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DYNAMIC ULTRASONIC VISUALIZATION OF BLOOD VESSELS AND FLOWS

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This paper describes a new method for dynamic real-time visualization of blood flows. This method uses a special signal processing system (SEC) which consists in cancellation of stationary echoes from the reflections of ultrasonic waves from soft tissues with continuous measurement of the phase of signals scattered in blood.

The stationary echo cancellation system has been designed on the basis of the properties of periodic filters using quartz delay lines. A level of stationary echo cancellation above 55 dB was achieved, which, when using an original detector of the phase of signals scattered in blood, permits real-time observation of blood flow velocity profiles. This, in turn, permits the internal diameter of a vessel (the degree of constriction) to be evaluated in the site under investigation.

1. Introduction

Ultrasonic methods for the visualization of the internal anatomical structures of man use the particular properties of ultrasonic wave propagation, permitting the detection of small differences in the density and elasticity of soft tissues. The density and elasticity of tissues determine the acoustic impedance of the biological medium investigated. Knowledge of the acoustic impedance of media permits the determination of the part of the energy of ultrasonic waves incident on the boundary between the two media which is partly reflected and returns to the receiving transducer and also of the part which penetrates into an adjacent medium. The higher the difference between the acoustic impedances of the two media is the greater is the energy on the boundary between them. The magnitude of the reflected energy also depends on the geometry of the ultrasonic beam and on the geometry of the system. For simplification, the case of the perpendicular incidence of ultrasonic waves on biological structures of different acoustic impedance has been assumed in the present discussion.

The development of the optimum system for the visualization of blood vessels requires the estimation of differences in the level of signals reflected from stationary structures and of signals scattered in blood. An exact solution of this problem would require some difficult in vivo measurements of the energy of signals reflected from all the biological structures on the path between the ultrasonic transducer and the blood vessel. Even in the latter case, however, the results obtained describe only the anatomical system related to a specific structure, thickness and distribution in the of tissues investigated ultrasonically.

In the case when the wave is reflected from a flat reflector with the energy coefficient of reflection R the intensity of the wave received by the transducer is [10]

$$I_0 = \frac{P_t A}{4x^2\lambda^2} R\exp(-4ax), \tag{1}$$

where P_t is the sound power of the signal transmitted, λ is the wavelength, a is the attenuation coefficient of the wave in the medium, A is the effective area of the transducer which is equal to its geometrical area when it works in a piston mode and x is the distance from the transducer.

In the case when the ultrasonic wave is incident on a single particle (target) which is small compared to the wavelength and the width of the ultrasonic beam, the scattered wave is isotropic and propagates in the form of a spherical wave.

In the latter case, the power of the scattered wave is calculated from the product of the effective scattering surface δ and the intensity of the incident wave. The intensity of the reflected wave is thus

$$I = \left(\frac{P_t A}{x^2\lambda^2}\right)\left(\frac{\delta}{4\pi x^2}\right) \exp(-4ax). \tag{2}$$

The first term represents the intensity of the wave incident on a particle with the active "scattering cross-section" δ which is at a distance x from the transducer, the other describes the propagation of a spherical wave.

In order to calculate the intensity of the isotropic wave scattered by many molecules, it is necessary to sum up the successive intensities from particular particles described by formula (2). The total active surface area of the system of particles is represented by the product of the blood volume and the effective scattering coefficient η. According to SHUNG *et al.* the coefficient η is a function of hematocrite [12]. The total intensity of the scattered signal can thus be expressed in the form

$$I_r = \frac{P_t A^2}{x^4 \lambda^2 4\pi} \frac{c\tau\eta}{2} \exp(-4\alpha x), \tag{3}$$

where $Ac\tau/2$ is the blood volume contained in the ultrasonic pulse field in the form of a cylinder with the base area A and height of $c\tau/2$; and τ is the duration of the pulse. The ratio of the intensities of the wave scattered from blood and of that reflected on the boundary between fat tissue and the wall of the vessel can be determined by way of dividing expressions (3) and (1) by each other and subsequently substituting into the quotient the respective numerical values,

$$\frac{I_r}{I_0} = \frac{A}{\pi R x^2} \frac{c\tau}{2} \eta. \tag{4}$$

Let the duration τ of pulses be 1 μs and the diameter of the transducer $2a$ be 5 mm. For 30 % hematocrite the scattering coefficient $\eta = 11 \cdot 10^{-5}$ cm^{-1} [12]. The acoustic impedances of fat tissue and the wall of the vessel are equal, respectively, to $1.38 \cdot 10^6$ kg/m^{2}s and $1.66 \cdot 10^6$ kg/m^{2}s. The energy coefficient of reflection, R, on the boundary of the tissues mentioned above is equal to $85 \cdot 10^{-4}$. For these numerical values the ratio I_r/I_0 is equal to $6 \cdot 10^{-5}$.

In practice this signifies that successive echoes whose amplitudes differ by almost theree orders of magnitude occur at the input of the receiver system. This prevents simultaneous observation of reflected and scattered echoes. It is necessary to bear in mind that large echoes reflected from structures lying at depth corresponding to a double or even triple repetition period of pulses can also occur. This phenomenon is described by the ambiguity function $\chi(t, v)$ for $v = 0$ (stationary target) [3, 6].

2. A survey of ultrasonic methods for imaging blood vessels

The easiest method for the investigation of geometrical dimensions of a blood vessel on the oscilloscope screen is the A-mode.

In the case of such structures as fat tissue, muscles, etc. echoes on the oscilloscope screen are stationary, while echoes from moving structures such as blood vessels or the heart change their position on the time base of the C.R.T.

In the case of arteries this motion is periodic, following the beat of the heart in that the walls of the vessel go away from each other for systole and approach for diastole. This effect can be observed on the oscilloscope screen where the echoes from the walls fluctuate, moving away and approaching each other. The A-mode has not been widely used in the visualization of blood vessels; it nevertheless has been the basis for a number of interesting works, particularly that of BUSHMAN [2] concerning changes in the diameter of a carotid artery in healthy persons and those with advanced sclerosis.

In the TM (time-motion) mode the time base on the oscilloscope screen is blank when no echoes occur and brightened up only by the echoes from structures in the ultrasonic field.

Echoes are thus projected along the time base in the form of bright points, which are mobile when the ultrasonic waves is reflected from a moving structure.

The TM technique has been applied mainly in the investigations of the moving structures of the heart; it has increasingly been used recently for the visualization of the abdominal aorta, particularly of its aneurisms.

Whereas the A- and TM-modes only show a geometrical, one-dimensional image of the distribution of biological structures along the ultrasonic beam, the so-called B-scanning (for brightness) permits two-dimensional visualization of the structures examined. The B-mode uses partly the basis electronic devices applied in the A-scanner apparatus. The essential modification is in the system of mounting the probe on a special gantry with two mobile arms. Carrying the probe along the structure investigated in translatory and rotary motion, the arms of the gantry change their position and accordingly the angles θ, $ø$ and α change between the arms of the gantry and the probe. Simple trigonometric relations permit the determination on this basis of the coordinates x, y of the position of the transmitted ultrasonic beam and of the one received by the ultrasonic transducer. This transformation, which changes the motion of the probe into the motion of the time base on the oscilloscope display, is implemented using a special electronic system, e.g. of the Metrop type [5]. When the ultrasonic wave encounters reflecting structures on its path, the resultant echoes are displayed on the time base of the oscilloscope tube in the form of bright points, while the brightness of these points depends on the strength of the echo. The B-mode has most widely been used in the investigations of the abdominal cavity in obstetrics, gynaecology and in the diagnostics of the eye and eye orbit.

Nevertheless, in the few recent years the technological developments which increased resolution and dynamics of apparatus permitted the application of B-scanning for the visualization of large blood vessels, particularly of aneurisms of the abdominal aorta.

In many cases B-scanning has greater diagnostic value than that of X-ray arteriography, in view of the harmlessness and repeatability of examinations essential in the monitoring of the distention of an aneurism. In arteriography examinations must not be too frequent, since they put an extra burden on the patient. Punctures of an artery, catherisation and the introduction of the bulk of contrast often damage the vessel and threaten a general shock. An additional advantage of B-scanning is the possibility of projecting the vessel in two planes, while arteriography only gives an image in one plane, which makes it difficult to interpret angiograms of aneurisms in which thrombi occur.

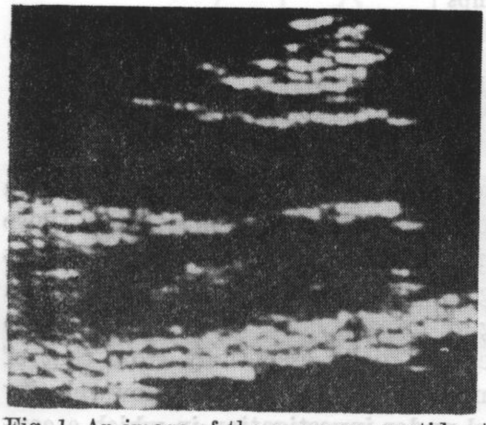
The other group of methods for the visualization of blood vessels includes different Doppler methods. Because of its simple apparatus and its easy clinical application, the so-called continuous-wave method (C.W. Doppler) has been widely used in the evaluation of the state of patency of vessels, in the localization of thrombi and fistulas in arteries and veins and also, more recently, in the forecasting of the level of amputation on the basis of exact measurements of the systole pressure in peripheral vessels. Among the numerous mutations of the C. W. method, the technique for the visualization of blood vessels elaborated in 1972 by Reid and Spencer [11] seems to be particularly interesting. In view of its similarity to the imaging of vessels in X-ray angiography, its authors have called it Doppler angiography.

Doppler angiography is widely used, particularly in the USA, where serial production of relevant apparatus has recently been commenced (Carolina Medical Electronics, Inc.). Its buyers are clinical units engaged in diagnostics of arteriosclerosis of carotid arteries, particularly with an impending stroke. Until recently the only method for monitoring the patency of blood vessels has been that of X-ray angiography, which involves the high risk of complications.

A preliminary ultrasonic examination, though it does not always defines unambiguously the degree of inductability of the internal and external carotid arteries, limits the use of X-ray angiography only to dubious cases [14].

The transition from qualitative investigations to quantitative ones has been made possible by the development of a pulse Doppler method which permits the measurement of blood flow rate selectively at varying depth [9]. As a result, the degree of constriction of the vessel investigated can be evaluated with high accuracy (usually better than 1 mm) [6, 7].

a)



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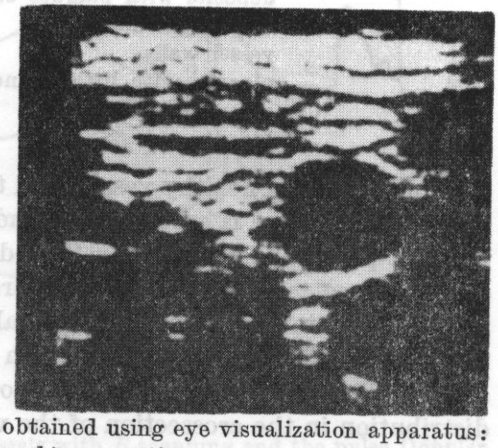

Fig. 1. An image of the common carotid artery obtained using eye visualization apparatus: a) longitudinal section, b) crossection

Several types of pulse Doppler apparatus are now produced in the world. Particular attention is due the British devices (Mavis-GEC Medical) which are equipped with 30 parallel channels permitting simultaneous registration of blood flow at different depth and French (Echovar Doppler Pulsée — Alvar Electronic) and Polish (UDP-30 — Z.D. Techpan) devices for automatic registration of blood flow profiles averaged in time [7].

Concluding this short survey of methods for the visualization of blood vessels, it is interesting to note the complementary technique which combines B-scanning with the pulse Doppler method. In Poland the first attempts to use this measurement technique were made in 1974 using a B-scanner for the visualization of the internal structures of the eye [4] and a prototype ultrasonic pulse Doppler flowmeter, UDP-30, developed by the author.

Fig. 1 shows an image of the carotid artery of a healthy, young man. The inner diameter of the vessel is displayed in the form of a blackened channel (longitudinal section) or a blackened hole (cross-section), when sharp, bright points arise on the boundary of a wall of the vessel. This results from the fact that the energy of ultrasonic waves reflected from walls of the vessel is greater by two orders of magnitude than the energy of waves scattered in blood (see Table 1).

It follows from Table 1 that the areas with fat deposits narrowing the effective inner diameter of the vessel and the areas filled with blood cause a reflection of ultrasonic waves on a similar level.

Table 1. Relative levels of ultrasonic echoes from sclerotic deposits of different type [1]

Tissue	Relative echo level [dB]
hard calcium deposits	+40
deposits with plaques or lumps of calcium deposit	0-20
vessel walls	0
plaques with little or no calcium deposit	from 0 to −40
blood	−40

In the case of healthy vessels and those narrowed as a result of hard calcium deposits the image is easy to read and interpret. With soft deposits (e.g. those with fat plaques) the level of reflected signals does not differ greatly from that of signals scattered in blood and accordingly the oscilloscope screen displays an image of less intensive brightness (making an impression of a blackened image) in the place of a constriction, which does not differ essentially from that of a vessel filled with blood. A simultaneous measurement of the blood flow velocity distribution in the crossection of the vessel under investigation in which a constriction of this type is suspected to occur can prevent the examiner from an erroneous interpretation of results.

3. The principle of dynamic visualization of blood flows by means of the moving target indicator (M.T.I.) technique

The application range of the methods described above is usually limited by structural features of apparatus, resolution of the system and time necessary for an image of the vessel investigated to be obtained.

In a continuous-wave visualization system the images obtained resemble those of vessels in X-ray angiography. Twodimensional projection permits the evaluation of patency of vessel only in a plane parallel to the surface of the skin over the vessel. The vessel can be projected in a plane perpendicular to the surface of the skin using a multi-channel pulse Doppler apparatus. The large number, however, of analyzing gates, necessary for good resolution and minimalization of the time of investigation of the vessel, complicates the apparatus and raises its price considerably.

The complementary system-of a real-time B-scan and of a single-channel pulse Doppler method — although it provides high resolution and rapid representation, is limited to only the visualization of straight sections of the vessel which lie in the plane of motion of the probe B (see Fig. 2). This limits the eva-

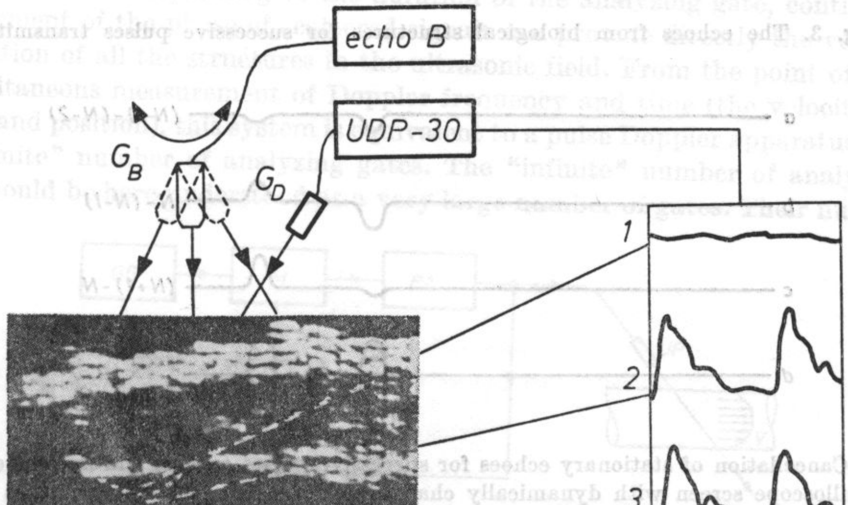

Fig. 2. Complementary visualization of blood vessels with B-scanning and the pulse Doppler method (UDP-30)

a) an image of a vessel part in B-scanning, where dashed line marks areas with soft calcium deposits; b) registration of flow velocity at different parts of a vessel cross-section; curves *1* and *4* show the lack of flow in an area with soft calcium deposits, G_B – ultrasonic probe for B-scanning, G_D – ultrasonic Doppler probe

luation of patency in vessel bifurcations, e.g. in the bifurcation of the common carotid artery into internal and external carotid arteries and in the bifurcation of the femoral artery into the femoral profunda artery.

Compared to the visualization methods discussed the method for the real--time detection of the phase of signals scattered in blood using the radar technique of stationary echo cancellation SEC [8] seems particularly attractive.

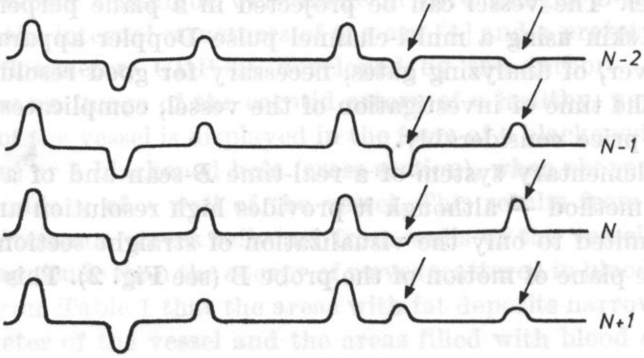

Fig. 3. The echoes from biological structures for successive pulses transmitted

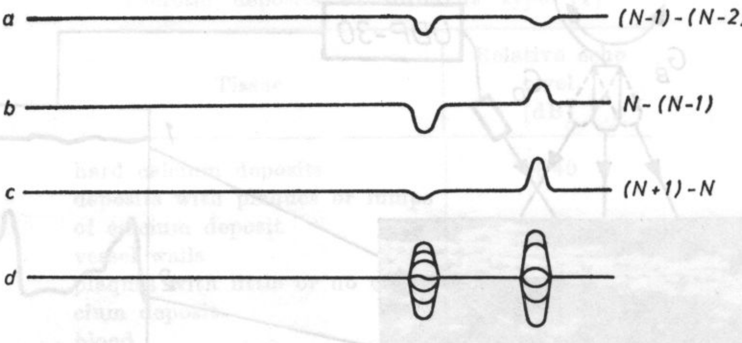

Fig. 4. Cancellation of stationary echoes for successive pulses (*a, b, c*) and an image on the oscilloscope screen with dynamically changing echoes from mobile structures (*d*)

It can be assumed that an ultrasonic signal in the form of a sequence of high-frequency pulses is introduced into the patient's body. For stationary structures both the phase and amplitude of the successive reflected pulses remain the same. For moving structures, even when assuming that the amplitude of the reflected pulses is constant, the phase changes as a result of change in the distance between the reflecting structure and the source of ultrasonic waves in the time from the transmission of the pulse N to the transmission of the pulse $N+1$.

It can be seen in Fig. 3 that most echoes do not change from pulse to pulse. They are thus reflected from stationary structures; some, marked with arrows, differ from one another because of change in phase upon reflection from moving structures. By way of signal processing, which consists in the subtraction of two consecutive echo sequences from each other, where the echo sequence from the earlier pulse is delayed by the time T_p equal to the repetition period of transmitted pulses, it is possible to achieve the effect of cancellation of stationary echoes (Fig. 4).

In a standard pulse Doppler method the ultrasonic transducer transmits a sequence of coherent high-frequency pulses with some specific repetition frequency $F_p (F_p = 1/T_p)$. Inside the body the ultrasonic waves are reflected on the boundary of tissues with different acoustic impedance ϱc and scattered by morphotic elements of blood, with the ratio of the amplitudes of reflected and scattered signals greater possibly than 100.

Thus it can be seen that strong signals reflected from stationary or slowly moving tissues (e.g. the walls of vessels) can mask weak signals scattered in blood which contain Doppler information about its flow rate. Whereas in the pulse method the flow rate of a blood layer at chosen depth is measured in short time intervals corresponding to the duration of the analyzing gate, continuous measurement of the phase of scattered signals can provide directly the velocity distribution of all the structures in the ultrasonic field. From the point of view of simultaneous measurement of Doppler frequency and time (the velocities of motion and position), this system is equivalent to a pulse Doppler apparatus with an "infinite" number of analyzing gates. The "infinite" number of analyzing, gates should be here understood as a very large number of gates. Their number

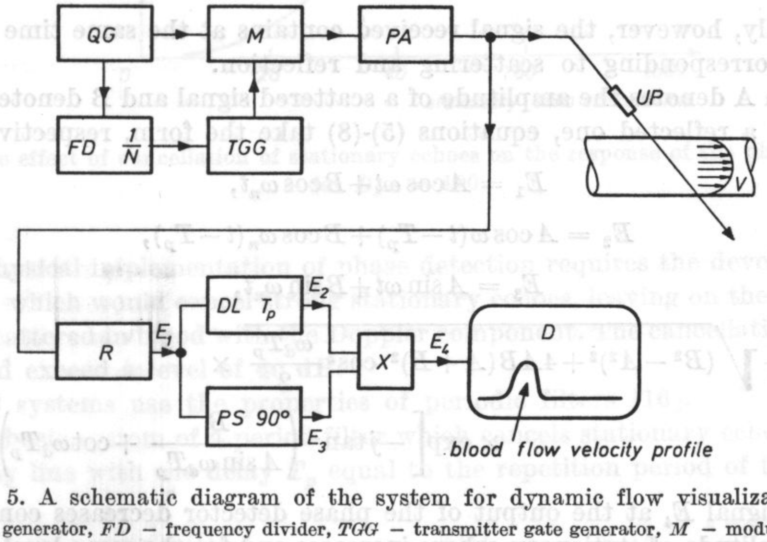

Fig. 5. A schematic diagram of the system for dynamic flow visualization
QG – quartz generator, *FD* – frequency divider, *TGG* – transmitter gate generator, *M* – modulator, *PA* – power amplifier, *UP* – ultrasonic probe, *R* – receiver, *DL* – delay line, *PS* – phase shifter, *X* – doubly balanced mixer, *D* – display

which it is difficult to determine analytically, depends mainly on the repetition period T_p and the frequency f_n of transmitted pulses. The notion of the number of analyzing gates should not be identified with the resolution of the system, which depends on the duration and shape of transmitted pulses.

An approximate number of analyzing gates corresponding to the SEC system can be determined from the expression $n \simeq T_p f_n$. For a repetition period of 64 μs and the frequency of the carrier wave $f_n = 4.37$ MHz a multi-gate system corresponding to the SEC system should contain about 280 analyzing gates. A schematic diagram of the system for the visualization of blood flow by continuous measurement of the phase of signals scattered in body is shown in Fig. 5.

In the case when only a signal scattered in blood occurs without additional stationary echoes the signals E_1, E_2, E_3 and E_4 in Fig. 5 can be represented in the form

$$E_1 = A \cos \omega t, \tag{5}$$

$$E_2 = A \cos \omega (t - T_p), \tag{6}$$

$$E_3 = A \sin \omega t, \tag{7}$$

$$E_4 = -\frac{A^2}{2} \sin \omega T_p = \mp \frac{A^2}{2} \sin \omega_d T_p, \tag{8}$$

where

$$\omega = \omega_n \pm \omega_d, \quad T_p = \frac{2\pi N}{\omega_n}, \quad \omega T_p = 2\pi N \pm \omega_d T_p.$$

Usually, however, the signal received contains at the same time the components corresponding to scattering and reflection.

When A denotes the amplitude of a scattered signal and B denotes the amplitude of a reflected one, equations (5)-(8) take the form, respectively,

$$E_1 = A \cos \omega t + B \cos \omega_n t, \tag{9}$$

$$E_2 = A \cos \omega (t - T_p) + B \cos \omega_n (t - T_p), \tag{10}$$

$$E_3 = A \sin \omega t + B \sin \omega_n t, \tag{11}$$

$$E_4 = \frac{1}{2} \sqrt{(B^2 - A^2)^2 + 4AB(A+B)^2 \cos^2 \frac{\omega_d T_p}{2}} \times \\ \times \exp\left[-j \tan^{-1}\left(\frac{B}{A \sin \omega_d T_p} + \cot \omega_d T_p\right)\right]. \tag{12}$$

The signal E_4 at the output of the phase detector decreases considerably as the amplitude of stationary echoes increases and for the same level of reflected and scattered signals ($A = B$), for example, the signal at the output of the phase detector reaches an amplitude of half the level for $B = 0$. For stationary echoes stronger by a factor of a hundred than scattered echoes it is possible to observe a decrease by a factor of fifty in the signal at the output of the phase detector. An additional decrease in the sensitivity of the phase detector is related to the limitation of the linear characteristic range of the receiver for strong signals. For transmitted pulses with an amplitude of 60 V echoes scattered in blood reach an amplitude of several dozens of μV, while the amplitude of echoes from stationary structures can exceed several dozens of mV.

At the output of a typical receiver, with amplification of 60 dB, the latter echoes can achieve a level of above 10 V, exceeding the linear range.

The effect of the degree of cancellation of stationary echoes on the amplitude of the output signal from the phase detector was calculated from expression (12). The assumption was made that the amplitude of stationary echoes is larger by a factor of a hundred than that of scattered echoes, $B = 100\,A$. The calculated results are plotted in Fig. 6.

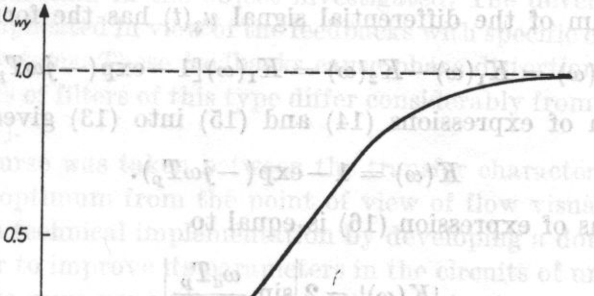

Fig. 6. The effect of cancellation of stationary echoes on the response of the phase detector for $B/A = 100$

A physical implementation of phase detection requires the development of a system which would cancel strong stationary echoes, leaving on the same level signals scattered in blood with the Doppler component. The cancellation efficiency should exceed a level of 40 dB.

SEC systems use the properties of periodic filters [16].

The basic system of a period filter which cancels stationary echoes consists of a delay line with the delay T_p equal to the repetition period of transmitted

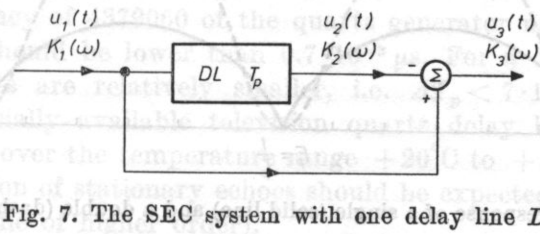

Fig. 7. The SEC system with one delay line *DL*

pulses and a differential amplifier. The frequency response of this system can be determined from its transfer function

$$K(\omega) = \frac{K_3(\omega)}{K_1(\omega)}, \tag{13}$$

where $K_1(\omega)$ and $K_3(\omega)$ are the spectra of the signals at the input and output of the filter.

The spectrum of the signal $u_2(t) = u_1(t-T_p)$ at the output of the delay line has the form, according to the theorem of the shift in the time domain,

$$K_2(\omega) = K_1(\omega)\exp(-j\omega T_p). \tag{14}$$

The spectrum of the differential signal $u_3(t)$ has the form

$$K_3(\omega) = K_1(\omega) - K_2(\omega) = K_1(\omega)[1-\exp(-j\omega T_p)]. \tag{15}$$

Substitution of expressions (14) and (15) into (13) gives

$$K(\omega) = 1-\exp(-j\omega T_p). \tag{16}$$

The modulus of expression (16) is equal to

$$|K(\omega)| = 2\left|\sin\frac{\omega_d T_p}{2}\right|. \tag{17}$$

The respective modulus of the transfer function of a double in series SEC system has the form

$$|K(\omega)| = 4\sin^2\frac{\omega_d T_p}{2}. \tag{18}$$

In both systems the detectability of stationary structures is a function of the Doppler frequency, this relation being sinusoidal for a single SEC system and proportional to a squared sine function for a double SEC system. It can be seen from Fig. 8 that not only echoes of zero Doppler frequency (stationary echoes) and multiple repetition frequency but also those whose frequency is

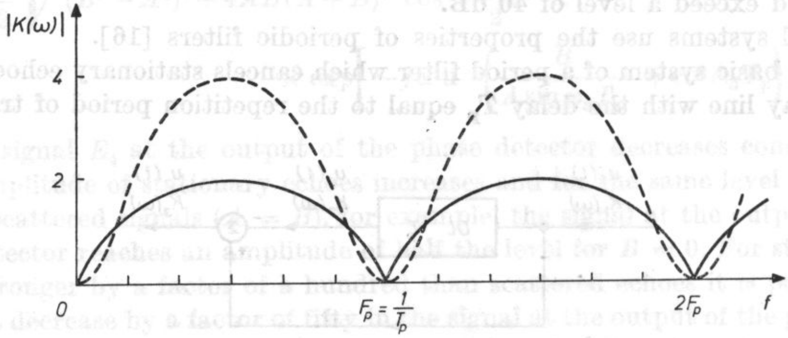

Fig. 8. The frequency response of a single (solid line) and a double (dashed line) SEC system

close to zero are cancelled. These signals usually come from slowly moving vessel walls and from blood flowing at vessel walls.

It was found experimentally that the amplitude of the Doppler signal changes according to the systole and diastole of the heart and increases for higher flow velocities (the systole phase). The probability of low velocities being detected is therefore low for the SEC system described, introducing error in the evaluation of the diameter of a vessel, particularly in the diastole. An ample theory has been developed for feedback periodic filters whose transfer characteristics are close to ideal ones, i.e. they show maximum cancellation of stationary and slowly changing echoes and a flat transfer characteristic in the velocity range occurring in the object investigated. The development of these filters is very complicated in view of the feedbacks with specific coefficients introduced into the systems. These feedbacks cause phase distortions and in general the characteristics of filters of this type differ considerably from those calculated theoretically [13].

A middle course was taken between the transfer characteristic of the SEC system which is optimum from the point of view of flow visualization and the possibilities of its technical implementation by developing a double SEC system in series. In order to improve its parameters in the circuits of undelayed signals, additional circuits were used for compensation of the frequency characteristic of the delay line and the additional delay over the range 0 to 40 ns. Since the cancellation of stationary echoes occurs for *r.f.* signals, good agreement is required between the repetition period of transmitted high-frequency pulses and the delay T_p of the delay line. This imposes severe requirements on the thermal stability of the delay line and of the high-frequency generator. A coherent sequence of high-frequency pulses with the repetition time T_p can be achieved by dividing the frequency of a signal generated by a quartz oscillator; thus the stability of the repetition period of transmitted pulses is equal to the stability of quartz.

Assumption that a 40 dB cancellation of stationary echoes is sufficient for correct performance of the phase detector for a single SEC system gives

$$\left|\sin \frac{\omega_n \Delta T_p}{2}\right| < 0.01. \tag{19}$$

For a small argument $\sin x = x$ and thus

$$\omega_n \Delta T_p < 0.02. \tag{20}$$

For a frequency of 4379060 of the quartz generator used the permissible changes in ΔT_p should be lower than $0.7\cdot 10^{-3}$ μs. For a double SEC system these requirements are relatively smaller, i.e. $\Delta T_p < 7\cdot 10^{-3}$ μs.

The commercially available television quartz delay lines have a delay drift of $5\cdot 10^{-3}$ μs over the temperature range $+20°$C to $+50°$C. In this range a 40 dB cancellation of stationary echoes should be expected only for a double SEC system (or one of higher order).

4. A schematic diagram and the performance principle of the device for the visualization of blood vessels with the SEC system

A prototype pulse Doppler UDP-30 flowmeter was used in developing a model device for blood flow visualization with cancellation of stationary echoes. In its schematic diagram two functionally independent parts can be distinguished. The first part is a modified pulse Doppler flowmeter which permits the

Fig. 9. A general view of an ultrasonic pulse Doppler UDP-30-TES flowmeter

measurement and registration of the instantaneous values of blood flow and of time averaged flow profiles. The second part is made of SEC systems and of phase detection of Doppler signals for the observation of blood flow dynamics in real time. The apparatus contains some common units: a high-frequency generator, a frequency divider, a power transmitter with a modulator, a limiter and a high-frequency preamplifier. A modification in these units with respect to a standard UDP-30 device consisted in replacing the generator LC with a quartz generator whose frequency f_n is equal to the integer multiple of the reciprocal of the delay T_p of the delay lines ($f_n = N/T_p$).

In the present device the frequency of the quartz generator $f_n = 4379060$ MHz, while in the frequency divider the division by $N = 280$ was used. The design of the high-frequency receiver was changed by introducing additionally into the existing time gain control (TGC) system a unit for gating off the preamplifier over a time of 3 μs following the transmission of the pulse. This cancels in the receiver the direct transmitted pulse and the pulses reflected in the focussing lens of the ultrasonic probe and from the surface of the skin.

The amplitudes of these signals usually exceed the linear characteristic range of the receiver. At the output of the delay line these signals undergo considerable distortion both in amplitude and in phase and cause an increase in the level of third echoes in the delay line, thus decreasing the efficiency of the SEC system.

A double in-series stationary echo cancellation system was built using delay lines designed for colour TV sets. In the selection of lines attention was paid to their temperature stability, minimum cancellation of direct echoes (delay T_p) with the highest possible cancellation of third echoes (delay T_p) and of the other spurious echoes in the line [8].

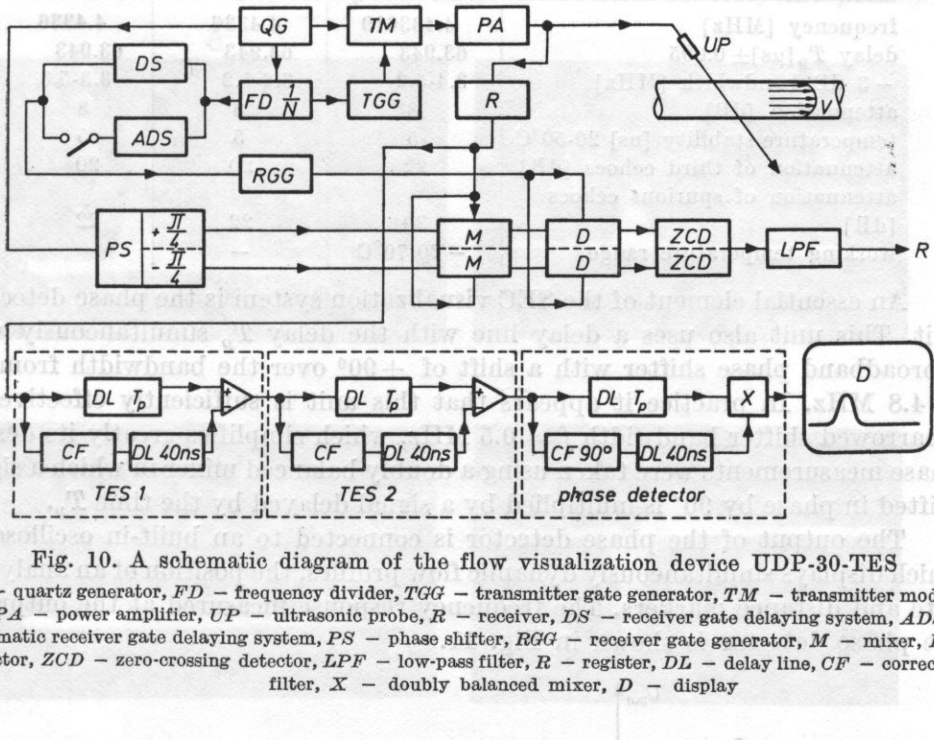

Fig. 10. A schematic diagram of the flow visualization device UDP-30-TES
QG – quartz generator, FD – frequency divider, TGG – transmitter gate generator, TM – transmitter modulator, PA – power amplifier, UP – ultrasonic probe, R – receiver, DS – receiver gate delaying system, ADS – automatic receiver gate delaying system, PS – phase shifter, RGG – receiver gate generator, M – mixer, D – detector, ZCD – zero-crossing detector, LPF – low-pass filter, R – register, DL – delay line, CF – correction filter, X – doubly balanced mixer, D – display

Among the delay lines now available on world market, similar parameters are characteristic of the lines manufactured by Telefunken, Andersen Laboratories Inc., Matsushita and Kinsekisha.

The delay lines from Anderson Laboratories Inc. used in the system are different from the others in their greater cancellation of third and spurious echoes, which influenced their choice.

As was mentioned above, the efficiency of the performance of the SEC system depends on the temperature stability of the delay T_p. In the delay lines it is better than 5 ns over the temperature range $+20°C$ to $+50°C$; differences in delay of up to 20 ns occurred, however, among particular lines. In view of this, both in the circuit of direct signals in both cancellers and in the phase detector, additional units for precise correction of the total delay were used over the range 0 to 40 ns. These systems used small-size delay lines with continuous delay adjustment. These lines compensate in addition for the differences in delay introduced by correction filters compensating the frequency response of quartz delay lines.

Table 2. Typical values of delay line parameters

Parameter	Andersen Lab. Inc. PDL 641E	Matsushita SFD-EN 645	Kinsekisha
frequency [MHz]	4.433619	4.4336	4.4336
delay T_p [μs] $\pm$ 0.005	63.943	63.943	63.943
-3 dB bandwidth [MHz]	3.4-5.2	3.6-5.2	3.3-5.3
attenuation [dB]	8	8	8
temperature stability [ns] 20-50°C	5	5	5
attenuation of third echoes [dB]	22	20	20
attenuation of spurious echoes [dB]	30	22	22
working temperature range	$-$20-70°C	—	—

An essential element of the SEC visualization system is the phase detection unit. This unit also uses a delay line with the delay T_p simultaneously with a broadband phase shifter with a shift of $+90^0$ over the bandwidth from 3.8 to 4.8 MHz. In practice it appears that this unit is sufficiently effective for a narrowed shifter bandwidth $f = 0.5$ MHz, which simplifies greatly its design. Phase measurements were taken using a doubly balanced mixer in which a signal shifted in phase by 90° is multiplied by a signal delayed by the time T_p.

The output of the phase detector is connected to an built-in oscilloscope which displays simultaneously dynamic flow profiles, the position of an analyzing gate and distance markers. The frequency response measured at the output of the phase detector is shown in Fig. 11.

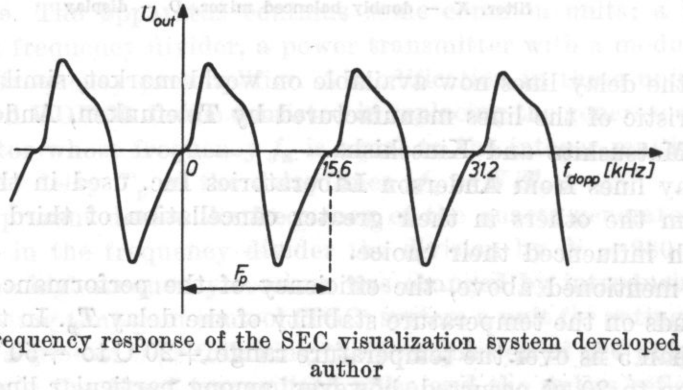

Fig. 11. The frequency response of the SEC visualization system developed by the present author

The system has a quasi-monotonous frequency-voltage characteristic over the range $-\frac{1}{4} F_p$ to $+\frac{1}{4} F_p (\pm 3.9$ kHz), while outside this range the phase of a signal varies with keeping the same sign of the Doppler frequency (discrimination of the flow direction) over the range $-\frac{1}{2}F_p$ to $+\frac{1}{2}F_p$, i.e. in the frequency range measured using a pulse Doppler flowmeter.

From this diagram it is possible to evaluate the degree of cancellation of stationary echoes (the frequency nF_p, where $n = 0, \pm 1, \pm 2, \ldots$) and of slowly moving objects whose velocity corresponds to Doppler frequencies < 300 Hz.

It is interesting to note the broadband character of the SEC system. In pulse Doppler flowmeters the maximum frequency measured cannot exceed half the repetition frequency F_p. This condition limits severely the possibility

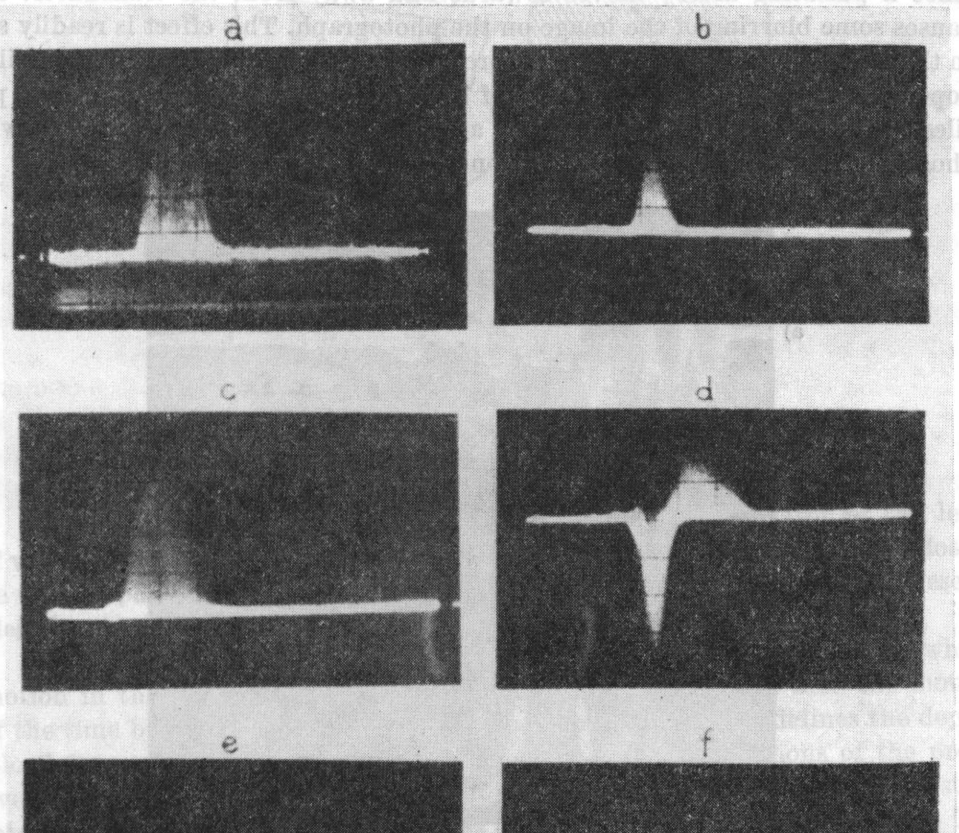

Fig. 12. Dynamic blood flow velocity profiles in common carotid artery (a), internal carotid artery (b), common femoral artery (c), simultaneously in jugular vein and carotid artery (d), subclavian vein (e) and femoral vein (f)

of evaluating the blood flow in a constriction of the carotid artery where, according to SPENCER *et al.* [15], the upper Doppler frequency may exceed 20 kHz. This restriction does not apply to the SEC visualization system whose characteristic covers a range of $\pm$ several dozens of kHz.

Fig. 12 shows photographs of typical signals corresponding to a dynamic representation of blood flow profiles on the oscilloscope tube display. The length of the time base on the oscilloscope screen corresponds to a depth of 5 cm in the body. In the case of arteries the base of the blood flow profiles displayed corresponding to the diameter of the vessel investigated is wider than in practice, since a pulsating artery moves towards and away from the transducer. This causes some blurring of the image on the photograph. This effect is readily seen in the course of observing blood flow profiles in real time, directly on the oscilloscope screen. A directional character of blood flow detection is evident: the profiles corresponding to blood flow in arteries have a positive direction, while those for flow in veins a negative one.

a)

b)

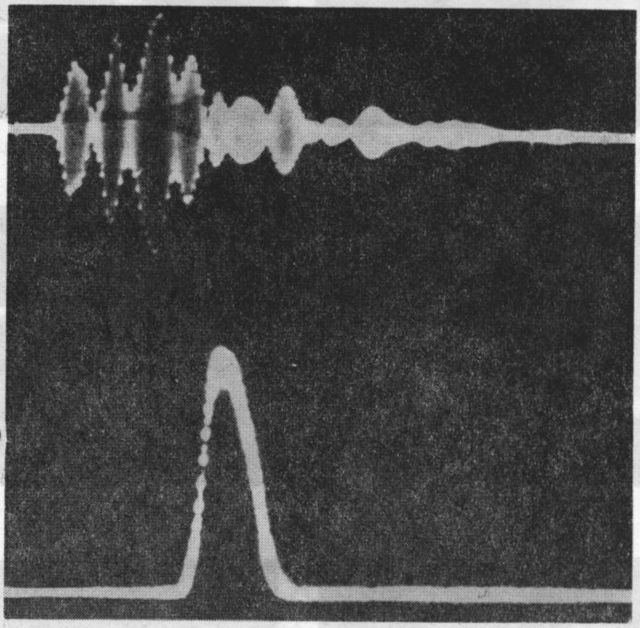

Fig. 13. Ultrasonic echoes at the input of the SEC system obtained in the investigation of blood flow in the common carotid artery (a) and a flow profile in this artery after cancellation of stationary echoes above 55 dB (b)

The efficiency of stationary echo cancellation can be evaluated on the basis of flow measurements in the common carotid artery. Fig. 13a shows echoes reflected from different tissues and walls of the artery measured at the input of the SEC system, while Fig. 13b shows a blood flow profile at the output of the phase detector after cancellation of stationary echoes in the SEC system.

The representation of blood flow profiles on the oscilloscope screen is essentially a modification of A-scanning where the coordinate x defines the depth and the coordinate y defines the blood flow velocity (in a typical A-scanning the coordinate y is proportional to the intensity of an ultrasonic signal reflected on the boundary of tissues with different acoustic impedance).

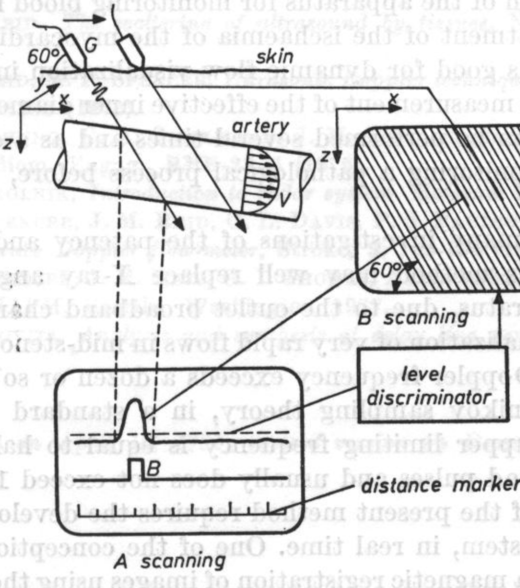

Fig. 14. The principle of the visualization of a vessel in the SEC system

A combination of a phase detection system, a discriminator of the level of velocity profiles and a system for spot brightness modulation on the oscilloscope tube display provides a system for visualization of blood vessels which resembles B-scanning (Fig. 14).

In this system the ultrasonic probe is mounted on a mobile arm whose motion in the x axis is transformed in the electronic system into the motion of the time base in the x direction. The coordinate z, in turn, defines the depth. Blood flow velocity profiles corresponding to successive positions of the probe over the vessel investigated are transformed in the level discriminator system into a signal modulating the brightness of spots on the oscilloscope screen. The time base on the screen is blackened when no blood flow occurs and brightened only by a signal corresponding to a velocity profile. The length of a section of the time base brightened corresponds to the diameter of a vessel at the site examined.

5. Conclusions

In view of limited material of clinical measurements, it is difficult to evaluate fully the SEC apparatus. It is now undergoing tests at the Clinic of Vascular Surgery CMKP, Warsaw (headed by Prof. Dr. Med. H. RYKOWSKI) and at the

Department of Pathophysiology of Circulation System, Institute of Pediatrics, Academy of Medicine, Warsaw (headed by Prof. Dr. Med. A. Chrościcki). The first results show that, apart from visualization of vessels, this method will be particularly useful in the evaluation of the dynamics of flows inside the heart in children's cardiology. A parallel direction of investigation concerns the adaptation of the apparatus for monitoring blood flows in vascular transplants in the treatment of the ischaemia of the myocardium.

The SEC system is good for dynamic flow visualization in real time, with the possibility of direct measurement of the effective inner diameter of the vessel. These investigations can be performed several times and as a consequence this method is good for monitoring a pathological process before, in the course of and after operation.

It seems that in many investigations of the patency and constriction of superficial arteries this method may well replace X-ray angiography.

The present apparatus, due to the outlet broadband characteristic of the system, is good for visualization of very rapid flows in mid-stenosis of the carotid artery for which the Doppler frequency exceeds a dozen or so kHz. According to the Shannon-Kotelnikov sampling theory, in a standard ultrasonic pulse Doppler method the upper limiting frequency is equal to half the repetition frequency of transmitted pulses and usually does not exceed 10 kHz.

A full aplication of the present method requires the development of a convenient registration system, in real time. One of the conceptions concerns the use for this purpose of a magnetic registration of images using the video recorder, a technique increasingly often used in the registration of images in echo-cardiotomography in real time.

References

[1] F. E. Barber, D. W. Baker, A. W. C. Nation, D. E. Strandness, J. M. Reid *Ultrasonic duplex echo-Doppler scanner*, IEEE Trans. On Biom. Engng., **BME 21**, 2 (1974).

[2] W. Buschman, *Ultrasonic imaging of arterial wall echoes*, Ultrasound in Med. and Biol., **1**, 33-43 (1975).

[3] L. Filipczyński, R. Herczyński, A. Nowicki, T. Powałowski, *Blood flows: hemodynamics and ultrasonic Doppler measurement methods* (in Polish), PWN, Warszawa — Poznań 1980.

[4] L. Filipczyński, G. Łypacewicz, J. Sałkowski, T. Waszczuk, *Automatic eye visualization and ultrasonic intensity determination in focused beams by means of electrodynamic and capacitance methods*, Proc. 2nd European Congress on Ultrasonic in Medicine, Munich 12-16 May, 1975; Excerpta Medica, Amsterdam 1975.

[5] L. Filipczyński, J. Groniowski, G. Łypacewicz, J. Sałkowski, T. Waszczuk, *Design of UG-4 ultrasonograph and its application in obstetrics and gynaecology* (in Polish), Probl. Techn. Med., **6**, 231 (1975).

[6] A. Nowicki, *Ultrasonic pulse Doppler method in blood flow measurement*, Archives of Acoustics, **2**, 4, 305-323 (1977).

[7] A. Nowicki, *Simplified automatic measurements of blood flow by the ultrasonic pulse Doppler method*, Archives of Acoustics, **4**, 4, 359-366 (1979).

[8] A. Nowicki, M. J. Reid, *An infinite gate pulse Doppler*, Ultrasound in Med. and Biol., **7**, 41-50 (1981).

[9] P. A. Peronneau, F. Leger, M. Xhaard, M. Pellet, P. V. Schwartz, *Velocimetre sanguine a effect Doppler a emission ultrasonore pulse*, L'onde Electrique, **50**, 5, 369-389 (1970).

[10] J. M. Reid, *The scattering of ultrasound by tissues*, N.B.S. Special Publ., **453**, 29-47 (1976).

[11] J. M. Reid, M. P. Spencer, *Ultrasonic Doppler technique for imaging blood vessels*, Science, **176**, 1235-1236 (1972).

[12] K. K. Shung, R. A. Sigelmann, J. M. Reid, *Scattering of ultrasound by blood*, IEEE Trans. on Biom. Engng., **BME-23**, 6 (1976).

[13] M. I. Skolnik, *Introduction to radar system*. McGraw Hill, New York 1962.

[14] M. P. Spencer, J. M. Reid, O. L. Davis, P. S. Paulson, *Cervical carotid imaging with a continuous wave Doppler flow-meter*, Stroke, **5**, 145-154 (1974).

[15] M. P. Spencer, J. M. Reid, G. I. Thomas, *Noninvasive cerebrovascular evaluation*, Special Publ. of IAPM, Seattle, Washington 1977.

[16] H. Urkovitz, *Analysis and synthesis of delay line periodic filters*, IRE Trans., **CT-4**, 41-53 (1957).

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