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AN ULTRASONIC METHOD FOR THE INVESTIGATION OF CORONARY GRAFT PATENCY

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A new method and system for the visualization of blood flow in coronary grafts is presented. This method is based on the technique of stationary echo cancellation (SEC), phase detection and integration of Doppler signals in real time, using broad-band analog CCD delay lines. Preliminary results of examinations of coronary graft patency for 15 patients are presented.

1. Introduction

Visualization of biological structures in real time is now the main direction in the development of ultrasonic diagnostic methods. With the unquestionable advantages of this new technique, particularly for dynamic imaging of the heart, it neglects the hemodynamic parameters related to the blood flow in the cavities of the heart, in the area of the mitral, aortic, pulmonary and tricuspid valves and in the coronary vessels.

The main aim of this paper is to develop a method and system for noninvasive visualization of blood flow in coronary grafts. At the same time, this method could be used in investigations of children's hearts.

2. Method and system

Of the many methods which are used for the measurement of the hemodynamic parameters, in practice only the so-called Doppler method permits transcutaneous noninvasive measurements of blood flow in the peripheral vessels and in the heart.

The position of the left, anterior coronary graft (LAD), which runs diagonally down from the aorta in the area of the right ventricle outflow track, causes the continuous wave method to be greatly unreliable, in view of the simultaneous effect of the flows in the two vessels (aorta + pulmonary artery) with the left-diagonal position of the probe in the second or third intercostal space and also in view of the motion of the pulmonary valve and, quite frequently, of the

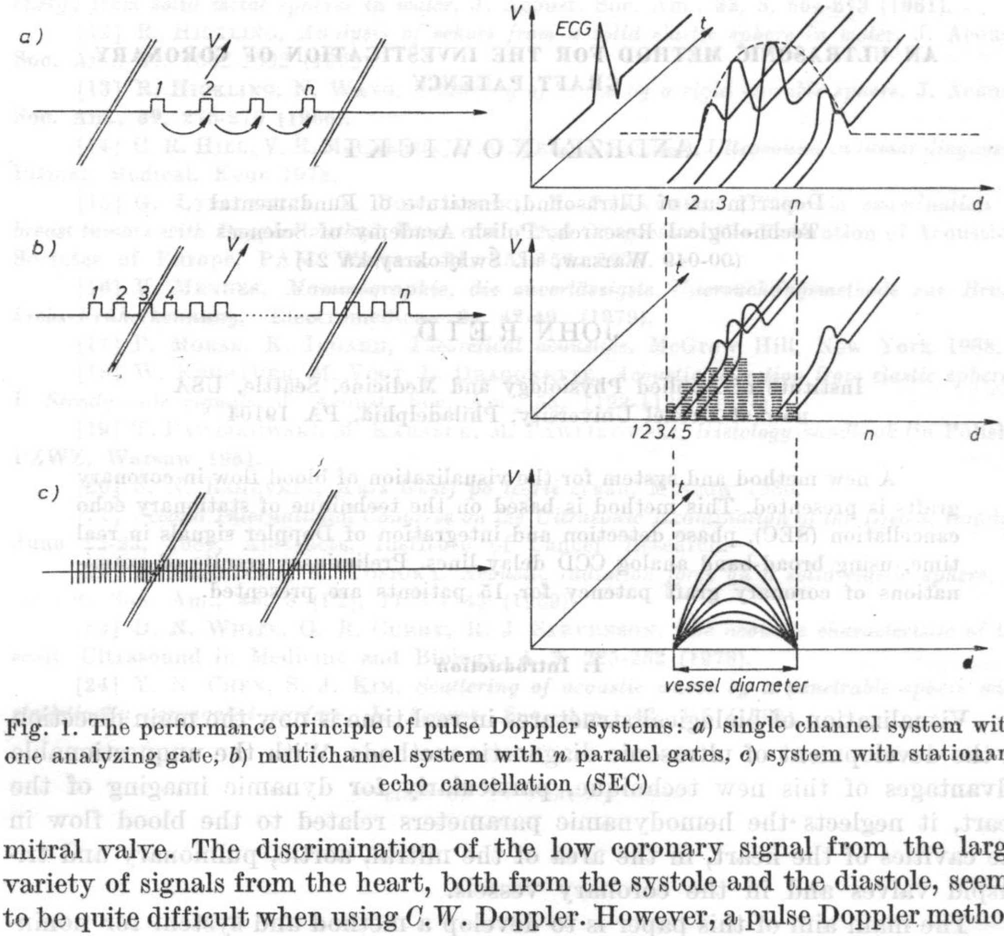

Fig. 1. The performance principle of pulse Doppler systems: *a*) single channel system with one analyzing gate, *b*) multichannel system with *n* parallel gates, *c*) system with stationary echo cancellation (SEC)

mitral valve. The discrimination of the low coronary signal from the large variety of signals from the heart, both from the systole and the diastole, seems to be quite difficult when using *C.W.* Doppler. However, a pulse Doppler method which combines the properties of the echo type and Doppler type equipment, thus providing both frequency and range resolution [2], offers a great promise in this field.

The recent years have brought large developments in Doppler pulse flowmeters, DPF, with three distinct varieties (Fig. 1),

a) Doppler pulse flowmeters with one analyzing gate [6] (e.g. UDP-30 produced by ZD Techpan);

b) Doppler pulse flowmeters with a large number of analyzing gates (e.g. Fish's system with 30 analyzing gates) or with serial digital data processing [1, 4, 5];

c) Doppler pulse flowmeters with stationary echo cancellation [7, 9]. The high cost of the multi-gate equipment, both analog and with serial digital data processing, has encouraged investigations of the third of the systems mentioned above. In addition this was to a large extent affected by the fact that the dynamics of stationary and Doppler signals from the motion of the heart and the blood flow frequently exceeds the range of the fast 8-bit A/D converters now available (with the real dynamics of the Doppler signals being evaluated at 60-90 dB).

The first model of equipment using stationary echo cancellation was developed in 1977-78; it had the coefficient of stationary echo cancellation > 50 dB and a range of $\simeq 5$ cm. Although the cancellation was sufficient, the range was too short for cardiological applications. It resulted from the use in the systems of stationary echo cancellers. They were periodic filters of the first order [7], of quartz delay lines with the delay $T = 64$ μs. This corresponds to a repetition frequency of 15.625 kHz and a range of $\simeq 5$ cm.

In order to increase the range, a new generation of the equipment was developed using CCD delay lines custom-made by Reticon USA in the construction of periodic filters, a phase detector, and an integrator with a positive feedback loop. Such a line consists of 295 "delay elements". The analog signal is controlled by a four-phase signal of the clock frequency f_c. The delay T_0 of the line is $295/f_c$.

The purpose of the equipment affected to a large extent the choice of the frequency f_n and the repetition interval T_p. These two parameters are related to the so-called ambiguity function [6] of Doppler pulse systems. This function describes the relations between the maximum range, resolution and the maximum measured velocity for given values of the parameters f_n and T_p.

In order to avoid the range "ambiguity", the repetition time T_p should be long enough so that the ultrasonic echoes could return to the receiver before the next pulse is transmitted. In view of this, the maximum penetration depth is defined by the expression $d = T_p c/2$, where c is the ultrasound propagation velocity in tissue.

In turn the minimization of the "ambiguity" in the measurement of the flow velocity $V_{\max}$ ($V = f_d c/2f_n$) imposes a converse condition. According to sampling theory, the maximum Doppler frequency f_d must be less than half the repetition frequency f_p ($f_p = 1/T_p$). These two conditions can be given by one expression which restricts the magnitude of the product of the maximum range $d_{\max}$ and the maximum velocity $V_{\max}$,

$$d_{\max} V_{\max} < c^2/8f_n. \tag{1}$$

In investigations of the heart the desired penetration depth should be greater than 10 cm. In a healthy heart the maximum flow velocity usually does not exceed several score cm/s, except the ascending aorta, where the flow velocity can exceed 1.5 m/s. In cases of heart defects this velocity increases greatly and can reach a value of several m/s, at the level of narrowed mitral and tricuspid valves. In coronary grafts the flow velocity is much lower, not exceeding dozen-odd cm/s.

The technical difficulties related to the accurate measurement of low velocities (of Doppler frequencies < 200 Hz), which are in turn related to the effect of the external interference and the "masking" of signals corresponding to slow flow by those caused by the motion of the heart walls (with the amplitudes of the latter being greater by more than 40 dB than the signals scattered in blood), and the attenuation of ultrasound in blood (this attenuation increases linearly as the frequency increases) restrict the frequency range to be applied in cardiology to 2-5 MHz.

Lower frequencies are usually used on adults; higher, on children.

In the present system the transmitter frequency $f_n = 4$ MHz and the repetition time of transmitted pulses, $T_p = 147{,}5$ μs were chosen. For this repetition time the frequency of the generator controlling an analog CCD delay line is 2 MHz.

It should be noted that the repetition time of high-frequency pulses, the high-frequency signal and the clock signal of the CCD line should be coherent so as to prevent phase drift in the channel of Doppler frequency detection and the drift of the delay of the line with respect to the time of processing transmitted pulses. The repetition interval T_p corresponds to the range $d = 11.4$ cm, under the assumption that the mean velocity c of ultrasound propagation in the body is 1550 m/s. At the level of narrowed valves and in the aorta the permissible range of measured velocities (< 66 cm/s) is too short; it ensures, however, unambiguous measurements in coronary grafts.

In view of the more than hundredfold difference between the level of echoes reflected from the walls of the heart and signals scattered in the blood flowing in coronary vessels and the cavities of the heart, in the solution proposed here the stationary echo cancellation SEC should be greater than 40 dB. This imposes the condition that in the construction of periodic filters SEC of at least the second order should be used. Elements of a theory which describes the performance of such filters were given in a previous paper [9].

Fig. 2 shows a schematic diagram of the device. In view of its similarity to the diagram of the UDP-30 and UDP-30 SEC devices described previously [3, 9], only this part of the system which implements stationary echo cancellation and phase detection and the integrator with a positive feedback loop were described in greater detail.

For this purpose, Fourier analysis was used to give the transforms of time signals at the characteristic points of the system. The signals $f_1(t)$ and $f_4(t)$, whose Fourier transforms are given by the following expressions, are obtained at the output of all the mixers M.

$$F_1(f) = \tfrac{1}{2}[F(f-f_n)+F(f+f_n)]; \tag{2}$$

$$F_4(f) = -\tfrac{1}{2}j[F(f-f_n)-F(f+f_n)]. \tag{3}$$

For a single SEC filter

$$F_{\text{out}}(f) = F_{\text{in}}(f)\exp(-j2\pi fT). \tag{4}$$

For a double SEC filter

$$F_2(f) = F_1(f)[\exp(-j2\pi fT)-1]^2; \tag{5}$$

$$F_5(f) = F_4(f)[\exp(-j2\pi fT)-1]^2. \tag{6}$$

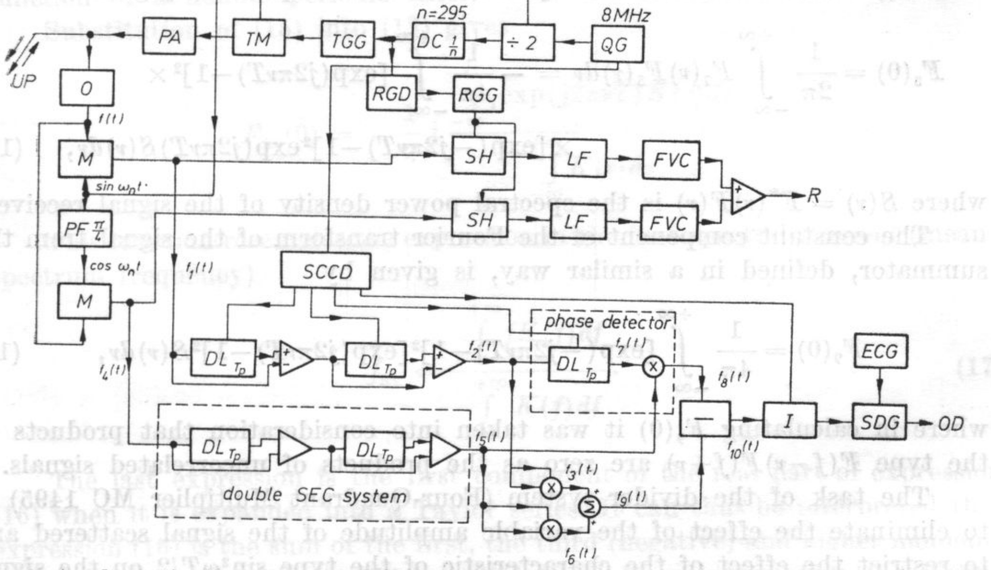

Fig. 2. A schematic diagram of the SEC device: *QG* — quartz generator, *DC* — frequency divider, *TGG* — transmitter gate generator, *TM* — transmitter modulator, *PA* — power amplifier, *UP* — ultrasonic probe, *RGD* — receiver gate delay, *RGG* — receiver gate generator, *SH* — sample and hold system, *LF* — low-pass filter, *FVC* — frequency/voltage converter, *R* — recorder, *O* — receiver, *M* — mixer, *PF* — phase shifter, *DL* — CCD delay line, *SCCD* — four-phase synchronization of CCD delay lines, *I* — integrator, *ECG* — ECG synchronization, *SDG* — systolic/diastolic gate, *OD* — oscilloscope display

According to the properties of Fourier transformation, the product of the signals in the time domain is equal to the convolution of its transforms in the frequency domain,

$$f_a(t)\cdot f_b(t) \leftrightarrow F_a(f)*F_b(f) = \frac{1}{2\pi}\int_{-\infty}^{+\infty} F_a(f-\nu)F_b(\nu)\,d\nu. \tag{7}$$

From (7),

$$F_3(f) = \frac{1}{2\pi}\int_{-\infty}^{+\infty} F_2(f-\nu)F_2(\nu)\,d\nu; \tag{8}$$

$$F_6(f) = \frac{1}{2\pi}\int_{-\infty}^{+\infty} F_5(f-\nu)F_5(\nu)\,d\nu. \tag{9}$$

The constant component of the Fourier transform of the signal at the output of the multiplier system (Four-Quadrant Multiplier IC 8013) has the form

$$F_8(f)_{DC} = F_7(f) * F_5(f) = \frac{1}{2\pi}\int_{-\infty}^{+\infty} F_7(f-\nu)F_5(\nu)\,d\nu. \tag{10}$$

Assuming the identity $F(f)_{DC} = F(0)$ and keeping in mind that $F(-\nu) = F^*(\nu)$, where $*$ denotes the operator of the complex conjugate function

$$F_8(0) = \frac{1}{2\pi}\int_{-\infty}^{+\infty} F_7(\nu)F_5(\nu)\,d\nu = -\frac{1}{4\pi}\int_{-\infty}^{+\infty}[\exp(j2\pi\nu T)-1]^2\times$$

$$\times[\exp(-j2\pi\nu T)-1]^2\exp(j2\pi\nu T)S(\nu)\,d\nu, \tag{11}$$

where $S(\nu) = F^*(\nu)F(\nu)$ is the spectral power density of the signal received.

The constant component of the Fourier transform of the signal from the summator, defined in a similar way, is given by

$$F_9(0) = \frac{1}{4\pi}\int_{-\infty}^{+\infty}[\exp(-j2\pi\nu T)-1]^2[\exp(j2\pi\nu T)-1]^2 S(\nu)\,d\nu, \tag{12}$$

where in calculating $F_9(0)$ it was taken into consideration that products of the type $F(f-\nu)F(f+\nu)$ are zero as the products of uncorrelated signals.

The task of the divider system (Four-Quadrant Multiplier MC 1495) is to eliminate the effect of the variable amplitude of the signal scattered and to restrict the effect of the characteristic of the type $\sin^2\omega T/2$ on the signal from the phase detector. Thus,

$$F_{10}(0) = \frac{F_8(0)}{F_9(0)} = -\frac{1}{2}j\frac{\int_{-\infty}^{+\infty}[\exp(j2\pi\nu T)-1]^2[\exp(-j2\pi\nu T)-1]^2}{\int_{-\infty}^{+\infty}[\exp(j2\pi\nu T)-1]^2[\exp(-j2\pi\nu T)-1]^2}\times$$

$$\times\frac{\exp(j2\pi\nu T)S(\nu)\,d\nu}{S(\nu)\,d\nu}. \tag{13}$$

Expression (13) can readily be reduced for one Doppler frequency only, i.e. in the case of one blood corpuscle or a group of corpuscles flowing at the same velocity. In such a case the signal received has the form $A\sin 2\pi\nu_1 t$, $S(\nu_1) = A^2$ and the product $[\exp(j2\pi\nu T)-1]^2[\exp(-(2\pi\nu T)-1]^2 = 16\sin^4\pi\nu T$.

After successive simplifications, the real part of expression (13) becomes

$$\mathrm{Re}\,F_{10}(0) = \tfrac{1}{2}\sin 2\pi\nu T. \tag{14}$$

The SEC visualizer, described previously [9], has a similar characteristic: a quasi-monotonous frequency response over the range from $-1/4\,F_p$ to $+1/4F_p$.

An essential difference, however, is the independence of the output signal of the amplitude of the signal received.

It is more difficult to interpret expression (13) for a wide spectrum of the Doppler frequency. Let

$$[\exp(j2\pi\nu T)-1]^2[\exp(-j2\pi\nu T)-1]^2 \; S(\nu) = K(\nu), \tag{15}$$

where $K(\nu)$ is the product of the power density function and the transmission function of a double periodic filter.

Substitution of (15) into (13) gives

$$F_{10}(0) = -\frac{1}{2} j \frac{\int_{-\infty}^{+\infty} \exp(j2\pi\nu T) K(\nu)\, d\nu}{\int_{-\infty}^{+\infty} K(\nu)\, d\nu}. \tag{16}$$

This function resembles the expression of the first spectral moment (mean spectrum frequency)

$$f_{\mathrm{av}} = \frac{\int_{-\infty}^{+\infty} f K(f)\, df}{\int_{-\infty}^{+\infty} K(f)\, df}. \tag{17}$$

The last expression is the first component of the real part of expression (16) when it is expanded into a Taylor series. It can thus be interpreted that expression (16) is the sum of the first, the third (negative) and higher moments of the Doppler spectrum. This signifies that in approximation (neglecting the terms of higher orders), a signal proportional to the mean flow velocity in all 295 gates is obtained at the output of the system. In other words, the signal at the oscilloscope *OD* corresponds to the "profiles" of the blood flow velocity.

The real frequency response of the system is shown in Fig. 3.

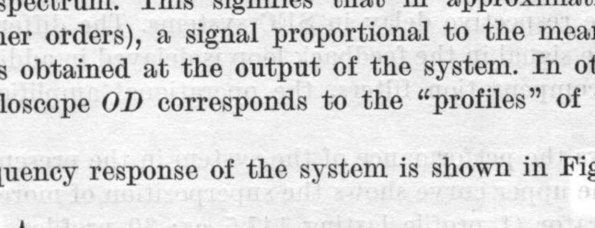

Fig. 3. The frequency response of the SEC device

The signal to noise ratio was greatly improved using an integrator with a feedback loop. The integrator also exerts a significant influence on the "legibility" of the dynamic flow profiles; it is known that instantaneous changes in the intensity of a scattered Doppler signal, which result from a stochastic distribution of red blood corpuscles in the ultrasonic field, affect to a large extent the signal from the stationary echo canceller and the phase detector. The integrator smooths out these changes and its function can be compared to the role of analog output filters in Doppler flowmeters.

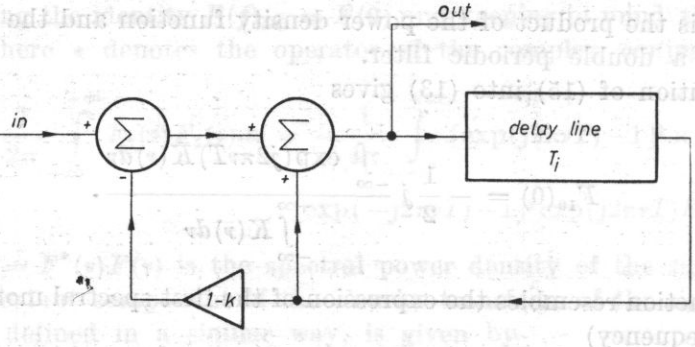

Fig. 4. A schematic diagram of the integrator

In this integrator the effective number of summed-up profiles is about $(1-k)^{-1}$, where $1-k$ is the amplification in the positive feedback loop. A stable performance of the integrator was achieved for a large value of the coefficient k, $k = 0.97$; i.e. about 30 profiles were integrated, increasing the signal to noise ratio by a factor of $\sqrt{30}$ [8].

The delay T_i of the CCD delay line in the integrator is shorter by about 350 ns than the respective delay in SEC systems. The difference results from the fact that the signal in the feedback loop is delayed in addition by the above values in the compensation filters, the operational amplifier $(1-k)$ and the summators.

Fig. 6 shows the performance of the system in the presence of large stationary echoes. The upper curve shows the superposition of more than 100 profiles from the integrator (1 profile lasting 147.5 μs; 30 profiles, 4.4 ms), received for a duration of 500 ms (the shutter opening time in a camera).

The improvement in the sensitivity of the system, particularly in the investigation of deeper structures, was caused by a large number of factors, including the role of the integrator mentioned above. A significant fact was also the use of special broad-band ultrasonic probes with relatively low internal losses. These probes were manufactured by Precision Acoustic Device; they provide multilayered, quarter-wave matching to the medium and very low, or none, damping by the back surface of the transducer.

In turn the identification of flow in coronary grafts was achieved using an electronic gate SDG controlled by the ECG signal of the patient. The performance of the system lies in the letting through or blocking of the output signal

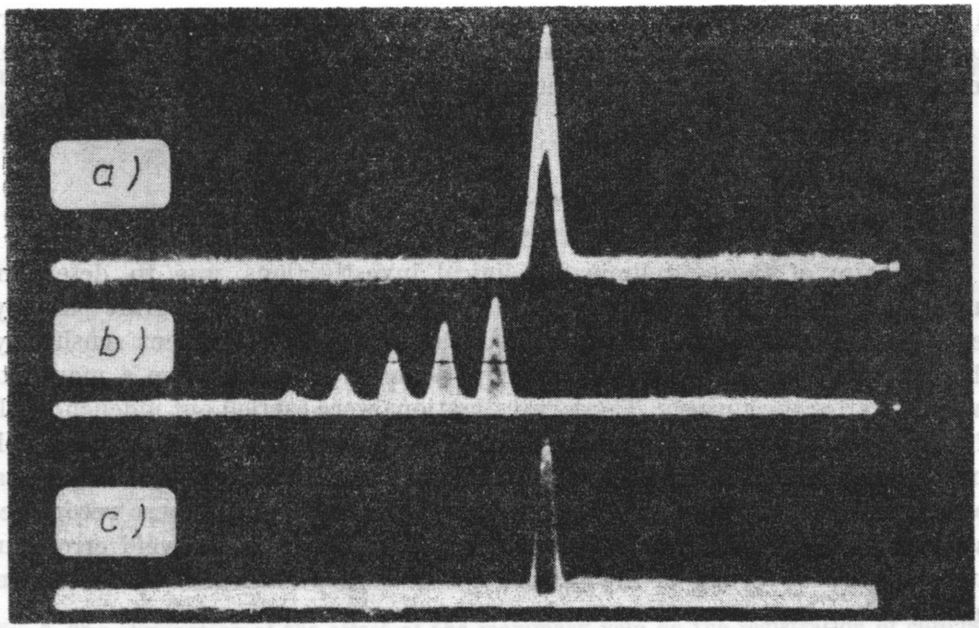

Fig. 5. The performance principle of the integrator: *a*) flow signal "profile" at the output of the integrator, $k = 0.97$; *b*) successive profiles under integration for a delay of the delay line less than the repetition time ($T_{int} < T_p$); *c*) signal at the input of the integrator for $T_{int} = T_p$

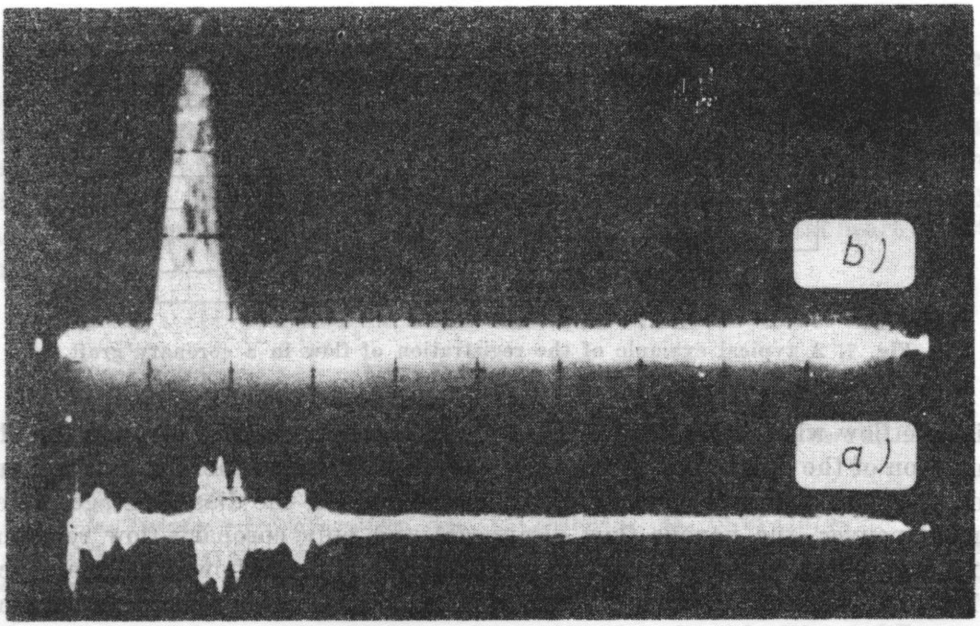

Fig. 6. Ultrasonic echoes: *a*) at the input of the SEC system; *b*) flow "profile" in the carotid artery. Stationary echo cancellation > 60 dB

in particular phases of the heart cycle. In the left, anterior coronary artery most flow occurs in the diastole, whereas the flows in the surrounding vessels (mammary, pulmonary artery) show a distinct systole component.

3. Results of clinical investigations

The aim of these preliminary clinical investigations was to determine the usefulness of the device for the investigation of coronary graft patency, i.e. a class of cases which require the highest possible equipment sensitivity. The examinations were carried out on patients with grafts, from a month to two years after the surgery. These investigations were carried out independently of arteriographic coronographic studies. In view of their readily accessible anatomical situation to the left of the chest in the second or third intercostal space, only left, anterior grafts were examined. Such grafts were recognized as patent which showed a distinct diastole flow. In order to avoid erroneous interpretation of results in the case of a flow superposition in the chest vein, the Valsalva maneuver was used.

After the identification of dynamic flows along the ultrasonic beam simultaneously at 295 gates, the flow velocity was registered at one gate located at a point where a distinct diastole flow occurred on the oscilloscope. An example of the registration of the coronary flow is shown in Fig. 7, where a distinct

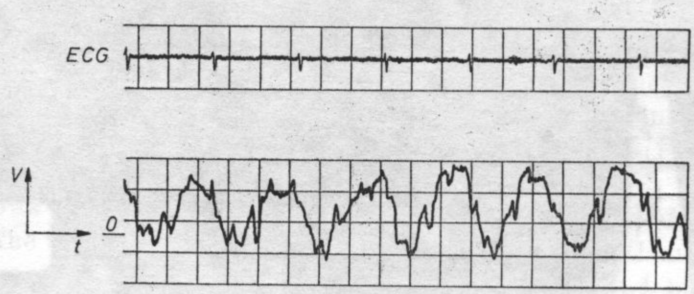

Fig. 7. A typical example of the registration of flow in a coronary graft

diastole flow with respect to the ECG signal can be seen. Fig. 8 shows the elimination of the flow signal in the mammary vein. Prior to the Valsalva maneuver, the signal consists of two distinct components in the diastole stage. After the maneuver, the venous flow disappears and only coronary flow remains. In a group of 15 patients examined arteriographically, 10 showed graft patency whereas 5 did not. In turn an ultrasonic examination showed graft patency in 6 patients; for the other 4 patients the signals registered failed to indicate a distinct diastole flow component.

It should be noted, however, that three of these patients were examined in the early stage of the verification of the method when the integrator, which improves greatly the sensitivity of the method, was not used.

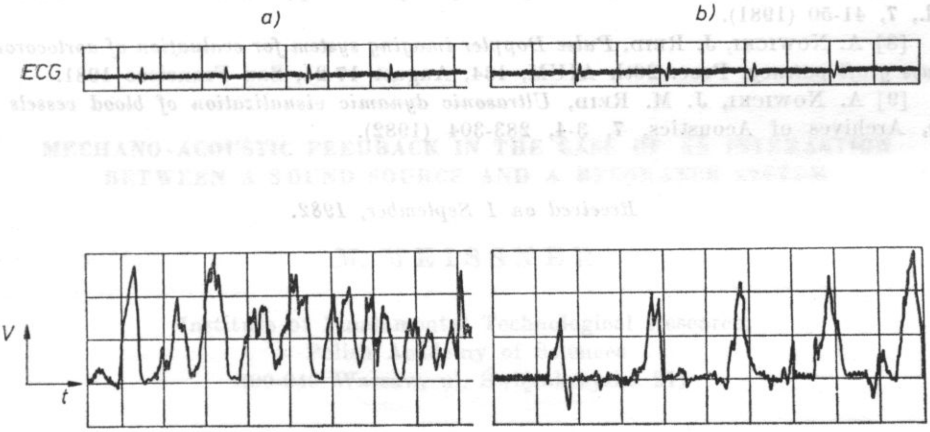

Fig. 8. The elimination of the effect of the mammary vein on the registration of coronary flow: signal before (*a*) and after (*b*) the Valsalva maneuver was used

4. Conclusions

The preliminary investigations of the present method have indicated its potential usefulness in direct evaluation of the dynamics of blood flows in real time both in small vessels and in the cavities of the heart. The noninvasive nature of the ultrasonic investigations makes them very attractive, particularly in the postoperational observation of coronary graft patency. A separate group of applications is the investigations of congenital heart defects in children.

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