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Real-Time Two-Dimensional Blood Flow Imaging Using Ultrasound Doppler

C. KASAI and K. NAMEKAWA

Aloka Co., Ltd., 6-22-1 Mure Mitaka, Tokyo 181, Japan

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A newly developed two-dimensional blood flow imaging system which uses ultrasound is described, in which blood flow in a given cross section of a live organ is displayed in real time. To perform this, an autocorrelation technique is used. Obtained blood flow data are displayed in color on a TV screen superimposed on black/white tissue images. The flow direction, speed and variance are expressed by differences in color and brightness. Experiments were conducted with a mechanical and an electrical scanner and good agreement with the theory was obtained. Studies on clinical significance have also been carried out for normal and diseased hearts with successful results.

§1. Introduction

Tomographic imaging that employs ultrasonic echoes has achieved outstanding advances in recent years, and today ultrasonic diagnostic equipment has become an indispensable tool for clinical use.

However, blood flows through the heart and blood vessels, and significant contributions to diagnostics could be made if the behavior of such flows could be anticipated. To do this, we developed a device that displays blood flow movements on a cross section of the heart or blood vessels in real time, using an autocorrelation technique: an outline of its operation was previously reported.[1,2] Details of the system, together with some clinical evaluations collected in a hospital were also described earlier.[3,4] In this paper we reiterate the principle and instrumentation of this real-time two-dimensional blood flow imaging system and furnish some clinical data. Additional detailed clinical data appears in reference.[5]

§2. Principle

In well-known B-mode instruments, only the amplitudes of echoes reflected at tissues are imaged. However, in blood flow visualization, the frequency change or phase shift, as well as the amplitude of echoes returning from corpuscles must be detected. Important aspects that provide the flow information necessary for diagrnosis are (1) flow direction, (2) mean flow velocity, and (3) flow disturbance.

(1) Flow Direction

In a pulse echo instrument employing one ultrasonic transducer, the flow component on the sound beam axis is measured. By detecting the polarity of the Doppler frequency shift of echoes with respect to the frequency transmitted, the flow direction (i.e., forward flow or reverse flow) is discriminated.

(2) Mean Flow Velocity

Mean blood flow velocity is estimated from the frequency spectra of echoes. When a sound with an angular frequency of ω_0 is transmitted into blood, the echo signal $e(t)$ received from the blood is described as follows:[6]

$$e(t)=R_e\{z(t)\,e^{j\omega_0 t}\} \tag{1}$$

where $z(t)$ is the complex envelope signal of $e(t)$ and is described as

$$z(t)=x(t)+jy(t) \tag{2}$$

Using a quadrature detector the real and imaginary parts of eq. (2) are obtained separately.[7] Denoting the power spectrum of $z(t)$ with $p(\omega)$, the mean angular frequency of $P(\omega)$ is expressed as[6]

$$\bar{\omega}=\frac{\int_{-\infty}^{\infty}\omega P(\omega)\,d\omega}{\int_{-\infty}^{\infty}P(\omega)\,d\omega} \tag{3}$$

Equation (3) gives the mean Doppler frequency shift due to the blood flow. The mean blood flow velocity $\bar{v}$ is estimated by the following equation:[7]

$$\bar{v}=\frac{\bar{\omega}}{\omega_0}\cdot\frac{c}{2\cos\theta} \tag{4}$$

where c is the velocity of sound and θ the angle between the sound beam and the blood flow vector.

In usual ultrasonic diagnostic instruments, the carrier frequency $f_0=\omega_0/2\pi$ is approximately in the range of 2–10 MHz, and the Doppler frequency $f=\omega/2\pi$ is normally less than 20 kHz.

(3) Flow Disturbance

The extent of disturbance in blood flow may be inferred from the variance of the spectrum.[8] Since the Doppler frequency directly relates to the flow vector (i.e., flow direction and speed) in an ultrasonic sample volume, the spectrum spread broadens in accordance with flow disturbance. However, in a laminar flow, the spectrum spread is narrow, since a uniform flow vector gives a singular Doppler-frequency shift.

Denoting the standard deviation of the spectrum by σ, the variance σ^2 may be represented by the following:[7]

$$\sigma^2=\frac{\int_{-\infty}^{\infty}(\omega-\bar{\omega})^2P(\omega)\,d\omega}{\int_{-\infty}^{\infty}P(\omega)\,d\omega}=\overline{\omega^2}-(\bar{\omega})^2 \tag{5}$$

Now we will find a way to measure the mean angular frequency and its variance using the autocorrelation function of the complex signal $z(t)$. By denoting the autocorrelation function with $R(\tau)$, the following relationship will pertain between $R(\tau)$ and $P(\omega)$ due to the Wiener-Khinchine theorem:

$$R(\tau)=\int_{-\infty}^{\infty} P(\omega)\, e^{j\omega\tau}\, d\omega \tag{6}$$

Expressing the primary and secondary differentials respectively by $\dot{R}(\tau)$ and $\ddot{R}(\tau)$ of $R(\tau)$ by τ, the following equations are derived from eqs. (4), (5), and (6):

$$j\bar{\omega}=\frac{\dot{R}(0)}{R(0)} \tag{7}$$

$$\sigma^2=\left\{\frac{\dot{R}(0)}{R(0)}\right\}^2-\frac{\ddot{R}(0)}{R(0)} \tag{8}$$

It is, of course, possible to measure the mean Doppler frequency and the variance using eqs. (7) and (8). However, direct computation of the equations is rather time-consuming with an actual instrument. Therefore, we will find a simpler method.

When the autocorrelation function is treated as

$$R(\tau)=|R(\tau)|\, e^{j\phi(\tau)} \tag{9}$$

the following approximations are derived:[4)]

$$\bar{\omega}=\dot{\phi}(0)=\frac{\phi(T)}{T} \tag{10}$$

$$\sigma^2=\frac{2}{T^2}\cdot\left\{1-\frac{|R(T)|}{R(0)}\right\} \tag{11}$$

where T denotes the emission interval of ultrasonic pulses. Equations (10) and (11) demonstrate the feasibility of obtaining the mean angular frequency and its variance from autocorrelation values and phases for $\tau=0$ and $\tau=T$.

Figure 1 shows the circuit of a complex autocorrelator for the real time computation of $R(T)$ and $\phi(T)$. A pair of complex Doppler signals in eq. (2) is split in two and fed to a conjugate complex multiplier where each of the two signals is input directly; the other two are supplied via a pair of delay lines with a delay time T. The conjugate complex multiplier executes the following computation

$$\gamma(T, t)=z(t)\cdot z^*(t-T) \tag{12}$$

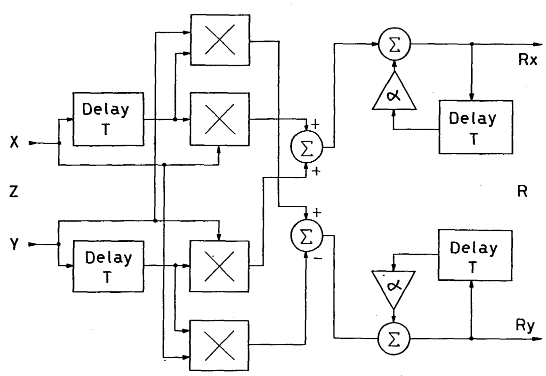

Fig. 1. Complex autocorrelator for calculating $R(T, t)$.

where

$$z^*(t-T)=x(t-T)-jy(t-T) \tag{13}$$

is the conjugate complex of function $z(t)$ that has been delayed by time duration T.

Equation (12) shows that Doppler echo signals from the same depths on a sound beam for two successive pulse transmissions are multiplied. Since sound travels with a speed of c, time lapse t from the pulse transmission corresponds to $l=ct/2$ in depth.

The autocorrelation is obtained by summing the signal $\gamma(T, t)$ at any depth l for several sound transmissions. This kind of summation can be done, for example, using a sweep integrator.[9)] Thus

$$R(T, t)=\sum_n \gamma(T, t)$$

$$=R_x(T, t)+jR_y(T, t) \tag{14}$$

$$|R(T, t)|=\sqrt{R_x^2(T, t)+R_y^2(T, t)} \tag{15}$$

where n is the successive number of sound pulses transmitted in the same direction, and hence nT becomes the integration time for a single-beam direction.

The phase $\phi(T, t)$ is obtained as the argument of $R(T, t)$.

$$\phi(T, t)=\tan^{-1}\frac{R_y(T, t)}{R_x(T, t)} \tag{16}$$

In addition, the autocorrelation value for $\tau=0$ is easily derived by the following equation:

$$R(0, t)=\sum_n z(t)\cdot z^*(t)$$

$$=\sum_n \{x^2(t)+y^2(t)\} \tag{17}$$

Thus the blood flow velocity is obtained using eqs. (4), (10), and (16) and the variance with eqs. (11), (15) and (17).

In an actual two-dimensional blood flow imaging, the number n is determined from the next equation:

$$nTNF=1 \tag{18}$$

where N is the number of raster lines composing one frame image, and F is the frame-rate per second.

§3. System

Described below is the two-dimensional blood flow mapping system that employs the subject autocorrelator.

Figure 2 shows a block diagram of the system, which is equipped with a conventional B-mode imaging unit and a one sample point pulsed-Doppler unit provided with a fast Fourier transform (FFT) spectrum analyzer. Images obtained by these units and flow mapping images are displayed simultaneously and superimposed.

The oscillator (OSC) is a high-frequency oscillator in which the output is divided to provide the clock pulses that trigger various units and the continuous wave that is required for the demodulation of Dopper signals.

Signals received through the transducer (TD) are first amplified by a pre-amplifier and a high-frequency amplifier and then conveyed to a pair of quadrature detectors, where the phase of the mixing reference frequency

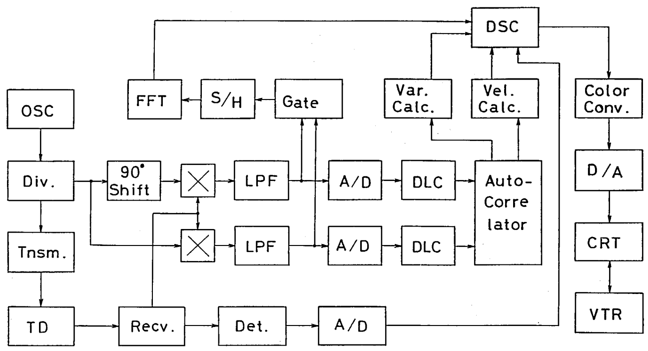

Fig. 2. Block diagram of real-time blood flow mapping system.

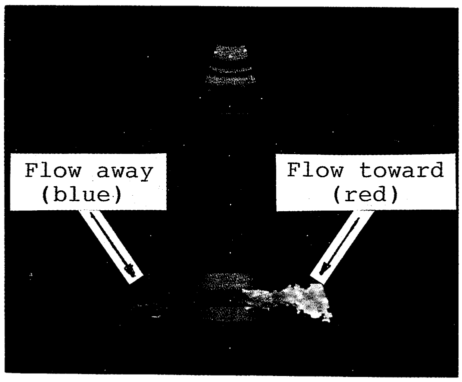

Fig. 3. Color-coded imaging of water flow. Water is flowing from right to left in a vinyl tube.

differ by 90°. Since this reference frequency is made identical to that of the transmitting burst wave, the outputs from the low-pass filter (LPF) become the complex Doppler frequencies that have been shifted by Doppler effects, and the pair of outputs also becomes complex signals with phases that differ by 90°. The pair of signals, after conversion to digital signals by analog-digital (A/D) converters, is passed through delay line cancelers (DLC)[9] and then supplied to the complex autocorrelator described in Fig. 1.

The delay line cancelers are comb filters that eliminate large echo signals from stationary or slowly moving tissues, which respectively have a zero or low Doppler frequency shift. In the autocorrelator, a sweep video integrator[9] is used to integrate correlated signals. The integration time is changed according to the display mode (i.e., B-mode or M-mode). Its output is conveyed to the velocity calculator and the variance calculator that respectively calculate the mean value and the variance of Doppler signals. The outcome is recorded in a digital scan converter (DSC). The DSC additionally records the B-mode or M-mode images that have been obtained with conventional equipment and FFT-analyzed spectra of the blood flow at any sampling point specified.

The color converter serves to convert the data stored in the DSC to chrominance signals. Firstly, with regard to the phase $\phi(T, t)$ that has been obtained with the velocity calculator, the color converter gives red if it is in the first or second quadrant (i.e., $0° < \phi < 180°$). This signifies that the Doppler frequency shift is positive, and therefore the blood flows toward the tranducer. If the phase is in the third or fourth quadrant (i.e., $-180° < \phi < 0°$), the color converter gives blue. This signifies that the Doppler frequency shift is negative, and the blood flows away from the transducer. The faster the blood flows, the brighter the color becomes.

Secondly, with regard to the variance, the larger its value, the greater the green blend ratio. Since variance represents the flow disturbance, the color hue changes according to its extent; that is, the red approaches yellow, and blue approaches cyan.

On the other hand, B-mode image, M-mode image, and FFT-analyzed blood flow data are all converted to black/white as in the conventional way.

The output from the color converter is transformed to analog signals by a D/A converter and is displayed on a color TV screen in real time.

§4. Experiments

Experiments were conducted to examine the performance capabilities of the equipment. The transmitting frequencies of the transducers were 3 and 3.5 MHz.

First, to study the relationship between the flowing liquid and the colors on the TV screen, a vinyl tube was laid horizontally inside a water tank and tap water was flown through it. The ultrasonic beam was scanned in a two-dimensional sector form over the tube. In the experiment, very small bubbles existing in the tap water were considred to act as echo sources.

Figure 3 shows the B-mode color image that was displayed on the TV screen. No coloring is shown in the central part of the vinyl tube where ultrasonic waves are incident almost at right angles with the water flow, which means that Doppler frequency shifts do not occur. However, colors are displayed in the other portions of the tube. Since water is flowing from right to left, the color at the right side is red and that at the left side is blue. Owing to a slight disturbance, there is some mixture of green. It is also recognized that color brightness increases according to the distance from the central part due to increase in Doppler frequency.

§5. Clinical Tests

Shown in this section are some clinical data which have been obtained at a hospital using the system.

Figure 4 shows blood flow mapping in the heart of a healthy subject. At left is the image in diastole where the inflow from the left atrium to the ventricle is displayed in red. At right is shown the image in systole where the outflow from the left ventricle to the aorta is displayed in blue.

Figure 5 is the M-mode flow image in the mitral valve direction of the same subject. Temporal flow variation accompanied by the opening and closing of the mitral valve can be seen clearly.

Figure 6 shows the flow mapping of a patient suffering from mitral stenosis and regurgitation. At left is a diastolic image which reveals a narrow jet spurt of inflow due to mitral stenosis. Since the jet speed is high and the Doppler frequency exceeds the Nyquist frequency, so-called color aliasing is seen where the color of the forward flow turns from red to blue. The right image is that

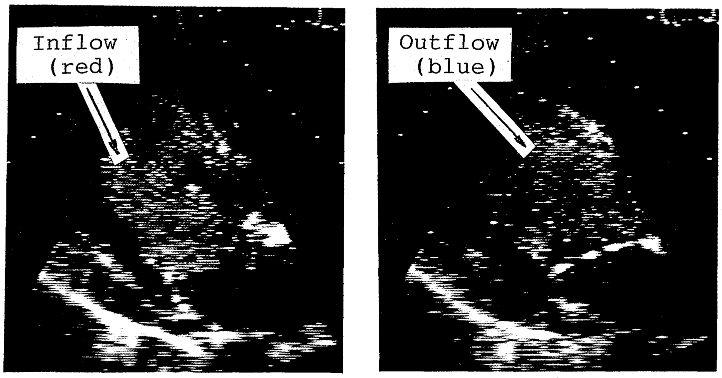

Fig. 4. Color flow images in the heart of a normal subject. Left: Inflow in diastole is shown in red. Right: Outflow in systole is shown in blue.

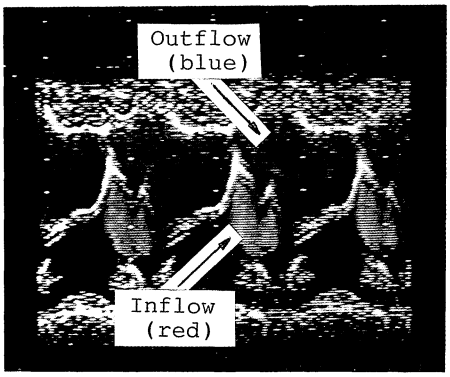

Fig. 5. The M-mode Doppler display of the normal subjet in Fig. 4. The inflow and outflow are shown in red and blue, respectively.

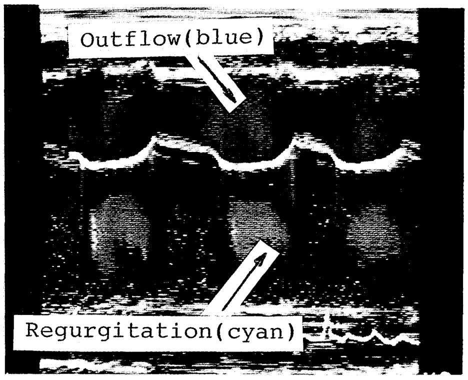

Fig. 7. The M-mode Doppler display of the patient with the valvular disease in Fig. 6.

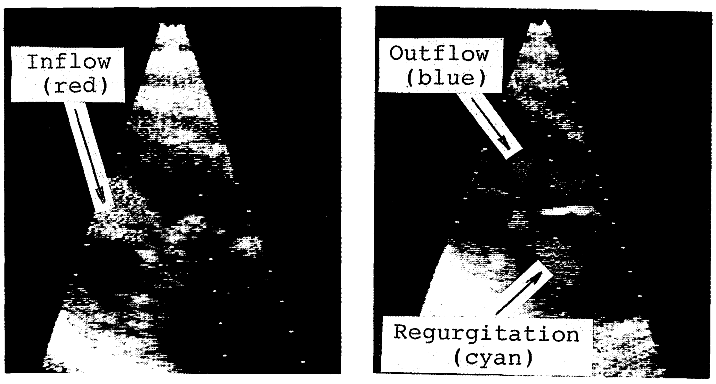

Fig. 6. Color flow images in the heart of a patient with a valvular disease of mitral stenosis and regurgitation. Left: Inflow in diastole is shown. Due to high speed jet, part of the color is aliased. Right: Systolic regurgitation as well as outflow are shown.

of systole, where regurgitant flow from the left ventricle to the atrium is also observed in addition to the outflow from the left ventricle to the aorta. By comparing the colors, it can be seen that the regurgitant flow is more cyanic, suggesting that it may be more disturbed.

§6. Conclusion

The principle and instrumentation for observing two-dimensional blood flow in the heart and blood vessels in real time have been described. This was accomplished by processing Doppler signals using an autocorrelation technique.

An experiment using a simulator has confirmed that the system employing the autocorrelator gives correct results.

Clinical results obtained in a hospital have verified the feasibility of the equipment, since valvular diseases, septum defects and other diseases have been diagnosed quite eqsily and noninvasively from real-time, two-dimensional information of blood flow.

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