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HIGH-FREQUENCY DOPPLER ULTRASOUND FLOWMETER

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This paper discusses a new Doppler device for measuring blood flow in small vessels. The ultrasound frequency applied is 20 MHz. Two different ultrasound probes were designed with transducer diameters of 0.9 mm and 1.6 mm. The transducers were located at the point of a catheter. As measured at 1.5 mm depth, the lateral resolutions were 1 mm and 1.7 mm, respectively. The flowmeter is very sensitive. Compared with standard flowmeters (from 2 to 8 MHz), signals from subcutaneous vessels are greater by 10-20 dB.

1. Introduction

The ultrasound Doppler measurement of the blood flow rate in small vessels is an important clinical issue both in the evaluation and diagnosis of the Raynoud disease as well as in the course of intraoperative vessel identification, e.g., during neurological operations and plastic surgery applied to hands and other organs. Because of the required high lateral resolution, the ultrasound beam should be narrow. The flows under study are often very slow; this, in turn, affects the choice of the ultrasound frequency, one which is several times higher than the one commonly used in standard flowmeters for the diagnosis of large peripheral vessels. Usually, flow rate measurements are taken for large peripheral vessels using 2-8 MHz frequencies.

HARTLEY and COLE [5], and the Green *et al.* [4] applied a pulsed Doppler method at 20 MHz for the intraoperative estimation of blood flow in small vessels. CATHIGNOL *et al.* [3] developed a device working at a similar frequency, but one with more limited measuring volume. PYNE [12] presented the results of blood flow rate in skin, an organ which is part of the thermoregulation system. In his research, Payne evaluated the usefulness of heat conductivity, xenon clearance, a Doppler laser flowmeter and the ultrasound Doppler working at 15 MHz and 30 MHz [11]. However promising, the results of the Doppler measurements showed that the

spectra of Doppler signals measured in small arterioles fall much below 50 Hz, making it difficult to process and, as a result, to evaluate them quantitatively. BERSON *et al.* [2] greatly enhanced the frequency used, up to as much as 113 MHz, expanding the spectra measured up to about 1 kHz, but limiting the probing depth down to that of several hundred microns.

2. Method and Apparatus

2.1. Frequency selection

The ultrasound reflected from moving red cells returns to the ultrasonic transducer at a frequency which varies depending on the transmitted wave frequency. The frequency difference is expressed by the formula

$$f_d = 2f_T \frac{V}{c} \cos\theta \tag{1}$$

where f_d is the Doppler frequency, f_T is the transmitted wave frequency, v is the ultrasound propagation in blood and θ is the angle between the ultrasonic wave propagation direction and the blood flow direction.

The selection of the optimum ultrasonic frequency is a compromise between attenuation by the medium being investigated and the wave energy being scattered by red cells. MC LEOD [8] gave a dependence describing the signal/noise ratio (SNR) as a function of scattering (an increase in signal with the fourth power of frequency), attenuation (an exponential signal drop with increasing frequency) and electronic noise which grows approximately linearly with increasing frequency.

$$\frac{S}{N} \div \frac{f_0^4}{f_0 d^2} e^{-2\alpha f_d d} \tag{2}$$

The optimum frequency f_{opt} (maximum SNR) can be calculated by differentiating expression (2) over frequency and zeroing the result

$$f_{opt} = \frac{3}{2\alpha d} \tag{3}$$

If attenauation α is denoted as k, in dB/MHz/cm, expression (3) becomes

$$f_{opt} = \frac{30 \log e}{k d} \tag{4}$$

The mean damping in soft tissue is about 1 dB/MHz/cm and, therefore, $f_{opt} = 15/d$ MHz. The experimental results [1, 3] indicate that in modern low-noise apparatus f_{opt} can be even twice as high. The ultrasonic frequency as determined from formula (4) falls between 15 MHz and 30 MHz for the wave penetration depth down to 1 cm.

2.2. Ultrasonic probe

As an effect of the technological limitations of polishing very thin piezoelectric (PZT 5) ceramic plates, the frequency of 20 MHz was chosen. At this frequency, the resonance thickness is about 0.1 mm. Attempts to produce thinner plates proved to be hardly repeatable.

Because of the intraoperative flow measurements in small vessels the probe should be as small as possible. In the C.W. method, the probe consists of two transducers, the transmitting and receiving ones, each having its own coaxial wiring to connect the transducer with the transmit-receive system. In the pulsed method, the ultrasonic probe has only one transmit-receive transducer connected with the transmitter and receiver equipped with the same wiring. It makes it possible to reduce greatly the external dimensions of the probe or catether.

HARTLEY and COLE [5] describe a pulsed Doppler 20 MHz system designed for intraoperative blood flow rate monitoring in dogs' coronary vessels. The transducer was fixed to a narrow and thin band wound round a vessel.

The basic problem which emerges as an intraoperative 20 MHz probe is built is the preparation of the transducer. The piezoelectric ceramics PZT-5A, which is universally used in medical diagnosis has good efficiency, and is readily available and cheap. It is

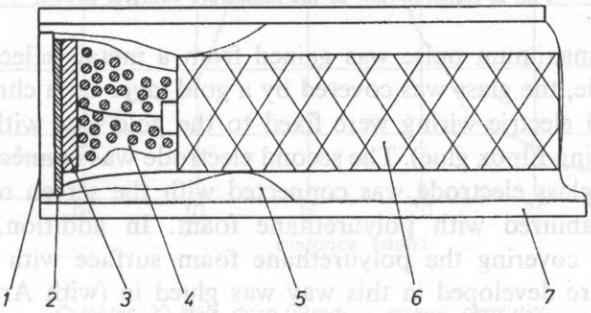

Fig. 1. Design of the tiny 20 MHz ultrasonic probe. 1) glass plate, 2) conducting glue layer, 3) ceramic transducer, 4) polyurethane foam, 5) glue, 6) coaxial cable, 7) pvc catheter.

necessary, however, to select the ceramics carefully, for at the required frequency of 20 MHz, its resonance thickness is lower than 100 μm and the grain thickness and brittleness of the ceramic sinter will begin to play an esential role. In successive operations, it is well-advised to eliminate the ceramics in which faults occur. What is particularly significant are the faults, revealed in the course of polishing, which can cause short circuits between electrodes, by filling the space with conducting glue. In the first instance, these transducers were polished to appropriate thickness using methods applied in optical glass processing. The result is an element which maintains polarization and has an electrode on one side. It was easiest to reduce the transducer diameter by gradual edge crushing until the desired dimensions were attained — 1.6 mm or 0.9 mm. The transducer was placed on a BK7, 0.2 mm thick glass plate, which was a substrate for the very delicate ceramic element. To improve the acoustic probe-water impedance matching, the glass plate was

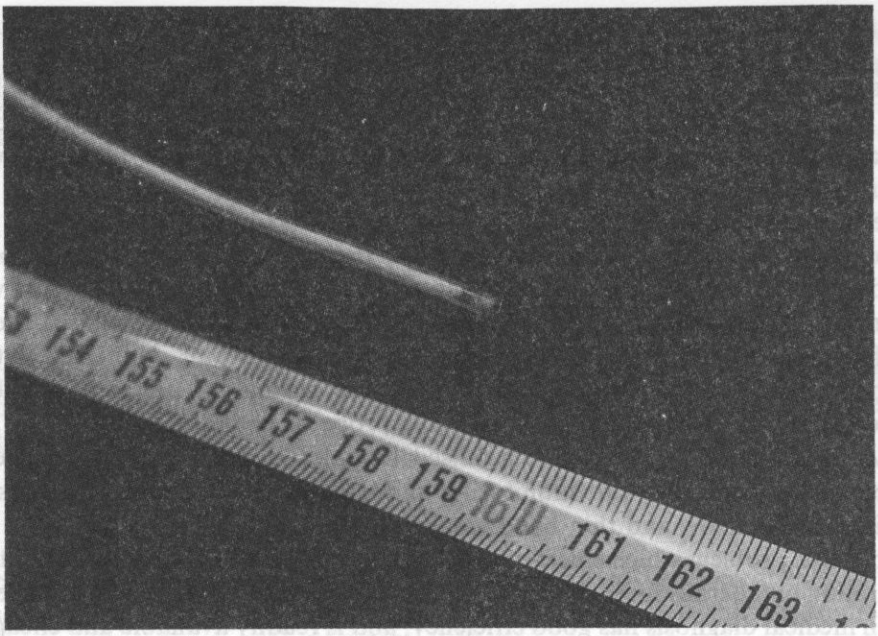

Fig. 2. General view of the miniature catheter probe.

polished until a maximum pulse was gained from a metal reflector submerged in water. On one side, the glass was covered by a gold layer on a chromium substrate. A transducer and electric wiring were fixed to the gold side without an electrode (with the conducting Elpox glue). The second electrode was connected with a coaxial 50 Ω cable. The glass electrode was connected with the screen of this cable. This structure was stabilized with polyurethane foam. In addition, the screen was supplemented by covering the polyurethane foam surface with conducting glue. Then, the structure developed in this way was glued in (with Araldit resin) inside a 1.6 m long PVC catheter, with an external diameter of 1.6 mm or 2.5 mm, depending on the transducer diameter. Over the section inside the catheter, a special concentric 50 Ω cable (Filotex Alcatel) was applied. At the site of connection with a standard RG 174 cable a parallel inductance was applied to compensate for the capacity of the transducer and cable. The probe and device were connected by a cable 1.95 m long.

The probe impedance was measured using a Network Analyzer HP 3577A bridge, comparing the measurement data with the impedance of a model whose parameters were changed so as to generate as similar curves as possible [6]. Figure 3 shows the modelling results. The differences between the measured and calculated model curves are caused by difficulties in defining the load on the back surface of the transducer and the thickness of the glass plate. A change in the layer thickness by several percent causes a very significant impedance change; and in the case of a layer with variable thickness, as a result of manual polishing, the resulting error cannot be calculated.

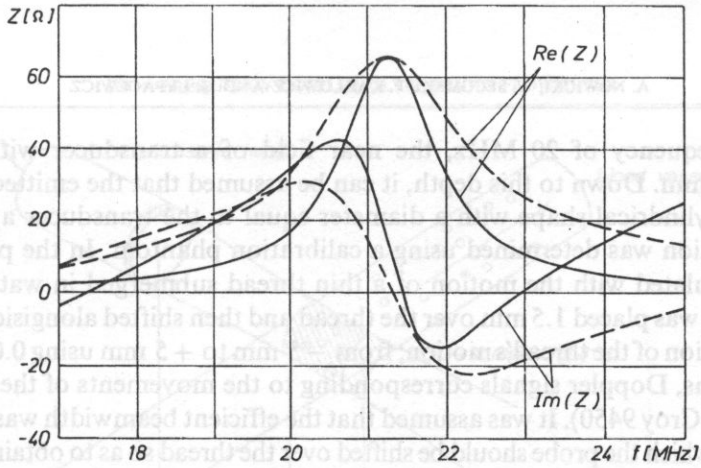

Fig. 3. Real and imaginary parts of the probe impedance measured using the HP 3577A bridge — solid lines; and those calculated for the model — dashed lines.

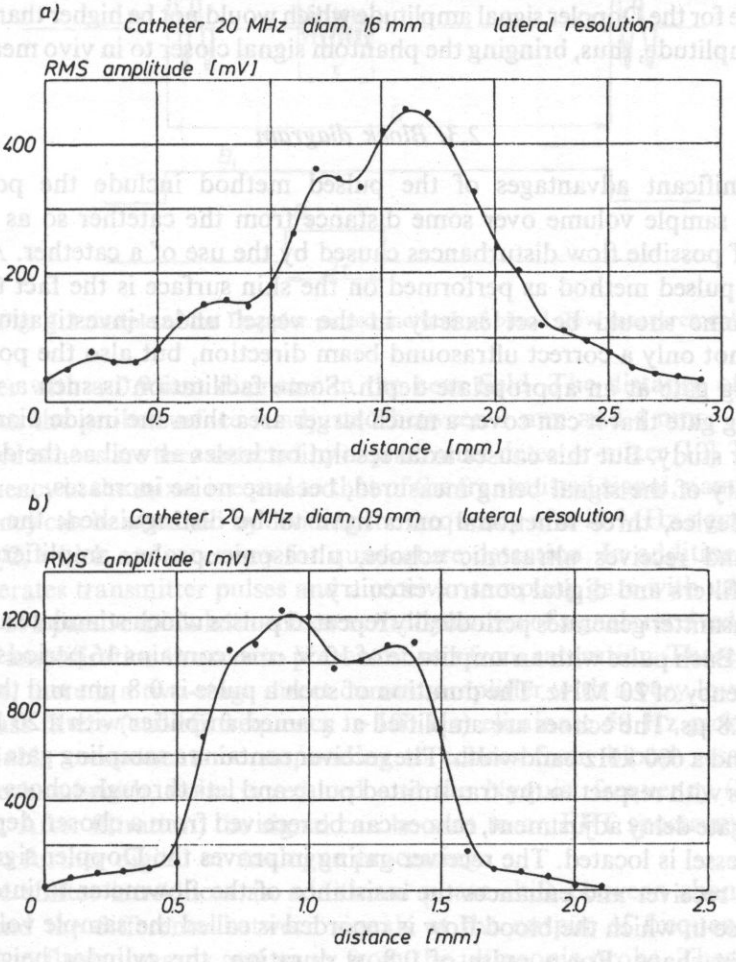

Fig. 4. Ultrasonic beam width (6 dB) measured using a thread phantom, a) transducer $\Phi = 1.6$ mm, b) transducer $\Phi = 0.9$ mm.

At the frequency of 20 MHz, the near field of a transducer with a 1.6 mm diameter is 8 mm. Down to this depth, it can be assumed that the emitted ultrasound beam has a cylindrical shape with a diameter equal to the transducer aperture. The lateral resolution was determined using a calibration phantom. In the phantom, the flow was simulated with the motion of a thin thread submerged in water.

The probe was placed 1.5 mm over the thread and then shifted alongiside in keeping with the direction of the thread's motion; from -5 mm to $+5$ mm using 0.05 mm steps. At all positions, Doppler signals corresponding to the movements of the thread were registered (Le Croy 9450). It was assumed that the efficient beamwidth was equal to the section along which the probe should be shifted over the thread so as to obtain a half of the maximum Doppler signal amplitude. As expected, the measured width (6 dB) was equal to the transducer aperture, that is 1.7 mm and 1 mm for a transducer with a diameter $\Phi = 1.6$ mm and, respectively, $\Phi = 0.9$ mm. The transmitter signal was decreased as much as to provide for the Doppler signal amplitude which would not be higher than 30 dB than the noise amplitude, thus, bringing the phantom signal closer to in vivo measurements.

2.3. Block diagram

The significant advantages of the pulsed method include the possibility of moving the sample volume over some distance from the catether so as to decrease the effect of possible flow disturbances caused by the use of a catether. A disadvantage of the pulsed method as performed on the skin surface is the fact that a small sample volume should be set exactly in the vessel under investigation. What it requires is not only a correct ultrasound beam direction, but also the positioning of the sampling gate at an appropriate depth. Some facilitation is such a widening of the sampling gate that it can cover a much larger area than the inside diameter of the vessel under study. But this causes axial resolution losses as well as the deterioration of the quality of the signal being measured, because noise increases.

In the device, three functional units have to be distinguished: the part which transmits and receives ultrasonic echoes, ultrasonic probe, amplifiers and low-frequency filters and digital control circuitry.

The transmitter generates periodically repeated pulses which stimulate an ultrasonic transducer. Each pulse with an amplitude of 10 V rms. contains 16 periods of sinusoid at the frequency of 20 MHz. The duration of such a pulse is 0.8 μm and the repetition period is 12.8 μs. The echoes are amplified at a tuned amplifier with a 20 MHz center frequency and a 600 kHz bandwidth. The receiver contains a sampling gate with a delay which varies with respect to the transmitted pulse and lets through echoes over 1.6 μs. Due to the gate delay adjustment, echoes can be received from a chosen depth at which the blood vessel is located. The receiver gating improves the Doppler signal-to-noise ratio of the receiver and enhances the resistance of the flowmeter to interference.

The space in which the blood flow is recorded is called the sample volume, and is cylindrical in shape. For a pulse of 0.8 μs duration, the cylinder height is about 0.5 mm. Its diameter depends on the ultrasound beam width, and is 1 mm for

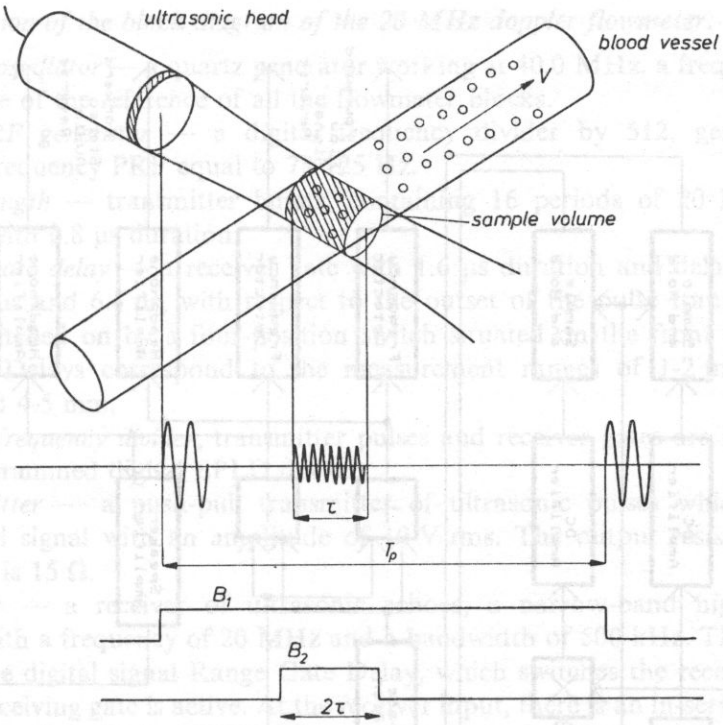

Fig. 5. Principle of the Doppler pulsed method of blood flow measurements.

a transducer with a 0.9 mm diameter in the near field. The distance of the sample volume from the probe surface is adjusted between 1 mm and 4 mm.

Amplified echoes are then detected by a quadrature detector-mixer [10]. The reference signal frequency in the mixer is equal to that of the transmitted signal, namely 20 MHz. This frequency can be obtained by dividing the frequency of a 40 MHz signals, shifted in phase by $\pi/2$, which are necessary for quadrature detection. In addition, this digital system generates transmitter pulses and a receiver sampling gate with variable delay.

The use of a passive modulator enhances the dynamics of signals received, but restricts the receiver amplification in order to prevent mixers from saturation. Therefore, directly after mixers, there is a two-stage direct current amplifier with a very low noise level.

High-pass filters (cut-off frequency >100 Hz) eliminate 50 Hz pick-up and the slowly variable components corresponding to echoes from blood vessel walls. By limiting the bandwidth, noise is reduced and the Nyquist frequency (39.6 kHz) is filtered aut. After filtration, the signal can be sent to an FFT spectrum analyser or subjected to further digital or analogue processing.

A phase shifter introduces an additional phase shift between channels by $\pi/2$. The sum and the difference between signals at the output of frequency detectors correspond to flows towards and away from the ultrasonic probe. These signals are also connected to a stereophonic earphone amplifier and a loudspeaker amplifier.

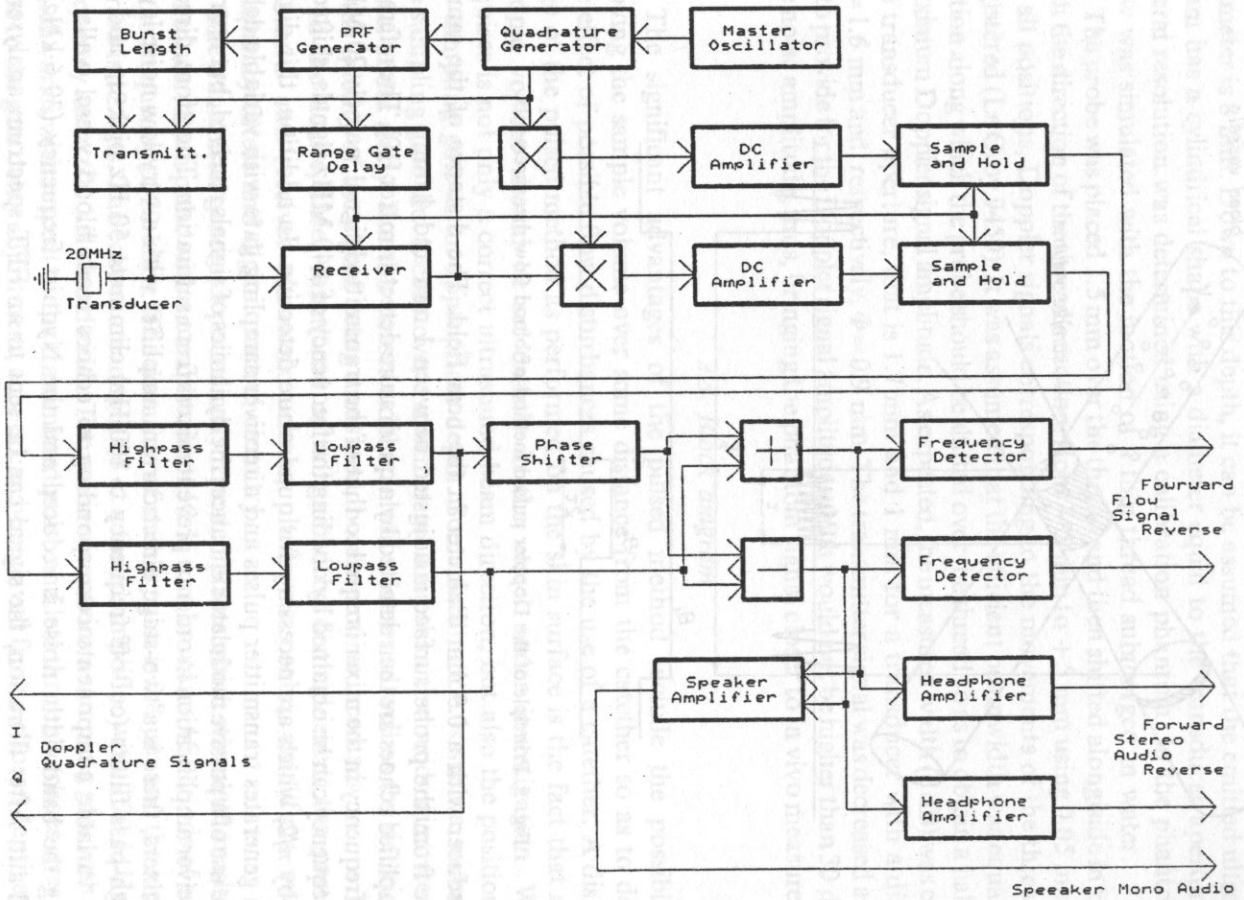

Fig. 6. Block diagram of the flowmeter.

Description of the block diagram of the 20 MHz doppler flowmeter.

Master oscillator — a quartz generator working at 40.0 MHz, a frequency which is the source of the reference of all the flowmeter blocks.

The *PRF generator* — a digital frequency divider by 512, generating the repetition frequency PRF equal to 78 125 Hz.

Burst length — transmitter bursts, containing 16 periods of 20 MHz carrier frequency with 0.8 μs duration.

Range gate delay — a receiver gate with 1.6 μs duration and delays of 1.6 μs, 3.2 μs, 4.8 μs and 6.4 μs, with respect to the outset of the pulse transmitted. The delay is switched on by a four-position switch situated on the front panel of the flowmeter. Delays correspond to the measurement ranges of 1-2 mm, 2-3 mm, 3-4 mm and 4-5 mm.

Digital frequency divider, transmitter pulses and receiver gates are implemented in one programmed digital EPLD device.

Transmitter — a push-pull transmitter of ultrasonic pulses which generates a sinusoidal signal with an amplitude of 10 V rms. The output resistance of the transmitter is 15 Ω.

Receiver — a receiver of ultrasonic echoes, a narrow-band high-frequency amplifier with a frequency of 20 MHz and a bandwidth of 500 kHz. The receiver is gated by the digital signal Range Gate Delay, which switches the receiver on only when the receiving gate is active. At the receiver input, there is an in-series resonance circuit with a diode limiter, which enhances the receiver sensitivity and separates the receiver from the transmitter during the transmission. The receiver consists of a two-gate MOSFET transistor and a monolithic buffer amplifier.

Quadrature generator — two digital frequency dividers by 2, controlled by a 40 MHz clock with inverse bias. As an effect, at their output there are 20 MHz signals with their phase shifted by $\pi/2$, which provides for quadrature detection.

DC amplifier — a post-detection signal is amplifed by a two-stage amplifier with a low noise level and full amplification of 27 dB.

High-pass filter — a first order high-pass filter which eliminates the D.C. component of the signal as well as the slowly variable components originating from blood vessel motions. The cut-off frequency of the filter is 80 Hz.

Low-pass filter — a Bessel fourth-order low-pass filter which eliminates the repetition frequency and prevents aliasing. The cut-off frequency of the filter is 16 kHz.

Phase shifter — a five-stage phase shifter which ensures the phase difference between channels I and Q equal to $\pi/2$ with an accuracy of 0.5% over the frequency range of 150 Hz-15 kHz.

Frequency detector — a zero crossing integrated detector, its output voltage is in direct proportion to the second moment of the input signal spectrum.

Headphone amplifier — a stereophonic headphone amplifier which makes it possible to hear separately flows towards and away from the probe.

3. Evaluation of the size of the signal scattered in blood

An attempt will now be made to estimate the effect of the sampling gate duration on the signal-to-noise ratio of the Doppler signal. If an ultrasonic wave is incident on a single blood cell, which is small compared with the wavelength, the scattered wave propagates in all directions as a spherical wave. A blood cell has the shape of a disk with an average thickness of 2.5 μm and a diameter of 8 μm. The average erythrocyte volume is about 90 μm^3. A sphere of this size would have a radius equal to 2.78 μm.

The strength of the scattered wave can be calculated from the product of the effective scattering surface δ and the incident wave intensity.

$$I_\delta = \left(\frac{P_t A}{x^2 \lambda^2}\right)\left(\frac{\delta}{4\pi x^2}\right) e^{-4\alpha x}, \tag{5}$$

where P_t is the transmitted signal power, α is the wave attenuation coefficient in the medium and A is the transducer surface area.

The first term is responsible for the intensity of the wave incident on a blood cell with the scattering surface δ, located at the distance x from the transducer. The second term accounts for the spherical wave propagation.

The determination of the power scattered by a large number of blood cells requires some discussion. The scattering cross section of a particle with dimensions much smaller than the wavelength can be determined from the expression given by MORSE and INGARD [9]

$$\delta = \frac{4\pi k^4 a^6}{9}\left[\left(\frac{\kappa_k - \kappa}{\kappa}\right)^2 + \frac{1}{3}\left(\frac{3\rho_k - \rho}{2\rho_k + \rho}\right)^2\right], \tag{6}$$

where $k = 2\pi/\lambda$ is the wave number, a is the blood cell radius, κ_k is the adiabatic compressibility coefficient of the blood cell equal to 34.1×10^{-13} m^2/dyne, κ is the adiabatic compressibility coefficient of plasma equal to 40.9×10^{-13} m^2/dyne, ρ_k is the blood cell density equal to 1.092×10^3 kg/m^3 and ρ is the plasma density equal to 1.021×10^3 kg/m^3 [13].

At 20 MHz, the scattering cross section δ(20 MHz) of one blood cell is equal to 9.24×10^{-16} m^2. 1% HMTC is equivalent to 1.07×10^{14} blood cells in 1 m^3, hence the scattering coefficient η (HMTC=40%, f=20 MHz) is $\eta = \delta(20\text{ MHz}) \times$ $\times 1.07 \times 40 = 3.96$ m^{-1}. Since for normal hematocrit (HMTC from 40% to 45%) the average distance between two blood cells is about 10% of their diameter; with such packing the blood cells cannoot be considered independent scattering sources. SHUNG *et al.* [13] showed that the scattering coefficient η grows linearly with increasing hematocrit only up to HMTC of 8%, and then an increase in η is slower, to reach a plateu for HMTC of 24-30%, and then to drop. From the scattering coefficients determined by SHUNG *et al.*, over the frequency range of 5 to 15 MHz, the value of η was approximated at 20 MHz, to obtain $\eta = 1.32$ m^{-1}, three times

lower than the value derived from formula (6). Since our purpose is only to estimate approximately the power scattered by blood cells, although some potentially large error can be made, it is justifiable to adopt the latter value in further calculations.

As a further step, the total intensity of the scattered signal can be determined as

$$I_r = \frac{P_t A^2}{x^4 \lambda^2 4\pi} \cdot \frac{c\tau}{2} \eta e^{-4\alpha x}. \tag{7}$$

In formula (7), the total radiation power, P_t, is unknown. The accuracy of measurements of the ultrasonic field distribution and the calculations of the power radiated as made on this basis depend on the dimensions of the active measuring component of the hydrophone. The available hydrophones include an active measuring component equal to about 0.5 mm, which is exceedingly large for transducer measurements taken in this case. Therefore, the scattering intensity was estimated only theoretically. The following calculation procedure was followed. On the basis of the measured impedance of the two probes, the coefficient of electromechanical coupling $k_t = 0.5$ and the dielectric constant $\varepsilon = 900$ were determined. These data were introduced to the equivalent model of piezoelectric transducer, acc. to MASON [7]; the acoustic pressures and powers as well as the intensities of the radiated wave and of the wave reflected from an ideal reflector at a distance of 1.5 mm from the transducer were calculated. The average skin absorption, $\alpha = 2.3$ dB/cm/MHz was adopted. When expressed in nepers, the total attenuation, after covering a distance twice as long as 1.5 mm, is 0.08 N/MHz. The calculated results are shown in Table 1.

Table 1. The calculated values of pressure, intensity, transmitted and received power and the receiving voltages at the transducer for two different 20 MHz piezoelectric transducers

	Transducer	
	$F=0.9$ mm	$F=1.6$ mm
Transmission voltage [V_{pp}]	30	30
Radiated acoustic pressure [N/m²]	0.79×10^6	0.46×10^6
Received acoustic pressure [N/m²]	0.15×10^6	0.89×10^6
Transmitted wave intensity [W/m²]	0.21×10^6	0.72×10^6
Received wave intensity [W/m²]	0.74×10^4	0.26×10^4
Radiated acoustic power P_1 [W]	0.13	0.14
Received voltage [V_{pp}]	0.865	0.854

For comparison, echoes from an ideal reflector were measured for the amplitude of the transmitted pulse, as in the simulation $V_{pp}=30$ V. The amplitude of an echo from an ideal reflector submerged in water was 1.5 V_{pp} (for the transducer with $\Phi=0.9$ mm). Taking into account the skin attenuation, instead of one in water, the echo in question would amount to 0.4 V_{pp}, slightly twice lower than the modelled one. Bearing in mind the fact that the model considered only the impedance of the transmitter and receiver, the twofold difference between the calculated and measured values can be accepted. In the scattering calculation, it was assumed that if the voltage 0.865 V_{pp} corresponds to the radiated power of 0.13 W, then the radiated power of 0.028 W corresponds to the voltage of 0.4 V_{pp}. Respectively, the intensity of the scattered wave received is 0.16 W/cm^2.

By substituting $\tau=0.8\times10^{-6}$ s, $\alpha=2.3$ dB/cm/MHz, $\eta=1.32$ m^{-1}, the transducer surface area $A=6.36\times10^{-7}$ m^2 (for a transducer with $\Phi=0.9$ mm) and $x=1.5$ mm, the intensity of the scattered wave $I_r=1.05\times10^{-4}$ W/cm^2, which is about 1520 times lower than that of a wave reflected from an ideal reflector. The voltage generated by the scattered wave would be almost 40 times lower, that is, 10 mV_{pp}.

In the pseudo-continuous mode, that is, without sample and hold gate, the noise signal detected in a time equal to the repetition period T_p is added to the Doppler signal from the vessel under study. In this case, the sampling gate is on throughout the time between successive transmitted pulses. During the repetition period $T_p=12.8$ μs the ultrasonic wave will travel, from here and back, a distance of 19.2 mm, corresponding to the penetration depth of 9.6 mm. The duration of the transmitted pulse τ is 0.8 μs.

Let the vessel under study be 0.75 mm wide and located at a distance of 1.5 mm from the probe. The ultrasonic pulse travels a distance from the anterior wall to the posterior wall of the vessel and back (when neglecting the angle between the beam and the vessel) over 1 μs. The duration of a signal scattered by a blood cell is then 0.078 of the repetition period T_p. In studies on small vessels, it is well-advised for the sampling gate to account for the whole vessel. Accounting for small displacements of a vessel in the course of measurements, in practical applications, the length of the sampling gate is slightly longer so that it can begin before and end after the wave has crossed a vessel. The length of the sample gate in the present system is 1.6 μs (0.125 T_p).

As estimated, the intensity of the wave scattered by blood cells is 1.05×10^{-4} W/cm^2 and, respectively, the voltage at the transducer is $U_r=10$ mV_{pp}. When refereed to its input, the effective voltage of the receiver noise is 6.6 μV rms. (18 μV_{pp}). The input signal is the sum of the Doppler component which corresponds to scattering and noise, $U_{in}=U_{dop}+U_{noise}$, with voltages equal respectively to 10 mV and 18 μV.

$$U_{dop} = \begin{cases} 10\ \text{mV} & \text{for } 2/10\ T < t < 3/10\ T \\ 0\ \text{V} & \text{for } t < 2/10\ T,\ t > 3/10\ T \end{cases} \quad (8)$$

These values correspond to those measured in a blood cell with an 0.5 mm diameter, located 1 mm under the skin surface. $U_{noise} = 18.5\ \mu V$ between the end of one transmitted pulse and the beginning of another, that is, practically, over the time T_p.

The Doppler component at the output of the integrating gate is equal to

$$\overline{U_{dop}} = \frac{1}{\frac{2}{16}T} \int_{\frac{2}{16}T}^{\frac{3}{16}T} 10\,\text{mV}\,dt = 5\,\text{mV} \quad (9)$$

on the other hand, the averaged noise voltage is

$$\overline{U_{noise}} = \frac{1}{\frac{2}{16}T} \int_{\frac{2}{16}T}^{\frac{4}{16}T} 18.5\,\mu\text{V}\,dt = 18.5\,\mu\text{V} \quad (10)$$

The signal-to-noise ratio, SNR, for such a signal is

$$\text{SNR} = 10 \log \frac{U_{dop}^2}{U_{noise}^2} = 48\ \text{dB}$$

The correct Doppler frequency measurement by the zero-crossing method (ZCC) requires SNR which is greater than 10 dB (a Doppler component three times greater than noise). The FFT spectrum analysis makes it possible to carry out measurements for smaller signal, SNR > 6dB, that is, when Doppler signal is twice as big as noise. The device is sufficiently sensitive even in the continuous mode, that is, when the sampling gate is open throughout the repetition period. In the continuous mode, without a sampling gate (or when it is open throughout the repetition time), the Doppler signal voltage is much lower (by a factor of five, in this case), because it is averaged over the time T_p, whereas the noise component remains at the same level. Even in this case, SNR is very high and exceeds 30 dB. In the continuous mode, it is much easier to locate a vessel. Figure 7 shows examples of Doppler signals recorded in the continuous mode from the digital artery, the radial artery and the upper palm vein. Therefore, e.g., the voltage of the signal scattered in a vein was about 2.7 mV_{pp}. With the sample and hold gate this voltage would increase by a factor of 5 up to 13 mV, the value being very close to 10 mV_{pp} theoretically calculated.

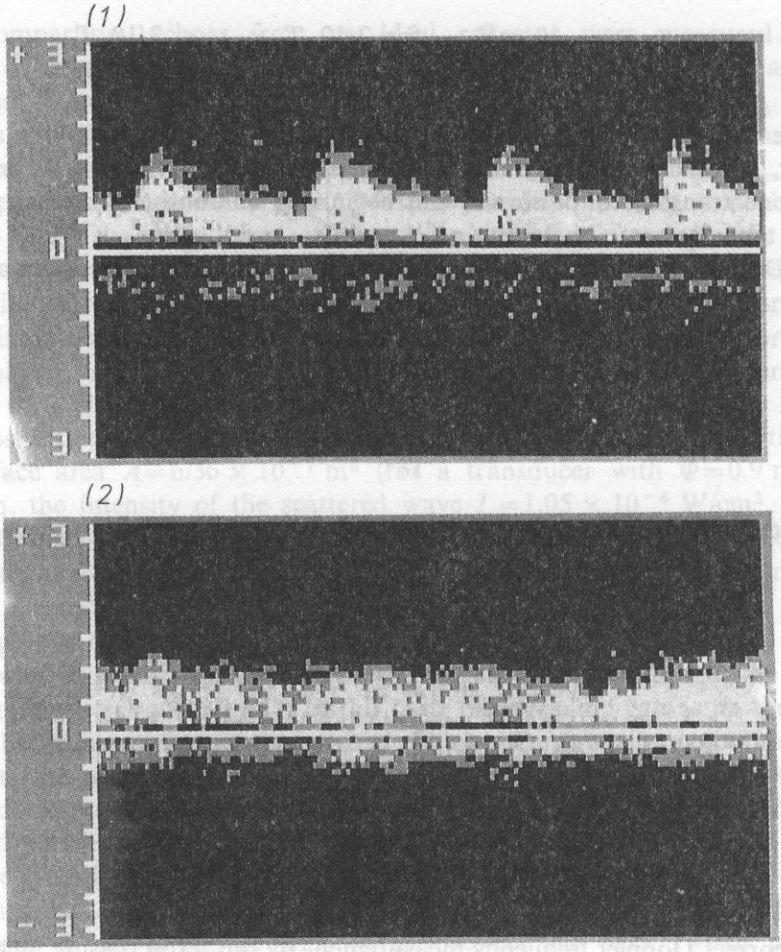

Fig. 7. Spectra of Doppler blood flow signals recorded subcutaneously at 1) digital artery at 2.5 mm depth, 2) hand vein at 2 mm depth.

4. Conclusions

Laboratory measurements confirmed the theoretically predicted, much larger amplitudes of Doppler signals compared with standard flowmeters working over the frequency range from 2 MHz. Special attention was paid to the design of the probe in order to achieve good lateral resolution. The total loss in the reflection of the radiated wave from an ideal reflector was lower than 26 dB.

The preliminary results of blood flow measurements in small vessels proved to be very promising. The high transversal and longitudinal resolution made it possible to locate and distinguish vessels under study, opening up a new application range of the Doppler technique in the intra- and post-operative control of the correctness of

vessel reconstructive surgery and the patency of the very narrow vessels sewn together in plastic surgery as well as in the course of and after bone replant operations in order to control the effect of perforators. Some other potential applications include the diagnosis of damaged flow in small hand vessels and intraoperative blood flow control in cranial vessels in the course of neurological operations.

References

[1] D.W. Baker, F.K. Forster and R.E. Daigle, *Doppler principle and technique, in:* Ultrasound, its application in medicine and biology, [Ed.] F.J. Fry Chapter 3, 161–287, Elsavier Sc. Publ. Company, New York 1978.

[2] M. Berson, F. Patat, Z.Q. Wang, D. Basse and L. Pourcelot, *Very high frequency pulsed Doppler apparatus*, Ultrasound in Med. and Biol., **15**, 2, 121–131 (1989).

[3] D. Cathignol, J.Z. Chapelon, J.L. Mestas and C. Furcade, *Description et application d'un velocimetre ultrasonore Doppler pour les petits vaisseaux*, Med. and Biol. Eng., **21**, 358–364 (1983).

[4] E.R. Greene, W.F. Blair and C. Hartley, *Noninvasive pulsed Doppler blood velocity measurements and calculated flow in human digital arteries*, ISA Trans., **20**, 2, 15–24 (1981).

[5] C.J. Hartley and J.S. Cole, *An ultrasonic pulsed Doppler system for measuring blood flow in small vessels*, Journal of Applied Physiology, **37**, 4, 626–629 (1974).

[6] G. Łypacewicz and E. Duriasz, *Design of ultrasonic probes for medical diagnostics*, Archives of Acoustics, **15**, 3–4 (1990).

[7] W.P. Mason, *Physical acoustics*, vol. 1, Part A, Academic Press, New York 1964.

[8] F.d. Mc Leod, *Multichannel pulsed Doppler techniques*, Cardiovasoular Res., 4, 428 (1972).

[9] P.M. Morse and K.U. Ingard, *Theoretical acoustics*, McGraw Hill, New York 1968.

[10] A. Nowicki, *Doppler Echography* (in Polish — Echografia dopplerowska), Polish Scientific Publishers PWN, Warszawa 1989.

[11] P.A. Paine, S.M. Jawad and R.Y. Faddoul, *A high frequency ultrasound pulsed Doppler system for the measurement of skin blood flow*, 4th Int. Symp. Bioeng. Skin, Besançon, France, Sept. 1983.

[12] P.A. Paine, *Measurement of skin blood flow*, Electronics and Power, March 30, 219–221 (1984).

[13] K.K. Shung, A. Rubens and J.M. Reid, *Scattering of ultrasound by blood*, IEEE Trans. Biomed. Eng. BME-24, **4**, 325–331 (1977).