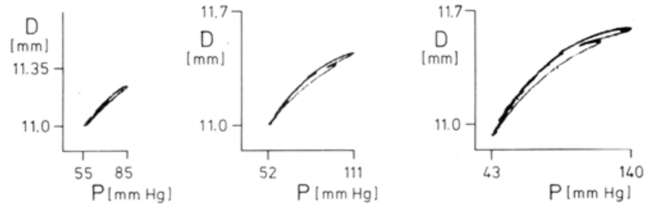

Fig. 2. Pressure-diameter diagrams for the right common carotid artery obtained from a 64-year-old male patient.

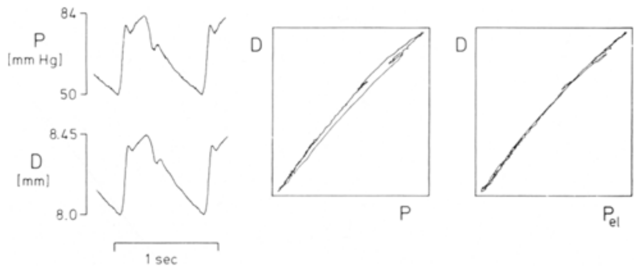

Fig. 3. Left: Pressure (P) and diameter (D) of the right common carotid artery of a 36-year-old male patient. Middle: Pressure-diameter loop of the pulse of left side. Right: Pressure-diameter relationship of the same pulse obtained by elimination of the loop by means of the procedure described in the text; P_{el} = elastic pressure.

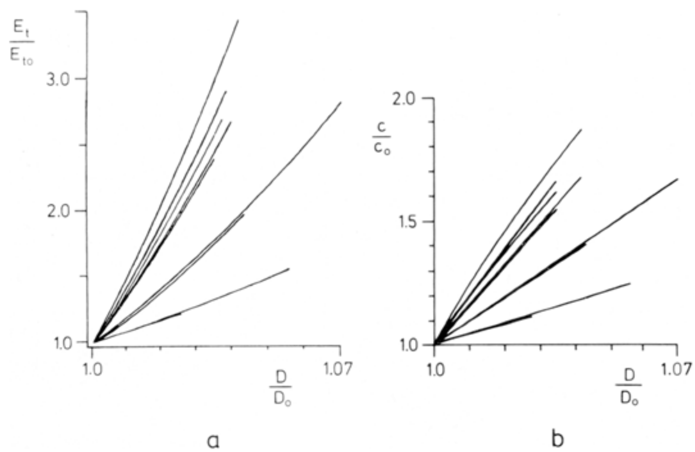

Fig. 4. a) Normalized tangential elastic moduli (E_t/E_{to}) plotted against the normalized external diameters (D/D_o) for the common carotid arteries of 9 patients. b) Normalized pulse wave velocities (c/c_o) plotted against the normalized diameters for the same carotid arteries. The suffixes "o" indicate the end-diastolic value. Age of patients 36 to 74 years. There is no systematic dependence of the slopes of E_t and c on age.

The ratio of the pulse wave velocities (c/c_0) was calculated as a function of D/D_0 by means of eq. (9) using the same polynomials as before. As can be seen from Figure 4b, the wave velocity increased by a factor between 1.1 and 1.9 during a pulse cycle.

Discussion

It is claimed in the literature, that surgical exposure of an artery removes the constraint of the surrounding tissues (2), so that the artery becomes stiffer. Our investigations show that the exposure of an artery does not significantly change the pulse wave velocity. From this we conclude that the influence of surrounding tissues on the mechanical properties of the arterial wall is too small to be measured as a change in the pulse wave velocity. We therefore believe that the objection to the effect that the mechanical properties determined in exposed arteries cannot be applied to unexposed arteries, is unfounded, provided that surgical exposure is carried out carefully.

Table 2 shows a survey of the pulsatile diameter changes of the human common carotid artery previously reported in the literature, including our own results. The figures obtained by *Greenfield* et al. (4) using a strain-gauge caliper are very small and would imply unrealistically high pulse wave velocities. It is probable that the caliper whose legs are sutured to the adventitia represents a mechanical load on the vessel wall.

The relative diameter changes of human carotid arteries as determined with the transcutaneous ultrasonic method by Arndt (1) are about 3 times as great, those determined by *Ungern-Sternberg* et al. (7) about twice as great, as our values. The discrepancy between the results obtained with the ultrasonic method and those we recorded with the photoelectric device cannot be explained by the difference in ages of the subjects examined. This can be seen by a comparison of the second group of patients of *Ungern-Sternberg* (Table 2) with our group. A plausible explanation of the high distensibility found with the ultrasonic method was given by *Mozersky* (6), who showed that even moderate contact pressure of the transducer head results in a relaxation of the arterial wall and, in consequence, in an increase in diameter pulsations.

A further important point of interest is the question of the linearity of the pressure-diameter relationship. *Kober* and *Arndt* (5) recorded the diameter changes of the common carotid artery in 8 conscious men using the transcutaneous ultrasonic technique and simultaneously the pressure in the brachial artery. The transmural pressure of the carotid was varied by means of a pressure chamber surrounding the neck. From their data, the authors concluded that, for the common carotid artery, the relationship between pressure and diameter is linear in the range from 50 to 150 mm Hg. In contrast, we found a marked nonlinearity of the P–D relationship at all pulse pressures of more than about 40 mm Hg. This nonlinearity has far-reaching consequences for the applicability of Fourier analysis since the Fourier components are, in contrast to a linear system, not independent of one another. For example, any Fourier component of diameter is not only related to the corresponding component of pressure, but also to the other harmonics of pressure. This problem has already been

Table 2. Survey of data on the human common carotid artery quoted in the literature (I–III) and in our studies (IV). D = enddiastolic diameter. In (I) and (IV), D is the external diameter, in (II) and (III) the external diameter minus one wall thickness. ΔD = pulsatile diameter change.

	Authors	Method	Subjects or Pat. number	age	Blood pressure (mm Hg)	Diameter (mm) D	ΔD	$\Delta D/D$ (%)
(I)	*Greenfield* et al. (1964)	surgical exposure, strain-gauge-caliper	13	28–69	130/86 (direct)	8.58 ± 1.2	0.08	0.9
(II)	*Arndt* et al. (1968)	transcutaneous ultrasonic echo technique	9	24–34	123/78 (indirect)	7.6 ± 0.5	1.08	14.3
(III)	*Ungern-Sternberg* et al. (1975)	transcutaneous ultrasonic echo technique	13 10	36–52 56–72	125/80 127/70 (indirect)	7.4 ± 0.4 7.9 ± 0.5	0.73 0.6	9.9 8.4
(IV)	*Busse* et al. (1978)	surgical exposure contact-free photo-electric device	15	36–74	112/60 (direct)	8.69 ± 1.05	0.49	5.6

discussed by *Gow* and *Taylor* (3) with respect to the aorta and the iliac and femoral arteries of the dog. Furthermore, in cases of marked nonlinearity the determination of the complex tangential elastic modulus from Fourier components of pressure and diameter must also be questionable. The same also applies to the concept of phase velocity. For this reason, we have avoided representing our results in the frequency domain and have restricted our data to the purely elastic properties of the arterial wall, which are independent of frequency. In this way, the nonlinearity of the pressure-diameter relation can be taken into account in the calculation of the elastic modulus and the pulse wave velocity. Viscous effects, which are of minor importance as compared with pure elasticity, will be considered in future work.

Zusammenfassung

An 15 Patienten im Alter zwischen 36 und 74 Jahren, bei denen während einer Halsoperation die A.carotis communis freigelegt war, wurden simultan der intraarterielle Karotisdruck (P) mit einem Katheterspitzenmanometer und an derselben Stelle der arterielle Außendurchmesser (D) mit einem berührungsfreien photoelektrischen Verfahren registriert. Die relativen Durchmesseränderungen, bezogen auf den enddiastolischen Außendurchmesser, betrugen im Mittel 5,6% bei Druckamplituden von etwa 50 mm Hg. Die durch die Viskoelastizität der Arterienwand bedingten Hystereseschleifen der pulsatorischen P–D-Beziehungen wurden mit Hilfe eines iterativen Verfahrens unter Verwendung eines Digitalrechners eliminiert, wodurch eine Trennung der rein elastischen und der rein viskösen Komponenten der P–D-Beziehung ermöglicht wurde. Aus den rein elastischen P–D-Beziehungen wurden der tangentiale Elastizitätsmodul und die Pulswellengeschwindigkeit als Funktionen des Außendurchmessers berechnet und in normierter Form, d. h. bezogen auf den jeweiligen enddiastolischen Wert, dargestellt. Mit wachsendem Durchmesser stieg der Elastizitätsmodul während jedes Pulszyklus auf das 1,2- bis 3,5fache des enddiastolischen Wertes an. Entsprechend nahm die Pulswellengeschwindigkeit auf das 1,1- bis 1,9fache des enddiastolischen Wertes zu.

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