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Evaluation of Toshiba GCA-901A gamma camera performance by using a material filter (Cu 0.125 mm) in Tc-99m conventional imaging

Dissertation submitted in partial fulfillment for the award of the Bachelor's Degree of Health Science in Medical Radiation

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2004

CERTIFICATE

This is to certify that the dissertation entitled

" Evaluation Of Toshiba GCA-901A Gamma Camera

Performance By Using Material Filter (Cu 0.125 mm)

In Tc-99m Conventional Imaging"

is the bonafide record of research work done by

Ms Ismaliza Bt Isa

During the period from August 2003 to March 2004

under my supervision.

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ABSTRACT

The effect of physical filter (Cu) on the performance of gamma camera (Toshiba GCA 901A) by reducing the scattered radiation has been investigated in planar imaging with Tc-99m radioisotope. The copper filter consisted of sheet of thickness 0.125 mm with 57 cm length and 42 cm width. It was placed on the face of the collimator. The Tc-99m is used to obtain the experimental data. A symmetrical photopeak energy window (20%) has been centered at 140 keV for collection of data. In this study, low energy parallel-hole collimators are employed. They are low energy general purpose (LEGP) and low energy high resolution (LEHR). The system's imaging ability has been investigated via the analysis of uniformity, sensitivity and spatial resolution parameters. Results are compared with copper filtered data and those obtained from without the material filter data. From the overall results, data shows an improvement in all tested parameters. In conclusion, the physical filter (Cu) used in this study may be advantageous in clinical studies.

INTRODUCTION

Planar imaging or conventional imaging that collect and present all data in a single plane. This is done by camera which is able to obtain an image of whole body with one acquisition. Planar acquisition of data are also the basis of many types of tomographic imaging.

A planar imaging system requires the informations about the direction in which the photons was traveling upon striking the crystal detector, the location of its interaction with the detector and the energy of the photons.

The direction of travel and point of interaction define a line somewhere along which the photons must have been emitted. Photons energy is used to discriminate those photons that have scattered between their site of emission and detection and those that have not (Bernier DR *et al*, 1997).

Scattered radiation is one of the major problems in Nuclear Medicine imaging (Volkmar Wirth, 1989). Normally, the scattered radiation reduces the contrast of the images, spatial resolution and sensitivity. Its generally treated as a corrupt or unwanted components which should be substracted from the detected data. While, theoretically, this is true, some fraction of those photons does contain useful information (Shah SI *et al*, 1998).

It is possible to differentiate scattered and unscattered gamma photons because some of the energy of gamma photons is lost in the process of scattering. If the energy lost by scattering is greater than the resolution width of the detection system, the scatter event can be rejected by energy discrimination by pulse height analysis.

Scattering of gamma photons depends on their energy and the energy of the scattered photons depends on the scattering angle through which they are deflected. The dominant interaction mechanism for gamma photons above 100 keV in tissue is Compton scattering. In the object, gamma photons scatter in all directions.

The useful (unscattered) photons are those that enter the holes of the collimator. Photons that are scattered multiple times or that are scattered at large angles lose enough energy to allow adequate rejection by means of pulse-height discrimination. Photons which scattered at small angle or moderate angle lose only a small amount of energy and contribute to the detector background in the sinogram and in the patient images. The collimator will reject events that do not line up with the holes (Mettler FA *et al*, 1998).

REVIEW OF LITERATURES

Since last three decades, physical filters have been used to improve the performance of scintillation gamma cameras. In 1975, Muehllehner applied this technique to positron emission tomography (PET) by employing a filter which consisted of 1.27 mm of lead, 0.76 mm of tin and 0.25 mm of copper. With this filter, a factor of five increase in useful count rate was reported. Ficke and Ter-Pogossian (1990), suggested that material filters might be advantageous in reducing low energy radiation originating in the field of view of the scanner. They analysed the spectrum from the NaI(TI) detectors by using 0.43 mm and 0.86 mm thick lead filter in PET. As a result of applying these filters low energy events were reduced by some 30-40% for the loss of 7-12% photopeak (unscattered) events.

Futher, Spinks TJ and Shah SI [1993] worked on the basis of the above ideas, to investigate the effects of lead (Pb) filter (0.5 mm and 0.1 mm thick) on the overall performance of a Neuro-PET multi-ring (CTI/Siemans 958B) Tomography. The aims were to determine the effects on (a) the singles rates (and hence dead time and the energy spectrum) and (b) the noise equivalent counts (NEC) which reflects the statistical quality of acquired data. [NEC is a function of measured true and random rates and the scatter fraction]. In their study, a significant reduction in low energy events (50-200 keV) was obtained, as well as a decrease in random rates and percent dead time. However, in spite of the reduction in dead time and low energy events in the spectrum with lead filters, no advantages was seen to be gained in the statistical quality of the data determined by NEC.

Harshaw Scintillation Phosphor [1975] has indicated that it could be possible to design an appropriate absorbing filter by combining various detector materials. Later, Strand and Larson [1978] published a paper in which they suggested that it may be possible to reduce the recording of undesirable (scattered) photons by means of specially made attenuating filters. In 1986, Pillay and his colleagues applied an alloy (composite filter, consisting of Pb, Zn and Sn, in single photon emission (planar) imaging. From various patient studies they claimed improvement in the quality (contrast) of images. Further, they mentioned in their paper that it could be possible to employ the technique in single photon emission tomography (SPET) as well.

More recently, Pollard *et al* [1992] employed lead sheets for imaging therapeutic doses of I-131 labelled with monoclonal antibody. They attached 1.6-6.4 mm thick Pb sheets to he front face of the gamma camera during patient scanning. A Monte Carlo study was also undertaken by them, simulating a gamma camera and a sheet of lead material mounting on it. They found that the system resolution is degraded with Pb sheet. However, they could not apply the technique in SPET imaging because of absorber's weight. Although they suggested the used of such absorber in single photon emission tomography to reduce the detection of unwanted photons from the data, the application was limited only to the situation where the camera system is able to accurately rotate the additional weight.

Shah SI. *et al* 2000 have investigated the GE 400 XC/T system's imaging ability via measurements of spatial resolution, modulation transfer function, single slice sensitivity and numerical analysis of a uniformly filled Tc-99m cylindrical phantom's reconstructed image.

OBJECTIVE OF THE STUDY

To investigate the performance parameters such as uniformity, sensitivity and spatial resolution of Toshiba GCA-901A gamma camera without and with material filter.

MATERIALS AND METHODS

Gamma Camera

Instrument used in Nuclear Medicine (NM) for the detection of gamma rays is known as the Gamma Camera. The model of digital gamma camera used in Nuclear Medicine Department, HUSM is Toshiba GCA-901A/HG. The basic components making up the gamma camera are collimator, detector crystal, photomultiplier tube array, position logic circuits and computer system.

A gamma camera converts photons emitted by the radionuclide in the patient into a light pulse and subsequently into a voltage signal. This signal is used to form an image of the distribution of the radionuclide.

Gamma cameras may be classified as either analog or digital types. An analog signal is used throughout the analog camera; this signal has an infinite range of values and is inherently noisy. A digital signal, on the other hand, only has a discrete number of values. Most of the newer cameras incorporate digital features like in Nuclear Medicine Department HUSM.

The main advantages of the digital cameras are that they are much faster, can interact directly with the computer, and generally require less maintenance. Even the most advanced digital cameras, however, start with the analog signal in the scintillation crystal and return to an analog for the CRT display of the image (Bernier DR *et al*, 1994).

Probably the most important portion of any gamma camera computer system is the software. The software used in Nuclear Medicine department is GMS-5500. The GMS-5500 software is shared by the medical image processing system GMS-5500A, the digital gamma camera models GCA-901A.

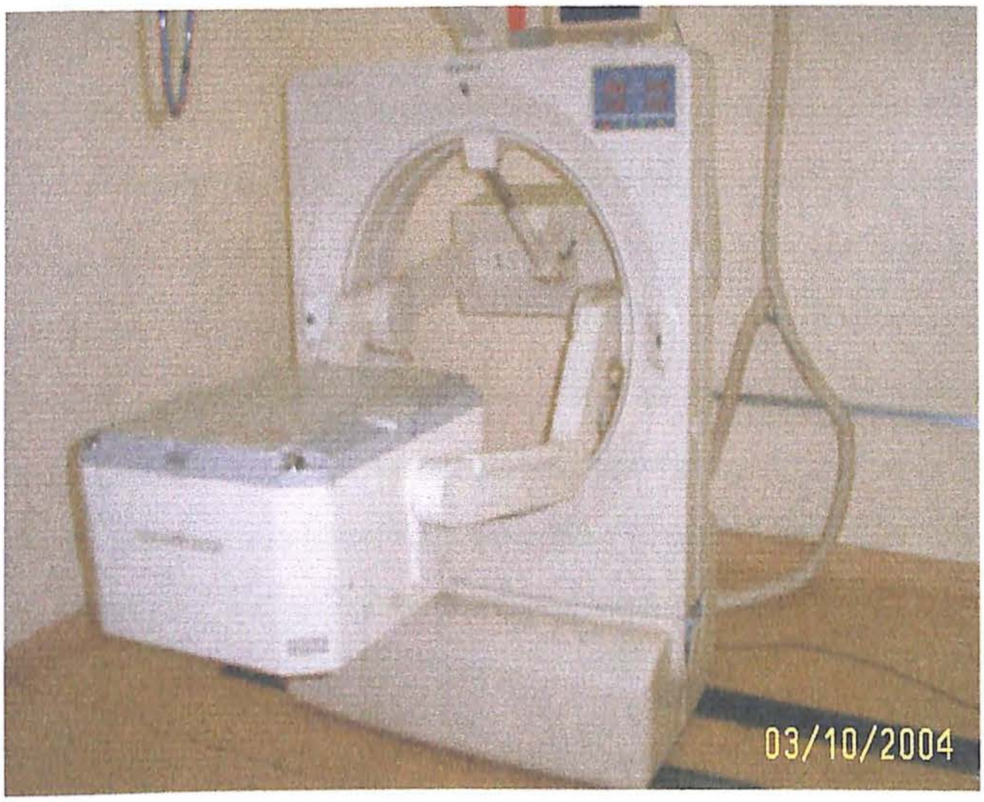


Figure 1 : Photograph of imaging unit used in this research (GCA-901A)

Collimators

To form the image of an object with a gamma camera, it is required to project gamma-photons from the radioactivity onto the scintillation detector of the scanning system. For imaging, collimators are mounted on the face of the detector of the gamma camera and are interchangeable depending on the type of the study and the energy of the radionuclide to be used. A collimator allows only those gamma-photons traveling along certain specified directions to reach the surface of the crystal of a scanning system. Those not traveling in a specified direction are not registered since they are stopped by the collimator before they reach the detector. In fact, this is the only way in which the camera can achieve reasonable spatial resolution in the image. This procedure inevitably results in a significant decrease in the number of detected gamma photons, resulting in reduced sensitivity, or increased acquisition time. The precise collimation of the gamma-photon emitting radionuclides, but also to eliminate at least some of the scattered gammas which would impair contrast.

Collimators play a crucial role in defining a systems extrinsic imaging characteristics. The energy rating of a collimator indicates the maximum energy of photons that can be efficiently handled by the collimator. This is usually defined as the energy at which less than 5% of the off-axis photons pass through the collimator. Low energy collimators are designed for a maximum energy of 140 to 200 keV, while medium energy collimators are effective up to 300-400 keV. The energy rating of the collimator also dictates septal thickness. Although tungsten absorbs photons more efficiently, most collimators are made of lead due to its lower cost.

The are four basic types of collimator ; Parallel-hole, diverging-hole, converging-hole and pin-hole (Shah SI and Leeman S, 1996). The parallel-sided holes are either circular or hexagonal (standard type of collimator) and are drilled/ cast in lead (Pb) or are shaped from Pb foils. However, square and triangular holes have also been used. The holes are separated by Pb walls (septa), whose thickness is selected to stop a significant fraction of the gamma-photos crossing from one hole to the neighbouring hole. The thickness of the septa is choosen according to the gamma-photo energy. Thus, low energy collimators have thin septa and hence more holes.

In this study, low energy parallel-hole collimators are employed, namely: low energy general purpose (LEGP) and low energy high resolution (LEHR). The reason for this choice was that these collimators are the most heavily used for clinical studies in nuclear medicine departments. As mention before, the reason to use the low energy collimator is because we used the Tc-99m with low energy (140 keV). Resolution is best when the object is placed against the collimator, but the sensitivity is independent of distance from the collimator.

Resolution and sensitivity of a collimator are inversely related. The best spatial resolution is achieved with a collimators with long holes of a small diameter because the angle of acceptance is smaller and more scatter is rejected.

The sensitivity of this low energy high resolution collimator (LEHR), however, is lower because fewer photons reach the crystal. Sensitivity, or efficiency, refers to the fraction of emitted photons which actually pass through the collimator and reach the detector. The sensitivity increases as the square of the hole size, and decreases as the square of the hole length. When spatial resolution is not critical and photon flux or acquisition time is limited, a high sensitivity collimator can be used.

The low energy general purpose collimator (LEGP) is a compromise between sensitivity and resolution. When using any collimator, spatial resolution decreases with increasing distance from the patient. As a rule of thumb, resolution falls about 1 mm for each additional centimeter that a patient is positioned from the face of a parallel-hole collimator. Thus, collimator resolution improves as the diameter of the collimator holes decreases, the effective length of the collimator holes increases and the object to collimator distance decreases.

Radionuclide Of Choice

This study was carried out by using Tc-99m. Technetium-99m was discovered in 1973 by Perrier and Segre in a sample of naturally occurring ^{98}Mo that irradiated by neutron and deuterons. It was introduced into nuclear medicine in 1975 with the development of the ^{99}Mo-^{99m}Tc generator at the Brookhaven National Laboratory. Technetium was first used clinically in 1961 at the University of Chicago and Heralded a new era for nuclear medicine. The widespread use of ^{99m}Tc as the radionuclide of choice for a variety of nuclear medicine imaging procedures has been based mainly on its physical properties (Sampson CB, 1990).

Tc-99m fulfills many of criteria of an ideal radionuclide and is used in more than 70% of nuclear imaging procedures in the United States. It has no particulate emission, a 6 hour half life, and a predominant (98%) 140 keV photon with only a small amount (10%) of internal conversion.

Technetium 99m is obtained by separating it from the parent ^{99}Mo (67 hour half life) in a generator system. A ^{99}Mo-^{99m}Tc generators consists of an alumina column on which ^{99}Mo is bound. The parent isotope decay to a radioactive daughter, which is a different element with a shorter half-life. Because the daughter is only loosely bound on the column, it may be removed or 'washed off' with an elution liquid such as saline. After the daughter is separated from the column, the 'build up' process is begun again by the residual parent isotope. Uncommonly, some of the parent isotope (^{99}Mo) or alumina will be removed from the column during elution and appear in the eluant containing the daughter isotope. This is termed as 'breakthrough'.

Physical Filter (Cu)

The dimensions of physical filter are 57 cm x 42 cm. It is made of Copper material. The thickness of copper filter is 0.125 mm and was choosen because the manufacturer was able to provide sheet Cu of only this thickness. The physical filter was placed on the face of detector.

By using physical filter, it is possible to reduce the scattered radiation (lower energy radiation) from photon beam coming from the patients body before they enter the crystal. A copper filter is designed for Tc-99m (140 keV) would not be useful for other radionuclide. That means, a thicker filter could be used with higher energy radionuclide (Muehllehner and Jaszczak, 1974).

Another reason, Copper has a number of other advantages in addition to its photon attenuation characteristic: it is easily available, machinable, of low cost and easy to mount on the face of gamma camera collimator.

Figure 2 : Material filter (Cu)

Flood Field Phantom

The flood field phantom is used to accurately determine field uniformity of scintillation cameras. In this study, we used the rectangular flood phantom which have :

- Dimensions: 19" x 25" x 1.25" thick (48.3 x 63.5 x 3.2 cm)
- Cavity: 15" x 21" x .5" (38.1 x 53.3 x 1.3 cm)

The Flood Phantoms feature extra strength side walls and clear lucite for easy positioning. The phantoms are leak proof and are excellent for transmission imaging It is usually water filled, and radioactive must be added. It is easy to fill and easy to drain. In addition, phantom is a sealable, flat, thin-walled container usually made of Lucite.

It has a cavity that can be filled and then sealed. Thorough mixing of the radionuclide in the flood phantom is essential, since any nonuniformity in the distribution of radioactivity in the phantom could be interpreted as a camera malfunction. When this problem is suspected on a flood-field image, the phantom should be rotated repeatedly and a second image should be obtained. A change in the pattern between images indicates a mixing problem in the phantom.

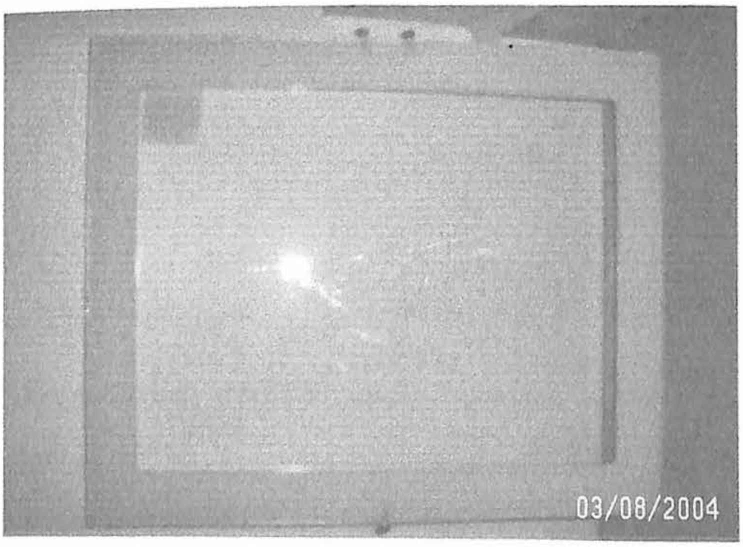


Figure 3 : Flood field phantom.

Scattering Medium (Acrylic)

The acrylic plates are equivalent to plastic water phantom and acts as a water equivalent scattering medium. The dimension size of this acrylic is 30 x 30 cm^2. This plastic water phantom, like most plastics, is flexible in nature and will creep (warp/bow/deform) under constant stress over time, even under its own weight. This deformation is not permanent. The slab will recover its original shape if placed back in its original geometry. An easy way to maintain slab flatness and minimize creep is to store the slabs, clean free of debris on or in between flat surfaces.

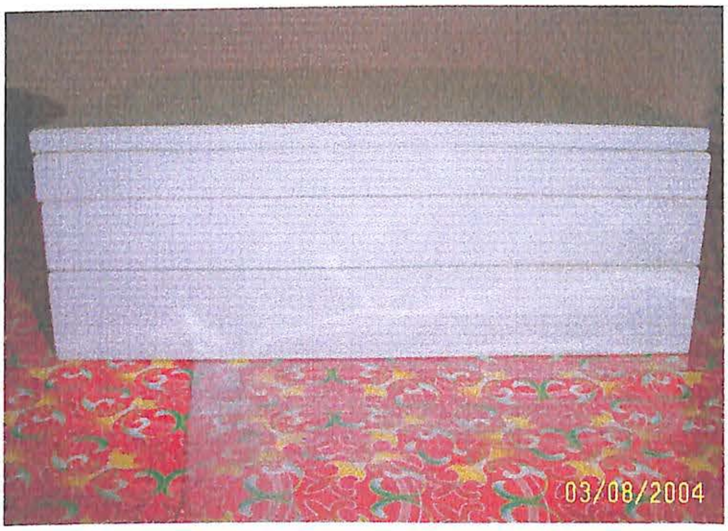


Figure 4 : Scattering medium (acrylic plates).

Data acquisition For uniformity, sensitivity and spatial resolution

The two thumb screws which located on the side of the flood phantom were removed. The phantom was filled with distilled water. About 14.193 mCi of Technetium-99m was carefully added. Then the thumb screws was tightened.

The phantom was rotated in such a manner as to allow the air bubbles to move out from the phantom. The two thumb screws were removed carefully again and the remaining volume filled with water. Then the phantom was ready to use.

The flood phantom was placed on the patient bed. The detector-collimator head was moved as close as possible to phantom.

A 128x128 matrix was selected and the total 20 million were recorded in the image. The planar images were collected by employing without and with material filter in combination with either a LEGP or LEHR Collimator. Standard energy window for Tc-99m was selected (126-154 keV).

Figure 5 : Experimental setup for extrinsic uniformity using LEGP collimator , with material filter (Cu).

For sensitivity acquisition, the 3.43 mCi of Tc-99m placed in enough solution that it fills 3 mm of the petri dish. The standard 100 mm diameter petri dish placed on the face of collimator and protected with absorbent paper. For the data acquisition, 128 x128 matrix size was choosen and the counts were preset at the total 1000 Kcounts. Then the data were acquired within a 20% window. The data were collected by employing a material filter in combination with either a LEGP or LEHR.

The material filter that we used was taped on the collimator face. The count per minute was determined. After that, the source was removed and the background was counted. The sensitivity expressed as net counts per second per milicurie. The data were recorded for comparison. A dose calibrator was used to measure the source and the empty syringe to receive the exact amount of activity used.

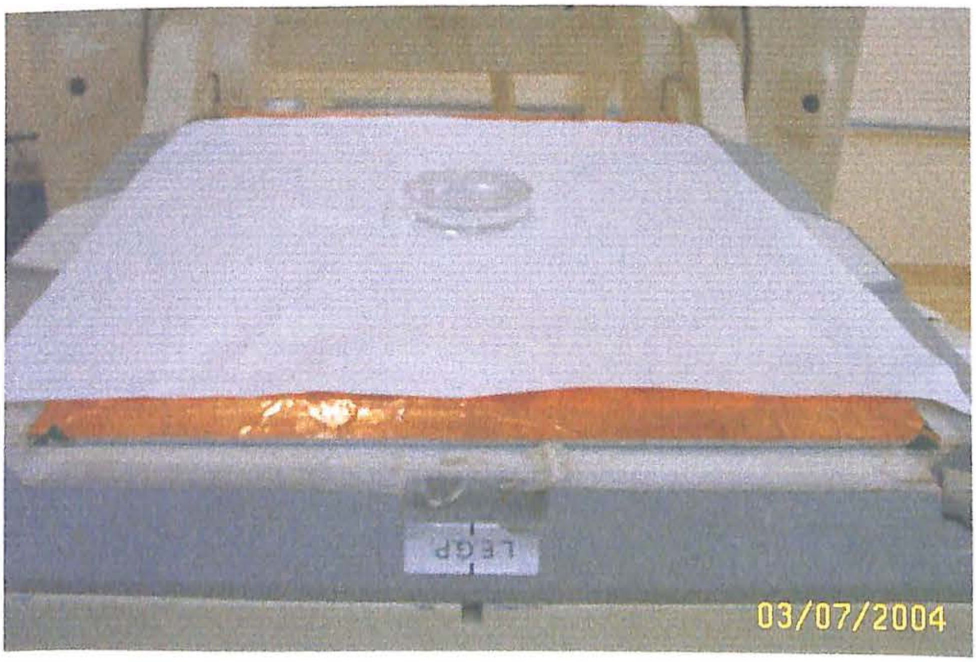


Figure 6 : Experimental setup for sensitivity using LEGP collimator, with material filter (Cu).

For spatial resolution acquisition, a polythene tube filled with 3.23 mCi (for LEGP collimator) and 16.3 mCi (for LEHR collimator) of Tc-99m solution of internal diameter 1.25 mm and 19 mm in length. The line tube was placed on the patient bed and it was protected by absorbent paper. The line source was scanned at various distances (1-10 cm with 1 cm step) from the surface of the gamma camera collimator. Then the planar images in a water equivalent scattering medium at different depth were obtained without and with material filter. The filter was carefully taped on the face of collimator.

The Acrylic plates (each 30 x 30 cm^2) were used as the uniform scattering medium. The plates were varied according to the required thickness of the scattering medium. Data were collected by employing a material filter in combination with either a LEGP or LEHR. A 20% energy window for Tc-99m (126-154 keV) was selected. The matrix size was 128 x 128. The total counts were maintained at 1 000 kcount (for LEGP collimator) and 2 500 kcount (for LEHR collimator). Care was taken to maintain the same position of the line source for the recording of each image.

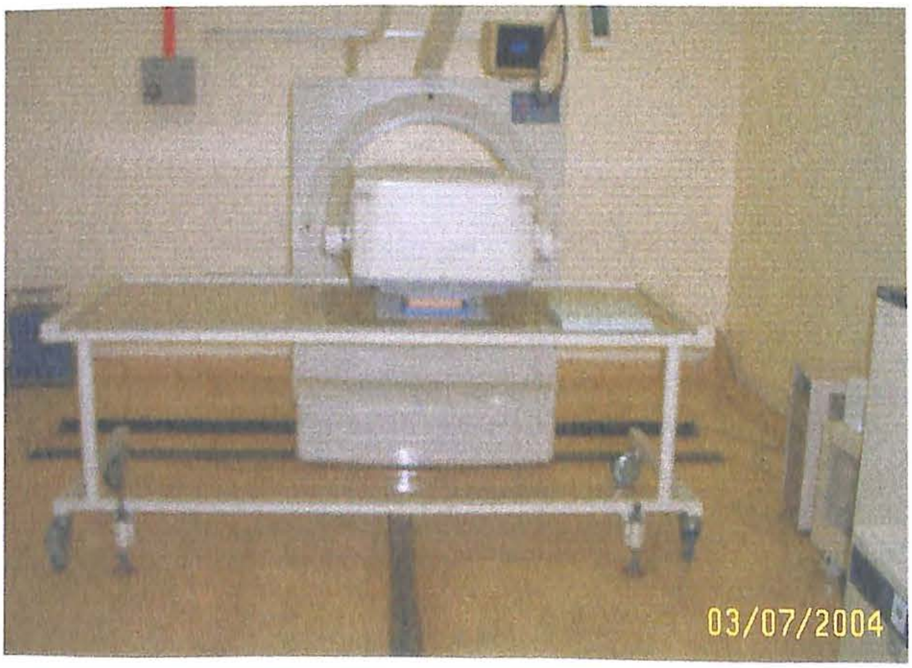


Figure 7 : Experimental setup for spatial resolution.

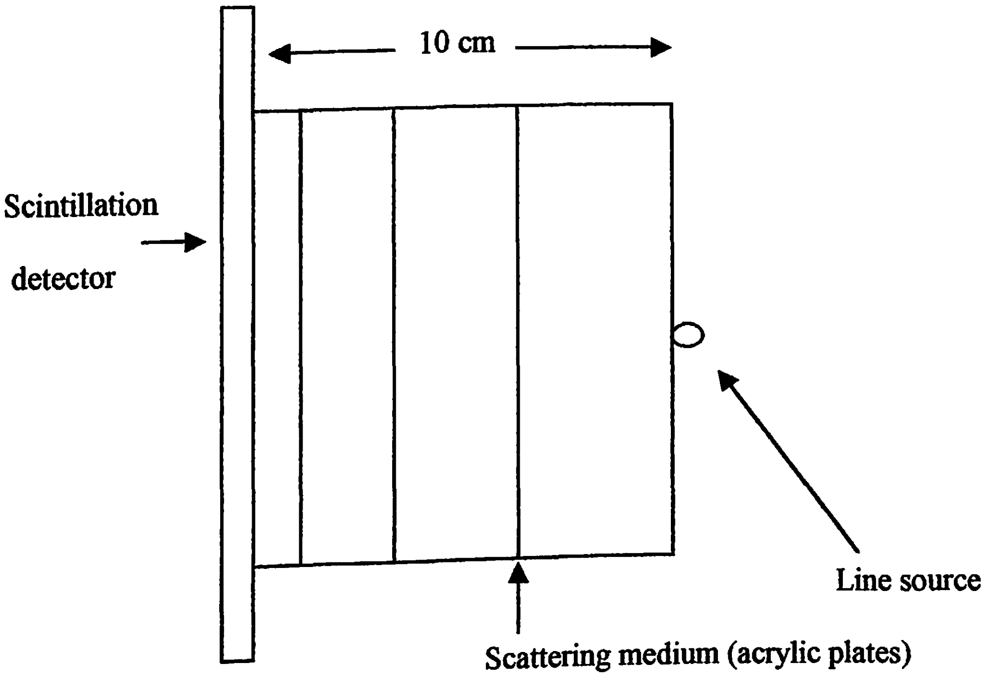


Figure 8 : Illustrates the experimental set-up for the measurement of line source in scattering medium (acrylic plates).

RESULTS

Uniformity

The flood field uniformity of gamma camera is defined as ability of camera to produce the uniform image when exposed to a homogeneous spatial distribution. Flood field uniformity with collimator known as system uniformity or extrinsic uniformity. It is absolutely essential to good diagnostic technique. This is measured by imaging a uniform flood source of photons, acquiring a large, statistically valid number of counts (>10,000 per pixel). Uniformity can be different for different radioactive and window setting (Mettler FA *et al*, 1998).

Flood field uniformity can be quantified in terms of the difference between the maximum and minimum counts obtained in any pixel divided by the average pixels counts.

Extrinsic flood field uniformity shall be expressed as "integral uniformity" (a maximum deviation) and a "differential uniformity" (a maximum rate of change over a specified distance, roughly slope). Both shall be measured for the UFOV (useful field of view) and the CFOV (central field of view).

Integral uniformity is a measure of the maximum pixel count deviation in the CFOV or UFOV. Differential uniformity is a measure of the maximum deviation over a limited range designed to approximate the size of a photomultiplier tube. The radionuclide used to measure extrinsic uniformity is to be Tc-99m. Any other radionuclide used shall be reported separately. The count rate shall not exceed 20,000 counts per second through a symmetric 20% photopeak window. Integral and differential uniformity can easily described as :

Integral Uniformity : For pixels within each area (CFOV and UFOV), the maximum and the minimum values are to be found from the smoothed data. The difference between the maximum and the minimum is divided by the sum of these two values and multiplied by 100%.

Integral Uniformity = + 100% ((Max - Min) / (Max + Min))

Differential Uniformity: For pixels within each area (CFOV and UFOV) the largest difference between any two pixels within a set of 5 contiguous pixels in a row or column shall be calculated. The calculation shall be done for the X and for the Y directions independently and the maximum change expressed as a percentage using the following equation :

Differential Uniformity = + 100% ((Max - Min) / (Max + Min))

	UFOV		CFOV	
	Integral Uniformity (%)	Differential Uniformity (%)	Integral Uniformity (%)	Differential Uniformity (%)
No Filter	8.8	(x) : 3.3 (y) : 4.0	6.9	(x) : 3.3 (y) : 2.9
Copper (0.125 mm)	8.6	(x) : 3.2 (y) : 4.1	6.7	(x) : 3.2 (y) : 3.0

Table 1 : Shows the uniformity results using LEGP collimator, without and with material filter.

	UFOV		CFOV	
	Integral Uniformity (%)	Differential Uniformity (%)	Integral Uniformity (%)	Differential Uniformity (%)
No Filter	8.8	(x) : 3.0 (y) : 3.8	6.8	(x) : 3.0 (y) : 2.6
Copper (0.125 mm)	8.1	(x) : 3.3 (y) : 4.3	6.7	(x) : 3.3 (y) : 3.1

Table 2 : Shows the uniformity results using LEHR collimator, without and with material filter.

Sensitivity

System sensitivity is the parameter of gamma camera with its collimator in place, characterizes its ability to efficiently detect incident gamma rays. The higher the sensitivity, the greater the fraction of the emitted photons that are detected. Poor sensitivity can only produced noisy and low resolution images.

The sensitivity and spatial resolution of radionuclide imaging units are two closely related parameters (Shah SI *et al*, 1996). The sensitivity depends upon the total solid angle subtended by the detector at the source. If the solid angle is made as large as possible, then the spatial resolution is likely to be poor. Sensitivity can be measured from a single slice and total phantom value, and expressed as (cps)/(μCi/ml) or (cps)/(MBq/ml). Counts are defined as interactions in the crystal that fall within the analyzer window, therefore field uniformity correction devices which alter the number of counts must be disabled.

There are several methods of measuring the sensitivity of an imaging systems, for instance by scanning point sources, flood sources or cylindrical phantoms. Scanning a cylindrical phantoms filled with a low radioactivity concentration is a widely used procedure. In this case, the effect of system dead time may be eliminated.

Collimator(s)	Sensitivity (cps / mCi/~~ml~~)	
	No Filter	Material Filter
LEHR	3496.50	3413.71
LEGP	6839.30	6776.27

Table 3 : Shows the sensitivity value obtained without and with material filter using LEHR and LEGP collimator.

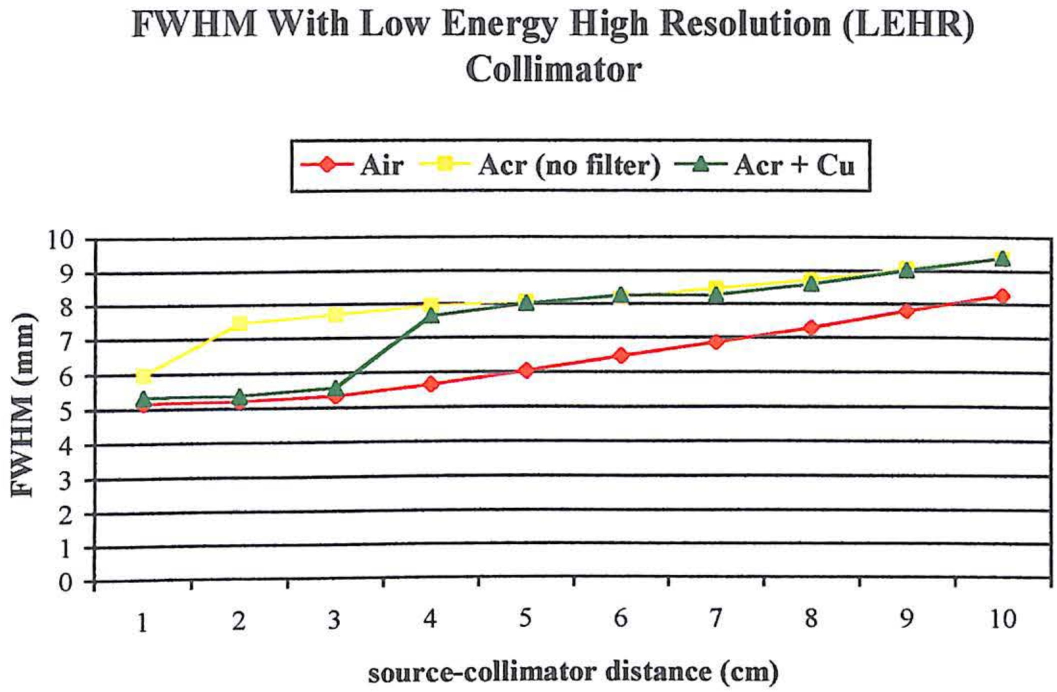
Spatial Resolution

Spatial resolution is a measure of the ability of the imaging system to distinguish two closely spaced objects and is a fundamental parameter for comparing systems. It is conventionally determined by the measurement of full-width at half maximum (FWHM) and full-width at tenth-maximum (FWTM) in the image of a point source (the point spread function, PSF).

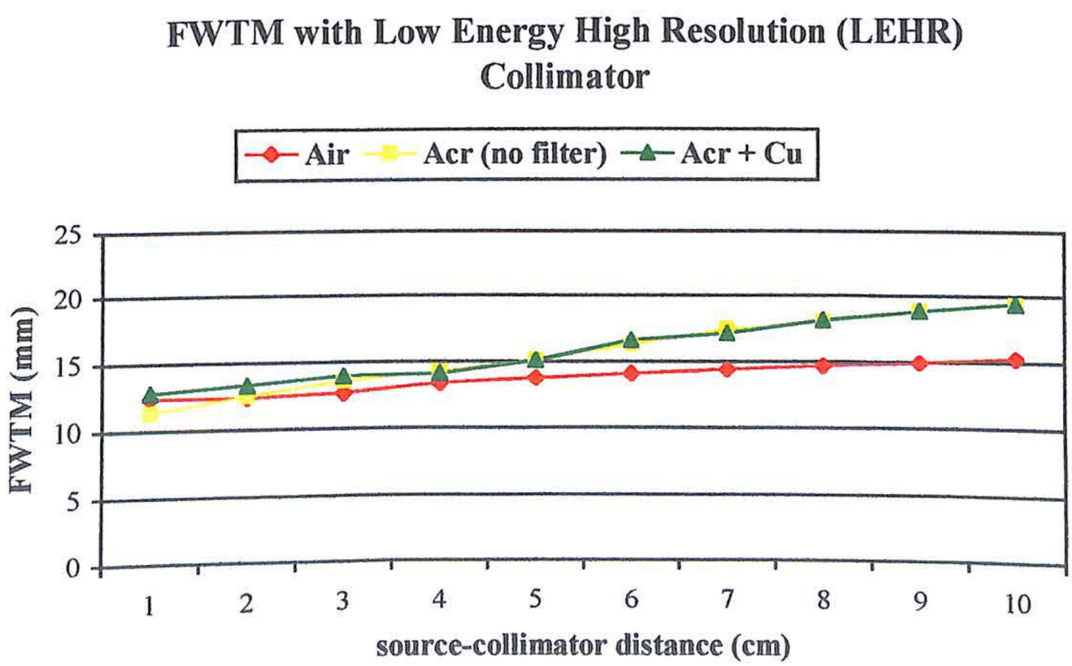
In practice it is difficult to produce a point source with sufficient activity concentration and adequate counting statistics. Thus, a line source is often used for resolution measurements and the line spread function (LSF) is obtained from an image plane intersecting the line at right angles (Shah SI *et al*, 1996). The LSF indicate the degree of blurring (loss of resolution of the camera).

Spatial resolution depends upon many factors, such as type of collimator, source to gamma camera distance and statistical fluctuations in the distribution of light photons among the PM tubes. The LSF of a single hole collimator depends on the divergence of the collimator, penetration of the collimator by photons and the scattering of photons into the cone of acceptance of the collimator. The later effect means that the LSF will depends on the scattering medium and the depth of the source in the medium.

To minimize the contribution of the scattered photons to the LSF, it is measured with the source in air. However, the divergence of the collimator results in the LSF still depending on the distance of the source from the detector. Thus in assessing the LSF, the location of the source relative to the detector should be noted.



Graph 1 : Shows FWHM values in air and in the presence of acrylic (without and with material filter) for LEHR collimator.



Graph 2 : Shows the FWTM values in air and in the presence of acrylic (without and with material filter) for LEHR collimator.