Jørgen Arendt Jensen. Maj.

A NEW VELOCITY ESTIMATOR FOR COLOR FLOW MAPPING

O. Bonnefous, P. Pesqué, X. Bernard

*LEP, 3, avenue Descartes, 94451 Limeil-Brévannes Cedex, France

Abstract

Recently, real-time blood flow imaging became possible thanks to the development of a velocity estimator based on the phase-shift measurement of the successive echoes obtained from moving blood [1]. However, this method suffers from the well known limitations of pulse-Doppler instruments. These drawbacks are mainly the conflict between velocity precision and axial resolution requirements, and the velocity limitation due to the aliasing phenomenon.
In this paper, a new principle for velocity estimation, based on time domain correlation [2], is shown to overcome these problems. A simplified version of the technique using only 1 bit signals has been derived. This processing scheme has been optimized through extensive computer simulations. Experiments carried out on recorded in-vivo data, have confirmed the feasibility of measuring velocity profiles accurately, without the high velocity limitation (aliasing). Moreover, the possibility of obtain instantaneous and accurate velocity profiles by processing a limited number of echo response (about 10 A-lines) should provide new potential modalities for clinical application of ultrasound.

I. Introduction

Successive developments of pulse-Doppler ultrasound techniques involved the measurement of instantaneous blood velocity profiles on the beam axis. The first approaches used multigate pulse-Doppler systems with sequential [3] or parallel processing [4] and the MTI (moving target indicator) method [5-7]. Recently, a multigate velocity profile reconstruction in real time, using analog ultrafast spectral analysis has been presented [8]. However, these techniques did not manage to produce two-dimensional imaging of the actual flow dynamics in real time.

Two-dimensional flow mapping in real time requires very efficient and robust processing. Indeed, the velocity profile estimation on the full beam axis has to be performed at the pulse repetition frequency (PRF equals a few kHz) and in comparison with conventional pulse-Doppler systems, the observation time (integration) is reduced to only a few repetitions (about 10) in order to satisfy real time frame-rate requirements. Despite these formidable problems, a first system has been developed which does produce two-dimensional Doppler flow maps [2] using an autocorrelation technique.

*LEP : Laboratoires d'Electronique et de Physique appliquée. A member of the Philips Research Organization.

In the latter system, the phase shift of successive echoes from the moving scatterers, in response to recurrent excitation, is detected according to the usual description of the Doppler effect. Consequently, the well-known limitations of existing pulse-Doppler processing arose. These are mainly :

- the maximum axial velocity which can unambiguously be measured :

$$V_{max} = \frac{c}{4} \frac{PRF}{f_o}$$

  where f_o is the central frequency of the transmitted signal and c is the velocity of sound.
- the increase of the variance of the velocity estimation if one use broadband excitation to improve range resolution.

As an alternative, we have presented in a previous paper [9] a new time-domain formulation which describes the pulse-Doppler effect on successive echoes from the moving targets as a progressive time shift due to displacement of the scatterers. This approach allows one to conveniently realize a powerful computer simulation tool. Various sets of rf A-lines reflected from a cloud of scatterers moving inside a vessel can thus be created, taking into account a velocity profile, the transducer design (directivity profile and bandwidth) as well as practical signal-to-noise ratios. Finally, strong stationary echoes representing, for instance, signals from the vessel walls, can be superimposed on the set of "moving" A-lines at a specified ratio. This modelling enables to take into account the random interference effects which characterize the echo signal from blood [10, 11]. Specifically, the simulated data allow an accurate and realistic evaluation of the different velocity estimators, in particular those suitable for real-time flow visualization.

As a consequence of the time-domain interpretation of pulse-Doppler, a novel class of blood velocity estimators are derived which are able to measure the time shift of the consecutive echoes. Among them, the cross-correlation function is a well-known technique for time-delay estimation used, for instance, in radar [12] and slurry flow measurements within pipes [13]. However, our application of time domain correlation is fundamentally different from the techniques developed in these areas or in previous works about random signal Doppler systems [14, 15]. The cross-correlation function is shown to provide an accurate and efficient blood velocity estimator. Moreover, it allows one to overcome the aliasing problem which appeared to be insoluble up to now [16]. This apparently "magic" feature can be easily understood if one realizes that there is no

0090-5607/86/0000-0855 $1.00 © 1986 IEEE

ambiguity in measuring a time shift instead of a phase shift. On another hand, it will be shown in this paper that cross-correlation of 1 bit-signals after stationary echo cancelling, provides a robust and performant velocity estimator.

The ability to overcome the classical velocity limitation of pulse Doppler systems, opens up new possibilities for clinical applications. Furthermore, the use of wideband transducers with this technique improves not only the axial resolution, but also the precision of the velocity measurement. These unique features indicate that instantaneous and accurate velocity profile measurements are possible.

II. Time domain formulation

Considering a recurrent excitation applied to a fixed transducer it is received successive echoes from the moving medium. Under some approximations and considering a pure, uniform axial velocity V_z of blood it can be shown [9] that the succeeding response are time-shifted according to

$$y_{k+1}(t) = y_k(t-\tau) \qquad (1)$$

where k is the rank of the excitation, y_k is the corresponding echo response, τ is the time-shift related to axial velocity V_z, speed of sound c, and excitation period T, by

$$\tau = \frac{2\,V_z}{c}\,T \qquad (2)$$

In other words, this simple relation indicates the distance a group of scatterers have moved between two excitation pulses.

However, this relation is only a first order approximation of practical situations because blood velocity is neither purely axial [10] or uniform. Thus, this relation has to be limited to a first understanding of velocity estimators and computer flow simulation should be used for more accurate performances evaluation.

III. Computer flow simulation

Computer simulation of blood flow provides an efficient tool to assess the robustness of velocity estimators to practical parameters such as signal to noise ratio, number of A-lines to be processed, quantization accuracy and sampling rate, cross-range velocity... Also the influence of beamwidth and bandwidth of the transducer can be easily evaluated.

The geometry of flow simulation is shown fig. 1. The details of the formulation are given in ref. 9 . Briefly, this model enables to take into account any velocity profile inside a vessel to synthetize a set of echographic A-lines. To match as close as possible to real situation, an A-line from fixed-strong scatterers can be added to the previous set of "moving" A-lines with a given ratio. Typically, a ratio of 40 dB is considered which corresponds to the ratio of mean reflectivity between soft tissues and blood. Finally a sequence of uncorrelated noise lines with a specified bandwidth can be also added. The chaining of the simulation is summarized in the flow chart shown in fig. 2.

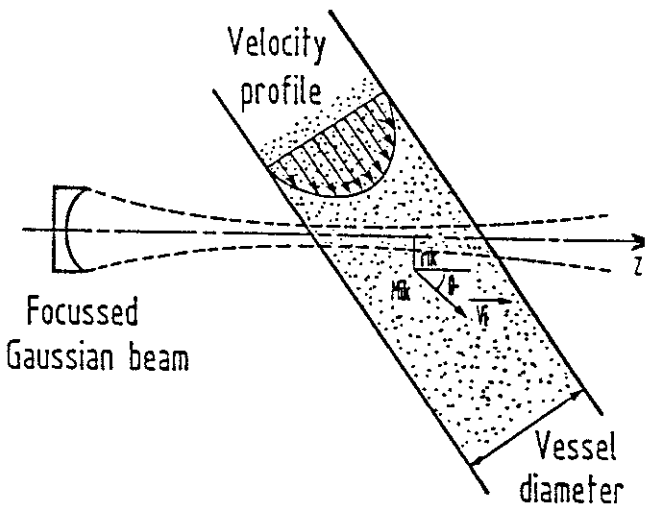

Fig. 1 : Geometry of the computer flow simulation

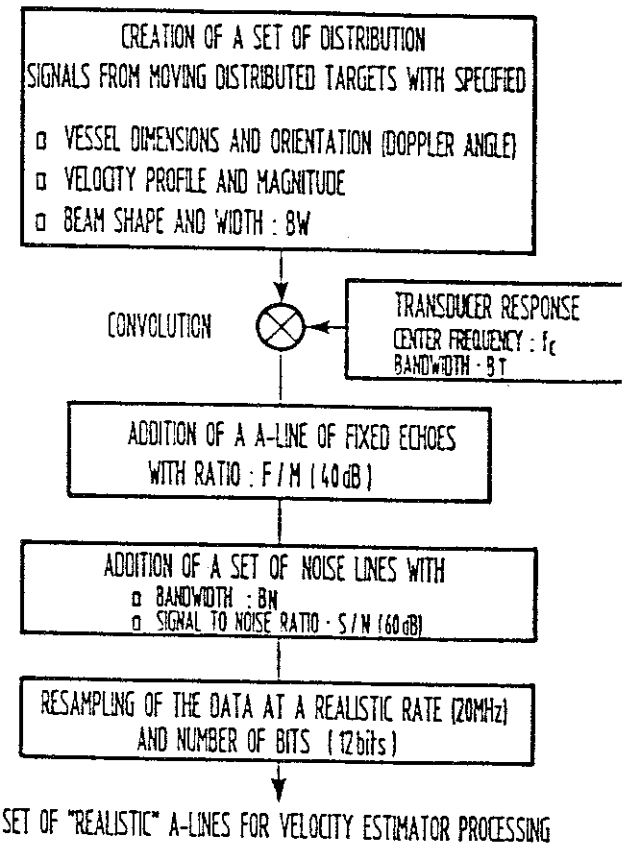

Fig. 2 : Flow chart of computer blood flow simulation

IV. Velocity estimator based on phase-shift or time-shift

IV.1. Phase-shift estimators

The time-shift mentioned in section II can be also interpretated as a phase-shift of a frequency component f_o, given by

$$\Delta\Phi = 2\pi f_o \tau \qquad (3)$$

where f_o is usually the center frequency of the echographic signal.

As a consequence, two main classes of blood velocity estimators suitable for pulse-Doppler applications can be derived : the usual ones measuring phase-shifts and a new one estimating time-shifts.

Main inherent problems in phase-shift measurements are :

- the bias introduced by frequency dependence of tissue attenuation which changes the center frequency of the return signals [17] ;
- the fluctuations of the local mean frequency of the A-line induced by random interferences from multiscatterers medium [18, 19] ;
- the ambiguity of phase measurements which leads to the aliasing effect.

These drawbacks are the well-known limitations of conventional pulse-Doppler instruments and are present on existing color flow mapping systems. Thus quantitative, accurate and instantaneous velocity profile measurements are almost impossible with estimators based on phase-shift.

IV.2. Time-shift estimators

In this section, we demonstrate that the limitations of conventional pulse-Doppler techniques discussed above can be overcome by an estimator working on time-shift measurement. As mentioned above, the basic idea of this processing is that at the first order, the blood flow induces a translation in time of the successive echo signal (Eq. 1). The time correlation technique requires that most of the scatterers stay within the beam between pulses so that the correlation between two consecutive range gated echoes remains high. In practice, this assumption is easily satisfied. As an example, let us consider a flow velocity V of 1 m/s, a pulse repetition frequency of 5 kHz, the displacement D of the cells between echoes is given by :

$$D = VT = 0.2 \text{ mm} \qquad (4)$$

This deplacement is actually relatively small compare to the typical size of the resolution cell of an ultrasonic system. Consequently, the axial flow velocity profiles can be estimated from the measurement of the echoes time-shift at any range along the beam.

As a first-order understanding of the working of the time correlation technique, let us consider Eq. (1) that relates two successive A-lines after stationary echo cancellation. The local cross-correlation between these two signals, defined for a specified depth $Z = ct/2$, is given by

$$C_k(t,u) = \int_t^{t+W} y_k(t')\, y_{k+1}(t'+u)\,dt' \qquad (5)$$

where the time $t = 2Z/C$ locates the range gate, W is the duration of the range gate and u is the shift parameters of cross correlation function. Introducing the results of Eq. (1), we obtain

$$C_k(t,u) = \int_t^{t+W} y_k(t')\, y_k(t'+u-\tau)\,dt' \qquad (6)$$

In this ideal situation (uniform axial velocity), it is clear that the time position of the maximum of the correlation function provides a measurement of the time-shift and therefore of the blood velocity at the corresponding range

$$V_k(t) = \frac{c}{2}\frac{\tau_k(t)}{T} \text{ with } C_k(t,\tau_k(t)) = \text{Max}_u[C_k(t,u)] \qquad (7)$$

A more accurate evaluation of the performances of the time correlation technique has been done using our computer-simulation tool. The results confirm the advantages expected from the principle of this estimator. First of them is the possibility to overcome the usual velocity limitation of pulse Doppler due to the aliasing effect. Indeed, a high range velocity will result in a significant time-shift. There is no ambiguity in measuring it since the correlation function presents only one absolute maximum. This is illustrated in figure 3. The two upper plots of this figure show the range gated echoes from the simulated moving A-lines of a rapid blood flow of 1 m/s. The lower plot presents the cross-correlation function of these range gated echoes and the location of the absolute peak does measure a time-shift greater than one period of the rf signal.

Moreover, there is no longer mutually exclusive requirements in reflectivity and blood flow imaging because the use of wideband transducers in the cross-correlation method improves both the range resolution and the accuracy of the velocity measurement. Indeed, the time-shift method can be understood as a measure of the range displacement of the targets between two excitations. Therefore, good range resolution leads to an accurate velocity estimate.

V. One-bit correlator

Cross-correlation processing is a well-known comparison method between two signals. In the case of time-shifted signals like the A-lines reflected by a moving medium, this comparison can be carried out also on the sign of the signals. This is a 1 bit-correlation technique.

The performances of cross-correlation processing are also valid with the one-bit correlator. These amazing features which are the aliasing suppression and the accuracy of velocity estimation with wideband signals are well depicted on fig. 4 and 5. Indeed, the velocity estimations on the same sets of simulated data with a phase-shift method illustrate the large variance (fig. 4) and the aliasing effect at 47 cm/s (fig. 5).

VI. Experimental results

In order to evaluate further the performances of our new blood velocity estimator we have built a dedicated Digital Acquisition System (D.A.S.) for in-vivo data recording. Velocity estimates were done using a SKY array processor.
This D.A.S. is organized around a VME bus controlled by a 68000 microprocessor running under VERSADOS operating system. The transmit-receive electronics was designed to operate with 5 MHz center frequency transducers. The receiver and the excitation have a fairly large bandwidth of 2.5 MHz, matched to the one of the transducers in order to obtain a good resolution on the estimated velocity profiles. The receiver output signal is digitized

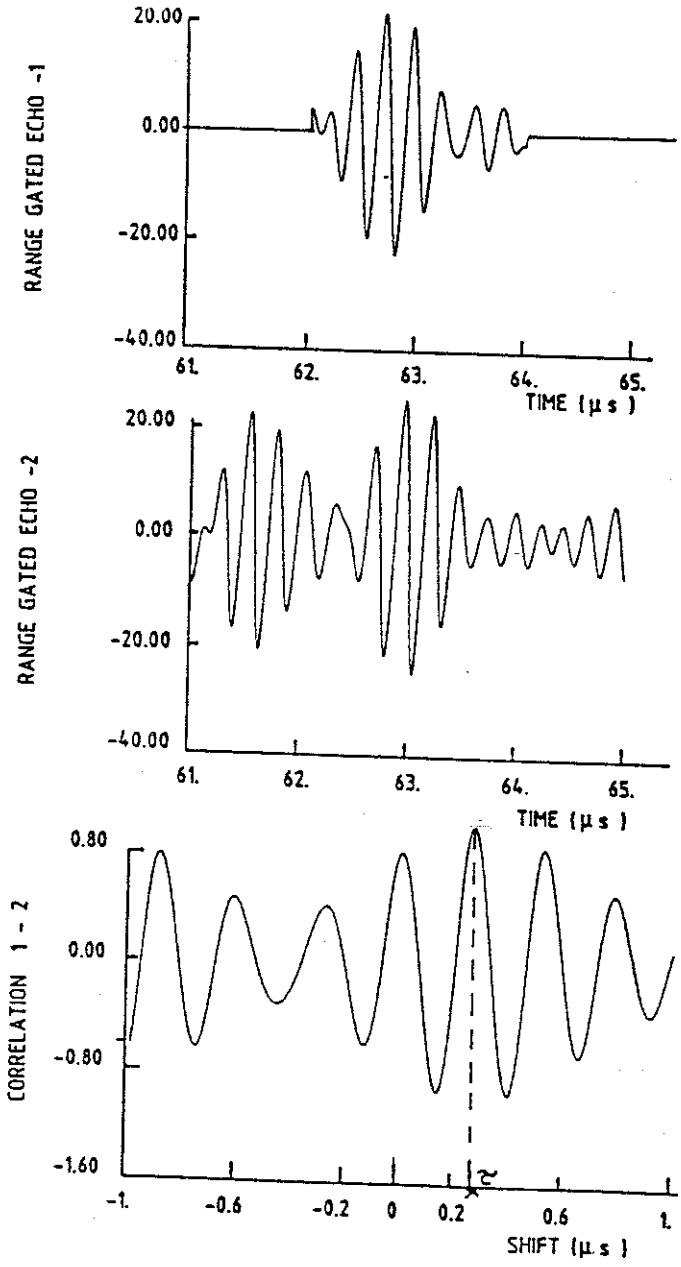

Fig. 3 : Example of cross-correlation function between successive range gated echoes from a fast flow (V_z = 1 m/s)

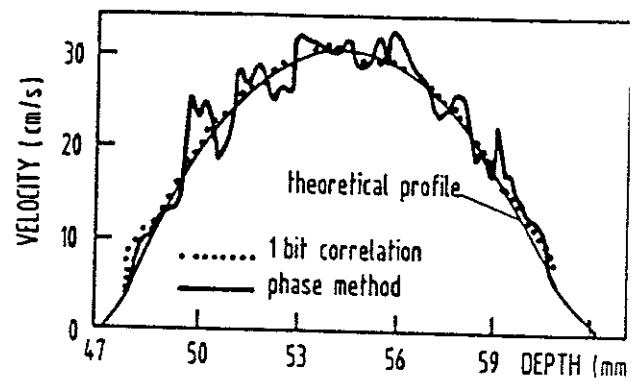

Fig. 4 : Estimated velocity profiles from simulate data. Larger variance of phase-shift methods

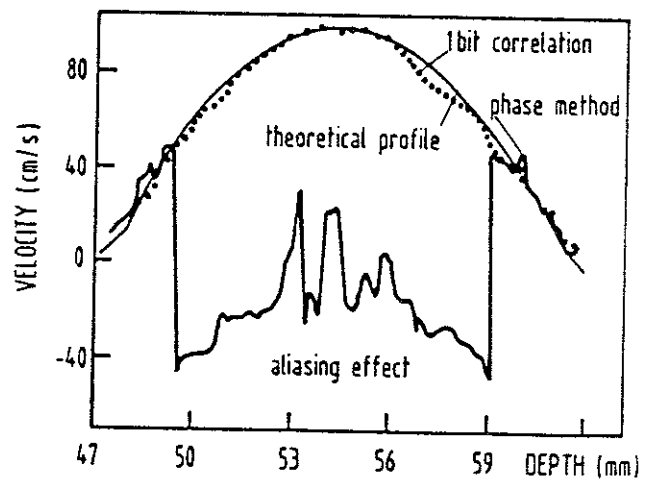

Fig. 5 : Estimated velocity profiles from simulated high velocity data. Aliasing suppression with time-shift methods
(PRF = 5 kHz ; F_c = 4 MHz)

at 20 MHz with 12 bits accuracy. The digitized data are then stored in a buffer memory. The timing logic board controlled the rate of transducer excitations (PRF), the number of excitation pulses, the sampling clock and data recording management. Specifically, this corresponds to acquisition start time and duration, number of consecutive A-lines recorded and dead time between sets of acquisition in order to cover a full cardiac cycle for instance. A block diagram of the D.A.S. is given fig. 6.

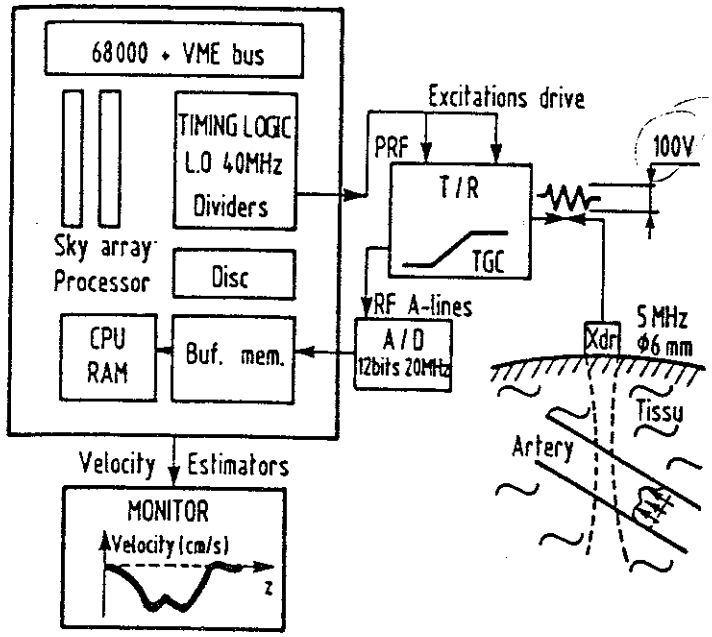

Fig. 6 : Block diagram of the Digital Acquisition System for in-vivo recordings

We have used this system to record sets of A-lines from various vessels in the body. Experiments were carried out with a 5 MHz Aerotech transducer 6 mm in diameter focused at 3 cm. The in-vivo velocity profile estimation confirm very well the advantages expected from our new velocity estimator. As an example, fig. 7 gives the instantaneous velocity profiles in the common carotid obtained with phased-shift based processing and correlation technique. The reduction of the variance on the velocity estimates, even by processing a limited number of successive echoes (here 14), evidence the ability of the correlation processing to provide with instantaneous and quantitative velocity profiles. Moreover, the suppression of the aliasing effect is demonstrated on fig. 8 ! It is shown on this figure the estimated velocity profiles again on the common carotid but during systolic flow. In this case, time-shift based methods are able to measure without ambiguity a peak velocity of 58 cm/s. On the same set of data, phase-shift based processing fails both in measuring the velocity gradient near the vessel walls and the high velocities inside the vessel. In the experiment with a PRF of 5 kHz and a center frequency of 5 MHz the aliasing occurs at 37.5 cm/s.

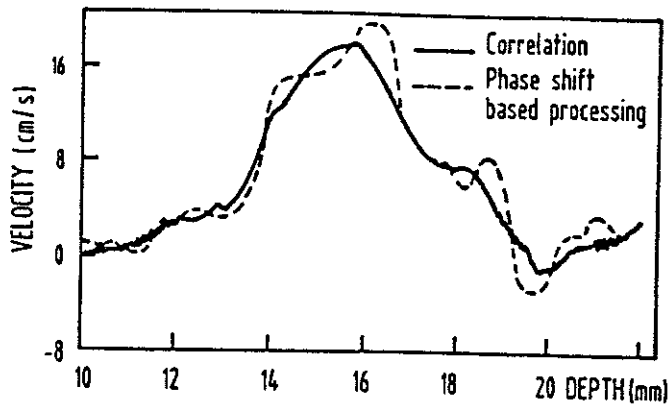

Fig. 7 : Instantaneous velocity profile in the common carotid (PRF = 5 kHz, F_c = 4 MHz, number of successive fixed echo cancelled signals processed NAL = 14). Larger variance with phase-shift method

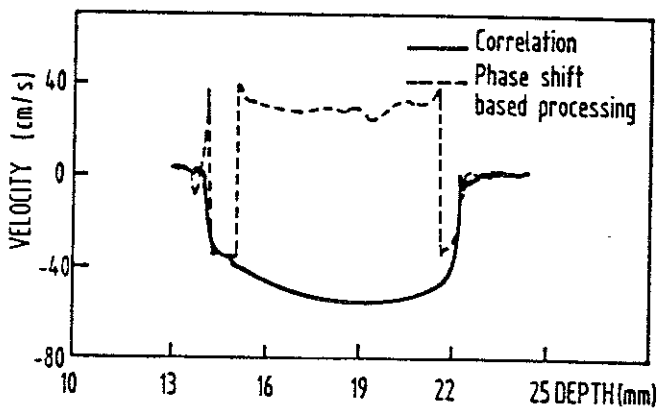

Fig. 8 : Instantaneous velocity profile in the common carotid during systolic flow (PRF = 5 kHz, F_c = 5 MHz, NAL = 14). Aliasing effect with phase-shift method

VII. Conclusions

We have presented in this paper a new blood velocity estimator. This technique has specific features such as the suppression of the aliasing effect and the improvement on the velocity estimation with broadband signals. These advantages have been confirmed both on simulated and in-vivo data. The good agreement between simulated and in-vivo results justifies a posteriori the validity of our computer flow model. On another hand, we will explore in the next future, the clinical potentialities of this tool which provides with instantaneous and accurate velocity profiles.

References

[1] C. Kasai, K. Namekawa, A. Koyano and R. Omoto
Real-time two-dimensional blood flow imaging using an auto-correlation technique
IEEE Trans. Sonics Ultrasonics, SU-32 (1985), 458-464.

[2] P.M. Embree, W.D. O'Brien
The accurate ultrasonic measurement of volume flow of blood by time-domain correlation
IEEE Ultrasonic Symposium 1985.

[3] A.P.G. Hoeks, R.S. Renaman and P.A. Peronneau
A multigate pulsed Doppler system with serial data processing
IEEE Trans. Sonics Ultrasonics, SU-28 (1981), 242-256.

[4] S.D. Ramsey, J.C. Taenzer, J.F. Holzemer, J.R. Suarez and S.P. Green
A real-time ultrasonic B-scan/Doppler artery-imaging system
IEEE Ultrason. Symp. Proc. (1975), 10-24.

[5] P.A. Grandchamp
A novel pulsed directional Doppler velocimeter, the phase detection profilometer
Ultrasonics in Medicine (Edited by E. Kazner, M. de Vlleger, H.R. Muller and V.R. Mc Cready), Excerpta Medica, Amsterdam (1975), 144-152.

[6] M. Brandestini
Application of phase detection principle in transcutaneous velocity profilemeter,
Ultrasonics in Medicine (Edited by E. Kazner, M. de Vlleger, H.R. Muller and V.R. Mc Cready), Excerpta Medica, Amsterdam (1975), 144-152.

[7] A. Nowicki and J.A. Reid
An infinite gate-pulsed Doppler
Ultrasound Med. Biol. 7 (1981), 41-50.

[8] P. Tortoli, G. Manes and C. Atzeni
Velocity profile reconstruction using ultrafast spectral analysis of Doppler ultrasound
IEEE Trans. Sonics Ultrasonics, SU-32 (1985), 555-561.

[9] O. Bonnefous, P. Pesqué
Time-domain formulation of pulse-Doppler Ultrasound and blood velocity measurement by cross-correlation
Ultrasonic Imaging , 8 (1986), 73-85.

[10] P. Atkinson and M.V. Berry
Random noise in ultrasonic echoes diffracted by blood
J. Phys. A : Math., Nucl. Gen. 7 (1974), 1293-1302.

[11] L.Y.L. Mo and R.S.C. Cobbold
A stochastic model of the backscattered Doppler Ultrasound from blood
IEEE Trans. Biomed. Eng., BME-33 (1986), 20-27.

[12] C.H. Knapp and C. Clifford
The generalized correlation method for estimation of time delay
IEEE Trans. Acoust., Speech Signal Proc. ASPC-24 (1976), 4-18;

[13] S.A. Jacobson, P.N. Denbigh and D.E.H. Naude
A new method for the demodulation of ultrasonic signals for cross-correlation flowmeters
Ultrasonics, 23 (1985), 128-132.

[14] P.J. Bendick and V.L. Newhouse
Ultrasonics random signal flow measurement system
J. Acoust. Soc. Amer, 56 (1974), 860-865.

[15] C.P. Jethwa, M. Kaveh, G.R. Cooper and F. Saggio
Blow flow measurement using ultrasonic pulsed random signal Doppler system
IEEE Trans. Sonics Ultras., SU-22 (1975), 1-10.

[16] P.N.T. Wells and R. Skidmore
Doppler developments in the last quinquennium
Ultrasound in Med. and Biol., 11 (1985), 613-623.

[17] M. Fink, F. Hottier and J.F. Cardoso
Ultrasonic signal processing for in-vivo attenuation measurement : short time Fourier analysis
Ultrasonic Imaging, 5 (1983), 117-135.

[18] M. Fink and J.F. Cardoso
Diffraction effects in pulse-echo measurement
IEEE Trans. Sonics Ultrason., SU-31 (1984), 313-329.

[19] R. Kuc
Bounds on estimating the acoustic attenuation of small tissue regions from reflected ultrasound
Proc. IEEE, 73 (1985), 1159-1168.