

Evaluation of Human Hemispheric Blood Flow Based on Non Invasive Carotid Blood Flow Measurements Using The Range-Gated Doppler Technique

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SUMMARY Common carotid blood flow (CCBF) was measured in 11 anesthetized patients without extracranial arterial disease (nine acute subarachnoid haemorrhages and two cases of head injury). The range-gated Doppler flowmeter with an adjustable range-gated time system and a double transducer probe was used to determine diameter, blood velocity, and blood flow of the common carotid artery. Values were, respectively, 5.9 ± 1.1 mm, 13.8 ± 6.1 cm \cdot sec⁻¹ and 387 ± 183 cm³ \cdot min⁻¹. Mean cerebral blood flow (rCBF) measurements were simultaneously made by ¹³³Xenon intra-arterial method, with a value of 22.28 ± 5.96 ml \cdot min⁻¹ \cdot 100 g. High correlation coefficient was found between CCBF and mean rCBF ($r = 0.73$, $p < 0.001$). The range-gated Doppler flowmeter thus provides a non invasive and easily duplicated method for monitoring cerebral blood flow in anesthetized patients.

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THE ¹³³XENON INTRA-ARTERIAL injection technique is effective for clinical evaluation of regional or mean cerebral blood flow.¹ The ultrasound Doppler technology has permitted extensive studies of blood flow velocities particularly in extracranial arterial diseases.²⁻⁷ Yoshida et al⁸ compared internal carotid Doppler velocity and rCBF values (internal carotid injection of ¹³³Xenon) in 98 patients who had mainly cerebral arteriosclerotic diseases. A fairly low correlation coefficient of 0.41 was found between the velocity and average rCBF values, while internal carotid flow (velocity \times surface of section) showed a correlation coefficient of 0.77 with rCBF. Risberg et al.⁹ compared simultaneous determination of cerebral blood flow (¹³³Xenon inhalation technique) in 30 subjects with normal arterial flow conditions. Fairly high correlation coefficients were found between rCBF values and end diastolic velocities in internal carotid ($r = 0.88$) and in common carotid ($r = 0.71$).

Previously reported results obtained by using continuous wave Doppler methods could not measure blood flow rates (cm³ \cdot min⁻¹) but only blood flow velocity (cm \cdot sec⁻¹) because a continuous wave system cannot measure the arterial diameter and the mean cross sectional velocity. Prior studies with a range-gated technique measured blood flow rates in radial,¹⁰ brachial,¹¹ and common carotid¹²⁻¹⁴ arteries. The latter studies did not compare the results to the cerebral blood flow. The purpose of this paper is to describe a range-gated Doppler blood flowmetry technique and to

compare blood flow in the common carotid artery with mean hemispheric cerebral blood flow as measured by the ¹³³Xenon intra-arterial injection technique.

Material and Methods

Patients

Twenty-one measurements were performed in 11 patients (5 females and 6 males) anesthetized for cerebral angiography. Mean age, weight, and body surface area were respectively, 39 years (range: 20 to 56), 61 kg (range: 48 to 85) and 186 cm² (range: 146 to 200). Diagnoses were nine acute subarachnoid hemorrhages and two cases of recent head injury. All patients were untreated and had cerebral angiography within 24 to 48 hours following the initial accident to assess subarachnoid haemorrhaging due to a ruptured intracranial aneurysm or to determine the extent of head injuries. All patients had normal angiograms of the extracranial circulation. In all patients, anesthesia was initiated by thiopentone (2.5 mg \cdot kg⁻¹) with phenoperidine (30 μ g \cdot kg⁻¹), and suxamethonium (1 mg \cdot kg⁻¹) was used for orotracheal intubation. Artificial ventilation (RPR Pesty Technomed) was used during the procedure with O₂ and N₂O (FiO₂:0.5), and anesthesia was maintained by bolus injections of phenoperidine (0.5 mg every thirty minutes), curarization was maintained by pancuronium bromide (0.06 mg \cdot kg⁻¹).

rCBF measurements by ¹³³Xenon intra-arterial injection

The ¹³³Xenon intra-arterial method has been previously described in detail.¹ In order to avoid contamination of the cerebral clearance curves with data from extracerebral tissues, the isotope was injected into the internal carotid, which was cannulated via the common carotid artery according to the Seldinger technique, by a small polyethylene catheter. The correct position of the catheter in the internal carotid artery was verified by fluoroscopic control. A bolus of 1 millicurie of ¹³³Xenon dissolved in 1 ml sterile saline solution was quickly injected. The wash-out of radio-

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activity was measured by a portable detector (Mecaserto MO 141) placed over the ipsilateral temporoparietal area of the skull. Collimation was provided by a cylindrical lead tube, 50 mm long and 20 mm in diameter, and a (1" × 1") NaI crystal. The detector was connected to a ratemeter with a linear writing potentiometer. The time-constant used in the ratemeter was one second for the first three minutes. The rCBF was calculated from the slope of the logarithmically displayed first two minutes of the clearance curve,^{15, 16} according to the following equation:

$$\text{rCBF} = 100 \cdot \lambda \cdot 0.693 \cdot T^{-1},$$

where T is the half-time of radioactivity and λ is the average blood-tissue partition coefficient. The rCBF of the temporoparietal region was considered as the hemispheric blood flow value.

Doppler blood flow measurements

Description of the range-gated Doppler flowmeter

Doppler flowmetry is based on the frequency change ΔF of the emitted wave after its backscattering by moving erythrocytes. ΔF is a function of the emission frequency (F), the velocity of the red cells (V) and the angle (θ), between the ultrasonic beam and the direction of blood displacement, according to the classical equation:

$$\Delta F/F = \frac{2V}{C} \cos \theta,$$

where C is the mean propagation velocity of ultrasound within tissues (1540 m/sec). The method was introduced by Satomura¹⁷ and Franklin,¹⁸ who used a continuous emission apparatus with one transmitting and

one receiving ultrasonic transducer. In addition to these classical Doppler features, the apparatus used (Echovar Doppler Pulse Alvar R*) in our investigations had (i) an adjustable range-gated time system and (ii) a double transducer probe which determined the angle between the ultrasonic beam and the vessel axis (Peronneau et al,^{19, 20} Baker²¹). A simple transducer was alternatively emitter and receiver (fig. 1). The ultrasound frequency used was 8 MHz, with emission duration of 0.5 μsec , and pulsed at a repetitive frequency of 15 or 30 KHz. Between emitted pulses, the transducer functioned as a receiver. An electronic gate was adjusted between pulses, t being the time delay and τ the gate duration. t and τ can be selected in order to analyse the Doppler signal in a definite sample volume at a definite distance from the transducer. With such a system, it is possible to determine exactly the distance (d) between the red cell and the transducer, according to the echographic relation $d = t \frac{C}{2}$. Under

these conditions, half a microsecond represents 0.4 mm (Peronneau et al¹⁹). The time delay t and the gate duration τ represent respectively, the depth and thickness of the sample volume along the beam axis. Furthermore, a small duration enables local velocity to be measured. This procedure can be applied to the determination of the diameter and the cross-sectional blood velocity of the vessel. The gate duration was chosen to obtain the smallest convenient sample volume size with significant energy (1 μsec for an emission duration of 1 μsec). However the smallest sample volume size is finite and yet not a point; errors may occur at the entry to and exit from the vessel because a part of the sample volume is out of the vessel lumen (Peronneau et al¹⁹). This can be minimized by using focused lenses

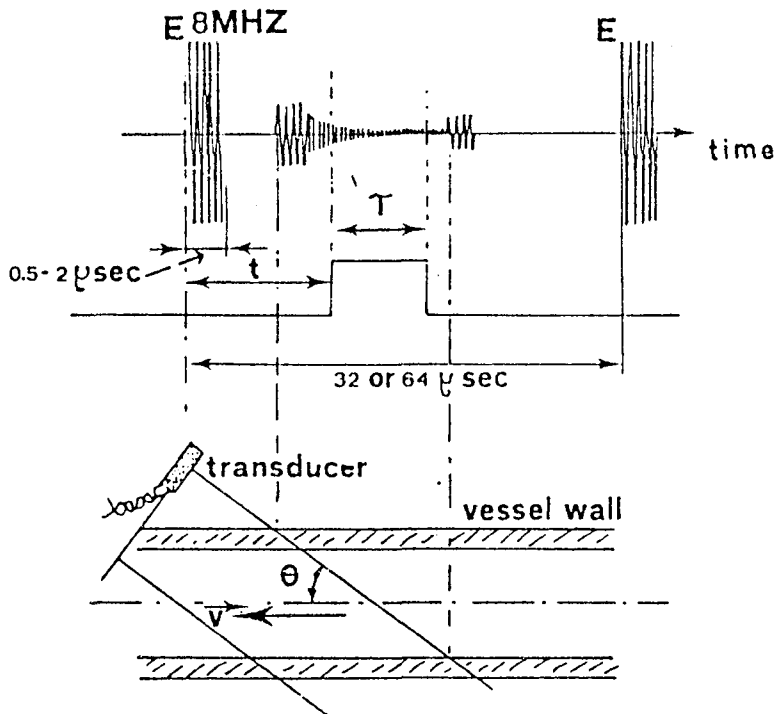


FIGURE 1. Diagram of the Doppler signal analysis: Emission (E) has an oscillation frequency of 8 MHz and a duration of 0.5, 1, 1.5 or 2 μsec . The time interval between two emission pulses is 32 or 64 μsec . The reception is defined by its time delay from the emission (t) and its duration (τ). Angle θ represents the angle between the ultrasonic beam and the velocity vector (V) of the blood column.

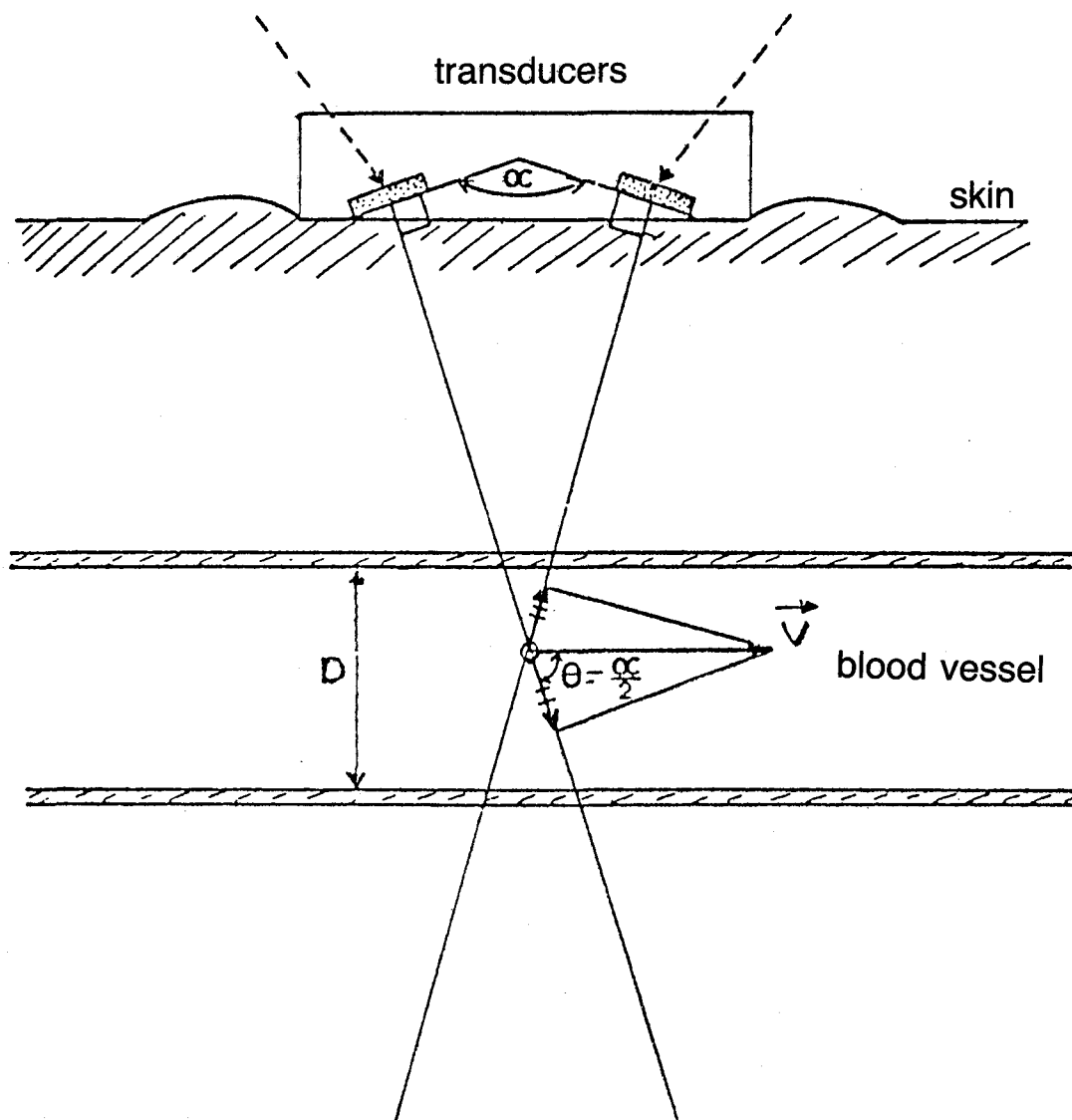


FIGURE 2. Bi-dimensional determination of the velocity vector. The value α represents the angle formed by the two transducers. The angle θ (angle between each ultrasonic beam and the vascular axis) equals $\alpha/2$ when the velocity components recorded by each transducer are equal in absolute values.

on the transducer to prevent divergence of the ultrasonic beam.

The probe includes a double transducer system (Peronneau²²) forming a fixed angle, α , as shown in figure 2. The probe was designed so that the intersection of the beams occurs at about the distance where maximal accuracy is needed. The two transducers are successively activated, and a simple calculation provides the longitudinal velocity within the plane defined by the ultrasonic beam and the vessel axis. In the present system, the α angle of the transducers was fixed at 120° .

CCBF measurements

The pathway was determined by palpation of the neck. An ultrasonic gel was used as a coupling medium between the probe and the skin. The Doppler signals were monitored by loud-speaker throughout the

examination, and simultaneously recorded on an ALVAR apparatus.

Determination of the ultrasonic beam incidence angle

An approximation of the common carotid artery location was first made by adjusting the time delay and gate duration.²¹ The time delay was fixed between 20 and 30 μsec , corresponding to the usual depth of the artery. The gate duration was adjusted progressively from 10 to 20 microseconds in order to obtain Doppler signals from the entire cross-section of the vessel. Under these conditions, two velocity curves, corresponding to the two transducers, were easily recorded. The probe position was adjusted so that the 2 velocity curves, successively recorded for a duration of ten cardiac cycles, were equal in absolute values (fig. 3), giving an ultrasonic beam incidence angle of 60° . Then the probe was kept in place over the artery by means of

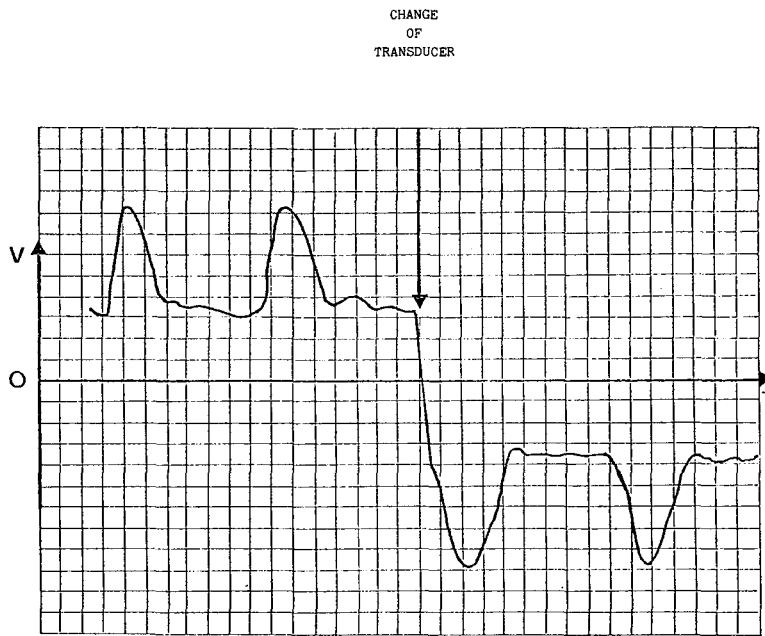


FIGURE 3. The figure represents the bi-dimensional instantaneous cross-sectional carotid velocity curve. The arrow indicates the change transducer. Velocity (V) is represented on the ordinate and time on the abscissa.

a flexible arm. This procedure was repeated throughout the examination before each velocity measurement.

Location of the arterial walls

With the two transducers correctly located at 60° incidence angle, emission duration and gate duration were adjusted to 1 microsecond. Arterial walls were located by moving the sample volume across the vessel lumen. The sample volume was advanced by using a constant step of $0.5 \mu\text{sec}$ (i.e., 0.4 mm). Since the accuracy of the location of the proximal and distal arterial walls equals the value of the step advance, t_1 and t_2 were the time delay of the first and the last Doppler signals, when the sample volume was close, respectively, to the proximal and distal walls relative to the transducer. The apparent echodiameter of the vessel D could be deduced from the difference $t_2 - t_1$ according to the formula

$$D = \frac{C}{2} (t_2 - t_1) \sin \theta.$$

After determination of D , the velocity over the vessel cross-section was obtained by adjusting the time delay to the proximal arterial wall and the gate duration to the diameter D . The calibration voltage of the apparatus corresponded to a velocity of 38 cm/sec for an incidence angle of 60° . Mean velocity (V_m) was calculated by electronic integration of the velocity curve and was the mean value of 10 successive cardiac cycles on each transducer. From the mean velocity (V_m), flow rate could be calculated according to the formula

$$Q = \frac{\pi D^2}{4} \cdot V_m.$$

Protocol

The study was performed on the right and left sides in 11 patients. In one case, rCBF was not carried out

because of difficulty in cannulating the carotid artery. The protocol was approved by INSERM (Institut National de la Santé et de la Recherche Médicale). Patients were placed in supine position on an examination table after control observations of blood pressure and blood gases during anesthesia infusion. The first CCBF measurement was made before carotid catheterization. $^{133}\text{Xenon}$ was injected a few minutes later. Half an hour after the recording of a clearance curve, a second CCBF measurement was performed. The protocol was the same for both sides. Serial angiography was performed after hemodynamic measurements.

Statistical methods

Means, standard deviations and correlation coefficients were calculated according to standard statistical methods. Regression analysis was performed using the least squares method.²³

Results

Table 1 shows the clinical characteristics of the patients and the arterial PO_2 , PCO_2 and pH during investigation.

Table 2 shows the rCBF and the range-gated Doppler measurements in the common carotid artery. The arterial diameter, the blood flow velocity and the common carotid blood flow (CCBF) were 5.9 ± 1.1 mm, $13.8 \pm 5 \text{ cm} \cdot \text{sec}^{-1}$ and $387 \pm 183 \text{ cm}^3 \cdot \text{min}^{-1}$ before catheterization and 5.6 ± 1.2 mm, $15.5 \pm 6 \text{ cm} \cdot \text{sec}^{-1}$ and $381 \pm 240 \text{ cm}^3 \cdot \text{min}^{-1}$, after. The mean value of rCBF simultaneously measured was $22.28 \pm 5.96 \text{ ml} \cdot \text{min}^{-1} \cdot 100 \text{ g}$.

TABLE 1 Clinical Characteristics and Blood Gases

| n = 11 | Age year | Weight kg | BSA m_2 | PH | PCO_2 kPa | PO_2 kPa |
|----------|-------------|--------------|---------------------|------|-----------------------|----------------------|
| mean | 39 | 61 | 168 | 7.49 | 3.7 | 23.7 |
| \pm SD | 13 | 13 | 6 | 0.06 | 0.7 | 9.7 |

TABLE 2 *rCBF and Range-gated Doppler Measurements in Common Carotid Artery Before and After Catheterization*

| n = 21 | rCBF ml · min ⁻¹ · 100 g | Diameter mm | | Velocity cm · sec ⁻¹ | | CCBF cm ³ · min ⁻¹ | |
|--------|--|----------------|-------|------------------------------------|-------|---|-------|
| | | before | after | before | after | before | after |
| mean | 22.28 | 5.9 | 5.6 | 13.8 | 15.5 | 387 | 381 |
| ± SD | 5.96 | 1.1 | 1.2 | 6.1 | 6 | 183 | 240 |

There was no significant difference between CCBF measured before and thirty minutes after carotid catheterization.

The relationship between CCBF and rCBF is shown in figure 4 ($r = 0.73$).

Discussion

Range-gated Doppler technique

One of the fundamental difficulties in quantitative measurements of blood flow is the lack of information concerning the angle between the ultrasonic beam and the vessel axis. The ultrasonic probe with two successively activated transducers permits determination of this angle. Knowing the value of θ it is possible to measure the vessel diameter, necessary for the flow rate calculation. The diameter of the common carotid artery as indicated by this apparatus, is similar to that measured by the ultrasonic B-Scan technique.²⁴ The shape of the blood flow velocity curves as measured by the range-gated Doppler method and the continuous-wave apparatus are very similar. Furthermore, the

range-gated Doppler technique is the only ultrasonic Doppler method capable of indicating the blood flow rate, since it can determine the angle between the ultrasonic beam and the vessel axis and simultaneously measure vessel diameter.¹¹ In a previous paper, one of the present authors established fairly good correlation between peripheral arterial blood flow as measured by range-gated Doppler and classical plethysmographic techniques.¹⁰

rCBF

The rCBF obtained in this study are approximately 40% lower than those measured in normal conscious subjects and in agreement with previously reported results for the same diseases.²⁵⁻²⁸ Anesthesia and, obviously, the pathology explain the reduction in mean value, and the high dispersion in flow from this mean value, with normal gasometric parameters.

CCBF

Our results in CCBF are very close to those reported by Olson (325 cm³ · min⁻¹),²⁴ Borodzinsky (530 cm³ ·

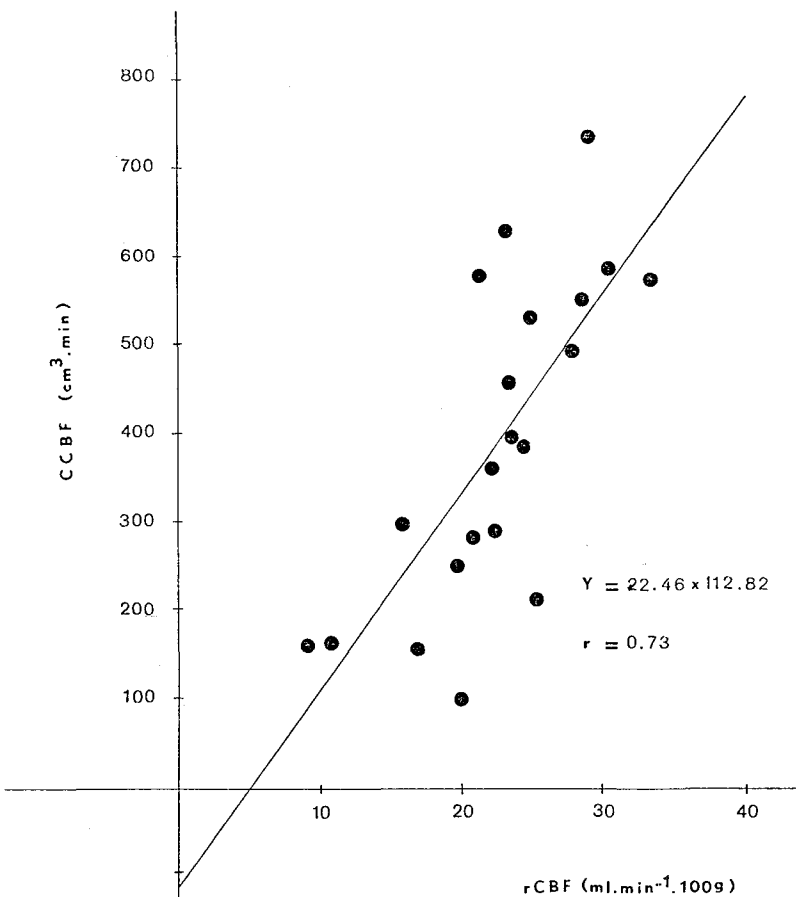


FIGURE 4. *Correlation between rCBF and CCBF.*

min^{-1})¹³ and Keller (300 to 480 $\text{cm}^3 \cdot \text{min}^{-1}$)¹⁴ who used similar range-gated Doppler technique. The system design differed by the angle between the ultrasonic beam and the vessel axis. Like these authors we calculated CCBF as the product of blood velocity (angle corrected) and average cross-sectional area. We assumed that arterial diameter variations during the cardiac cycle could be omitted. It is generally admitted that the pulsatile movement of the vessel walls averages less than 10% of the arterial diameter (Olson^{12, 24}). The absence of variation in CCBF before and after internal carotid catheterization allowed us to conclude that (i) the probe position remained unchanged during the investigation and that (ii) thirty minutes after introduction of the catheter, vessel diameter and blood flow velocity were not significantly changed.

Relationship rCBF/CCBF

The relationship between rCBF and CCBF was statistically significant: $\text{CCBF} = 22.5 \text{ rCBF} (\text{ml} \cdot \text{min}^{-1} \cdot 100 \text{ g}) - 113 (\text{cm}^3 \cdot \text{min}^{-1})$; ($r = 0.73$; $p < 0.001$). rCBF measurements are given as the perfusion of 100 g of brain tissue, inversely CCBF measurements are in absolute flow value depending on the organ weight. This difference may explain the slope of the linear relationship. The vertebral artery blood flow was not measured in the present study, due to technical difficulties in reliable recording of blood flow rate in deep and curved arteries. The negative intercept ($-113 \text{ cm}^3/\text{min}^{-1}$) observed in the CCBF-rCBF relationship may be related to the fact that (i) the two methods measure different parameters, (ii) the CCBF determination gives data only on flow to a portion of the brain (iii) the fraction of the vertebral blood flow and the anastomosis in rCBF measurement in this type of disease, are impossible to determine with our method. The value of the correlation coefficient ($r = 0.73$) is close to the one reported by Yoshida et al ($r = 0.77$)⁸, who evaluated the internal carotid blood flow as the product of the mean blood velocity obtained by continuous-wave Doppler technique and the square of the vessel diameter obtained a few days later from a frontal X-ray film during angiography. It was impossible to measure the internal carotid blood flow using our double transducer probe, for two main reasons: first, it is not always possible to distinguish which is internal carotid and which is external carotid; second, there is a high degree of variability in the configuration of the carotid bifurcation, and of the axes of the internal carotid. Thus, the reproducible probe-artery angle which is necessary for quantification of blood flow cannot be guaranteed for deep and curved arteries such as the internal carotid. However, we show that a non invasive method applied to the common carotid artery provides appropriate correlation of flow in that vessel with rCBF. Furthermore, blood flow and diameter measurements can be obtained quickly and easily by our Doppler method, allowing a non invasive means for monitoring of cerebral blood flow.

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