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Doppler technology

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Introduction

The use of ultrasonic waves for studying the human circulation was initiated in the late 1950s and early 1960s by Satomura (1957), Franklin et al. (1961) and Pourcelot (Georges et al., 1965). These pioneering works were based on the Doppler effect and led to the development of the first continuous-wave Doppler velocimeters. These systems were non-directional and used perivascular probes during acute and chronic surgical procedures. For the transcutaneous approach it was necessary to develop directional systems, which were first designed by MacLeod (1967) and Pourcelot (1969) and were based on quadrature phase detection of complex Doppler signals.

The first pulsed Doppler systems were developed almost simultaneously by Baker (1970), Peronneau and Leger (1969) and Wells (1969). These new velocimeters had the major advantage over CW Doppler to be range sensitive, but they were difficult to apply in routine without the support of an imaging technique (Duplex B-Doppler systems).

The first trials for real-time imaging the blood circulation by the Doppler technique (now called colour-coded Doppler), were successful in 1977 and the first commercial machine was on the market in 1984 (Aloka). Since these dates, numerous improvements have been made for the simultaneous display of structures and blood flow: colour velocity imaging, power Doppler imaging, Doppler tissue imaging, non-linear imaging (harmonic imaging, pulse inversion imaging), 3D colour-coded display of vascular structures, etc.

Continuous-wave Doppler

By insonating tissues where blood is flowing, and then studying the received signal, a non-invasive measurement of the blood velocity can be made.

Moving blood cells Doppler shift the frequency of the ultrasonic wave scattered by these cells. The simplest way to exploit the Doppler shift is to transmit a continuous sinusoidal wave and then to compare the received signal with the transmitted signal to detect the change in frequency. A system based on this principle is shown in Fig. 2.1. Two transducer crystals are used for transmitting and receiving the ultrasonic waves. Usually these ceramics are housed in a probe similar to a pencil, easy to use for clinical applications.

For transcutaneous examination, it is necessary to ensure that the ultrasound beams overlap over a relatively large distance in order to avoid selective range depth of flow detection.

One crystal continuously emits a sinusoidal ultrasound wave at a frequency f_0. A second crystal receives the backscattered waves, and the corresponding signal is multiplied by a signal of frequency f_0 to detect the Doppler shift f_d. The result is a signal containing frequency components equal to the sum and difference of the frequencies contained in the emitted and received signals. Band-pass filtering of the result is used for removing the frequency signal near $2f_0$ and also the DC component from stationary tissues (see below). The resulting signal contains the Doppler information and then enables the velocity of the moving reflectors to be determined. As the blood moves at different velocities dependent on the position in the vessel, the received signal consists of a continuum of frequencies (frequency spectrum corresponding to the velocity distribution across the area of the vessel).

Forward and reverse flow signals

For separating the forward and reverse flow signals, the received signal is mixed with two signals in quadrature: cos $(2\pi f_0 t)$ and sin $(2\pi f_0 t)$ in order to obtain the real and imaginary parts of the Doppler signal. By shifting the real part

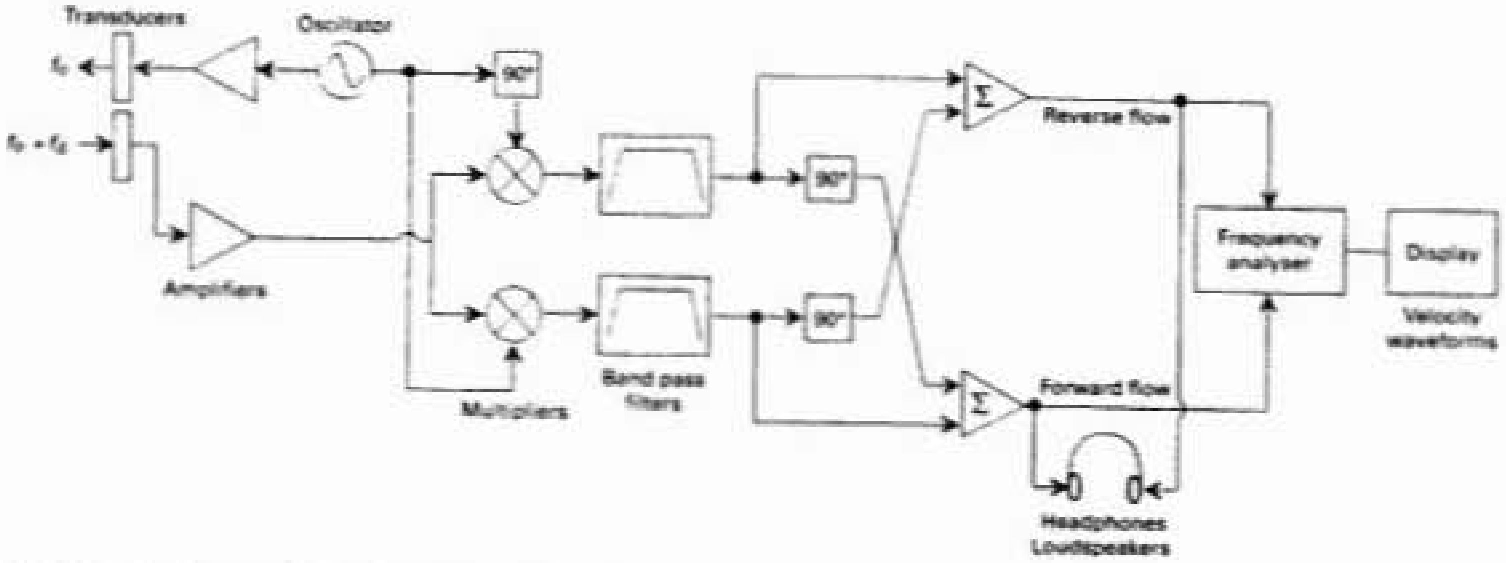

Fig. 2.1. Schematic diagram of a continuous-wave Doppler system.

90°, and adding that to the imaginary part, it is possible to obtain the forward flow signal (positive Doppler shift). This part of Doppler systems is relatively critical as the 90° phase shift must be very precise over a large range of audio frequencies. Some devices use an analogue circuit in which the input Doppler signal is shifted by −45° and +45°. Another approach can be made using a digital Hilbert transformation after sampling the Doppler signal. The reverse flow signal (negative Doppler shift) is constructed in a similar fashion by phase shifting the imaginary part −90°, and adding it to the real part. We can therefore obtain two channels, each representing reverse and forward flow, which can be fed to stereo loudspeakers or headphones for directional information. For the display of the velocity information, several signal processings have been proposed. The most useful ones were first based on the evaluation of the 'mean' velocity by zero-crossing detection of the Doppler signals. A major improvement has been introduced by the use of the frequency analysis of Doppler signals by fast Fourier transformation.

Analogue velocity waveform

In the first commercial Doppler systems in which the velocity waveform was recorded on a strip chart recorder, the Doppler frequency was determined by counting the zero-crossing rate of the signal. This method is relatively simple and gives an estimate of the dominant frequency when the velocity profile is relatively parabolic, or when the frequency spectrum is essentially monochromatic. For more complex flow conditions there is an overestimation of the mean frequency, which ranges between 5 and 15%. A considerable bias is seen for poor signal-to-noise ratios.

Spectral display

The frequency content of the Doppler signal reflects the velocity distribution of the blood cells across the diameter of the vessel. Through display of the frequency spectrum of the Doppler signal, it is possible to have immediate information concerning this velocity distribution and its evolution over successive cardiac cycles. The frequency analysis can be made by Fourier transforming the Doppler signal: this signal is divided into successive segments, and the density spectrum is calculated for each segment (using discrete or fast Fourier transformation). The result is displayed on a screen with frequencies on the y-axis, time on the x-axis, and amplitude of the spectrum on the z-axis (intensity). The amplitude of the spectrum is grossly proportional to the number of blood cells moving at a particular velocity. In the case of parabolic velocity profile, the grey scale is homogeneous over the entire spectrum as the number of red cells is the same for all particular velocities. The major advantages of the spectral display are the following:

- the positive and negative velocities are displayed simultaneously;
- the spectral display gives immediate information on the velocity distribution throughout the ultrasonic Doppler sample volume;
- the flow profile is easy to appreciate on the display, when the sample volume is large enough to cover the entire area of the vessel;
- it is possible to determine from the Doppler spectrum the maximum, minimum and mean velocities, these parameters being necessary for flow indices and flow volume calculation.

The combination of a continuous-wave Doppler with a spectral analyser is advantageous for the measurement of

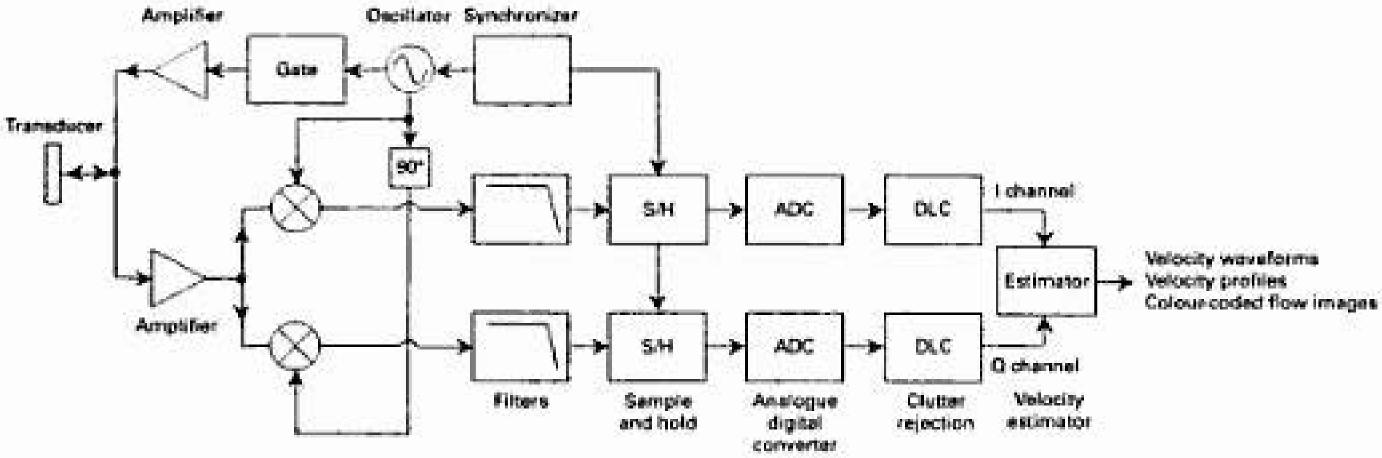

Fig. 2.2. Schematic diagram of a pulsed wave Doppler system with analogue quadrature sampling (DLC: delay line canceller).

high velocities (arterial stenoses, valvular disease), and for the rapid study of the circulation in superficial vessels using an easy-to-handle probe.

Pulsed wave (PW) Doppler

A major drawback of the continuous wave Doppler system is its inability to select a range at which the Doppler measurement can be made. To solve this problem, Doppler devices using ultrasound pulses for sampling the 'Doppler signal' at selected depths have been developed. A number of pulses is transmitted through the tissues and the backscattered signal received is sampled at the same depth (or the same delay after the pulse emission).

A schematic diagram of a pulsed wave system is shown in Fig. 2.2. Electric bursts of a few periods are sent to the transducer through a synchronized gate and an amplifier at a pulse repetition frequency (PRF) determined by the depth of exploration. The backscattered signal received by the same transducer is amplified and multiplied by the reference signal of the oscillator (centre frequency f_0). After low pass filtering which eliminates the sum frequency, the demodulated signal is sent to a sample and hold device and to an analogue-to-digital converter. For each pulse emitted and received, one sample is acquired at a selected delay Δt after pulse emission, so the depth of the sample volume in tissue is $D = \Delta t.c / 2$ (with c the velocity of the ultrasonic wave). The final received signal due to a number of pulse emissions corresponds to the sampling of discrete phase values between the reference signal and the backscattered signal. If the reflector is stationary, the sampled value will be constant. A moving scatterer will change its location and consequently change the phase value. The resulting signal has the same features as the CW Doppler signal. However, due to the sampling of the Doppler signal at the PRF, specific difficulties can arise when measuring high Doppler frequencies, e.g. high velocities.

The detection of blood flow velocity in pulsed Doppler systems can be based on several demodulation techniques :

- **analogue quadrature sampling** (as already described in the CW Doppler section), which has the drawback of hardware complexity and of balancing difficulty in the gain and phase between 'in' and 'quadrature' phase channels (I and Q channels)
- **RF quadrature sampling** (Fig. 2.3), in which the RF signal is directly sampled for obtaining the in-phase and quadrature Doppler signals (I and Q signals). In practice, two samples of the RF signal are obtained at the PRF rate at the preselected depth D, with a delay between the sampling times of $(2k+1).\ T_0/4$ (T_0 : period of the ultrasonic signal of frequency f_0, k : natural number). The principle is similar to that of the conventional pulsed Doppler system, except that two samples instead of one are taken at times $2D/c$ and $2D/c + (2k+1)\ T_0/4$ after each pulse transmission.

If the reflected signal is :

$R(t) = A(t)\ \cos\omega_0 t + U(t)\ \cos(\omega_0 + \omega_u t) + L(t)\ \cos\ (\omega_0 + \omega_l t)$,

the sampled signals are :

I signal $= U(t) \cos \omega_u t + L(t) \cos \omega_l t$

Q signal $= -U(t) \sin \omega_u t + L(t) \sin \omega_l t$ for $k = 0, 2, 4, \ldots$

$U(t) \sin \omega_u t - L(t) \sin \omega_l t$ for $k = 1, 3, 5, \ldots$

$U(t)$: signal of target moving towards the transducer.

$L(t)$: signal of target moving away from the transducer.

As in conventional pulsed Doppler systems, the direction of flow and the Doppler frequency estimation can be made by zero-crossing detection or by the more sophisticated autocorrelation technique.

This last technique has the advantage of being easily

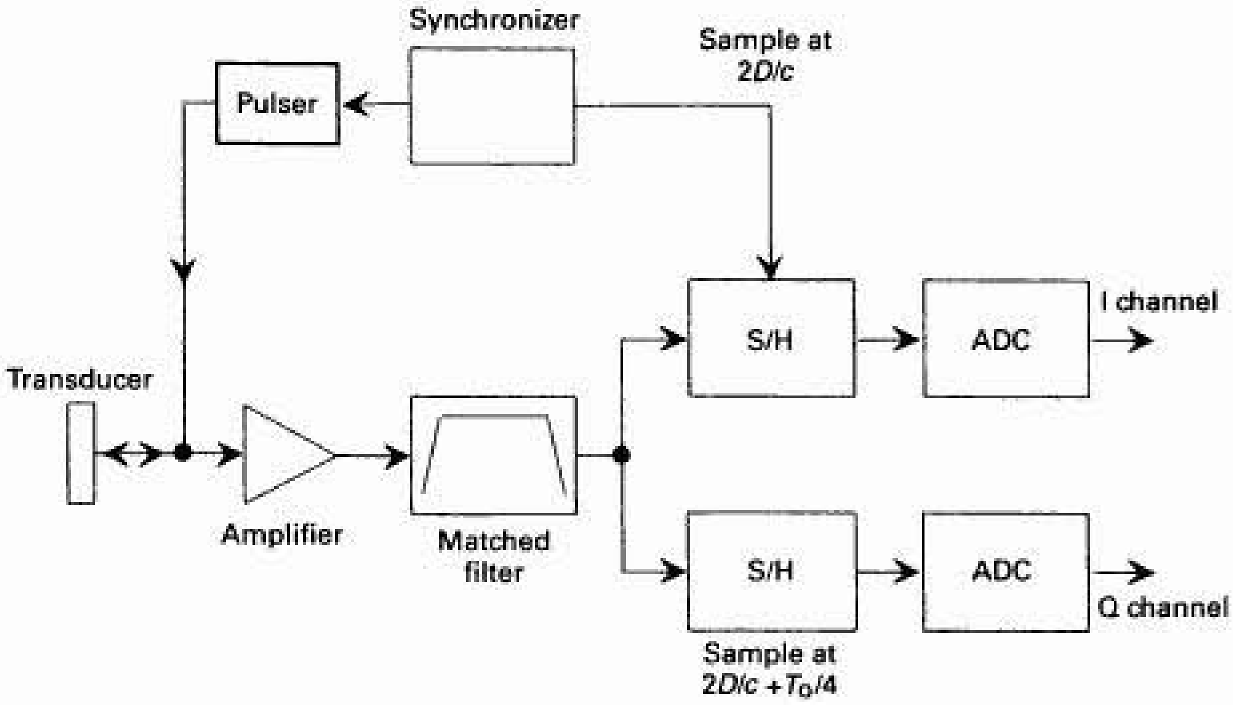

Fig. 2.3. Pulsed wave Doppler system with quadrature sampling of radio-frequency (RF) signal.

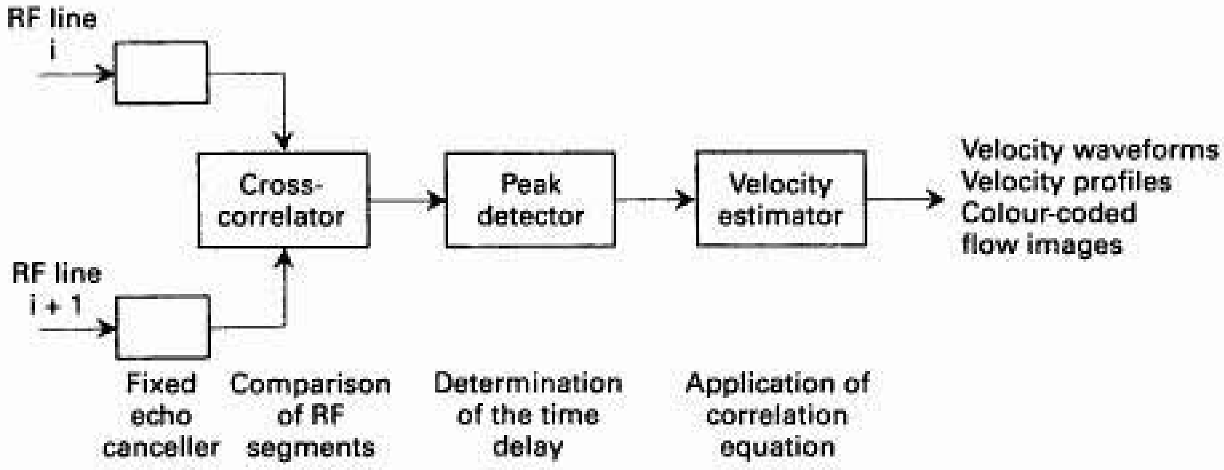

Fig. 2.4. Velocity estimation by time-domain cross-correlation technique.

implementable on multigate pulsed Doppler systems for real-time velocity profilometry and colour coded Doppler imaging, and also of reducing the complexity of the reception stages of pulsed Doppler systems.

- **time domain cross-correlation** (Fig. 2.4) in which we measure the velocity of the moving target by comparing the time delay between successive received RF signals. In theory this method does not suffer from aliasing problems and gives precise measurements. A number of echoes along an A-line are acquired and stored. Assuming that a given group of scatterers have a unique signature in ultrasonic echoes, the motion information can be extracted by tracking this signature at the pulse repetition rate T_{PRF} (Fig. 2.5). In practice, it is necessary to detect the time shift τ_0 between successive windowed section of two RF lines. By detecting the maximum of cross-correlation function, it is possible to measure τ_0 and also to detect the flow direction (Fig. 2.6). An interesting feature of the time domain cross-correlation technique is that the flow measurement is not affected by frequency-dependent attenuation of the ultrasonic pulses as we compare two signals segments which have both passed through the same intervening tissues. The relationship between velocity v and time delay τ_0 is given by the following formula : $v = c.\tau_0 \,/\, 2.\cos\theta.T_{PRF}$.

Stationary echo cancelling

Boundaries of the vessel and surrounding tissues produce high energy echoes which are superimposed on the flow signal and can be 10 to 1000 times larger than the Doppler signal. Current systems use high pass filters for removing the stationary structures echoes, but this leads to a poor

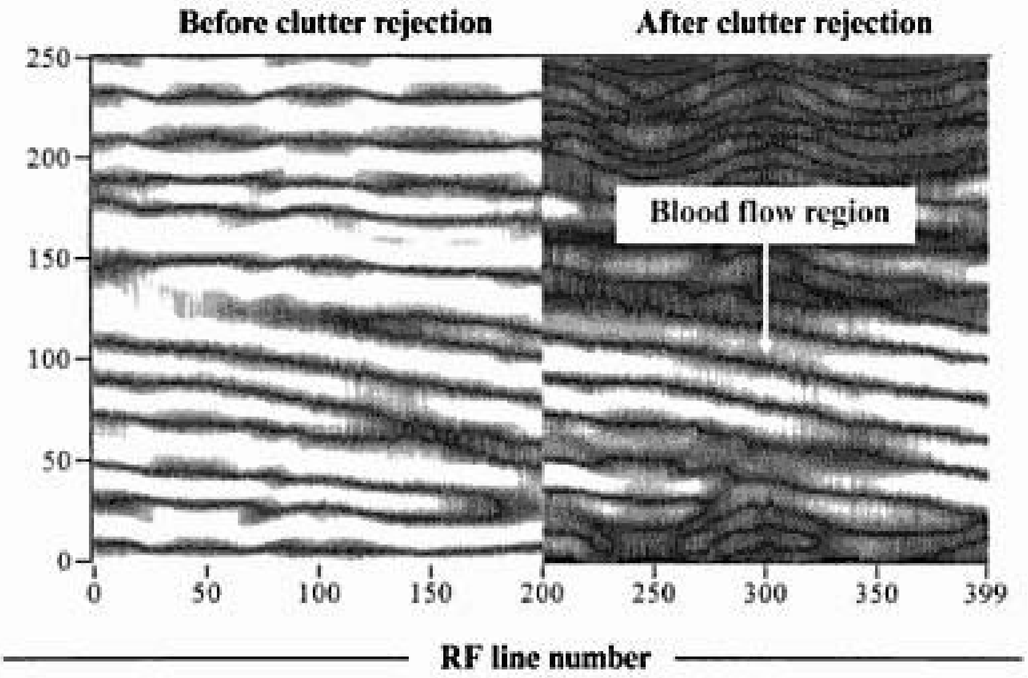

Fig. 2.5. M-mode RF image obtained by displaying all the RF lines side by side (axial position along the vertical axis and time along the horizontal axis). The data come from an erysipelas located on a human leg. In the right side of the image, a 180 μm diameter vessel appears after the fixed echo cancellation procedure. In the raw image (on the left side) this vessel is undetectable and no blood flow estimation can be made.

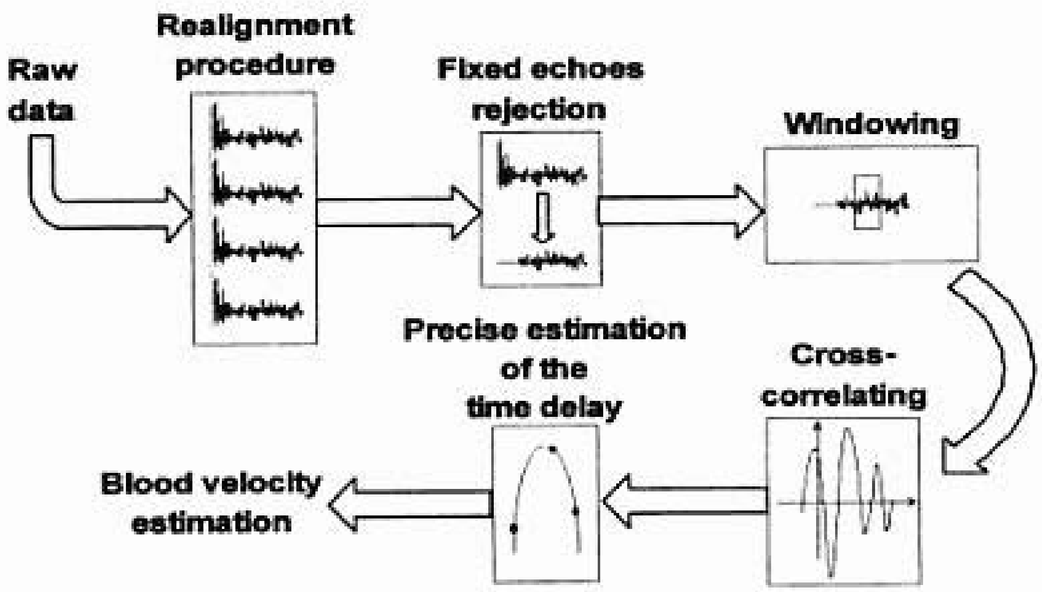

Fig. 2.6. Blood velocity estimation by cross-correlation method. The successive sequences include realignment of RF signatures, and precise determination of time delay τ_e.

detection of slow flow velocities. The removal of these echoes can also be done by filtering the acquired RF lines. A simple method for that filtering is to subtract two consecutive lines (this is a moving average numerical high pass filter). However, the consequence of the filtration of the flow signal is a reduction in amplitude and a distortion of the pulse spectrum that depends on velocity. In order to reduce that problem, it is better to subtract the acquired RF line from an average of preceding lines (Fig. 2.7).

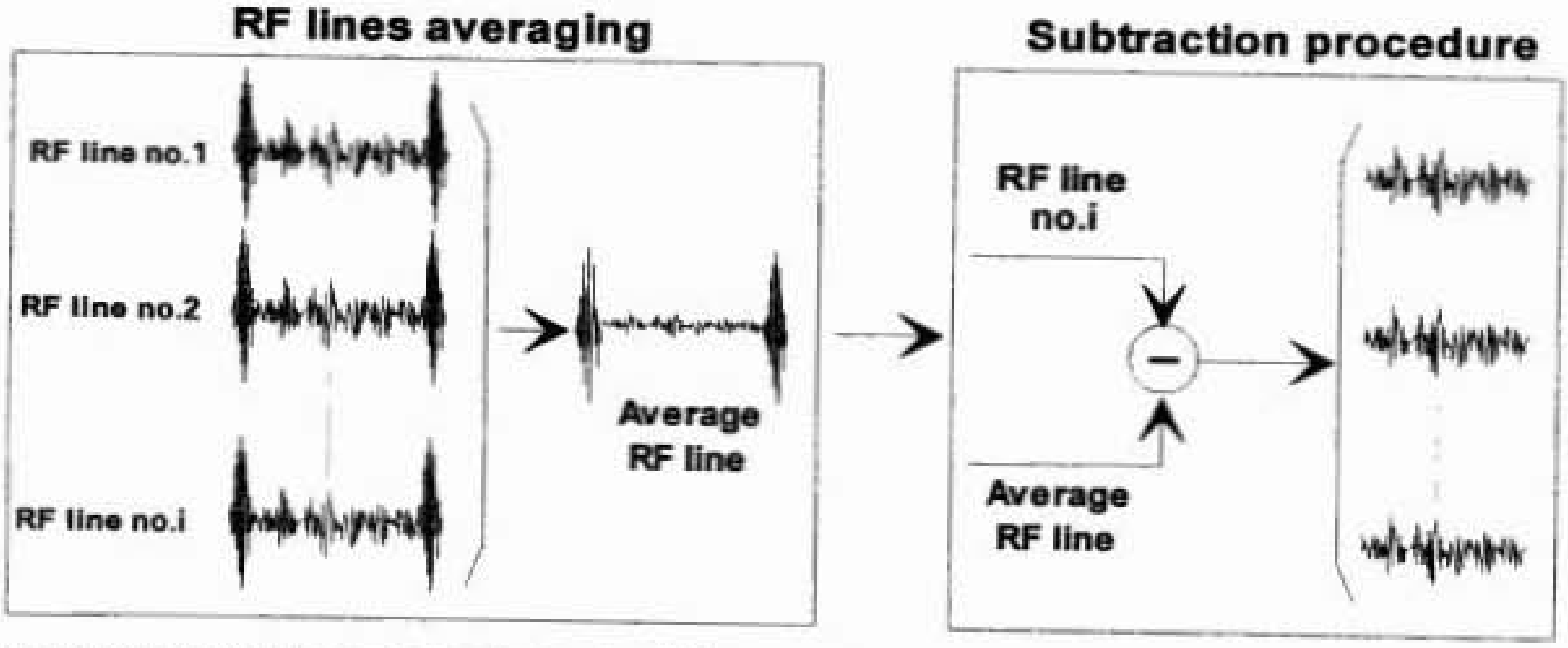

Fig. 2.7. Principle of stationary echo cancelling based on RF lines averaging and subtraction procedure.

Velocity range

The maximum detectable velocity v_{max} in a conventional pulsed Doppler system is due to the well-known range-velocity limitation given by the equation $v_{max} = c/4.f_0.T_{PRF}$.

This means, in practice, that a range of velocities over $2v_{max}$ can be estimated, and that interval can be placed where it gives the most useful information (baseline shifting).

It is possible to overcome this limitation by doubling the PRF, thereby doubling the v_{max}. In this case, a stationary component is superimposed on the received signal and can be removed by a conventional method. This technique is only applicable if no vessel is located closer to the transducer which could give a signal in the 'false' gate. In some cases it is also possible to solve the problem of aliasing by using a lower ultrasonic frequency f_0, or by increasing the angle between the flow vector and the ultrasonic beam.

In modern devices, using the cross-correlation technique and time domain processing of RF signals, higher velocities can be estimated without aliasing. As this method requires short ultrasonic pulses, it is very compatible with B-mode imaging, and has therefore become more and more developed for colour-coded Doppler systems.

Duplex B-mode Doppler and colour flow mapping

The pulsed Doppler system is a relatively blind system which needs the support of an image for the location of sample(s) volume(s) into vessels or heart chambers. It is quite easy to generate a Duplex B-mode Doppler device based on electronic linear or phased arrays transducers as common electronic circuits can be shared especially at the level of the pulsing and steering systems. The time for data acquisition is shared between the B-mode image and the sonogram acquisition sequences. For low velocity measurements or high PRFs it is possible to alternate the B-mode line and Doppler line acquisitions in order to have both informations in parallel. In most of the cases it is preferable to alternate B-mode images and Doppler acquisition sequences. After location of the sample volume in the area of interest, the B-mode image is frozen and the Doppler acquisition sequence is started.

The most advanced mean for studying the circulatory system is the display of a full map of both structures and blood flow in real time. Colour flow mapping (CFM) systems represent a major step toward this goal. Basically, in CFM, the blood flow velocity detection is based on a multigated system, whose lines of detection are regularly changed in position in order to form an image of flow velocities. The velocity image is superimposed on a B-mode image: the direction of the flow is coded in colour and the amplitude of the velocity is coded as colour intensity. Typically, a red colour is used for the display of flow toward the transducer and a blue colour for flow away from the transducer.

A colour flow mapping system is quite similar to the pulsed wave Doppler system. A whole measured line is divided into range gates and velocity estimation is made for each range, based on a limited number of RF lines. Then the measured line is moved and the whole process will apply again. After the complete scanning of the area of interest, a colour image of velocity is displayed.

In CFM systems based on phase shift estimation, the received signal from the transducer is multiplied by the emitted frequency, low pass filtered and then converted to a digital signal. The stationary echoes are removed by a delay line canceller numerical filter. The velocity of blood flow can be determined from the derivative of the phase of the received signal. In most cases estimation of the blood velocity is based on autocorrelation. It is also possible to detect the mean Doppler frequency from the power density spectrum, but it is too time consuming for application in CFM.

In CVI systems using time shift estimation, the transmitter emits a short duration pulse, and a time gain control amplifier adjusts the level of the received signal. A matched filter is used for noise reduction, and then the RF sampling is done, and stationary echo cancelling is performed by subtracting consecutive lines. Finally, the velocity is determined by cross-correlation estimation. The advantage of time shift estimation is that it needs few successive RF lines (in theory only two) for velocity measurement, and that the same RF signal can be used for both B-mode and Doppler images.

New development

Power Doppler imaging (DPI)

A more recent approach to velocity mapping was based on the display of the energy backscattered by moving structures, instead of amplitude and direction of the velocity vector. Two different systems have been developed:

- Power Doppler imaging (DPI) of blood flow velocity. The signal displayed is a function of the number of red cells in the sample volume. It has the major advantages of being unsensitive to aliasing and to the direction of flow (flow perpendicular to the ultrasonic beam can be detected), and also to show more sensitivity for low velocities than conventional colour coded Doppler systems. The principle is to determine the energy of the Doppler signal backscattered by the red cells by using a high pass filter, which eliminates echoes from stationary structures. The energy is therefore almost independent of flow velocity and depends only on attenuation into the tissues. The disadvantage of PDI is that it does not give information on flow direction. A combination of colour coded Doppler and power Doppler imaging has been recently developed (convergent Doppler) in order to draw advantages of both techniques (sensitivity, angular independance, flow direction).
- Doppler tissue imaging in which only low velocities corresponding with low frequency and high energy Doppler signals are detected by low pass filtering. This technique is useful in cardiology for the study of the dynamics of cardiac structures.

Parallel signal processing

A major limitation of the colour flow mapping (CFM) system is its low frame rate which can lead to a stroboscopic effect and to severe limitations for the display of transient flow velocity changes. Development of parallel signal processing of adjacent lines of exploration is a way of increasing the frame rate and for improving the signal to noise ratio. Another approach is the simultaneous acquisition of data from several transmitting–receiving zones of the transducer array. This technique is more difficult to apply due to the possible cross-talk between the reception areas and false detection by lateral lobes.

Angle independent velocity estimation

It is possible to determine amplitude and orientation of a velocity vector by combining the information obtained by several ultrasonic beams. Each beam determines the projection of the velocity vector along the axis of exploration. With two beams it is therefore possible to determine the resulting velocity vector in a plane if we know the angle between the beams. This technique is implementable on phased linear arrays.

Speckle tracking

Two dimensional flow estimation could be based, in theory, on the speckle tracking in successive B-mode images. A small region of the first image is correlated throughout a larger region of the following image. The displacement of the speckle between the frames is determined by the position of the peak in the covariance estimate or approximate function (absolute sum of the pixel difference). The velocity is then determined, knowing the time between the images. This method relies on the fact that speckle patterns are stable for modest translations, this result being applicable to a large variety of scatterers including red cells. However, it needs a high frame rate and a large amount of calculation.

Harmonic imaging

With current CFM it is difficult to visualize small vessels and low flow velocities close to vessel walls. A high pass filter used for echo cancelling severely attenuates low velocity signals. One solution to solve this problem is the use of ultrasonic contrast agents (gas bubbles) which improve the signal-to-clutter ratio and induce non-linear scattering. The re-emission by the gas bubbles contained in the blood of a relatively strong second harmonic signal is at the origin of the harmonic imaging systems. By using a wide-band transducer it is possible to transmit pulses at a centre frequency f_o and to receive signals centred on f_o and $2f_o$. A first harmonic removal filter is then used to eliminate a large amount of echoes arising from the vessel walls and surrounding tissues (frequency f_o) and to increase the ratio between velocity signal and clutter signal dramatically.

Conclusions

The Doppler technique has been in constant evolution since its early beginning, more than 40 years ago. It is now more and more integrated into the modern devices, thanks to the use of the same transducers for transmitting and receiving ultrasonic signals, which are processed for tissue and flow imaging and for the display of velocity waveforms and velocity profiles. Further developments are expected in the fields of absolute quantification of velocity vector, flow volume measurement, 3D display of vascular structures, measurement of tissue perfusion and study of microcirculation.

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