CHAPTER 3

THE RADIONUCLIDE VENTRICULOGRAPHY
WITHOUT EDGE ENHANCEMENT

3.1 INTRODUCTION

Recently the technique of multigated equilibrium scintigraphy has become widely utilized for clinical assessment of regional and global left ventricular function as well as right ventricular function both at rest and exercise.

Essentially the performance of multiple gated equilibrium cardiac blood pool scintigraphy involves
1) a scintillation gamma camera with a collimated detector to record the distribution of radioactive tracer in the chest.
2) a radiopharmaceutical that remains in the cardiovascular compartment.
3) a physiologic synchronizer or gating device to time the scintigraphic events with respect to the cardiac cycle.
4) an imaging protocol which calls for patient positioning in a variety of views to optimize the calculation of ejection fraction and the assessment of regional wall motion.
5) a computer as a means of data recording and analysis.

The modern scintillation Gamma camera has a Large Field Of View (LFOV) either in circular or rectangular form. The Gamma camera used for this work was "ELSCINT APEX4" and G.E. Max 4" with a circular NaI (Tl) crystal with a diameter of 28" with a thickness of 1 cm (3/8") with
37 and 61 PMTs each respectively. The detector being hygroscopic, it is hermetically sealed from all sides except the one viewed by array of Photo Multiplier Tubes (PMTs). The front face is covered by a lead collimator. The whole detector system is shielded by lead to protect it from undesirable radiations. The detector can, therefore, view radiations only from the front face through the collimator.

The lead collimator not only protects the detector but also prevents the gamma rays which are not falling perpendicularly in case of parallel hole collimator or in a specific direction for other type of collimators.

The open side of the detectors is tightly coupled to the light guide assembly so that there is no chance for the air/moisture to enter the crystal. The PMTs are coupled to guide assembly with an optical coupling grease made of silica. PMTs are connected to high voltage circuit.

When a photon transfers energy to the crystal scintillation is produced at a given location in the crystal. The point of interaction is defined in terms of spatial co-ordinates. The fraction of this light reaching the PMT varies inversely as the distance of each PMT from the point of interaction. So the nearest PMT will get maximum light and the farther one will receive less amount of light. The PMT converts the light into electrical signal, multiply and amplify them before sending them to pre-amplifier/amplifier. The signals are simultaneously sent to X,Y and pulse arithmetic circuits.

The X and Y signal amplitudes are in proportion to the spatial co-ordinates of the original scintillation.

In the circuit, the pulse summed to provide Z signal is proportional to the total energy absorbed in the crystal due to photon interaction.
Therefore it is essential to normalize the X and Y signals by dividing them by Z signal. This ensures the co-ordinate signals within the field of view of the camera.

The Z signal is passed on to pulse height Analyser (PHA) and if it falls within the preselected window only then the X, Y co-ordinate signals are allowed to record the event.

The pulse is used to start the digitization of the position signals.

There are many circuits involved in the recording, storing and processing the digitised signals. So all the modules should perform to its maximum level in co-ordination with other module for the good performance of the camera.

The most popular testing method used to evaluate the performance of the camera by the manufactures is the protocol developed by the National Electrical manufactures Association (NEMA 1986, 1994, AAPM 1981, TG-4 1995).

The first level of check is the acceptance test after the installation of the equipment.

The system is then subjected to regular Quality Assurance Protocol to check the performance parameters (Murphy 1987, Blust 1984).

The important parameters to be checked are as under.

1. Spatial resolution
2. Uniformity
3. Energy resolution
4. Spatial linearity
3.2 SPATIAL RESOLUTION

Spatial resolution is related to the smallest separation between two point sources which will permit them to be distinguished as two distinct sources, and is measured by imaging a point as point spread function or a line source as line spread function.

3.2.1 Point Spread Function (PSF)

Point spread function is the function used to assess the spatial resolution of an imaging system. Consider an object consisting of a perfect point. The image of this object will be at least one pixel wide. Normally it will consist of a spot of several pixels, brightest in the centre and progressively darker away from the centre. This image function is the point spread function. Obviously, the more concentrated the spot is, the better the resolution. If a profile through the spot is plotted we obtain a one-dimensional PSF as shown in Fig.3.1.

Now the resolution can be defined as the width within which the PSF drops to half the maximum value, called FWHM. The PSF need not be symmetrical, so there may be different spatial resolution in different directions.

The smaller the separation between two distinctly identifiable sources, the better the resolution. It is normally expressed in millimeters.
Fig. 3.1 Point spread function of a point source and effect a scattering material and full width at tenth maximum (FWTM) (Courtesy : Gerald 1988, Ref. No. 27)
The spatial resolution may be measured without collimator \( (R_t) \) and with collimator \( (R_c) \).

Although the image is always acquired with collimator in place but the intrinsic resolution has its significance.

The system resolution \( R_s \) for a source at a distance is

\[
R_s = \sqrt{R_t^2 + R_c^2}
\]

(3.1)

3.2.2 Intrinsic Resolution

It is the resolution of the crystal and PMT.

It is expressed as FWHM of the line spread function of a collimated line source placed on the surface of the detector. A 30-40 mm line source of \( ^{99m} \)TC with an activity of 37 MBq is collimated through a fine aperture of less than 0.5 mm.

10-15K counts are acquired for the peak channel of the LSF and with at least 10 data points on either side.

Then the line source is moved to various position to find out the intrinsic resolution.

The LSF of the system used for the study is shown in the Fig. 3.2. The line is moved to different position along the X and Y axis and the profile along the line is displayed.
Fig. 3.2 Profile of Line Spread Function Measured in a Flood Field in the X and Y axis.
3.2.3 System Resolution

The quantitative measurement of system resolution may be made with the help of parallel line sources placed on the surface of the collimator or at various distances from it with and without the scattering medium. The measurement of FWHM is similar as described in intrinsic resolution measurement.

A computer technique using multiple line source phantom may also be employed.

The mathematical parameter for spatial resolution is modulation transfer function. It is expressed as frequency function by taking Fourier transform of LSF (Sorenson and Phelps 1987, Chandra 1992).

The concept of MTF portray how much of the contrast at a specific resolution is maintained by the imaging process. In the optimal case, the MTF value is 1 meaning that object and image contrast are identical.

However, this information is not stated in the spatial domain with units in cm but in its Fourier domain with spatial frequency units of 1/cm. The MTF usually starts with a value 1 at 0 spatial frequency which represents a homogeneous background. It then drops in a system specific manner down to zero. By using the MTF, two systems can readily be compared. At each spatial frequency the system with higher MTF maintains better contrast.

The Fourier transforms are characterised by magnitude and a phase angle. This modulus is expressed mathematically as
Fig. 3.3  The Quadrant Bar Phantom Rotated to 3, 6, 9 and 12'O Clock Position.
MTF varies with frequency, lesser the frequency higher the value of MTF. The resolution is improved with higher MTF.

Apart from quantitative measurement spatial resolution by qualitative measurement may be done. A quadrant bar phantom with different spacing of lead bars in four quadrant is imaged with collimator and is shown in Fig.3.3. These phantoms with different bar width and spacing are available. A point source of Tc99m (30-40) MBq was used and placed on the central axis of the detector at a distance 2 m for intrinsic resolution. The image of the phantom was acquired for 2 million counts and stored for estimation of intrinsic resolution. Whereas the flood phantom filled with uniform source of $^{99m}$Tc 150 MBq was placed over the quadrant bar phantom for system resolution.

The bar phantom was imaged for different orientations. At least two images, one in a given position and the other by rotating it through 90° must be acquired, whereas as in our case. The phantom was rotated to 3'O clock 6'O clock and 9'O clock position and a total of 4 images were acquired and the same is shown in the Fig.3.3.

The measurement was taken with 20% PHA window centered on the photopeak. By visual inspection one can determine the smallest bar spacing that can be resolved in X and Y direction with the phantom on the surface of the collimator.
Fig. 3.4 Uniform Flood Field and Flood Field of Improper Gain of PMTs.
The FWHM was estimated by the method described in IAEA-TECDOC (1991).

\[
\text{FWHM} = 1.75 \times B
\]

where \(B\) is the smallest bar spacing the cameras could resolve. In our system it was found to be 3.25 mm. So the system resolution is \(1.75 \times 3.25 \text{ mm} = 5.69 \text{ mm}\).

### 3.3 UNIFORMITY

It is the most important performance parameter of the camera. The image was acquired by flooding the detector with a uniform source of the radionuclide with 20% window opening. Ideally the gain of all the PMT’s must be same. Due to non-linear response of the detector or different gain of the PMTs. The acquired image may not be uniform. The variation in gain will result due to the aging of the tube at its own rate. So it is necessary to readjust the gain of the PMT periodically.

Figure 3.4 shows the flood field uniformity of the system. Four different flood fields of the system. One of the flood fields show the uniform and acceptable performance of the system, all the other three flood fields show the erratic behaviour of the PMTs which need to be adjusted to get a uniform field.

The width of the photopeak depends upon the adjustment gains of the PMTs. The combined peak width from all PMTs should be such that the peak of each PMT should fall within the composite peak.

For this reason only 20% window is normally centred on the 140 KeV peak of \(^{99m}\)Tc or \(\pm 10\%\) so that this window will cover all energies from 126 to 154 KeV.
All scattered photons either from the detector or from the patients are included in this window. Some manufacturers designed cameras with asymmetric window which operate without loss of uniformity (Graham et al 1986, Halama et al 1988). Newer version Gamma cameras allow still narrow window by employing correction circuits whereas the Gamma cameras used for this research work, the minimum window possible was only 20% or ± 10% only.

Some of the latest version of the cameras contain a pulsed light source fiber optically fed to each PMT so that the individual PMT gain is adjusted every few seconds so that there will not be any drift in gain. The Apex Elscint camera also has a light source for auto adjusting but manual intervention is required to set the process of adjustment on and it is done only as and when required and not in every few seconds as in the latest version. This technique will be more useful in SPECT as it eliminates the PMT gain variation with orientation of the rotational SPECT (IAEA TEC DOC 1991).

### 3.3.1 Uniformity Measurement

The uniformity is a measure of the slightly different response of different areas of the detector to irradiation by uniform source. Although an image obtained on photographic film is adequate for daily assessment quantitative data from a digital image are necessary for comparison purpose. It is the variation in pixel count values which is analysed. It is usual to carry out the calculations for both the Geometrical Field Of View (GFOV). The whole usable field of view and the Central Field Of View (CFOV).

The CFOV is defined as an area, centred on the GFOV having linear dimensions of the GFOV scaled by a factor of 0.75. In practice, for ease of calculation: the CFOV is defined as a circle having a radius of 75%
Fig. 3.5 Uniformity Measurement with NEMA Protocol.
of the largest circle which it is possible to inscribe within the GFOV, as shown in Fig.3.5, it shows the NEMA methods adopted to measure integral uniformity of the fields for two different GFOV and CFOV.

For both GFOV and CFOV, the mean and standard deviation of the pixel counts are calculated. The Coefficient Of Variation (COV) of the pixel counts is given by $100 \times \frac{SD}{C_m}$ where $C_m$ is the mean pixel count.

The integral non-uniformity $U_i$, a measure of the range of pixel value, is given by

$$U_i(+) = 100 \frac{C_{\max} - C_m}{C_m} \%$$  \hspace{1cm} (3.3)$$

$$U_i(-) = 100 \frac{C_{\min} - C_m}{C_m} \%$$  \hspace{1cm} (3.4)$$

where $C_{\max}$ and $C_{\min}$ are the maximum and minimum pixel counts and $C_m$ mean pixel count respectively. The differential non-uniformity $U_d$, measuring the rate of change of pixel count values, is defined by

$$U_d = 100 \frac{c/m}{\%}$$  \hspace{1cm} (3.5)$$

3.4 ENERGY RESOLUTION

The energy resolution can be calculated with multichannel analyser (MCA). The peaks of various radionuclides may be checked at 20% window. The same was checked with three different sources. a) $^{99m}\text{Tc}$, b) $^{67}\text{Co}$ and $^{131}\text{I}$ sources. The energy resolution is expressed as a percentage of gamma ray energy. The energy resolution of for $^{99m}\text{Tc}$ gamma ray is around 11% and is about 7% for $^{137}\text{Cs}$.

For our system the energy resolution was found to be 6% for $^{99m}\text{Tc}$. 
3.4.1 Energy Correction

The spatial variation in energy response contributes significantly by to the non uniformity in scintillation camera. This results in a shift in photo-peak pulse height relative to the energy window selected. These variations are due to improper tuning of PMTs or changes in the electronic component.

Another cause may be the optical design where efficiency of light collection varies as a function of position (Muchllehner et al 1980). The manufactures of gamma cameras normally mention their method of energy correction in the manual. Most of the modern systems incorporate auto tuning for PMT gain during acquisition itself (Steidley and Kearns 1978, Halama et al 1988). The procedures slightly vary from each system (Simmons 1996), where as both the systems used for the research did not have the auto tuning facility.

3.5 SPATIAL LINEARITY

One of the major reason for the non uniformity is the mispositioning of the events by the electronics of positioning circuit (Wicks and Blan 1979). The camera non-uniformity is mostly because of mispositioning of events than the photodetection efficiency of the crystal.

3.6 COUNT RATE CAPABILITY

The count rate performance of the system can be evaluated both intrinsically and extrinsically.
3.6.1 Absorber Calibration

The absorbers are first calibrated with respect to their attenuation for $^{99m}$Tc gamma rays at ± 10% window without collimator and with lead mask at the centre of the detector housing. The background counts were recorded and then the source was positioned on the central axis of the detector at a distance of 1.5 m from it (IAEA TEC DOC 1991). Fifteen copper sheets of 10 x 10 cm with a thickness of 0.5 mm were selected. Absorbers 13,14 & 15 were kept on top of the source holder. The count rate was always kept at 2Kcps.

The absorber were added one by one and count rate with each added absorber were noted. The attenuation factor for each sheet was calculated.

The attenuation factor is the ratio of transmitted gamma rays to those incident upon the sheet.

After the calibration of the absorber, the source activity was increased. Counts were observed for 200 seconds. The time of the day was noted for the mid point measurement for applying decay correction and net count rate less background was recorded say $(C_0)$. At this low level the input $R_0$ and the observed $C_0$ must be equal. Then the absorbers were removed one by one and the counts for a preset 20 seconds was noted.

Let the net count rate after background subtraction be $C_r$.

This continued till the last three absorbers 13-15 alone were in place.

Initially with all absorbers in place $R_o = C_o$.

After removal of one absorber $R_i = C_o' / F_1$ where $C_o'$ is the observed count rate with one absorber removed and $F_1$ is attenuation factor for
absorber 1. The same value of attenuation can be used if all the absorbers were of same thickness. In our case all the absorbers were of the same thickness of 0.5 mm each. The results are given in Table 3.1.

A graph was plotted to show the relation between observed and input count rate Fig.3.6.

From the graph it is easy to determine that the maximum observed count rate is with 20% loss (C=0.8R).

This test should be done as an acceptance test but in our study we did this as a precursor to ensure that the system performance was not interfering with the processed result, which is described in latter chapters.

3.7 SYSTEM SENSITIVITY

This is one of the most important tests in the count rate performance of the system. This test is to see the response of the scintillation camera with a given collimator to a known activity of radionuclide in common use.

A plastic container with a dia of about 15 cm containing about 40 MBq (1 mci) of $^{99m}$Tc solution is placed over the collimator facing vertically upwards. The counts of 20K were acquired with 20% PHA window centered on the photo peak. The count rate and the exact time of the day corresponding the mid point of the experiment was recorded to apply decay correction to the isotope.

The background count rate was determined. The test for all the multihole collimators in use at the centre was repeated. Similar but a separate petridish may be filled with $^{121}$I source solution for medium energy collimator. The count rate was recorded at ±10% PHA window.
Fig. 3.6 Relationship between input and observed count rate
TABLE 3.1 CALIBRATION OF ABSORBERS

Background: 640 cps

<table>
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<tr>
<th>Identity of the added absorber</th>
<th>Time of day</th>
<th>Count rate without added absorber</th>
<th>Count rate with added absorber</th>
<th>Attenuation factor</th>
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<td>1470</td>
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</table>

0.616498
The count rate was corrected for background and also for the radioactive decay and then the system sensitivity for a given collimator was expressed as

Counts/sec/MBq or
Counts/min/mCi

The values for the system was found to be within ± 6%.

3.8 EVALUATION OF THE COLLIMATOR

There are many books, manuals and literature pertaining to the entire gamut of checks to be performed on a scintillation camera but the collimator evaluation was not given its due attention till recently. The collimator is the most important part of the imaging system and it is particularly vital in emission tomographic studies. Very few papers were published on this important aspect (Chang et al 1985, 1988, Busemann-Sokole 1987, Gerald 1988, Yoshizumi et al 1990).

The following are the very simple tests to be carried out for a parallel hole collimators.

1. Image of a point source with collimator.
2. Image of parallel line source.
3. Image of at least five point sources at minimum and maximum distance from the collimator.
4. Centre of rotation (COR) offset at minimum and maximum radius of rotation. This test is relevant only to the SPECT system. Our study, though was done only with planar imaging mode, this test was done as a whole some assessment of the overall performance of the system.
3.8.1 ADC Linearity

The quantitative assessment of the ADC linearity needs a pulse generator (Gerard J.G. and Alex T.E. 1992). However if the X-ADC and/or Y-ADC are non-linear the displayed images will show stripes and lines in digital image. Sub-optimal ADC performance may lead to the production of non-uniformity in the images. The value of the amplification factor can be high to a factor of twenty or greater (Todd-Pokropek 1983) close to the centre of the matrix. Less obvious losses of accuracy in quantitative measurements can also be produced (Todd-Pokropek 1983).

3.9 CRYSTAL HYDRATION

The NaI(Tl) crystal has to be hermetically sealed to protect it from getting hydrated. The air tightness of the scintillator canning may change due to aging. The canning may get damaged due to careless or rough handling and the crystal will start absorbing moisture. The hydrated crystal instead of being transparent, become yellowish. The affected portion can transmit only a small fraction of light due to change in colour. The yellow spots may be small initially and grow bigger in size with time and latter may affect the whole crystal. The image of a homogeneous flood source will appear as non-homogeneous. The hydration effect can be seen by off peak imaging technique (K-Landoori 1986). In refurbished or old cameras the hydration test should be done more frequently (Lukes et al 1983). So the cameras were subjected to very stringent quality assurance tests and the results are recorded and discussed in this chapter.

Once we were thoroughly satisfied with the performance of the Gamma cameras the patients were subjected to Radio-nuclide ventriculography.
For doing the radionuclide ventriculography a series of steps needs to be performed before the actual imaging and data acquisition is done. The radiopharmaceuticals most commonly used is $^{99m}$Tc in vivo labelled red blood cells, though many other methods are available and were discussed in the introduction chapter. To label the red blood cells it is necessary to give an injection of cold stannous pyrophosphate and then 20-30 minutes latter $^{99m}$Tc.

3.10 PREPARATION OF THE ISOTOPE

$^{99m}$Tc is produced from generator. All radionuclide generators work on the principle that a parent nuclide decays to produce a daughter nuclide, where the chemical nature of parent and daughter are quite different. This difference allows separation of the daughter from the parent. Thus generators allow short half-life radionuclides to be supplied to centres distant from the reactors.

The molybdenum/technetium generator consists of $^{99}$Mo absorbed on to an alumina-filled column, the amount of $^{99}$Mo being dependent on the activity rating of the generator. The $^{99}$Mo is present on the column in the form of $^{99}$MoO$_4^{2-}$. $^{99}$Mo decays to its daughter product $^{99m}$Tc in the form of pertechnetate.

$^{99m}$Tc is removed from the column in the form of sodium pertechnetate, Na $^{99m}$Tc O$_4$ by drawing a solution of sodium (0.9%) chloride through the column. This process is called the elution. This is only from the readymade sterile generator or column generator as it is called. This is supplied by the company and the elution process is very simple but costly.

The other cumbersome but the cheapest method is solvent extraction method. This was the procedure followed for this study. In our
centre we are following only solvent extraction method. Many of the other centres are switching to column generator method.

To elute $^{99m}\text{Tc}$ through the solvent extraction method the following procedures need to be followed:

a) The Technetium $^{99m}$ ($^{99m}\text{Tc}$) generator is washed thoroughly and rinsed with methyl ethyl ketone (MEK).

b) Sodium hydroxide (NaOH) and potassium chromate is mixed with 100 ml of distilled water and the solution is allowed to cool to the room temperature as the reaction is exothermic, it will produce heat.

c) 30 ml of methyl ethyl ketone is taken in the $^{99m}\text{Tc}$ generator through the top opening.

d) 30 ml of sodium hydroxide, potassium chromate along with $^{99m}\text{Mo}$ is sucked into the $^{99m}\text{Tc}$ generator through the side opening using a vacuum pump. The total volume of all i.e., sodium hydroxide, potassium chromate and $^{99m}\text{Mo}$ should be equal to 30 ml.

e) The above solution is mixed thoroughly using the vacuum pump and by means of a bellow for 3 minutes and the mixed solution is allowed to stand for another 3 minutes.

f) Through the toggle two way opening in the bottom of the generator the yellow coloured highly viscous solution is collected in a bottle and kept separately for the next day use.
g) The remaining colour less, light weight and free flowing liquid is collected in a beaker.

h) The beaker is heated to dryness until the liquid just evaporates. If the beaker is allowed few seconds extra will have the beaker charred. The beaker is cooled to room temperature.

i) Wash the walls of the beaker with 8 cc of 0.9% sodium chloride solution (Normal saline). The washing is repeated few times.

j) The solution in the beaker, if colorless, is collected in a vial. The solution will be brown or dark brown in colour if the beaker is heated too much and result in colloid formation and the labelling will not be proper.

k) The vial containing the NaCl is sterilised for half an hour in an autoclave or in a pressure cooker.

l) The vial is then cooled to room temperature and the isotope is ready for the preparation of cold or hot kits for administration. As this involves lot of procedure, the yield of the activity depends upon the expertise of the technical person and refinement of the technique used.

3.11 PATIENT PREPARATION

a) The chest of the patient is shaved or thoroughly cleaned with a scrubber to fix the ECG electrode as it serves as the physiological synchroniser.

b) The patient is given 5-10 mg of potassium perchlorate orally. This is done because if the labelling of the $^{99m}$Tc with the red
blood cells are not complete, the $^{99m}$Tc will remain as free technetium in the blood circulation and will go to thyroid and gastric mucosa and spleen and thereby interfere with the imaging process and also will contribute radiation dose to thyroid, G.I. tract and spleen. So the potassium perchlorate as a competitive inhibitor will occupy the places of free technetium and $^{99m}$Tc will have no space to occupy and to remain in the blood circulation alone even if it is not bound to R.B.C. as labelling agent.

c) Sn-PYP (Stannous-pyrophosphate) was the radiopharmaceutical used for all our studies as labelling agent (DTPA), Diethylene Triamine Penta acetic Acid may also be used as labelling agent.

Sn.PYP was mixed with saline and one kit was used for three patients. The Sn-PYP was given by I.V to the patients the reconstituted kit should be used within one hour from the time of reconstitution. Though the manufacturer insist that the kit should be used within an hour from the time of reconstitution, from our experience we found the BARC supplied Sn-PYP kit, if kept in the freezer, could be used even after 24 hours from the time of reconstitution without compromising the labelling efficiency.

d) 740 MBq (20 mCi) of $^{99m}$Tc in the form of sodium per technetate was given to the patient intravenously after 15-20 minutes from the injection of Sn-PYP.

To ensure optimum electrical contact between the skin and electrodes the following steps should be followed:
a) Shave any hair at the site of the electrode.

b) Skin should be rubbed until it appears pink with a fine sand paper or abrasive pad that has been saturated with alcohol. The alcohol prepares the skin by removing the surface oil while the abrasive pad removes the dead skin.

3.11.1 Electrode Placement

The most suitable electrodes have a foam backing and are pre-jelled and have been proven to minimise electrocardiograph interference. Always check if centre of the electrode is moist. Electrodes with dried gel or cracked edges will result in poor ECG tracings.

a) Always the edges of the Electrode to be applied first and then the centre of the electrode and then the electrode to be pressed firmly to the skin. If pressure is applied at the centre then the gel spreads uniformly and ensures better contact. If the centre is applied first and then pressed the gel will leak out of the electrode and cause inadequate bond.

The electrodes should be placed in such a way that they are not in the field of view. Leads may cause attenuation during data acquisition, if they are placed over the cardiac compartments.

Electrodes with plastic applicators may be put over the heart directly.

The electrodes are connected to ECG monitor through lead wires, a base line tracing need to be done to check for artifacts. The electrodes may be repositioned if necessary.
Leads II, avf and V₅ are most commonly used as they are more sensitive to ischemic S.T.changes.

In order to obtain proper gating one has to choose ECG lead with tall R-waves. If tall T waves are present they may produce double gating. If this is the case one must select a different lead.

3.12 IMAGING PROTOCOL

For rest imaging multiple gated equilibrium studies are performed in multiple views. Usually anterior (Straun, 1977) and 40-45° LAO views are obtained. 70° LAO, left lateral and LPO views may also be performed. The ventriculographic studies which permit the isolation of the left ventricle from the other cardiac chambers. The RAO view provides the largest ventricular silhouette. However in equilibrium radionuclide studies, since all chambers of the heart are filled with radioactivity and since the right ventricle is partially superimposed over the left ventricle in the RAO view, so RAO view presents a problem in interpretation, especially in the inferior portion of the left ventricle (Freeman et al 1981). In the anterior view this overlapping is minimised. The anterior view improves the resolution and increased count rate also as the collimator is placed very close to the heart. The 70° LAO view further improves the detection of inferior segmental wall motion abnormality, although it is less specific than the anterior view in determining the normal interior wall motion (Freeman et al 1981). In the light of the above reasons we restricted our study only to two views i.e., 45° LAO and Anterior view.

For determination of LVEF the 45° LAO view was utilized, since this projection best separates the left-ventricle from the other cardiac chambers as shown in Fig.3.7. This view is also used for visualizing segments of the left ventricular wall not seen in the anterior or RAO
Fig.3.7  45° LAO View Delineating Left Ventricle.
Fig. 3.8 16 and 24 Frames Display with 64 x 64 and 128 x 128 Matrix.
projection for doing this view it is important to choose a degree of obliquity that provides the least overlap of the ventricles.

In the majority of cases, in the 40-45° LAO the septum is viewed entirely on end, optimally separating the ventricles. Adding caudal tilt to the 40-45° LAO is helpful in separating the atria from the ventricles (Matin P and Kirss JP 1970). This tilt can be achieved either by angulating the detector or by employing a slant-hold collimator. In this study as we did not have a slant-hole collimator only parallel hole collimator was used and the detector head was angulated in the caudal direction.

The complete imaging protocol used in our laboratory was as follows. A radioactivity of 740 MBq (20 mci) of $^{99m}$Tc was given to the patient as discussed in detail already.

Synchronised multigated acquisition (SYMA) was performed in the anterior and 45° LAO projections and some cases left lateral view was also acquired but that view was not used in this study.

The mobile scintillation gamma camera ELSCINT APEX-4 and MAX4 SPECT Gamma camera equipped with low energy, high resolution parallel hole collimator, an R.wave gating device and a computer was employed. Our computer system was capable of acquiring 16, 24 and 32 frames with 64 x 64, 128 x 128, 256 x 256 and 512 x 512 matrix. Though literature recommends the 14 frames during the first two thirds of cardiac cycle (Maddahi J et al 1979), we do the image acquisition in the 24 frames mode at 128 x 128 matrix as a routine protocol for rest imaging. For exercise MUGA only the frames were reduced to 16 and matrix to 64 x 64, as shown in Fig.3.8.

A total of $6 \times 10^6$ counts were collected with ±10% PHA energy window centred on the 140 KeV photopeak of Technetium-99m.
Fig. 3.9 Method of Generation of Multiple Gated Images (Courtesy: Freeman 1980. Ref No. 24).
3.13 DATA ACQUISITION

The cardiac functional studies require an acquisition mode that has the following features (Liebermann D.E 1977, Bacharach S.L et al 1979).

1. high temporal resolutions
2. capability of displaying data during or immediately after acquisition with minimal manipulation and
3. Minimal memory storage requirement

Multiple gated acquisition employs a variable number of frames (14-100) depending upon the processing software and manufactures but now the majority of the systems has 16, 24, 32 frames etc., and a physiologic marker R wave as synchronizing signal. The scintigraphic data from the camera are channeled to a series of image frames located in the computer core memory.

Allocation of a count to a particular image is governed by the time delay between the R wave and the count immediately after the trigger, the data from the camera are placed in frame 1 for a fixed duration of time Fig.3.9.

This time duration is determined by the system for every patient. Once the ECG leads are connected and the acquisition is started the system waits for the patients ECG to stabilise and the time between two consecutive R-R waves are determined, the R-R interval in milli seconds is divided by the number of frames chosen for that acquisition.

For example if the R-R interval is 16 milli seconds this is divided by 24 if the no. of frames chosen for recording the data is 24.
Time duration for each frame is $16 \text{ ms}/24 = 0.66 \text{ ms}$.

So the data acquired for the first 0.66 ms will be placed in the first frame, when this time interval has elapsed, scintigraphic data are then channeled to frame 2. The time interval is also called as time per frame, this process continues up to the last frame (N). When the time interval has elapsed, the remaining data from the camera are discarded until the next trigger signal arrives. Data from the successive beats are added to the appropriate frames until the desired total count is achieved. At least 20 frames per cycle is required for accurate E.F. measurement (Hamilton G.W et al 1978). If a system is limited to 14 frames, these should be divided over the initial two thirds of the cardiac cycle to maintain good accuracy of E.F. An important aspect of multiple gated acquisition is that the gating signal be an accurate representation of the onset of systole (truly end-diastole) and that the interval between gating signals remain relatively constant during each study.

Proper selection of matrix type and size is another major consideration for acquisition and storage of multiple gated equilibrium scintigraphic studies. In order to assess the regional wall motion, the highest possible digital resolution should be achieved. As a general rule a minimum size of 64 x 64 matrix size is recommended.

3.14 QUANTITATIVE ASSESSMENT OF LVEF

Ejection fraction is considered to be the single most representative index of global ventricular function. This index represents the fraction of the end diastolic volume of the ventricle that is ejected with each beat, and is equal to the stroke volume divided by the end diastolic volume.

Though few other methods are also available for the assessment of LVEF the most preferred method of scintigraphic measurement of EF is
Fig. 3.10  A Normal Time Activity Curve.
area-counts technique. This is the proportion between the volume of the cardiac chamber and the number of counts emitted from that chamber. A background corrected ventricular activity versus time curve is a relative volume versus time curve of the corresponding ventricle (Green MV et al 1975 and Luig H et al 1974).

From this curve, EF can be measured as a function of the peak and the end systolic volume. A typical normal time activity curve is in the Fig.3.10.

$$\frac{EDC - ESC}{EDC}$$

where EDC and ESC are the background corrected counts in end-diastole and end-systole.

Left ventricular ejection fraction calculation (LVEF) is determined from the relative number of counts in the end diastolic and end-systolic frames and requires assignment of left ventricular region of interest and background correction.

### 3.14.1 Region of Interest (ROI)

In order to obtain counts from the ventricles, nuclear medicine computer systems enable the operator to select the small portions of an image and extract data concerning the relative concentration of radionuclide in that region. Areas selected for this purpose are called Region of Interest (ROI). Once defined the ROIs are used to

1. analyse the concentration of activity within the ROI and
2. derive the changes in concentration of activity in the ROI as a function of time and this is known as time activity curve.
3.14.2 Determination of Background

The background is the radioactivity which is counted in the ventricular ROI but did not originate from the ventricular chamber. This radiation comes from the blood in the tissue in front of and behind the left ventricle as well as from scattered photons originating in areas alongside the ventricle.

In our studies the range of values determined and used for deciding the left ventricular function in terms of percentage ejection fraction is as given below:

- Normal LV function → 45% & above
- Mild LV dysfunction → 40% - 45%
- Moderate LV dysfunction → 35% - 40%
- Severe LV dysfunction → below 35%

3.15 LEFT VENTRICULAR SEGMENTAL WALL MOTION

In all the patients who underwent the doppler and radionuclide ventriculography, the scintigraphic method of left ventricular segmental wall motion was also done. The anterior and 45° LAO views were utilised. From the combined view a total of 9 segments were assessed by both Doppler and radio nuclide ventriculography.

The segments were
1. Basal septal
2. Apical septal
3. Antero lateral
4. Antero basal
6. Apical
**MULTIPLE GATED BLOOD POOL IMAGING ANALYSIS**

<table>
<thead>
<tr>
<th>Segment</th>
<th>No</th>
<th>Score</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>ANTERIOR</strong></td>
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<td></td>
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<tr>
<td></td>
<td>3</td>
<td></td>
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<tr>
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<td></td>
</tr>
<tr>
<td></td>
<td>11</td>
<td></td>
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<tr>
<td><strong>LEFT ANTERIOR OBLIQUE</strong></td>
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</tr>
<tr>
<td></td>
<td>2</td>
<td></td>
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<td>7</td>
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<td>12</td>
<td></td>
</tr>
<tr>
<td><strong>LEFT LATERAL</strong></td>
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<tr>
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<td>5</td>
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<td>12</td>
<td></td>
</tr>
<tr>
<td><strong>LEFT POSTERIOR OBLIQUE</strong></td>
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</table>

**Scores**
- 1 = dyskinetic
- 0 = akinetic
- 1 = severely hypokinetic
- 2 = hypokinetic
- 3 = normal
- NV = not visualised

Evaluation form used for semiquantitative analysis of regional wall motion. Each of the 12 segments is scored on a 5-point scale, allowing categorization of regional left ventricular performance. This approach allows standardization of segmental analysis.

Fig.3.11. The scoring chart for the Regional wall motion abnormality.
The evaluation of chamber size and regional ventricular wall motion was performed by inspection of data in movie format.

We followed a scoring system which allowed semiquantitative assessment of segmental wall motion from the data as visualized in the movie format. The scoring chart followed in our laboratory is shown in Fig.3.11 with this system the scoring is in a 5 point scale and each segment is assigned a value in the score chart depending upon the mobility exhibited by that segment, a score of 3 represents normal wall motion (Normokinesis).

3 = normal
2 = hypokinetic
1 = severely hypokinetic
0 = akinetic
-1 = dyskinetic (Madalahi J et al 1978).

Quantitative methods for assessing segmental wall motion with multiple gated equilibrium scintigraphy were also studied in many centres (Freeman 1980, Verba et al 1979).

The results obtained are tabulated and given in the tables.