Doppler Echocardiography

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General Principles

Doppler principle allows determination of the velocity and direction of motion of the moving objects. One can understand this principle by imagining oneself stationed at a platform. If a train is approaching the platform and happen to whistle, then to the observer on the platform the whistle noise shall get louder as the train approaches the stationary observer Fig. 1A. However if the train is receding from the observer, then the whistle noise would get quieter Fig. 1B. This phenomenon is explained by the fact that sound waves received by the stationary object are smaller, i.e. of higher frequency, when compared to the frequency of sound waves emitted by the train. However, when the train is approaching the observer Fig. 1A Top Pannel, the sound waves received are larger, i.e. of lower frequency when compared to the frequency of sound waves emitted by the train Fig. 1B Top Pannel. Thus sound waves transmitted from a moving object are frequency shifted when received by a stationary object. Now let us assume that sound waves are transmitted by the stationary object and are reflected back to the observer by the moving object such as the train. In such a situation if train is moving away from the stationary object which transmits the sound wave on to the train, the reflected sound waves, which the observer receives, shall have comparatively larger frequency when compared to the frequency of sound that the observer transmitted. Similarly when a train or a moving object is approaching the stationary object which is transmitting the sound wave, the reflected sound waves from the moving object, which the observer receives, shall have comparatively higher frequency than the frequency of sound waves which were transmitted. Thus it is obvious that frequency shifts of sound waves, i.e. frequency difference between the transmitted and received sound waves indicates whether an object is moving away or towards the stationary object. Now we need only to imagine the moving train as the red cell and stationary object as the ultra sound transducer to understand how this principle is applied in the body. The transducer (stationary object) transmits a certain frequency of sound waves

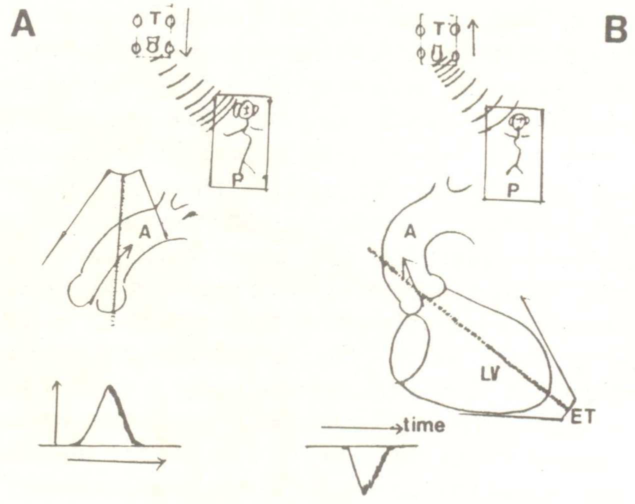

Fig. 1

When a train (T) approaches a stationary observer on the platform (P) the frequency of sound waves emitted from the train increases when received by the observer
.Receding train from the observer has an effect of decreasing the emitted frequency of sound waves as received by the stationary observer.
In real life, a transducer placed in the suprasternal notch records the flow in the ascending aorta (A) which is coming toward the transducer thus velocity spectrum is recorded above the base line while transducer placed at the cardiac apex, apex records flow in the aorta (A) which is flowing away from the transducer, thus velocity spectrum is recorded below the base line.

which are reflected by the moving object, i.e. train or a red cell. The frequency shift between the transmitted and received sound waves would tell us if the red cell is moving away from the transducer or is approaching it. Thus by applying the Doppler principle one can determine the direction and velocity of red cell movements within the vessel.

Doppler frequency shifts, when recorded by keeping the Doppler beam parallel to the flow direction, would best reflect the velocity of flow. For instance, if we want to record Doppler frequency shifts from suprasternal region and make the Doppler beam parallel to the ascending aorta then the flow of blood red cells would be towards the transducer which is at suprasternal notch Fig. 1A bottom pannel. The frequency shifts can be converted into velocity of blood flow by the Doppler equation. By convention blood flow velocity spectrum of the blood flow which is following towards the transducer is re-

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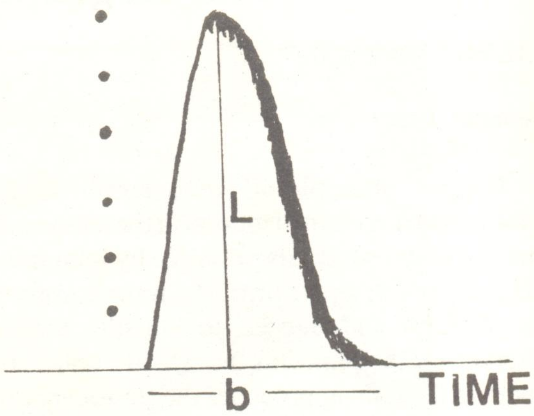
corded above the base line and flow away from the transducer is recorded below the base line, Fig. 1B bottom pannel. Frequency shifts and thus blood flow velocity can be recorded during the whole of cardiac cycle. Thus a timed velocity spectrum is obtained. One must further understand the genesis of this time-velocity profile. How is it obtained? As previously mentioned, reflected sound wave from one red cell would show frequency shifts between transmitted and reflected wave frequencies. There are millions of red cells in the blood vessels and these are moving in various directions. Thus when sound waves are reflected by a segment of blood vessels at any given time a very large number of frequency shifts would be available for analysis. In order to reduce the number of such measurements, only a small 'area or volume' of blood is earmarked (gated) from where reflected waves are recorded. Fast Fourier analysis is performed by the computer so that 'average velocity' i.e., average of all the frequency shifts from all of the red cells at any given time is computed. These computations are repeated sequentially for all the pulsed waves during the whole of systole and diastole. Obviously this is a very large data base which can only be analysed with the aid of a computer.

Laminar Flow

Most blood flows in the living vessels are laminar. It is a 'quiet' blood flow and the blood moves in layers which slide over each other. The layers which are closer to the vessel wall move relatively slowly than those layers which are located in the centre of the vessel. This is due to the fact that vessel wall offers friction which slows the flow. Now if one looks at the head of such a blood flow, then one sees a parabolic velocity profile, central velocities are higher, i.e., blood flow is faster in the centre than the periphery. This type of blood flows is orderly and 'most' velocities, i.e. red cells, are moving in one direction at a 'certain' speed. This profile characterises the laminar blood flow.

The velocity spectra in a time velocity spectrum of a laminar flow shows a fast early acceleration phase which has very narrow range of velocity scatter at each given time. After the acceleration phase is completed peak velocity is recorded, Fig. 2. A deceleration phase follows where the velocities scatter is increased so that at a given time red cells are moving with various velocities within a narrow range or band of velocity dispersion.



$$\bar{V} = \frac{1}{2} bL$$

Fig. 2

Timed velocity spectrum from a great vessel is recorded. Dots on Y axis show magnitude in m/sec and X axis shows time in seconds. L shows peak velocity in m/sec and b is the time from onset to the end of the velocity spectrum in seconds. Thus average velocity $\bar{V}$ can be determined from the equation.

Frequency Shift to Blood Flow Velocity to the estimation of Blood Flows

It is a simple matter of physics to use frequency shifts of the emitted and reflected sound waves from a moving object such as red cells, to calculate the blood flow velocity. Doppler equation provides such measurements:

fd = $(2f_o \times \text{Cos}\theta)/C$, Where

Cos θ = Angle between the ultrasonic beam and the direction of flow.

C = Speed of ultrasound in the medium.

fd = Difference of frequencies, (frequency shift).

fo = Frequency of emitted ultrasound.

Blood flow velocity can then be calculated by the following equation:

$$V = \frac{fd\ .\ C}{2ft\ (\text{Cos}\theta)}$$

Where fd = frequency shift; C is speed of ultrasound in the medium; ft = transmitted frequency; and Cos θ is the angle of the Doppler beam and the blood flow direction.

Blood Flow Calculation

Blood flow in a rigid tube is:

$$= \bar{V} \times A$$

Where $\bar{V}$ is the mean blood flow velocity (cm/sec) and A (cm^3) is the cross sectional area of the vessel.

Mean velocity can be derived from the time velocity spectrum.

Aortic and pulmonary arterial blood flow velocity tracings can be assumed to be triangles, so that mean velocity can be calculated as V = 1/2 bL Fig. 2 and that stroke volume, SV is $\bar{V}$ (cm/sec) x area (cm^3) which is II r^2 and cardiac output is: Co=SV x heart rate. An alternate method of obtaining the mean velocity is to do planimetry of the velocity spectrum and divide it by the base (time in sec). Area of a vessel, such as aorta, is $\overline{\Pi}$ r^2. Vessel diameter can be measured by imagining the pulmonary arterial or aortic root. The diameter is measured at the points of semilunar valve insertion. One half of the diameter is the radius. Cardiac output of pulmonary or aortic blood flow can then be calculated as:

$$\underline{\bar{V}(cm/sec) \ \times \ A(cm)^3} \ \times \text{ Heart rate/min.} = \text{ml/min.}$$

Calculation of Pressure Gradient at Stenotic Valves

When blood flow encounters a narrowed passage or an orifice the velocity of blood flow increases progressively towards the narrowed area so that at the ostium of the narrowing the velocity is at its maximum. As the blood enters the vessel distal to the obstruction the increased velocity of the blood flow is maximal in the central or laminar part of the stream (jet). The blood flow around the central jet becomes turbulent and small currents of blood flow develop, around the centre of the jet, in every possible direction. This continues for some distance in the distal vessel till the laminar flow is re-established, Fig. 3.

Fig. 3

When a narrowed area in a vessel is encountered by the flow its flow velocities increase progressively as the flow approaches the narrowed orifice. Distal to the narrowing a jet is ejected which has turbulent flow having velocities of flow in every possible direction. Central part of the jet shows high velocity flow which are reasonably directional. The jet axtends into the peripheral vessel for a variable distance before assuming laminar characteristics.

Continuous wave Doppler technique allows recording of the high velocity jets. The laminar portion of the jet (centre) velocities can often be recorded and to be certain that one is recording this area of the jet, one need to listen to the noise of the laminar jet velocities which is of high pitch and of smoother quality than when the turbulent portion of the jet is being recorded where the noise is harsher and of relatively lower frequency. The peak velocity can be obtained from the velocity profile and by using the Bernoulli's equation one can obtain the pressure gradient at the narrowed orifice:

Pressure gradient = 4 V^2 ;

Where V is the peak velocity.

Derivation of gradient of pressure equation 4 V^2 at the stenotic orifice by using the Bernoulli's equation.

Bernoulli Equation:

$$P_1-P_2 = \underset{\substack{\text{Convective}\\ \text{acceleration}}}{\overset{\text{I}}{1/2\,\rho\,(V_2^2 - V_1^2)}} + \underset{\substack{\text{Flow}\\ \text{acceleration}}}{\overset{\text{II}}{\rho\int^{2}\frac{DV}{Dt}\,DS}} + \underset{\substack{\text{Viscous}\\ \text{Friction}}}{\overset{\text{III}}{R(V)}}$$

ρ = Mass density of blood which is $1.06.10^3$ kg/M^3,
P_1 = Pressure, proximal to the obstruction,
P_2 = Pressure distal to obstruction,
V_1 = Velocity of proximal blood flow,
V_2 = Distal jet flow velocity.

$P_1 - P_2 = 1/2\,\rho\,(V_2^2 - V_1^2)$; Ignore II and III.

V_1 much less than V_2 so ignore V_1, observe that

$1/2\,\rho$ = Mass density of blood = $1.06.10^3$ kg/M^3

Because 133 Kilo pascels (K Pa) = 1mm Hg

So that $1/2\,\rho = \dfrac{1.06 \times 1000}{133 \times 2} = 4\text{mm Hg}$

Therefore simplified equation is $P_1 - P_2 = 4V^2$.

Atrio-ventricular Valve Stenosis

Hatle showed that the rate of velocity drop during diastole was proportionate to the severity of the stenosis of atrioventricular valves, Fig. 4. So that it was possible to calculate the valve area. Various other clinical studies have shown a good correlation between the echo derived atrioventricular valve area and that obtained by cardiac catheterization studies.

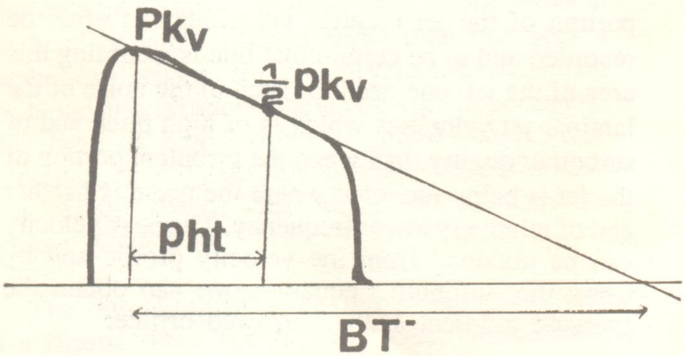


Fig. 4

Timed velocity spectrum of turbulent flow across the mitral valve is shown in a patient with mitral stenosis. A line is drawn superimposed on the EF slope. Peak velocity (PKV) is marked and the time taken for the peak velocity to decrease to 1/2 of the PKV value is the pressure half time (PHT). From the point on the base line where EF slope line transects the base, to the point on the base line which is at the time of peak velocity level, is the BT time (base time).

The method proposed by Hatle et al for calculation of valve area requires measurements of pressure half time which is the time taken for the peak diastolic velocity at the atrioventricular valve to reduce to 1/2 of its peak value. This can be calculated as:

$$\frac{Pk\ V}{\sqrt{2}} = \frac{Pk\ V}{1.4} \quad \text{Where } Pk = \text{Peak velocity.}$$

It is further observed that pressure half time of 220 mili sec. correlated with mitral valve area of 1.0cm². Thus mitral valve area can be calculated by dividing a figure of 220 by the pressure half time, thus:

$$MVA = \frac{220}{\text{Pressure half time}}$$

A further simplification of this method is to draw a line through the E wave slope till it crosses the base line and measure the base time, which is the time from the point of peak velocity at the base line to the slope-base line intersection point. This time is called the base time (BT), Fig. 4 and thus:

$$MVA = \frac{750}{BT}$$

A close correlation is shown between the two methods.

Assessment of Ventricular Performance

Systolic functions are measured by the peak velocity; acceleration time, velocity time integral and PEP/ET ratio, Fig. 5.

Peak acceleration slope of 19±5 metres per sec. corresponds to an ejection fraction (EF) of more than 60 per cent and peak acceleration of 12±2m/sec relates to EF of 41-60 per cent and peak acceleration of 8±2m/sec suggests EF of less than 40 per cent.

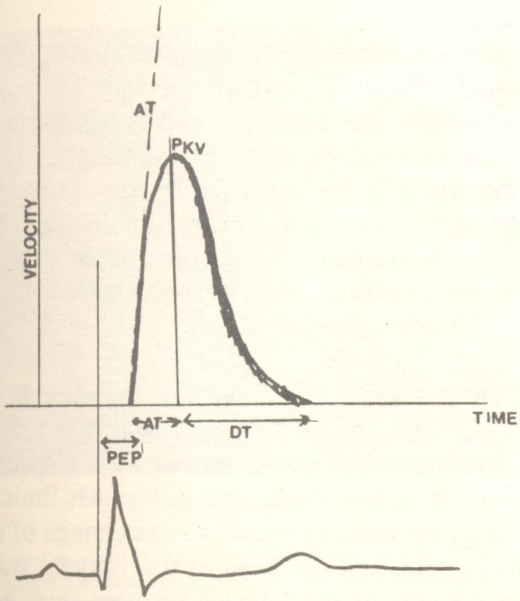


Fig. 5

Timed velocity spectrum from a great vessel is shown. Acceleration slope (AT) is drawn and the position of peak velocity (PKV) is marked. Acceleration time (arrows) is shown. Pre-ejection period (PEP) and ejection times (AT+DT) are shown.

Ventricular Diastolic Functions

E/A ratio normally is 1.7±0.4 - 2.5±0.9, Fig. 6, E area is normally 65±4 per cent of total diastolic flow area. A wave area is 20±7 per cent of total diastolic area. These indices are indirect measures of ventricular diastolic functions. The flow across the mitral valve depends upon the relative pressure gradient between the ventricular diastolic pressure and atrial pressures, ventricular diastolic compliance, normality of atrioventricular valve apparatus and heart rate; all of which contribute to the flow dynamics of the atrio-ventricular valves. Considering these factors it seems that only clinical use of these indices may eventually decide their usefulness as a measure of ventricular diastolic function.

DOPPLER VELOCITY PROFILE OF NORMAL FLOWS

Great Vessel Flows

Pulmonary blood flow: The blood flow velocity across the pulmonary valve can be recorded by using the Doppler technique and by placing the sample volume in the pulmonary artery in a left parasternal short axis view. In such a position the blood is expected to be flowing away from the transducer, thus velocity profile is recorded below the base line. During acceleration phase of the velocity spectrum the spectral broadening is minimal, however during the deceleration phase the velocities are spread along a narrow band (dotted lines), Fig. 7A.

Peak velocity is recordable at the point of maximum velocity during ejection. The time lapse from the beginning to the end of the velocity spectrum is the ejection time. As has been pointed out previously the pulmonary blood flow can be calculated by the equation; blood flow ml/min=mean velocity (cm/sec) x area of the pulmonary artery IIr^2(cm^3) x heart rate (beats per minute). The error involved in this method is the error one makes in measuring the diameter of the pulmonary artery. The diameter is measured at the level of pulmonary valve at its annulus. Since square of 1/2 of the diameter (radius) is used in calculation, small errors make large difference in blood flow estimation.

Aortic Flow Velocity Profile

Aortic flow velocity can be recorded from the suprasternal notch with transducer in coronal plane in an anterior tilt. In this position blood flow is directed toward the transducer so that velocity profile is recorded above the base line. The velocity spectral dispersion pattern is similar to that obtained for the pulmonary blood flow. Aortic flow velocity can also be recorded from apical position.

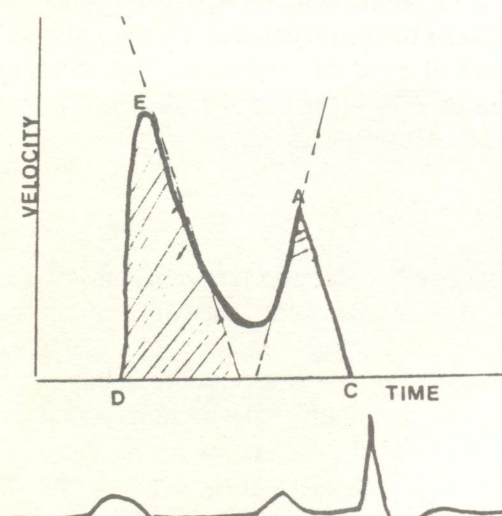


Fig. 6

Velocity spectrum of blood flow at the mitral valve is shown. E and A points are marked and the area under E and A waves are shown. D is beginning point of mitral flow and C is the end of diastolic flow.

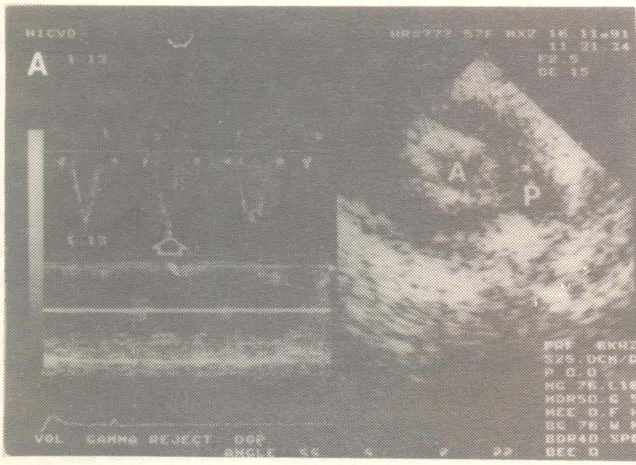


Fig. 7A

Velocity profile of pulmonary arterial flow is shown, in the left pannel open arrow points to the peak velocity. Note, dispersion spectrum is much pronounced on the deceleration phase of the tracing. The sample volume (=) is placed in the pulmonary artery (P) in short axis plane, in the right pannel; A = Aorta.

Atrioventricular Valve Flow Velocity Spectrum

The flow velocity profile resembles an M shape with initial early rapid diastolic flow E wave and late diastolic A wave, i.e., slow filling phase. Early diastolic peak velocity (E) is higher than the peak velocity of the A wave. Atrial contraction contributes to the enhanced flow during late ventricular filling. The diastolic velocity spectrum when recorded distal to the atrioventricular valves shows a narrow band of spectral dispersion. For measurement purposes it is suggested that the darkest part of this zone be used, Fig. 8A.

Pulsed Wave Doppler

Pulsed wave Doppler uses bursts of sound waves for a unit of time and there is a pause after each impulse emission. During this pause, reflected sound waves are received. Thus it is possible to pulse only certain number of impulses per minute depending upon the depth of the object to be interrogated. Obviously longer the distance of object from the transducer, longer is the time required for transmission and receiving of the impulses smaller will be the number of possible impulses per minute. The second factor which determines the frequency of pulses is the speed of ultrasound in the medium.

Aliasing

It is a common observation that if one observes a wheel with spokes, which is rotating in a certain direction and the rotation speed is being gradually increased so that when a critical speed of rotation of the wheel is exceeded the spokes start to apparently move in the opposite direction. This phenomenon by which a fast moving object apparently seems to change direction, when its speed exceeds a certain critical limit, is called aliasing.

We observe this phenomenon in pulsed Doppler echocardiography. If an object, which is reflecting the emitted sound wave, moves with a speed that exceeds a certain limit, that is Nyguist limit, then the Doppler velocity record shows change of direction, Fig. 8A and the head of the velocity tracings is transected and is projected inverted from the top of the tracings, i.e., in the opposite direction to the initial tracing. In colour flow mapping the direction of the blood flow is colour coded so that when aliasing occurs there is a change to opposite colour when the speed of blood flow exceeds Nyquist limits so that one sees blue trapped within the red and vice versa.

The way to avoid aliasing is to increase the pulse repetition frequency, but there is a limit to that. The second option is to reduce the depth of the observation which may not eitherbe possible or desirable. Thus in order to study high speed blood flows, the

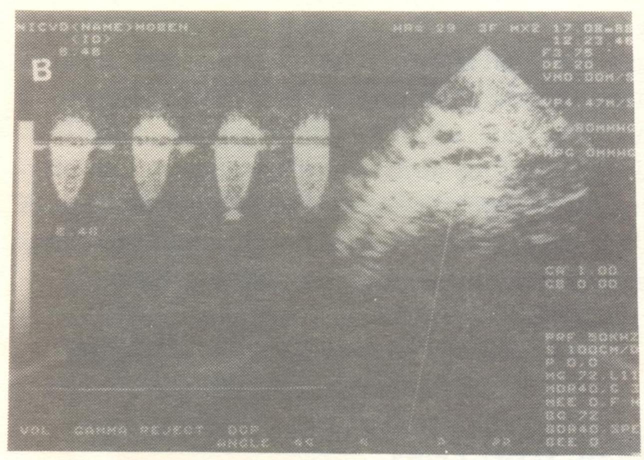


Fig. 7B

Shows velocity profile of stenotic pulmonary valve flow, left pannel. Note the velocity dispersion is so gross that the whole of the record shows wide scatter so that the entire spectrum is whitened. This is characteristic of disturbed or turbulent flows.

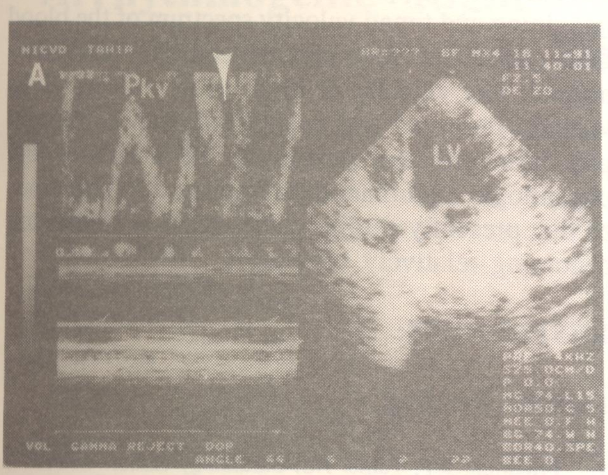


Fig. 8A

Sample volume (=) is placed close to mitral valve in the left ventricle, from a transducer placed at the apex (right pannel). Normal velocity profile of mitral blood flow is recorded (left pannel). Note a narrow band of dispersion more on deceleration phases of E and A waves.

technique of pulsed Doppler can not be used. Continuous wave Doppler is utilized to record high speed flows.

Continuous Wave Doppler

This technique differs from the pulsed Doppler technique in that the sound waves emitting unit is separate from the reflected sound waves receiving unit. Thus sound waves are projected on to a fast moving object continuously and the reflected waves are recorded continuously so that this technique allows recording of very high velocity blood flows, Fig. 7B, 8B.

Turbulent flows in the form of a jet are very high velocity flows and this technique allows recording of such flows which are generated at the stenotic orifices either of great vessels or cardiac valves such as aortic or pulmonic valve stenosis. The disadvantage of this technique is that the depth (position) of the object from where reflections are being received can not be determined and additionally all objects within the path of transmitted beam would reflect the sound waves which will be recorded.

In conclusion, imaging, Doppler derived blood flows and colour coding allows haemodynamic measurements and visualization of physiologic parameters which with conventional physiology could only be measured mathematically in numbers.

Colour Flow Mapping

In this technology cardiac imagining and Doppler blood flows are recorded separately and are eventually matched so that Doppler information is placed within the images. In the black and white Doppler echocardiography the imaging of cardiac structures are projected on the video screen and visualised, however, Doppler flows obtained by pulsed or continuous wave Doppler could only be recorded separately on the screen as time velocity spectra. Moreover Doppler information was usually obtained from the site within the cardiac chamber, i.e., from one sample volume by the gating mechanism previously discussed. In the colour flow mapping technique multiple gates are used within each echo beam. Thus Doppler frequency shifts are recorded from these 'gates', i.e., sample volumes, in each beam and sequentially from all the echo beams as they sweep from beginning to the end of the sector angle. Obviously large amount of data is thus obtained. The data analysis is done not by measuring each Doppler frequency shift and calculating individual velocity points in each echo beam and echo sector but instead slopes of frequency shifts are calculated and from the computer memory 'frequency shifts trend' matching is done and the nearest match is reported as the measurement. This technique allows rapid 'measurements' to be made over a fraction of time period.

The second important innovation is that direction of blood flow is colour coded, generally red

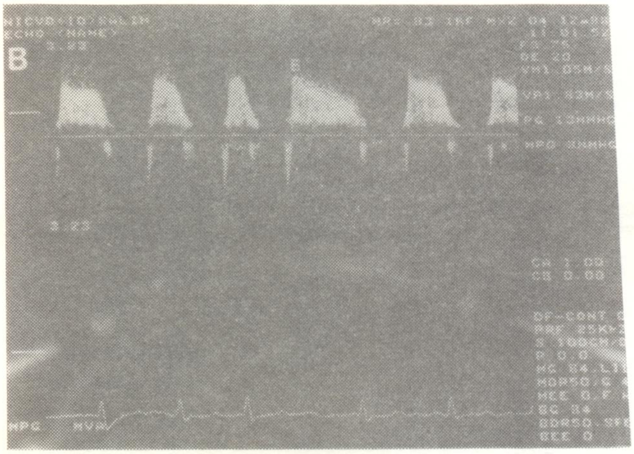


Fig. 8B

Disturbed flow across the mitral valve is recorded in a patient with mitral stenosis and atrial fibrillation. Note turbulent flow pattern and slow E slope (E).

colour is displayed if blood flow is toward the transducer and blue colour is displayed when blood flow is directed away from the transducer. The magnitude of blood flow is shown in the brightness of the colour. These colour displays consist of pure colours when normal laminar flow is present which is within a certain velocity range, however, if blood flow velocity is exceeded then aliasing occurs so that one observes a blue centre of a red coded flow which means that a blood flow is towards the transducer but due to high velocity apparent reversal of 'colour' is observed.

The colour flow mapping technique allows 'resolution of aliased velocities' by introducing yellow colour code for the turbulent flow. So that high velocity flow appears as combination of red, blue, yellow so that multicolour flows suggest high velocity turbulent flows. This remarkable technological feat has allowed qualitative observation of blood flows and is a great advance in studying the haemodynamics of flow disturbances within the heart and great vessels. Additionally quantitation of flows is possible and can accurately be done at the area to be interrogated by incorporating the pulsed or continuous wave Doppler technique, so that colour flow imagining and timed velocity spectrum of the blood flow can be recorded simultaneously. This allows normal, low velocity or high velocity flows to be visually observed and quantitated. Echocardiography now provides details of the anatomic and physiologic disturbances in various malformations so that presently cardiac catheterization studies are becoming relatively infrequent.

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Further Reading

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