CHAPTER 2

LEFT VENTRICULAR ASSESSMENT WITH DOPPLER TECHNIQUE

2.1 INTRODUCTION

It is one of the easiest non-invasive method to assess the left ventricular function.

The Austrian Mathematician physicist Johann Christian Doppler, 1803-1853, first observed the general phenomenon that the frequency of a wave form is dependent upon the wave. The effect is applicable to any kind of wave including sound waves.

In Doppler Sonography, the Doppler effect is used to measure the blood flow velocity. Ultrasound reflected from red blood cells will change frequency according to the blood flow velocity. It is one of the easiest non invasive method to assess the Left Ventricular function.

2.2 DOPPLER EQUATION

The mathematical relationship between the Doppler frequency shift and blood flow velocity is given by

\[ f_D = f_e - f_o = 2f_o \frac{v \cos \alpha}{c} \quad (2.1) \]
\[ f_\text{D} = \text{Doppler frequency shift} \]
\[ f_e = \text{Frequency of reflected sound or echo} \]
\[ f_Q = \text{Frequency of transmitted sound} \]
\[ v = \text{The velocity of blood flow} \]
\[ \alpha = \text{The angle between the Doppler ultrasound beam and the direction of blood flow.} \]

The Doppler equation may be derived from the following. Assume a situation in which blood flow is directed straight towards the Doppler transducer, i.e. Doppler angle is 0. The transducer transmits ultrasound with a frequency

\[ f_o = \frac{c}{\lambda_o} \quad (2.2) \]

where \( c \) is the propagation velocity and \( \lambda_o \) is the wavelength. The R.B.C. are flowing with a velocity of \( c \) and meet ultrasound waves with a relative velocity of \( (c+v) \). The RBCs encounter the ultrasound wave fronts with frequency

\[ f_r = \frac{(c + v)}{\lambda_o} \quad (2.3) \]

which is higher than \( f_o \) arrived from equation 2.2. The RBCs at this point be considered to be transmitters of ultrasound, transmitting with a frequency \( f_r \) straight to the transducer and the wavelength of the reflected ultrasound is shortened by a distance equal to the one travelled by the RBCs. The shortened wavelength of the reflected ultrasound is thus

\[ \lambda_e = (\lambda_r - v.t_r) \quad (2.4) \]

The frequency of the echos

\[ f_e = \frac{c}{\lambda_e} = \frac{c}{(\lambda_r - v.t_r)} \quad (2.5) \]
The above equations can be expressed as Doppler equation

\[ f_r = \frac{c + v}{\lambda_o} = \frac{c + v}{c} = f_o \frac{c + v}{c} = f_o \left( 1 + \frac{v}{c} \right) \] (2.6)

The frequency of the echos \( f_o \) is written as a function of \( f_r \)

\[ f_o = \frac{c}{\lambda_r - ut_r} = \frac{c}{\lambda_r - \frac{v}{f_r}} = \frac{c}{\lambda_r \left( 1 - \frac{v}{c} \right)} \]

\[ = f_r \frac{1}{1 - \frac{v}{c}} \] (2.7)

Then the equations 2.6 and 2.7 are combined

\[ f_o = f_o \left( 1 + \frac{v}{c} \right) \frac{1}{1 - \frac{v}{c}} = f_o \frac{\left( 1 + \frac{v}{c} \right)^2}{\left( 1 - \frac{v}{c} \right) \left( 1 - \frac{v}{c} \right)} \]

\[ 1 + 2 \left( \frac{v}{c} \right) + \left( \frac{v}{c} \right)^2 \]

\[ = \frac{\left( \frac{v}{c} \right)^2}{1 - \left( \frac{v}{c} \right)^2} \] (2.8)

The blood flow velocity \( v_1 \) will always be much lower than the ultrasound propagation velocity. So the terms \((v/c)^2\) may be omitted.
Fig. 2.1 Doppler Angle

$\alpha$: Doppler angle, $V$: True blood flow velocity in vessel, $V_5$: Velocity component oriented along Doppler ultrasound beams (dotted lines).
The Doppler shift is the change in frequency from transmission to echo reception

\[ f_s = f_e \left(1 + \frac{v}{c}\right) \]  \hspace{1cm} (2.9)

This Doppler equation 2.10 is arrived by assuming the angle of incidence of ultrasound beam is zero.

When there is an angle of incidence, \((\alpha)\) as shown in Fig.2.1, the velocity in the probe is the velocity component multiplied by \(\cos\alpha\). So \(v \cos\alpha\) will be the true flow velocity. Substituting \(v\) with \(v \cos\alpha\) in the equation 2.10 gives the general Doppler equation as shown in equation 2.1.

2.2.1 Doppler Angle

In the Doppler blood flow velocity measurement, the Doppler instrument sees only the blood flow velocity component being directed straight towards the transducer. This component is equal to \((v \cos\alpha)\) where \(v\) is the true blood flow velocity, \(\alpha\) is the Doppler angle. Owing to the cosine relationship between the relative and true blood flow velocity, the consequence of an error in measurement of the Doppler angle on the estimation of true blood flow velocity, increases with increasing Doppler angle. At Doppler angles less than 45°, a 5° over estimation of Doppler angle will give over estimations of true blood flow velocity less than 10%. At Doppler angles larger than 70°, the consequences will be dramatic.
To measure the flow velocity, it is necessary for the specialist to know the sensitive area. This is done by coupling the Doppler system with real time. "B" mode ultrasound imaging system.

Most system combine pulsed Doppler with real time sector imaging.

In general, a real time image of the anatomy of interest is obtained and the image is frozen on the viewing screen. The Doppler mode is switched on and the cursor is positioned in the location and axial length from which the Doppler signal will be obtained by locating appropriate cursors on the frozen real-time image.

2.3 RESOLUTION

When we talk of the resolution of the equipment, we are concerned with the ability of the equipment to resolve two closely placed objects or organs in different depths and also in the horizontal direction. These two are called as axial and lateral resolution. So before we ventured into the left ventricular functional assessment a phantom was designed and fabricated for the Quality Assurance (QA) of the ultrasound Doppler equipment. The quality assurance programme was to ensure that the diagnostic information provided by the machine is maintained at the maximum attainable level for that instrument and technology.

This quality assurance tests should be performed on a routine basis to detect deviations from the base line minimum limits.

The single most versatile and complete test object that can be used to perform these studies, amongst all, is AIUM standard 100 mm test object.
2.3.1 Method

The test phantom was fabricated with the locally available materials such as 0.75 mm dia stainless steel rods, transparent perspex sheets of dimension of 200 x 200 x 5 mm and spacer bolts with 50 mm length.

In the 200 x 200 mm square perspex plate the markings and drillings were made with 0.8 mm bore following a specific pattern, so as to place the rods in the ultrasound beam to test both the axial and lateral resolution. Care is taken to ensure the distance between the stainless steel rods is constant. At a latter stage we found that the stainless rods can be replaced with suture thread. The echogenicity of the suture thread is comparable with that of the stainless steel rods.

It has many distinct advantages over the stainless steel rods. The rods over a period of time may get rusted as the phantom is kept under water.

This problem does not surface with suturing thread. The rods may bend and the distance between them may change and there is no more a standard distance and the results from the ultrasound machine tested using this phantom can not be relied on. Where as in the case of suturing thread it can be tightened and once it is done it will not sag. So that the space between the thread lines will remain constant for ever.

Replacement of the rods, if some deviation has taken place, will not be possible, where as the suturing thread can be replaced at any time without much difficulty. The results with suture thread in place of stainless rods are commendable. The Fig.2.2 shows the results with suture thread instead of the stainless steel rods.

The test object was used to calibrate the equipment.
Fig. 2.2  The Ecogenicity of Suture Thread in Ultrasound Resolution Phantom.

Fig. 2.3  The Results of the Ultrasound Test Phantom with 3 MHz Transducer.
Fig. 2.4 The Effect of Increased Ultrasound Gain with 3 MHz Transducer.

Fig. 2.5 The Effect of Zoom with 3 MHz Transducer.
The axial and lateral resolutions were measured for all most commonly used transducer probes viz., 3 MHz, 3.5 MHz, 5 MHz and 7.5 MHz, the effect of focussing, zooming and defocussing were done.

2.4 TRANSDUCER

2.4.1 3 MHz Transducer

Three different observations were made in this case. The focus was kept constant at 8 cm. The configuration of the reflectors in the test phantom is seen in the Fig.2.3. Within the 8 cm range the image is sharp.

The presentation is good and the bottom of the tank is seen at about 20 cm depth. The gain was kept at minimum and all the reflectors were resolved very clearly.

The effect of increased gain is seen in Fig.2.4. Though the images are bright they are distorted. The images of the reflectors placed close to the probe are smudged and fused. The reverberation artifacts are also seen.

The result of zooming is shown in Fig.2.5 for the same 3 MHz transducer at a focus of 8 cm. The zooming results in loss in resolution, so the images look distorted. The reflectors placed at 0.1 mm distance are fused and look like a single image because of the loss in axial resolution. The most important observation made here is that the distance between the reflectors does not change even if the resolution is lost.

2.4.2 5 MHz Transducer

If the frequency is increased the axial and lateral resolution improves, as the beam can be focussed more precisely on the required region. This goes with some degree of compromise. We know that with the
Fig. 2.6 The Resolution of Ultrasound Beam with 5 MHz Transducer.

Fig. 2.7 The Effect of Reduced Focal Length with 5 MHz Transducer.
Fig.2.8  Reduced Penetration and Increased Resolution with 7.5 MHz Transducer.

Fig.2.9  Measurement of the Root of Aortic Valve.
increase in frequency the amplitude and intensity of the beam increases and the attenuation also increases. Due to increased attenuation the penetration is reduced. So the deep seated structures can not be seen.

This is demonstrated in the test phantom also the results are shown in Fig.2.6, the bottom of the tank is not seen in this case, on the other hand the image is very sharp within the focussed area.

The above figure shows that the focus is 3.5 cm. The image lying within this limit are very sharp and beyond this the images are distorted.

In Fig.2.7 the focal length is reduced to 2.2 cm. The images of the reflectors lying within this limit are very sharp and those lying away are distorted.

2.4.3 7.5 MHz Transducer

As we discussed in the section 2.4.2 here also the quality of the image improved due to increased frequency with reduced penetration or restricted area of viewing, as with the increased frequency the sharpness of the ultrasound beam improves, the lateral and axial resolution is much better but only in a limited imaging area. The focus is at 2 cm depth. The images near the surface are very clear and very sharp. The finer details can be found. When we go farther away from the focus here also the image gets distorted as shown in Fig.2.8.

So the equipment was calibrated before we proceeded further to carry on the study of the assessment of the left ventricular function. The results were very convincing and the performance of the Doppler equipment was upto the expected and prescribed level of functioning.
2.5 LEFT VENTRICULAR FUNCTION

Left Ventricular Ejection fraction is generally considered to be the single most representative index of the global ventricular function. This index represents the fraction of the end-diastolic volume of the ventricle that is ejected with each beat.

The cardiac Doppler is one of the simple and non invasive method to quantify the cardiac output. To determine the cardiac output we need two prime ingredients. The first is the heart rate, this can be determined by physical examination of the patient or by all automated ECG analysis. The second is the stroke volume and this can be determined by Doppler. Conceptually the stroke volume that appears in the aortic root is a cylindrical plug with the base being the cross section of the aortic root and the height being the distance that blood travels up the aortic root during a systolic ejection. The area of the aortic root is easily determined by two dimensional echo cardiography or by M-mode.

The distance blood travels during a systolic ejection is determined by Doppler velocity information. Acceleration, velocity and distance are related. If one can determine the area under an acceleration curve, velocity is measured, where as the area under velocity curve determines the distance. By taking the area under the Doppler systolic profile, we measure the distance the blood travels up the aortic root during a single systolic ejection.

By taking this distance and multiplying by the area of the aortic root one can calculate the stroke volume or the volume of the cylindrical plug that is ejected from the left ventricle during systole.
2.5.1 Cardiac Output (CO)

\[ \text{CO} = \text{Stroke volume (S.V)} \times \text{Heart Rate (H.R)} \]

Stroke volume (SV) = Cross sectional area (CSA) x flow velocity integral (FVI)

It is very difficult to make precise measurement of the cross sectional area. So utmost care is to be taken in doing the measurement. If the cross sectional area measurement is slightly inaccurate, the output is affected significantly.

In addition all methods of Doppler derived cardiac output make the following assumptions:

1) The cross sectional area does not change during the time the flow velocity integral is measured.
2) Flow is laminar and blunt.
3) During disturbed flow due to valvular disease or shunt, flow is neither, in the ascending aorta nor in stenosis.
4) Flow becomes parabolic rather than blunt down stream from an opening.
5) The flow velocity integral and cross sectional measurement are obtained from precisely the same place.

Careful attention and technique are needed to get the cross sectional area.

2.5.2 Cross Sectional Area Calculation

1. Obtain a diameter measurement where flow is obtained. It is recommended, to take an M-mode from both a long-axis and a short-axis and then average the two distances.
2. Since we are assuming the structure is circular the formula is 
\[ \pi \left(\frac{D}{2}\right)^2. \]

3. So the cross section area is, if a dia of 3 cm is obtained from Step 1, 
\[ (1.5)^2 \times \pi = 1.5^2 \times 3.14 = 7.065 \text{ cm}^2 \]

### 2.5.3 Flow Velocity Integral

a. The flow velocity integral is the column of blood moving through this structure during one R.R. interval. It is recommended to obtain the highest velocity possible. Do this by attempting to place the transducer parallel to the flow rather than correcting for the angle. When angle correction is done one may introduce more errors than correcting the error.

b. Measure the distance in centimeter. See from the baseline to the peak. Peak flow velocity in this application is measured at the centre of the Doppler flow spectrum at the time of maximum blood flow velocity.

c. Measure the time in seconds from the onset of systolic ejection to the end of systolic ejection.

d. Multiply the results of steps b and c.

e. Divide the result from step d by 2.

f. Add 0.3 to this result, will give you the flow velocity integral.

### 2.5.4 Cardiac Output Calculation

a. Multiply the cross sectional area by the flow velocity integral to obtain the stroke volume in cubic centimeters.

b. Multiply this result by the heart rate. The resulting cardiac output will be in ml/minute.
The Ejection fraction is expressed in terms of percentage of the cardiac output. It is expressed as

\[
E.F. = \frac{\text{End diastolic volume - end systolic volume}}{\text{End diastolic volume}} \times 100
\]

with the help of the Doppler Echocardiography equipment. The left ventricular ejection fraction and regional wall motion abnormality for 200 patients were done.

Fig. 2.9 shows the measurement of the root of the aortic valve.

Fig. 2.10 shows the aortic valve Doppler signal from a normal 50 years old individual. Cursor showing the markings of the beginning of the systolic and end systolic and the distance the blood travelled.

Fig. 2.11 shows the Aortic Doppler signal of 30 year old adult with normal function and output. Peak velocity 71 cm/sec, H.R. = 56 and FVI = 280 m sec.

Fig. 2.12 shows the Aortic Doppler of a 53 year old patient who had a mild Ischemic Heart disease involving a small segment of left ventricle (apex) and the global function was normal.

Fig. 2.13 shows the Aortic Doppler signal of a sixty year old patient, who has mild hypertension. The Doppler does not show any variation due to B.P. The accuracy of the results in Echocardiography depends upon the technique used.

Echocardiography is carried out in the supine or left posterior oblique positions employing the usual transducer positions. An ALOKA-870 colour Doppler flow imaging system was used for our study.
Fig. 2.10 Doppler Signal of a 50 Years Female.

Fig. 2.11 Aortic Doppler Signal of a Normal 30 Years Old Male.
Fig. 2.12 Doppler Signal of a Male who had IHD.

Fig. 2.13 Aortic Doppler Signal of a 60 Years old Male with Hypertension.
2.5.5 Transducer Positioning

Positioning of the transducer plays an important role in the diagnosis of cardiac disorders. The normal ventricle is shaped like a rugby ball with right ventricle wrapped around part of its surface. Consequently, the interventricular septum being approximately at right angles to the plane of the outlet septum.

Because angiocardiographic projections carry an inherent disadvantage. This seems to be an advantage in Echocardiography, it permits the study of the entire ventricular septum from a number of different views with greater ease and without risk.

The study of the ventricular septum is one of the finer points of Echocardiography, Sutherland and colleagues have provided an excellent description of the techniques necessary. In brief, these consists of a combination of short axis, long axis and four chamber views of the heart.

Two basic transducer positions are used during a routine cardiac examination. In the first position the transducer is placed on the chest so as to produce a field of view in the longitudinal or long axis of the left ventricle. In the second position the transducer is rotated through 90° to the transverse or short axis of the left ventricle. It is usually obtained, at the level of aortic valve, mitral valve, papillary muscle and ventricular apex.

2.5.6 Doppler Examination

It is critical that the operator remembers that the best Doppler information is obtained when the Doppler beam is oriented so that it lines up as closely parallel to blood flow as possible. This will ensure that the strongest Doppler signals to be reflected back to the transducer and thus
maximum peak velocity is obtained. In order to achieve the goal of a small intercept angle to the blood flow, the operator must try a wide variety of acoustic windows. There are few standard acoustical windows. The most commonly used windows are

a. Suprasternal window
b. Right parasternal window
c. Left parasternal window
d. Left apical window and
e. Sub costal window

Of all these acoustical windows listed it is usually the best to begin using the apical window. When in the apical four chamber view, slight superior angulation of the plane will allow the operator to encounter the left ventricular outflow tract. An operator can always obtain Doppler flow data from the ventricular side of the aortic valve. The transducer must be placed at the point of maximal impulse at the cardiac apex. Continuous wave can be used to quickly scan the left ventricular outflow tract and find the best angle of incidence and the best flow velocities. Pulse mode Doppler can then be used to sample the flow in the areas of interest.

The diameter of the aorta is measured at the root of the aorta, i.e. at the aortic valve, as it is almost a smooth circle and it is not changing its shape during the systole or diastole. The diameter of the aortic valve is measured at different places and the value is averaged. Any error in measurement of diameter will be squared while calculating the cross sectional area. Therefore utmost care is to be taken.

Peak velocity is measured with much care and precision. The cursor and the sample volume is placed at different places of the aorta and the maximum velocity is calculated.
The time between the onset of the ejection and the end of the stroke is calculated by placing the cursor at both sides of the pulse. As time is measured in milliseconds, there is much scope for error, therefore the ejection time is calculated for different strokes and averaged.

To make a candid and statistically significant study 200 patients were drawn randomly from the out patient and inpatient departments of the hospital. They mainly consisted of the patients who turned up to the cardiology department with chest pain, discomfort or with past incidence of coronary artery diseases or the patients referred from other departments for cardiac opinions.

The inpatients were drawn from the wards and were already on treatment for cardiac ailments. Some of the patients were old infarcted cases, treated for the same and reported for follow up. The patients were all from different walks of life. There were patients who do strenuous work, which results in straining the cardiac system and there were some other patients who do white collar jobs which did not involve much of a physical work.

The age groups of the patients also varied widely but all the patients were above twenty years old except two cases. The following table gives the age wise distribution of the patients.

All the above patients were also subjected to radionuclide ventriculography using Elscint Apex 4 gamma camera.

The above patients were referred to the department mainly for the assessment of left ventricular function and also for the assessment of the segmental wall motion abnormality.
The L.V. function depending upon its performance is graded into four main categories. They are

a) Normal L.V. function  
b) Mild L.V. dysfunction  
c) Moderate L.V. dysfunction  
d) Severe L.V. dysfunction

The range of values used in deciding left ventricular function in terms of percentage Global Ejectors fraction is as given below for Echo.

Normal L.V. function ≥ 75%  
Mild L.V. dysfunction 70% - 75%  
Moderate L.V. dysfunction - 60% - 70%  
Severe L.V. dysfunction - ≤ 60%

The segmental wall motion abnormalities are divided into five main categories. They are

Normal  
Mildly hypokinetic  
Severely hypokinetic  
Akinetic and  
Dyskinetic

In the case of Echo no predesigned chart is used as in the case of MUGA, whereas the same norms are used in all three modalities for assessing the segmental wall motion abnormalities.