A NOVEL TECHNIQUE TO IMPROVE THE RESOLUTION & CONTRAST OF PLANAR NUCLEAR MEDICINE IMAGING

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A NOVEL TECHNIQUE TO IMPROVE THE RESOLUTION & CONTRAST OF
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ABSTRACT

Nuclear medicine images have limited spatial resolution because of the limitations in the radiation detector and the associated electronics. Due to these limitations, the image of a point source of radiation is blurred. The degree of this blurring is called the point spread function of a gamma camera, the imaging device used in nuclear medicine. The aim of this study is to evaluate a technique to increase the spatial resolution and contrast by improving the point spread function of the system.

The basic idea of the proposed technique is to restrict radiation so that overlapping point spreads due to adjacent point sources could be isolated. This was achieved by using a special lead mask, having a particular pattern of apertures, with the existing technique. The new technique was implemented in two variations by using special lead masks with different sizes, shapes and patterns of apertures in order to investigate the degree of improvement in the spatial resolution. The images obtained with the proposed technique were compared to those obtained with the conventional technique. Qualitative comparisons were made by visually inspecting the images obtained by the two techniques while quantitative comparisons were made by statistically testing their modulation transfer functions for significant differences. Both the comparisons indicated that the proposed technique was successful and gave images with increased spatial resolution and contrast.
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CHAPTER I
INTRODUCTION

Medical Imaging provides a non-invasive technique to look at the functional and structural information of internal organs and structure. There are many different types of modern medical imaging techniques like Computed Tomography (CT), Magnetic Resonance Imaging (MRI) and Nuclear Medicine to name a few. This work focuses on the emission imaging technique – Nuclear Medicine.

1.1 Nuclear Medicine

Nuclear Medicine is a branch of medical imaging that makes use of radioactive tracer in diagnosis and therapy. In a nuclear medicine study, a compound labeled with a radionuclide is administered to the patient and it localizes in the organ of interest. This radioactive compound is called as a radiopharmaceutical. As the radionuclide decays, gamma rays are emitted, which are detected by a gamma camera. This radiation detection consists of mapping the distribution of the radiopharmaceuticals in the organ of interest to form an image of the organ [1]. Such imaging is also referred to as radionuclide imaging or nuclear scintigraphy. Some of the more common nuclear imaging studies currently in use include cardiovascular imaging, bone imaging and hepatobiliary imaging.
Nuclear medicine imaging has broad applications in medicine. It is useful for evaluating abnormal anatomy and function of many of the body’s organs, detecting occult tumors or infections, detecting vascular abnormalities such as aneurysms and abnormal blood flow to various tissues.

There are two broad classes of nuclear imaging: single photon imaging and positron imaging (PET) [1]. Single photon imaging uses radionuclides that decay by gamma-ray emission. A planar image is obtained by mapping the distribution of the radionuclide in the patient from one particular angle. This results in an image with little depth information but which can still be diagnostically useful [2]. For Single Photon Emission Computed Tomography (SPECT), data are collected from many angles around the patient. This provides cross-sectional images of the distribution of the radionuclide to be constructed, thus providing the depth information missing from planar imaging. The principal imaging instrument used in nuclear imaging is called the gamma camera. For this study, only planar images were used.

1.2 Purpose of the study

Nuclear Medicine provides exquisitely sensitive measures of a wide range of biological processes in the body like tissue perfusion, glucose metabolism, somtostatin receptor status of tumors, density of dopamine receptors in brain and gene expression [1]. Other imaging modalities such as MRI, X-ray imaging and CT provide good anatomical information but limited biological information [1]. Nuclear medicine is a very sensitive technique. The radiopharmaceuticals used are in the nanomolar or picomolar
concentration range as compared to the molar or millimolar range of the magnetic resonance methods [1]. Also, the available radiation detectors can easily detect very small amounts of radioactivity [1]. Due to all the above advantages, nuclear medicine has the potential of producing good images with a very small dose of radiation to the patient. However, the quality of nuclear medicine images is limited by the radiation detection and the imaging system. In particular, the gamma camera has limited spatial resolution and contrast. The purpose of this study was to evaluate a technique for increasing the spatial and contrast resolution of planar nuclear medicine imaging by improving the point spread function (PSF) of the nuclear imaging system.

1.3 Specific Aims of the Study

The aims of the study were:

1. To design and build the hardware required for the study, like the special lead mask with the pattern of apertures and the XY motion table.

2. To develop a protocol for data acquisition and acquire phantom images using the protocol.

3. To develop an algorithm that does the necessary processing on the acquired images.

4. To qualitatively compare the images obtained from the proposed technique and those obtained by the conventional method.

5. To quantitatively compare the spatial resolution of the proposed technique with that of the conventional method.
1.4 Statement of Hypothesis

As explained in section 2.7.3, the modulation transfer function (MTF) is the most quantitative measure of the spatial resolution of an imaging system. Hence, the MTFs with the conventional and proposed techniques were estimated and tested for statistical difference. In addition to the qualitative (visual) analysis, the following null hypotheses were tested using the Kruskal-Wallis test, at a significance level of 5% (α=0.05), followed by a multiple comparison procedure.

1.4.1 Null Hypotheses

\( H_{01} \): There is no significant difference between the MTF values obtained by the conventional technique and the proposed technique using pattern A apertures.

\( H_{02} \): There is no significant difference between the MTF values obtained by the conventional technique and the proposed technique using pattern B apertures.

1.4.2 Alternate Hypotheses

\( H_{11} \): There is a significant difference between the MTF values obtained by the conventional technique and the proposed technique using pattern A apertures.

\( H_{12} \): There is a significant difference between the MTF values obtained by the conventional technique and the proposed technique using pattern B apertures.
2.1 Overview of the gamma camera

The scintillation camera or gamma camera, also known as the Anger camera in honor of Hal O. Anger is the imaging device that is most commonly used in nuclear medicine. Anger invented the gamma camera in the late 1950s [3]. The patient is injected with a radioactive tracer, which has a very short half-life. The resulting gamma ray emissions from the patient contain the functional information of the organ being imaged. The components making up the gamma camera are the collimator, detector crystal, photomultiplier tube array, position logic circuits, and the data analysis computer (Figure 2.1) [4]. The collimator is the first object encountered by the gamma ray after exiting the human body. A collimator is a pattern of apertures in a gamma ray absorbing material like lead [4]. It allows only the gamma rays traveling in a particular direction to reach the detector crystal while those not traveling in the proper direction are absorbed [1]. The detector crystal is generally Thallium activated Sodium Iodide (NaI (Tl)) [3]. The gamma ray photon interacts with the detector crystal to produce free electrons by the Photoelectric effect or Compton scatter. These free electrons further interact with the crystal lattice to produce light photons by a process called scintillation, which is explained
later. Photomultiplier (PM) tubes are used to amplify this light signal. The working of a photomultiplier tube is explained in the next section. The position circuitry consists of the summing matrix circuit (SMC) [1]. It allows the determination of the position of the scintillation event in the detector crystal. When hundreds of thousands of gamma rays are detected from a particular viewpoint, a planar image of the organ of interest is created. To create cross-sectional (SPECT) image of an organ of interest, a planar image provides raw data for one set of projections. Many such projections, taken at different angles by rotating the camera around the patient, allows a computer algorithm to create a cross-sectional image of the distribution of radioactivity within the organ.

Reference: http://www.physics.ubc.ca/~mirg/home/tutorial/hardware.html

Figure 2.1: Gamma Camera
2.2 Collimators

Gamma rays are emitted isotropically in all directions. Simply using a detector would not result in an image, as there would be no relationship between the position at which the gamma rays hit the detector and that from which they were emitted from the patient [2]. To get an image using the gamma camera, it is necessary to project gamma rays from the source onto the detector. However, gamma rays cannot be focused as they are high-frequency radiation. So a much cruder device, the collimator, must be employed. The collimator allows only those gamma rays travelling along certain directions to reach the detector. Gamma rays not travelling in the proper direction are absorbed by the collimator before they reach the detector. This technique of projection is an inefficient method because most of the potentially useful radiation traveling towards the detector is absorbed by the collimator. This is one of the major reasons for the relatively poor spatial resolution of radionuclide images [2].

2.3 Scintillator Detector

When radiation from radioactive materials interacts with matter it causes ionization and/or excitation of atoms and molecules in the matter. When the excited atoms undergo de-excitation, energy is released. Most of the energy is released as thermal energy, however, in some materials a portion of the energy is released as visible light. These materials are called scintillators and the detectors made from them are called scintillation detectors. The amount of light produced in the scintillators is proportional to the energy deposited by the incident radiation in the scintillator. The amount of light
produced is also very small, typically a few hundred to a few thousand photons for a single gamma ray interaction within the energy range of interest for nuclear imaging (70-511 keV) [1].

The most commonly used scintillator for detectors in nuclear imaging is NaI(Tl). It is a type of inorganic scintillator. Some reasons for the wide use of NaI(Tl) detectors are as follows [1]:

1. It is a good absorber and a very efficient detector of gamma rays because of its high density (3.67 g/cm³).
2. It is a relatively efficient scintillator, yielding one visible light photon per approximately 30 eV of incident radiation.
3. It is transparent to its own scintillations emissions. So, there is little loss of scintillation light caused by self-absorption.
4. It can be grown relatively inexpensively in large plates.
5. The scintillation light is well-matched in wavelength to the peak response of the PM tube cathode.

2.4 Photomultiplier (PM) tubes

PM tubes are electronic tubes that produce a pulse of electrical current when stimulated even by very weak light signals, such as the scintillation produced by a gamma ray in a scintillation detector. The working of a PM tube is illustrated in Figure 2.2. The entrance window of the PM tube is coated with a photo emissive substance like cesium antimony (CsSb). The photo emissive surface is called the photocathode and it
ejects photoelectrons when scintillation light photons are incident on it. Typically, 1 to 3 photoelectrons are emitted per 10 visible light photons striking the photocathode. A short distance from the photocathode is a metal plate called a dynode. The dynode is maintained at a positive voltage (about 200-400 V) relative to the photocathode and attracts the photoelectrons ejected from it. A focusing grid directs the photoelectrons towards the dynode. A high-speed photoelectron striking the dynode surface emits several secondary electrons from it. Secondary electrons emitted from the first dynode are attracted to a second dynode, which is maintained at a 50-150 V higher potential than the first dynode. This electron multiplication process is repeated through many additional dynode stages (9 to 12), until finally a shower of electrons is collected at the anode. The total electron multiplication factor is very large, for example $6^{10}$ for a 10-stage tube with an average multiplication factor of 6 at each dynode. Thus a relatively large pulse of current is produced when a tube is stimulated even by a weak light signal. The amount of current produced is proportional to the intensity of the light signal incident on the photocathode and thus also to the amount of energy deposited by the radiation event in the crystal [1].

2.5 Limitations on the intrinsic spatial resolution

The limit of spatial resolution achievable by the detector and the electronics, ignoring additional blurring due to the collimator, is called the intrinsic spatial resolution of the camera. For NaI(Tl) detectors, the intrinsic resolution is limited primarily by the
random statistical variations and non-uniformities in the events leading to the formation of the output image [1]. These variations and non-uniformities are as follows [1]:

1. Statistical variations in the number of light photons produced in the crystal when two gamma rays of the same energy fall on the crystal. For $N$ light photons recorded at a certain location in the crystal, the recorded number from one event to other vary with a standard deviation of $\sqrt{N}$. So for a narrow beam directed at a point on the detector, the position of each event as calculated by computer algorithm is not precisely the same, but is distributed over a certain area.

2. Statistical variations in the number of photoelectrons released from the photocathode.

3. Statistical variations in the electron multiplication factor of the dynodes in the PM tubes.
4. Non-uniform sensitivity to the scintillation light over the area of the PM tube cathode.

5. Non-uniform light collection efficiency for light emitted from interactions at different locations on the detector crystal.

6. Nonlinear energy response of the crystal, such that the amount of light produced by the lower-energy Compton electrons in multiple events generate a different number of light photons than are produced by a higher energy photoelectric event, even when the total energy deposited in the crystal in the same.

7. Fluctuations in the high voltage applied to the PM tubes.

8. Electrical noise in the PM tubes.

Due to these factors, the amplitude of the signal generated from one event to the next, of precisely the same energy, is different [1]. So when a point source of radiation is exposed to the gamma camera, the image generated is not a point but a 2D blur of intensities, which determines the point spread function of the gamma camera.

2.6 Performance Parameters of Gamma cameras

This section describes some of the major factors that determine gamma camera performance.

2.6.1 Spatial Resolution

The spatial resolution of a gamma camera is a measure of the ability of the device to faithfully reproduce the image of an object, thus clearly depicting the variations in the
distribution of radioactivity in the object [3]. In other words, spatial resolution is the minimum distance between two points in the image that can be detected by the gamma camera. The overall spatial resolution \( (R_o) \) of a camera is the combination of the intrinsic resolution \( (R_i) \), the collimator resolution \( (R_g) \) and the scatter resolution \( (R_s) \), given by [3]:

\[
R_o = \sqrt{R_i^2 + R_g^2 + R_s^2}
\]

The smaller the \( R_o \), the better the resolution of the gamma camera. The intrinsic resolution \( (R_i) \) arises due to the non-uniformities in the working of the different components of the detection system, as explained in the next section. Most modern cameras have intrinsic resolution of about 4 mm full width at half maximum (FWHM) for Technetium-99m [3]. The collimator or geometric resolution \( (R_g) \) depends on the type and design of the collimator used. There are four major collimator types: parallel-hole, pinhole, converging and diverging. Parallel-hole collimators are most routinely used in nuclear imaging. In collimator design, there is a tradeoff between collimator resolution and collimator efficiency. The typical spatial resolution for a general purpose parallel-hole collimator is about 8 mm [3]. The scatter resolution \( (R_s) \) arises from the fact that radiations scattered by interaction with tissues in the patient, interact with the detector and are registered as counts in the wrong location on the image. The scatter resolution depends on the composition of the scattering medium, the source configuration and the pulse-height analyzer (PHA) settings [3].
2.6.2 Sensitivity

Sensitivity of a gamma camera is defined as the number of counts per unit of time detected by the device for each unit of activity present in the source [3]. Sensitivity depends on the geometric efficiency of the collimator, detection efficiency of the detector, PHA discriminator settings and the dead time of the system [4]. A general-purpose collimator has sensitivity on the order of 1 to $1.5 \times 10^{-4}$ cps/Bq [1].

2.6.3 Uniformity and Linearity

For obtaining a faithful image of the object being scanned, it is desirable that the gamma camera has a uniform response over the entire field of view. In other words a point source should register the same number of counts irrespective of its position in the field of view [3]. But it is found that even properly tuned gamma cameras produce images with count variations of up to 10% in different regions of the camera (Figure 2.3) [3]. This detector non-uniformity arises due to the different factors like variations in the response of the PM tubes when exposed to a point source, nonlinearities in the calculation of the X, Y coordinates of the pulses in the field of view and edge packing. The edge packing artifact arises from greater light collection efficiency near the edge as compared to the central regions of the detector [1]. This results in a bright border around the edge of the image.
It is also desirable that the gamma camera reproduce linear images i.e. straight-lined objects should not appear curved. However, non-linearity is frequently observed in nuclear medicine images. The two types of non-linearities observed are: pincushion distortion or the inward curving and the barrel distortion or the outward curving of line images (Figure 2.4) [1]. These result as the X and Y position signals do not change linearly with the distance of displacement across the face of the detector [1]. So when a source is moved from the edge of a PM tube towards its center, more counts are registered at the center than the edges resulting in pincushion distortion in the areas of the image lying directly in front of the PM tubes and barrel distortion between them [1].
2.6.4 Contrast

Image contrast refers to the differences in intensity in parts of the image corresponding to the areas of different levels of radioactive uptake in the patient [1]. Objects of interests in an image like lesions or tumors appear as either hot spots (areas of increased uptake) or cold spots (areas of decreased uptake) [3]. So an image with a good contrast helps physicians to detect lesions easily. The contrast of an image can be calculated as the ratio of signal change over a lesion or abnormality with respect to the surrounding normal tissue. If $R_o$ is the counting rate over the normal tissue and $R_l$ is the counting rate over a lesion, then the contrast ($C_l$) is calculated as [1]:

$$C_l = \frac{R_l - R_o}{R_o}$$
2.7 Methods of Evaluating Spatial Resolution

Spatial resolution can be evaluated by qualitative or quantitative means. Qualitative evaluation is done by visual inspection of images. However, by qualitative evaluation different observers might give different interpretations of image quality. Hence, quantitative methods are required for evaluation of spatial resolution.

2.7.1 Bar Phantoms

Bar phantoms can be used for the qualitative evaluation of the spatial resolution of an imaging device. A quadrant bar phantom consists of four sets of parallel lead bars arranged perpendicularly to each other in every quadrant (Figure 2.5 (a)). The width and the spacing between two consecutive bars is the same within every quadrant but differ in different quadrants [3]. Another type of bar phantom is the Hine-Duley phantom (Figure 2.5 (b)). It consists of five groups of lead bars with different thicknesses and spacings arranged in parallel [3].

The bar phantom is placed over the detector of a gamma camera with a flood source placed on top of it and an image is acquired. This image is visually inspected and the spatial resolution is equal to the width of the smallest lead bar that can be distinguished in the image. This technique does not give a quantitative measure of the spatial resolution and is also subjective up to some extent [3].
2.7.2 Line Spread Function Curve

The point-spread function (PSF) describes the response of a gamma camera to a point source. Ideally, the image of a point source of activity should also be a point. However, due to various reasons, described previously, the image undergoes degradation as it is processed through the camera. So the image of the point source is spread (blurred),

Figure 2.5: Bar Phantoms (a) Quadrant bar phantom [3] (b) Hine-Duley bar phantom [3]

thus degrading the spatial resolution of the camera. A line-spread function (LSF) is the integration of a number of point spreads and is used to estimate the spatial resolution quantitatively. A plastic tube filled with radioactivity is exposed to the detector [3]. The gamma camera registers the counts from the tubing which are used to generate the LSF. The counts obtained at incremental distances are plotted against the distance from the center axis of the collimator to give a bell-shaped LSF curve [3] (Figure 2.6 (a)). The Full Width at Half Maximum (FWHM) of this curve is used in nuclear medicine as a measure of a gamma camera’s spatial resolution. The spatial resolution determined by this method
varies with the type of collimator used. Also, the FWHM may not represent the true
spatial resolution as the scatter and septal penetration components fall at the tail of the
LSF and are not accounted for [3]. Hence, the Full Width at Tenth Maximum (FWTM) is
a better estimate of the spatial resolution. Both the FWHM and the FWTM of the LSF is
shown in Figure 2.6 (b).

2.7.3 Modulation Transfer Function

A more comprehensive and quantitative representation of the spatial resolution is
given by the modulation transfer function (MTF) [3]. The method of determining the
MTF is illustrated by the Figure 2.7. Suppose we have a source with a sinusoidal
distribution of radioactivity. Such a distribution gives a spatial frequency (cycles per
centimeter or cycles per millimeter) [1]. We can obtain a curve by calculating the MTF at
changing spatial frequencies.

Reference: http://webvision.med.utah.edu/imageswv/KallSpat13.jpg

Figure 2.6: Line Spread Function (a) LSF curves of two line objects (b) FWHM &
FWTM [2]
If $I_{\text{max}}$ is the maximum activity and $I_{\text{min}}$ is the minimum activity in the source, then the source/input modulation ($M_{\text{in}}$) is given by [1]:

$$M_{\text{in}} = \frac{I_{\text{max}} - I_{\text{min}}}{I_{\text{max}} + I_{\text{min}}}$$

An ideal imaging system would image the source faithfully, that is the image would depict the same distribution with $I_{\text{max}}$ and $I_{\text{min}}$ as in the source. However, due to its limitations the system reproduces an image with $O_{\text{max}}$ and $O_{\text{min}}$ as maximum and minimum intensities respectively [4]. So the image/output modulation ($M_{\text{out}}$) is given by [1]:

$$M_{\text{out}} = \frac{O_{\text{max}} - O_{\text{min}}}{O_{\text{max}} + O_{\text{min}}}$$

The ratio of the output to input modulation is the MTF of a spatial frequency ‘$k$’ [1]:

$$MTF(k) = \frac{M_{\text{out}}(k)}{M_{\text{in}}(k)}$$

An imaging system reproduces an image faithfully if it has a flat MTF curve with a value near unity [1]. Good low-frequency response is needed to display coarse details of the image, so that large but low-contrast objects in the image can be detected. Good high-frequency response is necessary to display fine details and sharp edges, so that small lesions can be imaged faithfully [3]. Practically, the MTF is not evaluated by the sinusoidal activity sources as described above. Instead, MTFs are determined by the mathematical analysis of the PSF or LSF. Specifically, the MTF of a system is given by the amplitude of the Fourier transform (FT) of the LSF [1].
We know that the FWHM of the LSF does not account for the scatter and septal penetration components, however, the MTF takes those into consideration and hence provides a comprehensive description of the spatial resolution of the system [3]. The MTF method also simplifies the evaluation of the system spatial resolution by taking into consideration different components of the system. For example, if the MTF of the intrinsic resolution of the camera is $MTF_{int}(k)$ and that of the collimator is $MTF_{coll}(k)$, then the overall system MTF is obtained by the element-by-element multiplication of individual MTFs [1]:

$$MTF_{sys}(k) = MTF_{int}(k) \times MTF_{coll}(k)$$

Hence, the overall MTF of the system is the product of the individual MTFs of all its components.
2.8 Techniques to improve spatial resolution

Many different approaches have been used to improve the spatial resolution of the gamma camera. One of the approaches is improving the different components of the camera, as explained:

1. PM tubes: Modern cameras use a large number of smaller, efficient PM tubes and have better optical coupling between the crystal and the PM tubes [1].

Disadvantages
   a. There is a limit to the reduction in the size of the PM tubes.
   b. As the number of PM tubes increases, the system becomes expensive.

2. Crystal Thickness: Intrinsic resolution depends on the crystal thickness. Thicker crystals result in greater spreading of scintillation light before it reaches the PM tubes. There is also a greater likelihood of detecting multiple Compton-scattered events in thicker crystals [1]. Hence, thinner crystals are used to get better intrinsic resolution.

Disadvantages
   a. Detection efficiency of the crystal decreases as high energy gamma rays can penetrate the crystal and go undetected.

3. Collimator Design: The spatial resolution can be improved by having smaller, tighter collimator apertures and by increasing the thickness of the collimator.

Disadvantages
   a. The sensitivity of the system decreases as a large number of gamma rays are absorbed by the collimator.
b. The scan time of the examination has to be increased in order to compensate for the reduced sensitivity.

Another approach to improve the spatial resolution is to develop new detector configurations. Many different detector configurations have been developed over the years. The SPRINT II system developed at University of Michigan and the CERASPECT system developed by Digital Scintigraphics Inc., Cambridge, MA are examples of such systems. Both the systems were designed for brain imaging applications and provide images with better resolution, about 8 mm at the centre of the brain and 5 mm at the edges, than a conventional SPECT, which has a resolution of about 12.5 mm FWHM [1]. The FASTSPECT, developed at the University of Arizona is an example of a completely stationary SPECT. It provides images of the brain with a resolution of about 4.8 mm [1]. Also, as there are no moving parts, it can be used for rapid dynamic SPECT studies. The details of the design and working of the system can be found in [6] for SPRINT II, in [7] for CERASPECT and in [8] for FASTSPECT. A commercially available system for high resolution cardiac imaging is the Cardiarc®. The common drawback of all these systems is that they can be used to image only specific parts of the body.

Another approach to improve spatial resolution is through the use of iterative reconstruction algorithms. In essence, these algorithms start with an initial estimate of the image and approach the true image by successive approximations. These algorithms can be incorporated with models of factors that degrade spatial resolution, like point spread function and scatter radiation [9]. So these algorithms can be used to obtain images with
high resolution. The major drawback of iterative reconstruction algorithms is that they are computationally intensive and hence, need longer reconstruction times. For this reason they are not yet being used extensively in nuclear medicine [9].
CHAPTER III
MATERIALS AND METHODS

3.1 Materials

The data for this study were collected at The Imaging Center, Stow, OH using a dual-head gamma camera manufactured by Trionix Inc. The gamma camera consisted of two NaI (Tl) detectors placed 180º apart from each other. This section describes the materials used for the study.

3.1.1 Special Lead Masks

Two special lead masks, which were lead plates of size 12”x17”x1/8”, were used for the study. One had a pattern of 2 mm diameter circular apertures (henceforth called pattern A). Figure 3.1 shows a single section of this pattern, which is repeated over the entire area of the lead plate. The center-to-center distance of the adjacent apertures was 20 mm. The other plate had a pattern of 1 mm square apertures (henceforth called pattern B). Figure 3.2 shows a single section of this pattern, which is repeated over the entire area of the lead plate. The center-to-center distance between adjacent apertures was 5 mm. The size of the apertures was selected to be smaller than the spatial resolution of the
conventional gamma camera. The apertures were spaced such that the point spreads due to any aperture did not overlap or minimally overlapped with those of the adjacent apertures. These lead masks were fastened to the surface of the conventional collimator (Figure 3.3).

Figure 3.1: A section of the pattern A apertures

Figure 3.2: A section of the pattern B apertures
3.1.2 Phantom

A deluxe Jaszczak phantom was used for the study. The phantom is widely used for evaluation of SPECT system performance and research purposes. The phantom contained technetium-99m, so it was the radioactive source that was imaged by the gamma camera. The image was acquired as samples by placing the phantom on a calibrated XY-table. 10 samples were collected with pattern A apertures and 25 samples were collected with pattern B apertures in order to acquire the entire image.

3.1.3 XY Table

The XY table was a device capable of precise movements in two directions, for moving the phantom. The table was custom made using sets of channels and sliders. Two channels were placed horizontally parallel to each other and four sliders were mounted on them. Then two more channels were mounted vertically on these sliders and a 1/8” thick plexiglass sheet was mounted on the sliders of the top channels. The plexiglass sheet was
mounted such that it was very close to the surface of the lead plate on the gamma camera, which ensured that the distance between the source and the collimator was minimum. Two micrometers were used for precise movements in both directions. The movement of the table was measured by a dial indicator, which was calibrated in inches and had a sensitivity of 0.001”.

Movement in only one direction was needed with pattern A apertures due to their particular arrangement. However, for the pattern B apertures movement in X and Y directions was required.

3.2 Data Acquisition

Data were acquired in two stages: as phantom images and as line source images for evaluating the spatial resolution of the techniques under comparison.

3.2.1 Phantom Data

The final set-up used for the data acquisition is shown in the Figure 3.4. The radioactive phantom was placed on the XY table, so that it can be moved freely without disturbing its orientation. For pattern A, the data was acquired as 10 samples by shifting the phantom 2 mm vertically after every sample, for a total vertical movement of 2 cm. As an example, Figure 3.5 shows the sampling process when the phantom is moved vertically downwards with the samples indicated by the numbers on the left side. The acquisition can be done by moving the phantom horizontally or vertically. This pattern is
repeated over the area of the plate, thus ensuring that the image of the phantom is sampled completely.

Figure 3.4: Final set-up for data acquisition

Figure 3.5: Sampling of the image using the pattern A apertures
For pattern B, the data was acquired as 25 samples by first shifting the phantom vertically in five steps of 1 mm each and repeating this for shifts in five horizontal steps of 1 mm each. As an example, Figure 3.6 shows the sampling process when the phantom is moved vertically downwards for first five samples, then horizontally to the left side to get the next five samples and so on until 25 samples are collected. As the pattern is repeated over the area of the lead plate, the entire image is sampled.

![Figure 3.6: Sampling of the image using pattern 2 apertures](image)

After these acquisitions, the XY-table was taken off the gamma camera but the lead mask was left in place and an image of a flood source, which is a uniform source of radioactivity, was acquired for 40 minutes. This image was used for calibration of the exact co-ordinates of each aperture on the image. Finally, an image of the same phantom was acquired by the conventional nuclear imaging technique for 20 minutes.
3.2.2 Data for spatial resolution estimation

The spatial resolution was estimated by the modulation transfer function method (MTF), since it is the most quantitative representation of the spatial resolution, as described in section 2.7.3. The MTFs for the proposed and the conventional techniques were calculated by the conventional line source method as described in [10]. In order to ensure that the diameter of the source does not affect the result, it is recommended that the diameter of the line source should be no more than the spatial resolution of the gamma camera [3]. So a capillary tube with an inner diameter of 0.5 mm was filled with Tc-99m and was used as the line source of radiation. The data was acquired by sampling both with the pattern A and pattern B apertures and also by the conventional technique. The methodology of data acquisition with the proposed technique was the same as described in the previous section. Once the sampled data was processed, as described in the next section, three line source images were obtained.

A profile taken through the centre of the line source in a direction perpendicular to it yielded the line spread function (LSF) of each of the images. Then the Fourier Transform of each of the LSFs was calculated and its amplitude was normalized to 1, as shown by the following equation. These normalized values were plotted to obtain three MTF curves.

\[
MTF(k) = \frac{\sum_{x=1}^{n} L_x e^{j2\pi kx}}{\sum_{x=1}^{n} L_x}
\]

where k is the spatial frequency, n is the number of pixels covering the LSF and \( L_x \) is the count rate in the \( x \)th pixel.
3.3 Data Processing

1. The images acquired by the gamma camera were saved in the interfile format. It is a format widely used for the exchange of nuclear medicine information. These images were transferred to a personal computer and were read using the Image Processing Toolbox of MATLAB (version 7.4) for further processing. All the images were 512 x 512 (16 bit) grayscale images. For pattern A the pixel size was 0.66 mm and for pattern B it was 0.33 mm.

2. The image of the flood source was processed first to find the co-ordinates of the apertures on the image. This was accomplished by morphologically opening the image first. This broke any bridges between the point spreads and made them distinct. Then the approximate centroids of each aperture were determined. These co-ordinates were further corrected by an algorithm which found the centroid of each point spread by considering the values of pixels in the circle with a fixed diameter around the approximate co-ordinates.

3. Once the centroids were found, a circle with a fixed diameter was marked around each of them and the intensities of the pixels in the circle were summed. Ideally, all the total counts at each aperture should have been the same because a uniform source of radioactivity was used and the diameter of all the apertures was the same. However, there were variations in the total counts because of slight non-uniformities in the size of the apertures as well as the variation in the positional sensitivity of the gamma camera. A sensitivity correction multiplier was
calculated for each aperture. The list of centroids and sensitivity correction factors was saved for further use.

4. Then the phantom images, which were a set of 10 images for pattern A and 25 images for pattern B, were read into MATLAB. One image was processed at a time and the total counts in an area around each centroid were calculated as described above. Then each of these total counts was multiplied by the corresponding sensitivity correction factor. This process was repeated for each of the 10 images for pattern A and 25 images for pattern B.

5. After this, a new image covering the same field of view but with pixel size 2 mm, for pattern A apertures and 1 mm for pattern B apertures, was reconstructed to put each of the corrected total counts in the corresponding pixel. The pixel size changed because all the pixels in one point spread were replaced by their total counts at a single pixel, equal to the size of the aperture used, in the new image. This was repeated for each of the 10 sets for pattern A and 25 sets for pattern B of the total counts, hence we could form a set of 10 for pattern A and 25 for pattern B new images.

6. For pattern A, each of the 10 images were shifted down by one row of pixels with respect to the previous image and then summed together to get the final image.

7. However, for pattern B, set of every 5 images (e.g. 1 to 5, 6 to 10 etc.) were shifted in the appropriate Y direction with respect to the previous image and summed together to get a set of 5 images. These 5 images were further by shifted
in the appropriate X direction with respect to the previous image and summed together to get the final image.

8. The final images were viewed with pixel sizes of 0.66 mm for pattern A and 0.33 mm for pattern B, so that they could be compared with the respective images obtained by the conventional method.

3.4 Statistical Analysis

The aim of the statistical analysis was to look for significant differences between the MTF values for the conventional technique and the proposed techniques. Because of the non-parametric nature of the data, a Kruskal-Wallis test was performed to test the two null hypotheses. The three MTF curves were compared together as a single experiment. The Kruskal-Wallis test statistic (H) was compared with the chi-square value ($\chi^2_{0.95}$) for two degrees of freedom at a 5% level of significance.

A multiple comparison procedure was carried out to find which of the three groups had significant differences, as demonstrated in [11]. To use the procedure, the mean ranks of each data set, obtained from the Kruskal-Wallis test, were used. Then, an experiment wise error rate ($\alpha$) was fixed as 0.15. The experiment wise error is an overall level of significance and is usually determined by the number of samples involved. A bigger error value is fixed for greater number of samples. Finally, both sides of the following inequality were determined [12]:

$$|R_1 - R_2| \leq Z_{(1-\alpha/k(k-1))} \times \sqrt{\left(\frac{N(N+1)}{12}\right)\left(\frac{1}{n_1} + \frac{1}{n_2}\right)}$$
where $R_1$, $R_2$ are the mean ranks and $n_1$, $n_2$ are the number of observations for the two data sets under comparison, $N$ is the number of observations in all samples combined, $k$ is the number of samples involved. If the left hand side (LHS) of the above inequality is greater than the right hand side (RHS) then the difference is declared significant at the $\alpha$ level.
CHAPTER IV
RESULTS

Experimental results of the study are presented in this chapter. The results are categorized based on the pattern of apertures used. The results obtained by using the pattern A apertures (Section 3.1.1) are presented in section 4.1 while those obtained by using pattern B apertures (Section 3.1.1) are presented in section 4.2. Both the results are compared with the images obtained by the conventional method.

4.1 Results with Pattern A apertures

The data acquisition protocol used was as follows:

1. Type of acquisition: Planar
2. Radiotracer used: Tc-99m, with photon energy of 140 keV.
3. Collimator: Low Energy Ultra-High Resolution (LEUR)
4. Pixel size: 0.66 mm x 0.66 mm
5. Time of acquisition:
   - Calibration data: 40 minutes
   - Proposed technique: 20 minutes @ 2 minutes / sample
   - Conventional technique: 20 minutes
Figure 4.1: Image obtained by the conventional technique

Figure 4.2: Image obtained by the proposed technique using the pattern A apertures
Figure 4.3: Horizontal profiles of the images for the two techniques

Figure 4.4: Flood source image used for calibration with pattern A apertures
Figure 4.5: Raw phantom image (sample 1) obtained using pattern A apertures

Figure 4.6: Processed phantom image (sample 1) obtained using pattern A apertures
A profile is a plot of pixel intensities for one row/column of the image matrix. Figure 4.3 illustrates the horizontal profiles through the center of images in Figure 4.1 and Figure 4.2. The horizontal profiles were plotted so that the difference between the minimum and maximum intensities in the two profiles could be compared. This difference in intensities is the contrast of the image and greater the difference means better contrast.

Figure 4.4 shows the image of the flood (uniform) source which was used for calibration as explained in section 3.3. Figure 4.5 shows the first of the ten sample images obtained by the sampling process explained in section 3.2.1 and is called as the raw image. Figure 4.6 shows the first of the ten images obtained after processing each raw image as explained in section 3.3.

4.2 Results with Pattern B apertures

The data acquisition protocol used was as follows:

1. Type of acquisition : Planar
2. Radiotracer used : Tc-99m (140 keV)
3. Collimator : Low Energy Ultra-High Resolution (LEUR)
4. Pixel size: 0.33 mm x 0.33 mm
5. Time of acquisition:
   - Calibration data : 40 minutes
   - Proposed technique : 100 minutes @ 4 minutes / sample
   - Conventional technique : 20 minutes
Figure 4.7: Image obtained by the conventional technique

Figure 4.8: Image obtained by the proposed technique using pattern B
Figure 4.9: Flood source image used for calibration with pattern B apertures

Figure 4.10: Raw phantom image (Sample 1) obtained using pattern B apertures
Figure 4.11 shows the image of the flood (uniform) source which was used for calibration as explained in section 3.3. The pattern B of apertures, as described in section 3.1.1, can be clearly seen in the image. Figure 4.10 shows the first of the twenty five sample images obtained by the sampling process explained in section 3.2.1 and is called as the raw image. Figure 4.11 shows the first of the twenty five images obtained after processing each raw image as explained in section 3.3.

4.3 Results for spatial resolution experiments

As explained in section 3.2.2, a line source method was used to evaluate the spatial resolution of the proposed technique. A plot of the pixel intensities through the center of the image of the line source in a direction perpendicular to it gives the LSF. The
LSF curves obtained for the three techniques were used to calculate their FWHM, FWTM and MTF (Sections 2.7.2 & 2.7.3). The FWHM and FWTM values obtained are presented in Table 4.1 while the MTF curves are plotted in Figure 4.8. The images of the line source are included in Appendix A and the measurement procedure of FWHM and FWTM from the LSF curves is included in Appendix B.

Table 4.1: FWHM and FWTM values obtained for the three techniques under comparison (Appendix B)

<table>
<thead>
<tr>
<th></th>
<th>Conventional Technique</th>
<th>Pattern A apertures (2mm in size)</th>
<th>Pattern B apertures (1mm in size)</th>
</tr>
</thead>
<tbody>
<tr>
<td>FWHM</td>
<td>3.83 mm</td>
<td>2.04 mm</td>
<td>1.35 mm</td>
</tr>
<tr>
<td>FWTM</td>
<td>7.48 mm</td>
<td>3.67 mm</td>
<td>2.65 mm</td>
</tr>
</tbody>
</table>

4.4 Results of the statistical tests

The MTF values for the three techniques were tabulated (Appendix C). The results for the Kruskal-Wallis test are shown in Table 4.2.

Table 4.2: Kruskal-Wallis test results at a significance level of 5% ($\alpha = 0.05$)

<table>
<thead>
<tr>
<th>Conventional technique data set</th>
<th>Pattern A data set</th>
<th>Pattern B data set</th>
<th>$H$</th>
<th>$\chi^2_{0.95}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rank Sum</td>
<td>Mean Rank (R₁)</td>
<td>Rank Sum</td>
<td>Mean Rank (R₂)</td>
<td>Rank Sum</td>
</tr>
<tr>
<td>366</td>
<td>14.07</td>
<td>162</td>
<td>32.4</td>
<td>267</td>
</tr>
</tbody>
</table>

$H > \chi^2_{0.95}$
The results of the multiple comparison procedure are shown in Table 4.3. Using the decision rule stated in section 3.4, it can be concluded that there exists significant differences between the MTF values of the conventional technique and those of the proposed technique for both pattern A and B apertures.

Table 4.3: Multiple comparison procedure results

| \( |R_1 - R_2| \) \( (LHS_1) \) | Significance value \( (RHS_1) \) | \( |R_1 - R_3| \) \( (LHS_2) \) | Significance value \( (RHS_2) \) |
|-----------------|-----------------|-----------------|-----------------|
| 18.32 \( (LHS_1) \) | 11.19 \( (RHS_1) \) | 15.59 \( (LHS_2) \) | 8.86 \( (RHS_2) \) |

\( LHS_1 > RHS_1 \) \( LHS_2 > RHS_2 \)

Reject \( H_{01} \) \( \) Reject \( H_{02} \)
CHAPTER V
DISCUSSION & CONCLUSION

As discussed in the previous chapters, the spatial resolution of the planar images obtained by the current technique of nuclear imaging is limited by the point spread function of the gamma camera. This study attempts to improve the point spread function so as to increase the spatial resolution of the gamma camera images. To validate this technique, two variations (Pattern A and Pattern B apertures) were developed and tested. The images obtained were compared with images obtained by the nuclear imaging technique currently in use.

One of the aims of this experiment was, to qualitatively (visually) compare the images obtained with the proposed technique to those obtained with the conventional technique. First, we compare the results of the new technique, using the pattern A apertures, to those of the conventional technique (Figure 4.1 vs. 4.2 & Figure 4.3). The improvements in image quality due to the proposed technique are listed below:

1. The boundaries around the dark circles are well defined with the proposed technique, especially, in the section with the rods of the smallest diameter.
2. The contrast is better because the intensities in the dark circles are very low as compared to those in the conventional image.
3. The profile in the Figure 4.4 varies smoothly. This is because the statistical noise in the pixel intensities is reduced by the proposed technique. Thus, the image with the proposed technique better reproduces the true distribution of radioactivity in the phantom.

4. The difference between the peak and the lowest intensities is 268 for the proposed technique against the 165 for the conventional technique, which is a 62.5% improvement in the contrast. As explained in section 2.6.4, contrast is the difference between the intensities in parts of the image corresponding to the areas of different levels of radioactive uptake.

Next, we compare the results of the new technique, using the pattern B aperture, to those of the conventional technique (Figure 4.7 vs. 4.8). The image in fig. 4.8 shows vast improvements over the image in Figure 4.7, as described below:

1. The spatial resolution is better as evident from the section with the smallest rods. In Figure 4.7, these small dark circles are difficult to see, however, in Figure 4.8 they are clearly visible with well defined edges.

2. The image with the proposed technique is bright with very good contrast while the image with the conventional technique is dull.

Figure 4.11 illustrates the MTF curves for the three techniques. As stated in section 2.7.3, for faithful representation of the original image it is desirable that the MTF curve of the imaging system be flat and near unity. It can be seen that the MTF curve for the conventional method approaches zero very quickly (at a spatial frequency of about 0.3 lp/mm). In comparison, the two MTF curves for the proposed method fall very slowly...
over the same spatial frequency range indicating that the proposed technique better reproduces the original image. Also, it can be seen that although the MTF curves for pattern A and pattern B apertures are similar at lower spatial frequencies, as the frequency increases the difference between them increases. The curve for pattern A falls a little faster than the curve for pattern B. This indicates that pattern B (with aperture size 1 mm) provides the best spatial resolution among the three methods.

Thus, conclusions drawn from the qualitative comparisons, as listed in the previous paragraphs, are well supported by the quantitative comparisons done between the FWHM, FWTM and the MTFs of the three techniques (Tables 4.1 to 4.3). The following conclusions were drawn from the results of the statistical tests (Tables 4.2, 4.3):

1. We rejected the null hypothesis $H_{01}$. Thus it can be concluded that there is a significant difference in the MTFs of the conventional technique and the proposed technique with pattern A apertures.
2. We rejected the null hypothesis $H_{02}$. Thus it can be concluded that there is a significant difference in the MTFs of the conventional technique and the proposed technique with pattern B apertures.

5.1 Salient features of the proposed technique

1. It improves the detector’s spatial and contrast resolutions.
2. Spatial resolution is not limited by the point spread function of the detector but is equal to the size of the apertures on the special collimator.
3. The hardware required is inexpensive and easy to manufacture using the standard collimator manufacturing techniques.

4. Allows the trade-off between spatial resolution, contrast and image acquisition time. For example, using the proposed technique with the pattern B apertures would give a high resolution image but it will also increase the scanning time. So we can use the pattern A apertures to decrease the scan time and accept decreased resolution, which is still better than the conventional technique.

5. The use of solid state detectors, like Cadmium Zinc Telluride, can be greatly increased with this method. The field of view of these detectors is limited by their small size due to the high cost of fabrication of the necessary electronics for each element. With the use of the proposed technique, larger detectors can be fabricated, but the effective size of each detector element can be made as small as desired using the special collimator described in the technique.

6. The technique is ideally suited for small animal imaging because there are no limitations on the radionuclide dose and the imaging time.

7. The technique can be easily modified for routine clinical scans. In a clinical environment, it may not be practical to move the patient for acquiring all the samples of the image as the patient maybe in pain. In this case, the technique can be implemented by moving the special collimator after every sample of the image.

5.2 Conclusions

This thesis presents a novel technique to increase the spatial resolution and contrast of the gamma camera. Based on the results from the visual comparisons and the
statistical tests performed, it was concluded that the proposed technique was successful and gave better spatial resolution and contrast images when compared to the conventional technique.

5.3 Future Work

1. The current method can be refined by improving the process of acquisition of the image samples, for example, the movements of the XY table can be made automatic and more precise by using a micro-controller.

2. A correction factor can be developed and used for correcting the overlap of adjacent point spread functions, especially, when collimators with smaller and tighter apertures are used.

3. Since the size of the images is reduced post-processing, a technique to increase the image size to the original size with no or minimal loss of resolution and contrast can be developed.

4. The technique can be extended to tomographic (SPECT) imaging combined with the use of iterative reconstruction algorithms because of the limited nature of the dataset. Investigations as to the number of angles needed for sufficient reconstruction data and the number of bins to be used for the reconstruction can be made.
REFERENCES


APPENDIX A

LINE SOURCE IMAGES USED FOR MTF CALCULATION

Figure A.1: Line source image obtained by the conventional method
Figure A.2: Image of the line source obtained with pattern A apertures

Figure A.3: Image of the line source obtained with pattern B apertures
APPENDIX B

FWHM & FWTM CALCULATIONS

Figure A.3: LSF for the conventional technique

<table>
<thead>
<tr>
<th>Peak Intensity (M)</th>
<th>M/2</th>
<th>X₁</th>
<th>X₂</th>
<th>FWHM (X₂ – X₁) * 0.33 mm</th>
<th>M/10</th>
<th>X₃</th>
<th>X₄</th>
<th>FWTM (X₄ – X₃) * 0.33 mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>370</td>
<td>185</td>
<td>32.07</td>
<td>20.46</td>
<td>3.83 mm</td>
<td>37</td>
<td>38.22</td>
<td>15.53</td>
<td>7.48 mm</td>
</tr>
</tbody>
</table>
Figure A.4: LSF for the proposed technique (pattern A)

<table>
<thead>
<tr>
<th>Peak Intensity (M)</th>
<th>( \frac{M}{2} )</th>
<th>( X_1 )</th>
<th>( X_2 )</th>
<th>FWHM</th>
<th>( \frac{M}{10} )</th>
<th>( X_3 )</th>
<th>( X_4 )</th>
<th>FWTM</th>
</tr>
</thead>
<tbody>
<tr>
<td>23960</td>
<td>11980</td>
<td>4.47</td>
<td>5.5</td>
<td>2.042 mm</td>
<td>2396</td>
<td>4.06</td>
<td>5.9</td>
<td>3.674 mm</td>
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</tbody>
</table>
Figure A.5: LSF for the proposed technique (Pattern B)

<table>
<thead>
<tr>
<th>Peak Intensity $(M)$</th>
<th>$\frac{M}{2}$</th>
<th>$X_1$</th>
<th>$X_2$</th>
<th>FWHM $(X_2 - X_1) \times 1, \text{mm}$</th>
<th>$\frac{M}{10}$</th>
<th>$X_3$</th>
<th>$X_4$</th>
<th>FWTM $(X_4 - X_3) \times 1, \text{mm}$</th>
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</thead>
<tbody>
<tr>
<td>2672</td>
<td>1336</td>
<td>8.14</td>
<td>9.5</td>
<td>1.356 mm</td>
<td>267.2</td>
<td>7.23</td>
<td>9.9</td>
<td>2.659 mm</td>
</tr>
</tbody>
</table>
## APPENDIX C

### MTF VALUES FOR THE THREE TECHNIQUES

<table>
<thead>
<tr>
<th></th>
<th>Conventional technique</th>
<th>Pattern A apertures</th>
<th>Pattern B apertures</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Frequency</strong></td>
<td><strong>MTF</strong></td>
<td><strong>Frequency</strong></td>
<td><strong>MTF</strong></td>
</tr>
<tr>
<td><strong>(lp/mm)</strong></td>
<td></td>
<td><strong>(lp/mm)</strong></td>
<td><strong>MTF</strong></td>
</tr>
<tr>
<td>0.0000</td>
<td>1.0000</td>
<td>0.0000</td>
<td>1.0000</td>
</tr>
<tr>
<td>0.0588</td>
<td>0.7946</td>
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<td>0.9286</td>
</tr>
<tr>
<td>0.1765</td>
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<td>0.2353</td>
<td>0.0386</td>
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<td>0.7798</td>
</tr>
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<td>0.2941</td>
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<td></td>
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</tr>
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<td>0.3529</td>
<td>0.0126</td>
<td></td>
<td>0.3529</td>
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<td>0.4118</td>
<td>0.0107</td>
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