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ULTRASONIC SYSTEM FOR NONINVASIVE MEASUREMENT OF HEMODYNAMIC PARAMETERS OF HUMAN ARTERIAL-VASCULAR SYSTEM

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This paper presents the working principle of an ultrasonic system constructed for the simultaneous noninvasive measurement of the blood flow velocity and, the diameter of the blood vessel. A bi-directional c.w. Doppler flowmeter was used to measure the blood flow velocity. The echo method was used to measure the blood vessel diameter and its changes. The values of the parameters measured were transfered to the computer connected on line with the ultrasonic measuring system. A programme was elaborated for computer analysis of a number of hemodynamic parameters determined from the measured blood flow velocity and the instantaneous diameter of a blood vessel. They are the blood flow rate, the blood pressure, the vascular input impedance and the elasticity of arterial vessel walls. Connected to a computer, the ultrasonic measuring system was used in examinations of the carotid arteries.

W pracy przedstawiono zasadę działania opracowanej aparatury ultradźwiękowej do równoczesnego, nieinwazyjnego pomiaru prędkości przepływu krwi i średnicy naczynia krwionośnego. Do pomiaru prędkości przepływu krwi użyto dwukierunkowego dopplerowskiego przepływomierza fali ciągłej. W pomiarze średnicy naczynia krwionośnego i jej zmian zastosowano metodę echa. Wartości mierzonych parametrów przesyłane były do komputera, który połączony był „on line" z ultradźwiękowym aparatem pomiarowym. Opracowano program analizy komputerowej szeregu parametrów hemodynamicznych wyznaczanych na podstawie mierzonej prędkości krwi i chwilowej średnicy naczynia krwionośnego. Są to prędkość objętościowa krwi, ciśnienie krwi, wejściowa impedancja naczyniowa, elastyczność ścianek naczyń tętniczych. Połączony z komputerem ultradźwiękowym system pomiarowy został zastosowany do badań tętnic szyjnych.

1. Introduction

The recent years saw a serious increase in the number of diseases of the vascular system. According to 1983 data, about 40% of all deaths is caused by diseases of the circulatory system. Of these, 40% are caused by atherosclerosis [15]. As a result of atherosclerotic vessel diseases, the lumen of the blood vessel narrows or closes. Most often, this occurs in large or medium arteries most significant for the human organism, namely the coronary, cranial and renal arteries, the aorta, the arteries of the lower limbs and mesenteric arteries. Atherosclerosis can develop for a long time, originating in the early years of life. Often clinical symptoms appear very late when the disease is much advanced and it is too late to treat its results. Therefore, there is the constant need for doing research on the methods and diagnostic equipment permitting the early identification of pathological changes in man's arterial vessels. In this direction, much progress was due to the introduction in medical diagnosis of the noninvasive ultrasonic technique, in particular ultrasonic Doppler flowmeters permitting the estimation of blood flow rate in blood vessels. They were widely applied, e.g., in examinations of patency of extracranial carotid arteries [22] and the arteries of the lower limbs [11].

The ultrasonic Doppler method does not ensure full diagnosis of man's vascular system. This results mainly from the fact that blood flow rate is just one of the many hemodynamic parameters describing the state of the circulatory system. The equally important parameters include the blood pressure, the complex vascular input impedance and the elasticity of arterial vessel walls.

An example of the new approach to diagnosis of the vascular system is the noninvasive ultrasonic method and system constructed at the Department of Ultrasonics, Institute of Fundamental Technological Research, Polish Academy of Sciences, for the noninvasive examination of the blood flow rate the vascular input impedance and elasticity of the carotid arteries [18, 19].

The vascular input impedance is defined by the ratio between the blood pressure and the blood flow rate for successive harmonic frequencies of the work of the heart [1, 4, 5]. In the method in question, the blood pressure is determined from displacements of the arterial blood vessel [18, 19]. These displacements are measured with accuracy up to 0.03 mm over the same time and vessel cross-section as the blood velocity. The impedance is calculated by the discrete Fourier transform of the time courses of the blood pressure and flow rate. It is implemented on a MERA–60 computer connected on line to the ultrasonic measuring system.

Measurements of displacements of the arterial vessel walls can also serve for evaluation of the elasticity of arteries. It is very significant from the point of view of the diagnosis of the complex of diseases of blood vessels, called arteriosclerosis. This group includes all changes in arteries which lead to fibrosis of part or whole of the arterial wall, and, as a result, to a change in its elasticity.

The purpose of this study is to present the principal part of the system constructed for the examination of the blood flow rate, vascular input impedance and elasticity of arterial walls. It is an ultrasonic meter of blood velocity, wall displacements and blood vessel diameter. This paper also discusses the basic assumptions and dependencies adopted in computer analysis in the determination of the vascular input impedance and elasticity of the carotid arteries from blood flow velocity and the instantaneous diameter of a blood vessel.

2. Method and system for digital measurements of displacements of blood vessel walls

The noninvasive ultrasonic echo method was used to measure displacements of blood vessel walls. Information on the displacement amplitude of a vessel wall is obtained by measuring the distance between the ultrasonic probe and the examined wall of a blood vessel. Changes in the diameters of peripheral arterial vessels under increased blood pressure are of the order of 0.1 mm. The investigation of so small displacements require high accuracy in tracing and measuring the position of the echo detected from the vessel wall.

In 1972 HOKANSEN [10] proposed an idea of tracing and measuring displacements of a blood vessel wall by means of the ultrasonic echo method. Applying an analog system of his own construction, he measured movements of walls of the femoral artery. In 1985, HOEKS [9] proposed a different conception of the measurement. For this purpose he used a multi-gate pulsed Doppler flowmeter. Using it, he measured the relative changes in the diameter of the common carotid artery. Moreover, this method raises large objections about the accuracy of representing vessel wall displacements.

This study presents a method for digital measurements of displacements of walls of a blood vessel and its diameter. It is an extension of the measurement conception proposed in 1982 by GROVES, POWAŁOWSKI and WHITE [7]. The general idea of this method is shown in Fig. 1.

An ultrasonic probe set perpendicularly to the blood vessel (see Fig. 4) emits towards it at intervals T_p impulses of the ultrasonic wave. The measurement of the position of the vessel wall with respect to the ultrasonic probe consists in counting clock impulses over the time between the trigger impulse initiating the transmitted impulse and the rising slope of the echo detected from the vessel wall. The time measured by the digital method is the basis for calculating the instantaneous distance d between the ultrasonic probe and the surface of the blood vessel wall. For a successive nth measurement cycle, this distance is

$$d_n = \frac{c N_n}{2 f_z}, \tag{1}$$

where N_n is the number of clock impulses counted in the nth measurement cycle, f_z is the frequency of the clock impulses and c is the ultrasonic wave velocity in the medium investigated.

A level comparator was used to determine unambigously the time when the echo slope occurs. In the comparator, the echoes detected from biological structures are transformed into a series of rectangular signals. The measurement of the position of a chosen echo slope is repeated for every cycle of work of the impulse transmitter. It is assumed in the further description of the measurement that the traced slope of the echo E2 overlaps the end of the preceding echo E1. These two echoes displace in

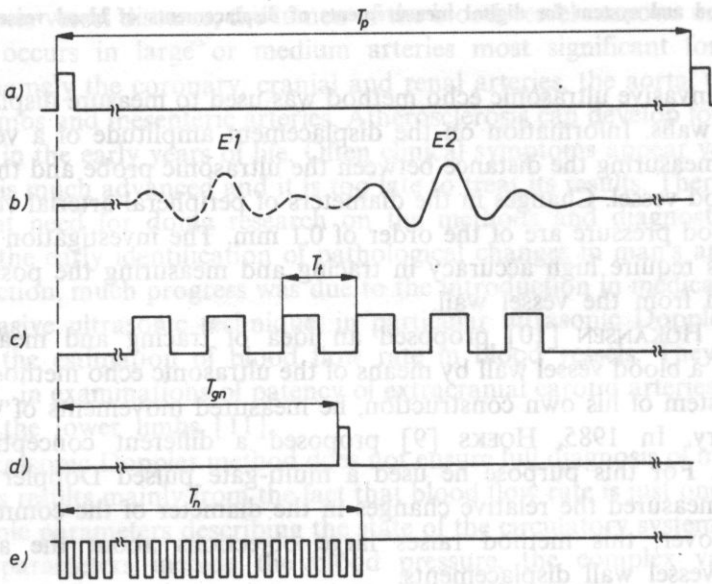

FIG. 1. The principle of the digital measurement of the time variable position of the blood vessel wall: a) trigger impulses, b) echos detected from the external and internal blood vessel walls, c) echos at the output of the level comparator, d) tracing gate, e) clock impulses

the same direction at the same velocity. A situation resembling the above one can occur for echos detected from the external and internal surfaces of a blood vessel wall.

To identify the chosen echo slope, a tracing gate is generated before it. The logical unit of the system stops the time measurement with the clock impulse which occurs after the appearance of the first rising echo slope after the tracing gate. In a current measurement cycle, the position of the tracing gate depends on the position of the echo slope traced in a preceding cycle, and is

$$T_{gn} = T_{n-1} - t_0, \tag{2}$$

where T_{n-1} is the time measured digitally between the trigger impulse and the traced echo slope in the $(n-1)$th measurement cycle, T_{gn} is the time delay of the tracing gate with respect to the trigger impulse in the nth measurement cycle and t_0 is a constant time shift. The time shift t_0 is implemented digitally in the form of N_0 clock impulses.

Fig. 2 shows two conceptions of the detection of the traced echo slope after the appearance of the tracing gate. If the tracing system responds to the positive level of the echo at the comparator output (Fig. 2a), the time shift t_0 of the tracing gate can

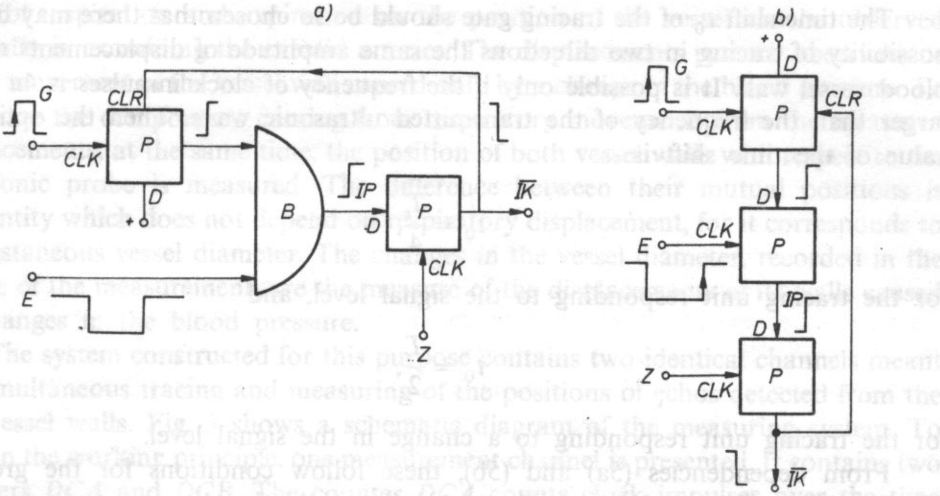

FIG. 2. Two ways of the identification of the positive echo slope by the tracing system: a) the unit responding to a positive echo level, b) the unit responding to changes in the echo level from low to high. B – *AND* gate, P – D Flip Flop, E – echo of the comparator output, G – tracking gate, Z – clock

be contained within the following limits:

$$T_z \leqslant t_0 < \frac{T_t}{2}, \tag{3}$$

where T_t is the ultrasonic wave period and T_z is the clock signal period.

The version of the tracing unit of this type was proposed by GROVES et al. [7]. There is, however, another solution in which the time shift t_0 can be twice as much. This applies to the tracing system which responds to a change in the level from a low to a high one of the echo signal at the comparator output (Fig. 2 b). Then, the time shift of the tracing gate can be

$$T_z \leqslant t_0 < T_t. \tag{4}$$

The difference between these two solutions can readily be explained by assuming that the tracing gate is shifter with respect to the traced echo slope by the time $T_t/2 < t < T_t$. It often occurs in the course of the positive phase of the echo preceding the traced slope. At the comparator output, when the tracing gate appears, there is a high level. For the first system (Fig. 2 a), it is false information about the occurrence of the traced echo slope, and the tracing unit generates the impulse IK which ends the time measurement. In the second solution (Fig. 2 b), the tracing unit does not respond to the high level of the comparator. It gives a signal for ending the measurement only when at the comparator output there is a change in the signal level from a low to a high one, corresponding to the appearance of the traced echo slope.

The time shift t_0 of the tracing gate should be so chosen that there may be the possibility of tracing in two directions the same amplitude of displacements of the blood vessel wall. It is possible only if the frequency of clock impulses is $4n$ times larger than the frequency of the transmitted ultrasonic wave. Then the optimum value of the time shift is

$$t_0 = \frac{T_t}{4}, \tag{5a}$$

for the tracing unit responding to the signal level, and

$$t_0 = \frac{T_t}{2}, \tag{5b}$$

for the tracing unit responding to a change in the signal level.

From dependencies (5a) and (5b), these follow conditions for the greatest possible displacements of the traced vessel wall over the time between two successive impulses from the transmitter of the measuring system. They are equal respectively to 1/8 and 1/4 of the wavelength of the transmitted ultrasonic wave in the medium under study. For the frequency of the transmitted wave of the order of MHz, these are very small displacements of the order of hundredths or tenths of a millimetre. Hence, there follows a general condition which should be satisfied by the measuring system in tracing the displacements Δu occurring over the time Δt:

$$\frac{cf_p}{2mf_t} \geqslant \left(\frac{\Delta u}{\Delta t}\right)_{\max}, \tag{6}$$

where f_t is the frequency of the transmitted wave, f_p is the repetition frequency of the transmitted impulses and m is a factor whose value depends on the time delay t_0 and is 4 for $t_0 = T_t/4$, or 2 for $t_0 = T_t/2$. The satisfaction of the above condition is restricted in the range of selecting the frequency ratio f_p/f_t. For, on the other hand, the values of the two frequencies are conditioned by the necessity of obtaining the needed resolution and range of measurement.

In the measuring system described here, meant mainly for the examination of the carotid arteries, the frequency of the transmitted ultrasonic wave is 6.75 MHz, whereas the repetition frequency of the transmitted impulses is 18 KHz. The maximum velocity of wall displacements caused by the blood pressure in these arteries does not exceed a dozen or so mm/s. From condition (6), the measuring system makes it possible to map fully the movements of vessel walls if the two above ways of detecting the traced echo slope are applied. On the other hand it should be noted that the tracing unit responding to a change in the signal level permits the tracing of displacements which are twice as fast compared with the unit responding to the signal level. Due to this, it ensures more stable measurement, in particular if there are additional displacements of the blood vessel with respect to the ultrasonic probe. One of the factors which cause the changes is the respiratory motion. The effect of inspiration and expiration on the position of the vessel can be observed distinctly in examining the carotid arteries. The displacements perturb observations of the movement of the vessel wall caused only by a change in the blood pressure. To eliminate the components coming from respiratory movements from the measured displacements, at the same time, the position of both vessel walls with respect to the ultrasonic probe is measured. The difference between their mutual positions is a quantity which does not depend on respiratory displacement, for it corresponds to the instaneous vessel diameter. The changes in the vessel diameter, recorded in the course of the measurement, are the measure of the displacements of its walls caused by changes in the blood pressure.

The system constructed for this purpose contains two identical channels meant for simultaneous tracing and measuring of the positions of echos detected from the two vessel walls. Fig. 3 shows a schematic diagram of the measuring system. To explain the working principle, one measurement channel is presented. It contains two counters *DCA* and *DCB*. The counter *DCA* counts clock impulses over the time interval between the trigger impulse releasing the transmitter of the impulse wave and the echo front detected from the blood vessel wall. At the same time, in the counter *DCB*, clock impulses are subtracted from the content of the counter *DCB* recorded in a preceding cycle. The difference value at the output of the counter *DCB* is compared with the programmed quantity N_0 in the digital comparator C_d. The comparator C_d generates the tracing gate *G* as soon as a numerical value equal to N_0

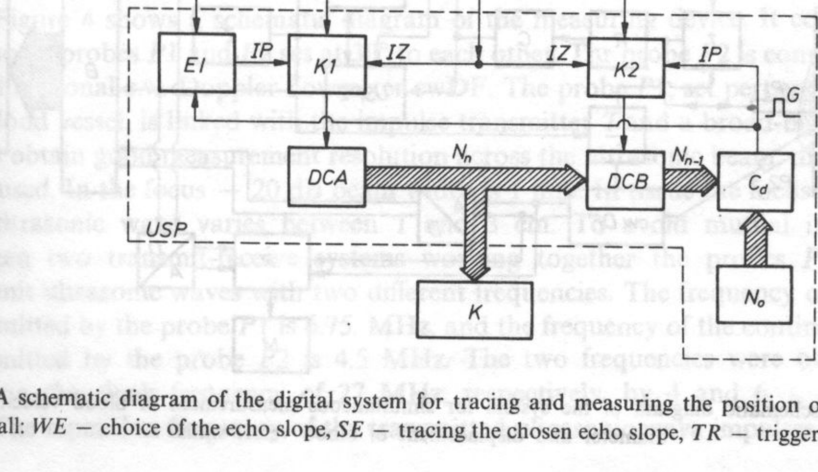

FIG. 3. A schematic diagram of the digital system for tracing and measuring the position of the blood vessel wall: *WE* — choice of the echo slope, *SE* — tracing the chosen echo slope, *TR* — trigger, *Z* — clock

appears at the output of the counter DCB. At the moment that the rising slope of the gate G appears, the sum of the clock impulses counted by the counter DCA is smaller by N_0 than the number of clock impulses counted by this counter in a preceding cycle. This means that, in accordance with the measurement method adopted (formula 2), the tracing gate G is displaced with respect to the position which the traced echo slope took in a preceding cycle.

The gate G is supplied at the input of the echo identification unit EI, to which echos detected from the blood vessel walls are supplied through the level comparator C. The unit EI generates the impulse IP (see Fig. 2), when the first echo slope after the gate G occurs. This impulse eliminates by the logical keys $K1$ and $K2$ the clock impulse from the inputs of the counters DCA and DCB. After rewriting the contents of the counter DCA into the computer K and the counter DCB, the counter DCA is reseated. A new measurement cycle begins with another trigger impulse IZ. The digital data obtained at the output of the counter DCA are transformed in the computer into information about the instantaneous position of the surface of the blood vessel wall with respect to the ultrasonic probe.

The tracing of the position of the chosen echo slope requires that the tracing gate G should be set up before it. It is only then that the above-described process of automatic tracing of the chosen echo slope can take place.

The initial position of the tracing gate G with respect to the echo slope is set digitally by introducing into the counter DCA of such a number of impulses which corresponds to the position of the chosen echo slope with respect to the trigger

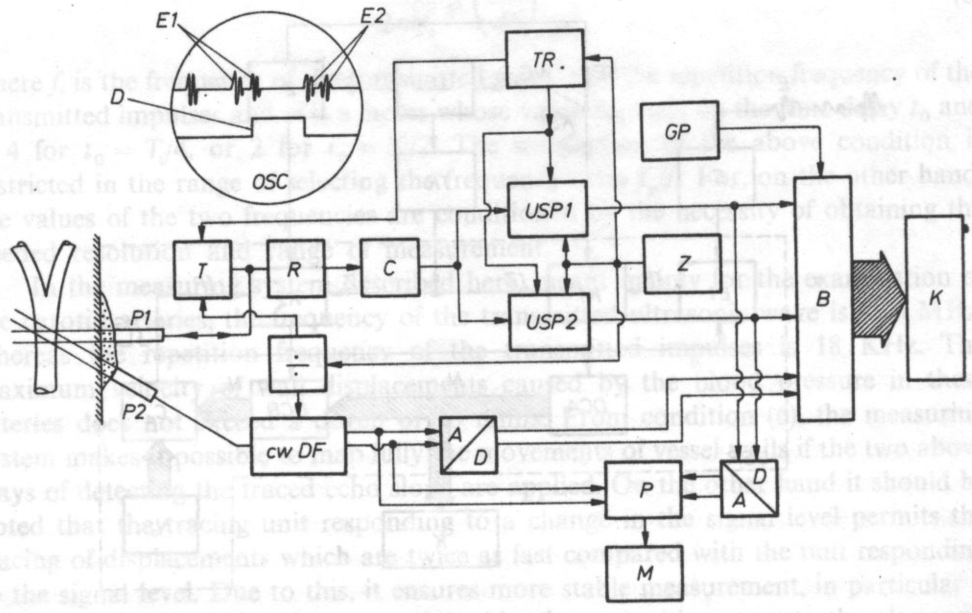

FIG. 4. A schematic diagram of the system for simultaneous measurement of blood velocity, vessel diameter and displacement of blood vessel walls

impulse. The current position of the gate is controlled by its presentation along with echos on the measuring system's screen. To set precisely the position of the tracing gate G, into the counter DCA, through the key $K3$, clock impulses are replaced by slowly variable ones. The time their introduction is manually controlled, with the echo identification unit EU switched off. The rewriting of the current numerical value from the counter DCA to the counter DCB and the generation of the tracing gate G by the comparator C_d take place in the same way as in the course of the automatic tracing of the echo, i.e., synchronically to the trigger impulses.

To measure simultaneously the displacements of the two walls of the blood vessel and the vessel diameter, two symmetrical channels for tracing and measuring the positions of the echo, called later USP, were applied. Each of them contains two 12-bit counters DCA and DCB, the digital comparator C_d and the echo identification unit EI. The inputs of these two channels are connected in parallel with the level comparator C. The measured data from the outputs of the two channels are assigned to the buffers from which they are then entered into computer for further processing and computations (see Fig. 4).

In the system in question, the delay time between the trigger impulse and the echo detected from the vessel wall is measured by counting clock impulses of 27 MHz. In effect, this permits the representation of the amplitude of displacements of the blood vessel walls an accuracy up to 0.03 mm.

3. Meter of blood velocity wall displacements and blood vessel diameter

The previously described system for the measurement of displacement of blood vessel walls was connected with an ultrasonic c.w. Doppler flowmeter into one joint set-up for simultaneous measurement of the blood velocity, the diameter of the blood vessel and its changes over the work cycle the heart. The device was designed to be applied in examinations of the human peripheral arteries, in particular the carotid arteries.

Figure 4 shows a schematic diagram of the measuring device. It contains two ultrasonic probes $P1$ and $P2$ set at 30° to each other. Thr probe $P2$ is connected with a bi-directional c.w. Doppler flowmeter cwDF. The probe $P1$, set perpendicularly to the blood vessel, is linked with the impulse transmitter T and a broad-band receiver R. To obtain good measurement resolution across the ultrasonic beam, the probe $P1$ is focused. In the focus $-$ 20 dB beam width is 1 mm. In tissue the focusing zone of the ultrasonic wave varies between 1 and 3 cm. To avoid mutual interference between two transmit-receive systems working together the probes $P1$ and $P2$ transmit ultrasonic waves with two different frequencies. The frequency of the wave transmitted by the probe $P1$ is 6.75. MHz, and the frequency of the continuous wave transmitted by the probe $P2$ is 4.5 MHz. The two frequencies were obtained by dividing the clock frequency of 27 MHz, respectively, by 4 and 6.

The repetition frequency of the transmitted ultrasonic wave impulses is 18 kHz.

This provides the measurement range of 4.2 cm into the patient's body. The echos obtained in the course of the measurement at the output of the receiver R are transformed into rectangular signals by the level comparator C. The echos formed in this way are fed to two digital units for tracing and measuring the position of the echo from the front wall (the channel USP 1) and from the back wall (the channel

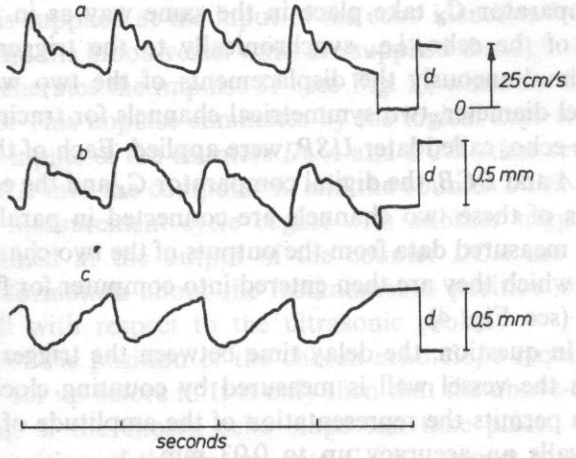

FIG. 5. Time courses of the blood velocity (a) and displacements of the back (b) and front (c) vessel walls recorded in the course of measurement in the common carotid artery in a 40-year-old man

USP 2) of the blood vessel. The 11-bit data obtained at the outputs of the two systems are fed to the output buffers B of the unit, along with the two 8-bit pieces of information about the blood velocity measured simultaneously in two directions. These data are then fed into the computer K, with the sampling frequency FP imposed by the generator GP. This frequency is adjusted depending on the working frequencies of the examined patient's heart. It is so chosen that there are 64 data of each of the measured quantities for the mean period of the work period of the heart. This condition results from the 2^n – point fast Fourier transform applied in the further computer data analysis.

Data on the blood velocity are obtained from the measured difference (Doppler) frequency between the frequency of the transmitted wave and that detected from the flowing blood by the probe P2 of the flowmeter cwDF. The mean Doppler frequency f_{zc} measured by the zero-crossings method is proportional to the mean velocity v_s of the blood flow through the cross-section of the blood vessel, according to the dependence

$$V_s = a\frac{c f_{zc}}{2 f_n \cos\theta}, \tag{7}$$

where c is the ultrasonic wave velocity in the medium under study, f_n is the frequency of the transmitted ultrasonic wave, θ is the angle between the direction of the transmitted and detected ultrasonic wave and the axis of the blood vessel, and a is the proportionality coefficient.

The proportionality coefficient in formula (7) depends on the velocity profile of the blood flow studied and on the ratio between the ultrasonic beam width and the blood vessel diameter [6, 17]. Assuming that the mean profile of the blood flow rate in the cross-section of the blood vessel is contained between a parabole and a flat profile, and that the ultrasonic beam is wider than half the vessel diameter, in the first approximation the coefficient a would be 0.85 [6, 17].

In determining the blood velocity from the Doppler frequency f_{zc} measured by the flowmeter, it was assumed that the angle θ in dependence (7) is 60°. This follows from the constant angle 30° between the probes $P1$ and $P2$, and from the assumption that the probe $P1$ is set perpendicularly to the axis of the vessel. It was assumed that the perpendicular setting of the probe $P1$ with respect to the blood vessel would be indicated by obtaining the maximum amplitude of the echoes from the two walls of the vessel. In the course of the measurement these echoes are obtained on the oscilloscope screen *OSC* of the device. At the same time, the screen shows the measured diameter of the vessel under study in the form of a gate.

The data obtained in the measurement of the blood velocity, the diameter of the blood vessel and displacements of its two walls are presented in the form of analog courses on the recorder P or the memory monitor M. The analog recording serves for controlling the measurement data obtained before they are fed into a computer for further analysis. The analog-to-digital (A/D) and digital-to-analog (D/A) converters were applied to transform the measured quantities into digital and analog values (Fig. 4). Figure 5 shows, as an example, the blood flow rate and displacements of the front and back walls of the artery recorded in the course of measurement in the common carotid artery.

The measurement of the internal diameter of the blood vessel requires good resolution of the transmit-receive impulse system. For this purpose, apart from the previously mentioned focusing of the probe $P1$, in the measurements, a narrow transmitted impulse was used with a duration of 0.3 μs (2 high-frequency cycles of the transmitter), corresponding to its length of 0.45 mm in tissue. As a result it made it possible, in the case of the common carotid arteries, to obtain single echoes from the external and internal surfaces of the blood vessel walls. Fig. 6 shows the measured displacements of the internal and external displacements of the surface of the back wall of the common carotid artery with its thickness. In the case studied it was 1 mm.

The single echos obtained from the two surfaces of the vessel walls made it possible to measure its internal diameter. Having at the same time, information on the blood velocity across the measured internal cross-section of the vessel, it is possible to determine on this basis the blood flow rate. The ultrasonic impulse Doppler method applied so far for this purpose does not permit the so precise measurement of the internal diameter of the blood vessel. This mainly results from the fact that the information on the vessel diameter is taken from the measured spatial distribution of the velocity profile of blood flow [2, 6] which for low velocities close to the vessel wall is falsified as a result of the filtration of the Doppler signal coming from the displacing blood vessel walls. A general view of the measuring system in question is shown in Fig. 7.

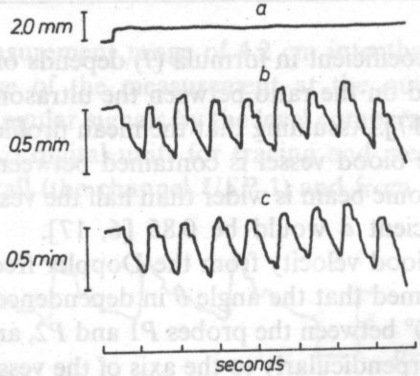

FIG. 6. The thickness of the carotid artery wall (a) and displacements of the external (b) and internal (c) surfaces of the wall studied, recorded in the course of the measurement. The mean internal diameter of the vessel examined was about 8 mm

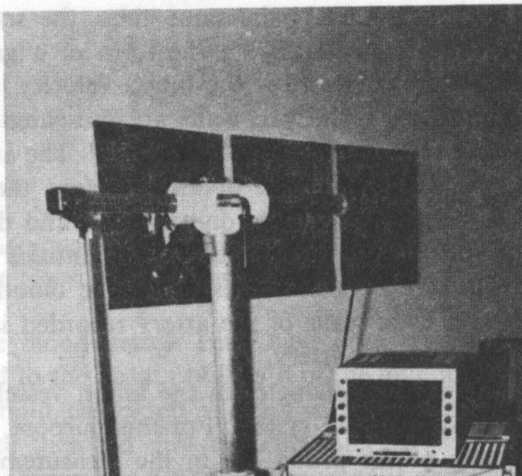

FIG. 7. A general view of the ultrasonic system

4. Computer analysis of hemodynamic parameters

The ultrasonic measuring system described above was used in examination of the carotid arteries. The data obtained at the output of the ultrasonic device are fed into a MERA-60 computer (corresponding to PDP-11) and written on a floppy disk. The number of data recorded each time in the course of the measurement contain information on 640 successive values of each of the four parameters measured simultaneously. They are: the digitally measured times t_{n1} and t_{n2} between the transmitted impulses and the echoes from the front and back walls of the blood vessel and the Doppler frequencies f_{2c1} and f_{2c2} measured for blood flow away from and towards the ultrasonic probe.

According to earlier determinations, the sampling frequency with which data are fed into a computer is so set that 64 data on each of the measured quantities correspond to one mean work cycle of the heart. This means that the number of recorded data comes from about 10 work cycles of the heart.

Along with the above-mentioned data, the value of the sampling frequency F and those of the systole pressure P_s and the diastole pressure P_d measured with a cuff in the brachial artery, with the patient in supine position, are recorded on a computer disk. The data set formed in this way is the basis for further computer analysis of the following hemodynamic parameters:

a) the blood flow rate Q,

b) the blood pressure P,

c) the input vessel impedance Z,

d) the relative change in the blood vessel diameter over the work cycle of the heart $\Delta D/D$,

e) the pulse wave velocity c_v,

f) the coefficient of rigidity of the blood vessel wall α.

The flow diagram of the computer analysis is shown in Fig. 8. From n successive measured data, respectively from dependencies (1) and (7) the instantaneous numerical values of the blood flow velocity $v_1(k)$ and $v_2(k)$ away from and towards the ultrasonic probe, and the distances $d_1(k)$ and $d_2(k)$ between the ultrasonic probe and the internal surfaces of the front and back walls of the blood vessel are determined. In turn, these data serve to determine k successive values of the internal vessel diameter $D(k)$ and the blood flow rate $Q(k)$.

Further computer analysis is performed separately for particular work cycles of the heart. The indicator which identifies a successive cycle is the maximum value of the rate Q occurring in the systole phase. The beginning of the studied cycle K_0 is established at the beginning of the systole phase. For each cycle, the beginning of the analysis is equally shifted with respect to the maximum value of the blood flow rate Q.

The programme of computer analysis assumed the possibility of determine the blood flow rate $Q(k\Delta t)$ and the blood vessel diameter $D(k\Delta t)$ on the basis of a chosen number IL of successive work cycles of the heart. Programmatically, this is implemented according to the following algorithms:

$$Q(k\Delta t) = \frac{1}{IL}\sum_{m=1}^{IL} \frac{\pi v_m(k\Delta t)\, D_m^2(k\Delta t)}{4}, \tag{8}$$

$$D(k\Delta t) = \frac{1}{IL}\sum_{m=1}^{IL} D_m(k\Delta t), \tag{9}$$

where $k = 1, 2, \ldots, 64$ is the number of a successive datum in the work cycle of the heart, m is the number of a successive cycle, $\Delta t = 1/FP$ and $v_m(k\Delta t)$ is the studied linear blood velocity in the mth cycle of the heart.

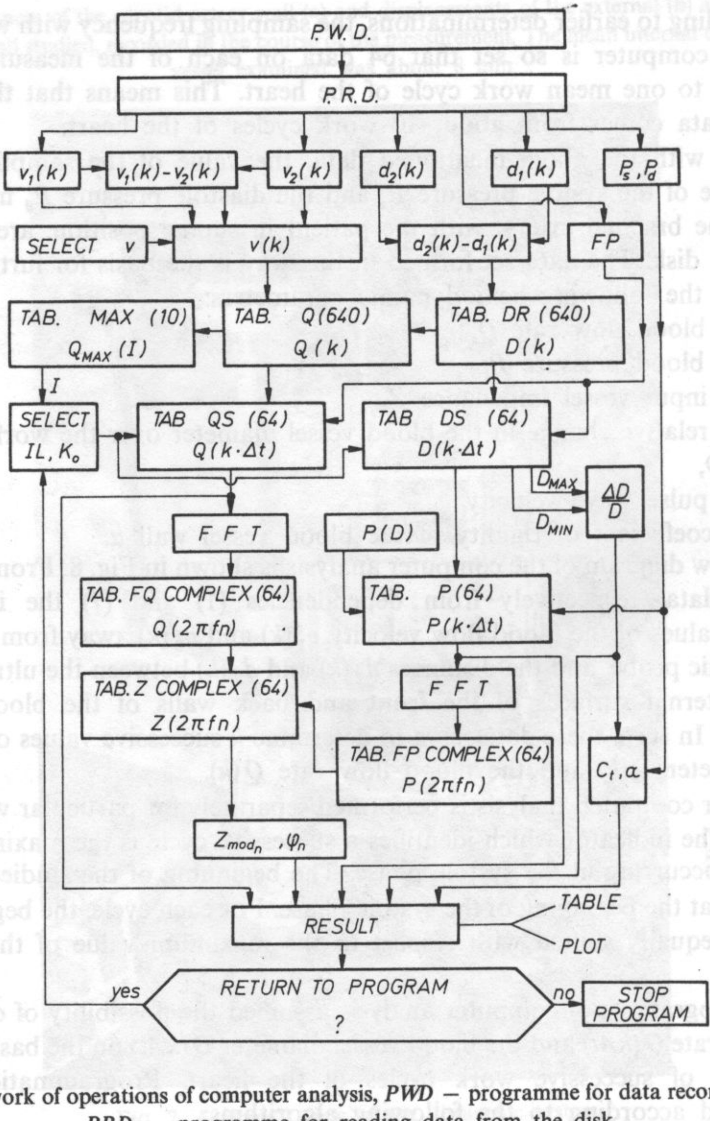

FIG. 8. The network of operations of computer analysis, *PWD* — programme for data recording on a disk, *PRD* — programme for reading data from the disk

According to the adopted network of operations the quantities $Q(k\Delta t)$ and $D(k\Delta t)$, described by formulae (8) and (9) form the basis for determining the mean values of the other hemodynamic parameters.

The blood pressure is another parameter determined from measurement data. It is calculated from instantaneous changes in the blood vessel diameter over a work cycle of the heart. The calculations assumed an exponential dependence between the blood pressure P and the blood vessel cross-section area S, given by the following function:

$$P = P_0 \exp(\gamma S), \tag{10}$$

where P_0 and γ are constant coefficients.

After transformation, this function becomes

$$P(D) = P_d \exp\left[\frac{D^2 - D_d^2}{D_s^2 - D_d^2} \ln \frac{P_s}{P_d}\right], \tag{11}$$

where D_s and D_d are the vessel diameters for the systole pressure P_s and the diastole pressure P_d. The pressure determined in this way is calibrated in absolute units by the systole pressure P_s and the diastole pressure P_d measured in the brachial artery. The values of the two pressures are respectively subordinated to the maximum $D_{\max}$ and minimum $D_{\min}$ diameters of the blood vessel. The blood pressure over the work cycle of the heart, calculated on this basis, is expressed by the following dependence

$$P(n\Delta t) = P_d \exp\left[\frac{D^2(n\Delta t) - D_{\min}^2}{D_{\max}^2 - D_{\min}^2} \ln \frac{P_s}{P_d}\right]. \tag{12}$$

The instantaneous blood pressure $P(k\Delta t)$ and the blood flow rate $Q(k\Delta t)$ are the basis for determining the impit vessel impedance Z. Assuming linearity of the vascular system studied, the impedance Z is calculated as the ratio between the Fourier transforms of the above-mentioned discrete time courses over the work cycle of the heart:

$$Z(2\pi fn) = \frac{P(2\pi fn)}{Q(2\pi fn)} = Z_{\mathrm{mod}_n} e^{j\varphi_n}, \tag{13}$$

where $n = 0, 1, 2, \ldots, k$, f is the frequency of the heart rate, φ_n is the phase and Z_{mod_n} is the modulus. In the calculating, the algorithm of the fast Fourier transform FFT [3] was used. In its final form, the input vessel impedance is represented by the modulus Z_{mod_n} and the phase φ_n for successive harmonic n frequencies of the heart rate. For $n = 0$, the impedance represents the mean resistance of the vascular system in question.

Another group of the investigated parameters are related to the elasticity of the walls of arterial blood vessels. One of them is the relative change in the vessel diameter $\Delta D/D$ over in the work cycle of the heart. It is defined in the following way:

$$\frac{\Delta D}{D} = \frac{D_{\max} - D_{\min}}{D_{\min}} \cdot 100\%, \tag{14}$$

where D_{max} and D_{min} are the maximum and minimum blood vessel diameters over the vessel cycle of the heart.

The parameter described by formula (14) depends on the elasticity of the vessel wall, but it cannot be its measure, since it does not into account the increase in the blood pressure causing changes in the vessel diameter. In the analysis, it was introduced solely for cognitive purposes.

A more objective index of the elasticity of arterial vessel walls is the pulse wave velocity c_t determined from the volume elastic modulus K [1]

$$c_t = \sqrt{\frac{K}{\varrho}} = \sqrt{\frac{1}{\varrho}\frac{(P_s - P_d)}{S_s - S_d} S_d}, \tag{15}$$

where ϱ is the blood density, S_s and S_d are the cross-section areas of the blood vessel for the systole pressure P_s and the diastole pressure P_d.

The pulse wave velocity c_t is an index very generally used in the literature for evaluation of the elasticity of the human arterial-vascular system [8, 12]. This velocity depends on the rigidity of blood vessel walls and increases with a person's age. c_t is measured by the method of two sensors set usually at two mutually distant points of the vascular system. This permits only an overall evaluation of the elasticity of the vascular system.

According to some authors [12, 16], the rigidity of the walls of the vascular system, and, thus, the pulse wave velocity, too, are affected by the blood pressure in the vascular system. This fact can be explained by the existence of a nonlinear function between the vessel wall displacements and the blood pressure which causes them. So far in the literature, there has been no agreement about the degree and character of this nonlinearity [13, 14]. In this situation in the evaluation of the vessel wall elasticity an additional index α was introduced. It results from the previously adopted exponential dependence between the blood pressure and the transverse dimensions of the blood vessel (formula (10)). The coefficient α determined from dependence (10) has the following form:

$$\alpha = \frac{S_d}{S_s - S_d} \ln (P_s/P_d), \tag{16}$$

where S_s and S_d are the vessel cross-section areas for systolic P_s and diastolic P_d pressures.

The hemodynamic parameters mentioned so far were preliminarily investigated in the common carotid arteries for a group of 43 healthy persons aged between 9 and 64 years. The age of the persons examined was divided into five groups. The results of the measurements are given in Table 1. In addition, a statistical analysis of the parameters studied, performed without a division into age groups, permits the following conclusion to be drawn:

1) The mean blood flow rate Q_{med} over the cardiac cycle in the common carotid artery in adults aged between 19 and 64 varied between 400 and 620 ml/min. Its mean value was 499 ± 75 ml/min.

2) The relative change in the diameter of the common carotid artery $\Delta D/D$ over the cardiac cycle was greatest in children and decreased with the age of the patients. Investigation of the change $\Delta D/D$ as a function of age by means of linear regression gave the following dependence:

$$\Delta D/D = 15.507 - 0.189\,x \quad [\%], \tag{17}$$

Table 1. The hemodynamic parameters determined noninvasively in the common carotid arteries of healthy persons in five age groups the mean blood flow rate Q_{med}, the mean blood pressure P_{med} from formula (12), the relative change in the artery diameter $\Delta D/D$, the pulse wave velocity c_t formula (15) and the coefficient of rigidity α formula (16). The systole pressure P_s and the diastole pressure P_d were determined with a manometer in the brachial artery

Age group (years)	9–16	19–30	32–40	41–50	52–64
Number of persons	7	9	8	8	11
Mean age	12.3	24.4	36.8	44.9	56.2
S.D. (years)	2.3	4.6	3.2	2.9	3.9
P_s	104.3	117.8	116.9	110.0	122.7
S.D. (mmHg)	10.6	7.9	7.0	10.0	13.8
P_d	60.7	77.2	80.0	70.6	80.0
S.D. (mmHg)	6.1	7.1	5.9	8.6	8.7
P_{med}	82.2	97.3	97.2	90.5	101.5
S.D. (mmHg)	6.4	6.1	5.8	9.2	10.5
Q_{med}	364.3	497.8	511.3	540.0	459.0
S.D. (ml/min)	53.2	87.9	74.9	59.8	61.9
$\Delta D/D$	13.89	11.0	7.54	6.85	5.24
S.D. %	1.83	2.02	1.79	1.05	1.14
c_t	4.52	4.97	5.84	6.26	7.44
S.D. (m/s)	0.61	0.38	0.71	0.49	0.74
α	1.84	1.87	2.55	3.18	3.98
S.D.	0.43	0.37	0.68	0.51	0.62

S.D. – Standard deviation

where x is the age of the persons examined in years. The coefficient of correlation for this linear dependence was 0.877.

3) The pulse wave velocity c_t determined from formula (15) increased linearly with the age of the patients according to the regression line:

$$c_t = 3.457 + 0.0677\,x. \tag{18}$$

The coefficient of correlation for this dependence was 0.877

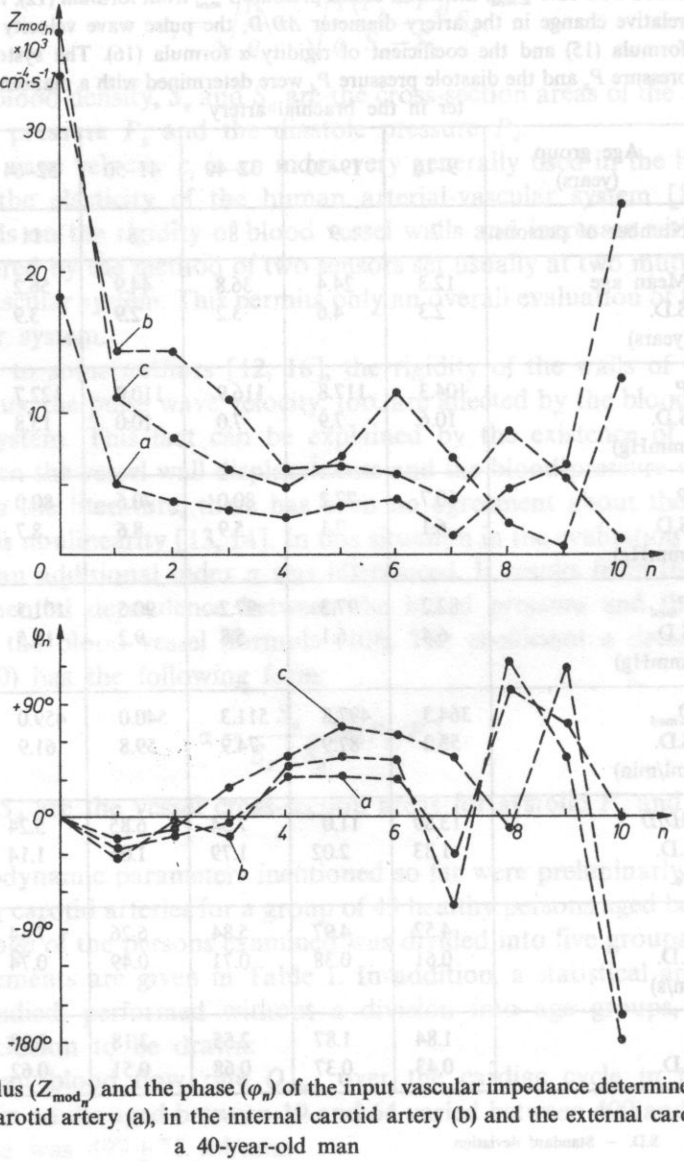

FIG. 9. The modulus (Z_{mod_n}) and the phase (φ_n) of the input vascular impedance determined noninvasively in the common carotid artery (a), in the internal carotid artery (b) and the external carotid artery (c) in a 40-year-old man

4) The coefficient of rigidity α increased linearly as a function of the age of the persons examined. It is described by the regression line:

$$\alpha = 0.8581 + 0.0523\,x. \tag{19}$$

The coefficient of correlation for this function was 0.832.

Using the measuring system described here, preliminary investigations of the input vessel impedance were also performed in the extracranial carotid arteries (Fig. 9). They indicate that for the first few harmonics the modulus and phase of the impedance strongly depend on the inertia, compliance and resistance of the vascular system studied. The first step towards the determination of these values was a computer simulation of the input vessel impedance in the common carotid artery by means of the impedance of a substitute circuit containing elements representing inertia compliance, vascular resistance and peripheral resistance [19]. Further research is under way on the interpretation of the vascular impedance [21].

The preliminary research results presented above indicate the large usefulness of the measuring system for noninvasive evaluation of the human arterial-vascular system, in particular for diagnosis of the carotid arteries.

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