Revival of a Gamma Camera

Adaptation of new Electronics and conversion of the Camera to a Demonstrator in Medical Imaging

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Abstract

Gamma camera is the most commonly used device in radionuclide imaging. A gamma camera of type Starcam Mobile 300 A/M manufactured by General Electric Medical Systems was donated to KTH by Danderyds Hospital, Nucl. Medicine Dept.

The aim of this thesis is to convert the gamma camera to be a demonstrator in a new laboration in the course of Medical Imaging at KTH.

The most of the original analogue and digital electronics were removed and the analogue control board was modified with 4 newly built stretchers. The data acquisition system has also been changed and now consists of a dedicated ADC card in a new PC with Windows XP and a data acquisition program.
1 Introduction

1.1 History

The discovery of natural radioactivity by Henry Bequerel in 1896 was the start that led to the development of nuclear medicine [1]. The year before Wilhelm Röntgen discovered X-rays and as early as 1897, it was concluded that X-rays could be used for therapeutic as well as diagnostic purposes [2,3]. The use of X-ray radiation for patient therapy became a clinical routine in the early 1920s.

The 1920s and 1930s was the time of rapid development of nuclear biology and medicine. Radioactive phosphor ($^{32}$P) was administrated to animals and the possibility to study a metabolic process in a living animal was demonstrated for the first time. Shortly after, $^{32}$P was used to treat patients with leukaemia. In the late 1930s thyroid physiology was studied using radioactive Iodine ($^{131}$I).

After a series of discoveries of various radioactive materials in the late 1930s, the radionuclide technetium-99m becomes the most important substance in the nuclear medicine. This artificially made radioactive element was discovered in 1938. The $^{99m}$Tc, because of its short half-life, low radiation dose and chemically reactive nature, was considered to be ideal for human imaging. Still today, this radionuclide is the most widely used in nuclear medical examinations [4].

Another major achievement of the 1930s was the development of an ion accelerator, the cyclotron [5, 6]. The cyclotron could bombard a target with high-energy particles, enable to create new radioactive nuclides that had not existed earlier. Several new artificial isotopes appeared as new radiopharmaceutical after the implementation of cyclotrons [7].

At the beginning of the 1940s, the first dedicated cyclotron for medical use appeared, and as early as 1941, patients were being treated with radioiodine.

The 1950s represent a remarkable growth. Several new pharmaceuticals were proposed and implemented and in the mid 1950s even organ-specific radiopharmaceuticals began to appear.

The 1950s saw two major milestones in the nuclear medicine development. As noted above, $^{99m}$Tc had been produced since 1938. However, because of its short half-life (~6 hours), it was not routinely available. A group at Brookhaven National Laboratory developed a system for local production of technetium that is still used today. In this generator system $^{99m}$Tc is eluted from the much longer-lived $^{99}$Mo (half-life ~66 hours) [4, 8].

The other major development of the 1950s was Hal O. Anger’s work on the development of the gamma camera for medical imaging. In 1952, Anger first announced a gamma camera with a scintillation crystal acting as an image intensifier for a film [9]. The first electronic gamma camera with multiple photomultiplier tubes was reported in 1957 and presented by Anger 1958 [10]. The development of the gamma camera made it possible to show in real-time, the blood flow in a patient, watch the kidneys function, examine the liver as it generates the bile etc.

The combination of the development of the technetium-99m generator and the use of the gamma camera started a new era in nuclear medicine.
The camera, described by Anger became the predecessor of the present gamma cameras where the input is stored in a computer memory and the output image can be manipulated and presented in many different ways. Tomographic image techniques are developed and with this nuclear medicine technique, the medical doctors are able to view sections of an organ or allow three-dimensional imaging (as in Single Photon Emission Computed Tomography, SPECT).

1.2 Radionuclides for Imaging

Various pharmaceuticals can be labelled with radionuclides to form radiopharmaceuticals [8, 11]. Administration of these radiopharmaceuticals brings the radionuclide to the designated tissue or passageways. This administration is performed through injection, inhalation or orally. The small amount of radiopharmaceuticals used, is in the order of nanograms and has little effect on the biochemical and physiological functions of the patient.

Radionuclides desirable for nuclear medicine imaging should have physical characteristics as, suitable physical half-life, absence of particle emission, decay via photon emission with energies high enough to penetrate the body with minimal tissue attenuation, but low enough to minimise the thickness of collimator septa.

The radionuclide that fulfils most of the desirable criteria is $^{99m}$Tc, which is used in more than 90 % of all nuclear medicine studies. The short half-life and the absence of $\beta^+$- and $\beta^-$-emission results in a low radiation dose to the patient. It is also easy to produce this radioactive source in a separator, known as $^{99m}$Tc generator.

1.2.1 The $^{99m}$Tc Generator

The generator is based on a separation of the $^{99m}$Tc daughter from the $^{99}$Mo mother nuclei [4, 8, 11]. The mother-daughter mixture contains an increasing amount of the $^{99m}$Tc daughter isotope that can be eluted for radiochemical use. The $^{99}$Mo isotope is produced mostly in nuclear reactors by neutron capture when irradiating $^{98}$Mo with thermal neutrons.

$$^{98}\text{Mo}(n,\gamma)^{99}\text{Mo}$$

The $^{99}$Mo is then transported to medical cliniques for extraction of the $^{99m}$Tc radioisotope. The parent $^{99}$Mo is absorbed by an aluminium column ($\text{Al}_2\text{O}_3$) inside the generator. $^{99}$Mo has a half-life of 66 hours and decay with electron emission:

$$^{99}\text{Mo}\rightarrow^{99m}\text{Tc} + e^- + \bar{\nu}_e$$

$^{99m}$Tc is a metastable radioisotope, with a half-life of 6.02 hours decaying to the stable nucleus $^{99}$Tc by emitting gamma radiation of 140 keV. The $^{99m}$Tc doesn’t bind to aluminium and thus can be eluted from the alumna column with saline. The problem associated with this method is that part of $^{99}$Mo and aluminium ions are also eluted with $^{99m}$Tc by saline. This so-called “break through” should be avoided to ensure purity of the daughter radioisotope for minimising the patient dose.
Figure 1-1 The principle of a generator for production of $^{99m}$Tc
2 The Anger Gamma Camera

The Anger type of a modern gamma camera consists of a camera-head, a gantry, an electronic processing unit for position and energy determination and a computer for data acquisition and image reconstruction.

2.1 The Camera-Head

The camera-head contains the detector assembly, consisting of a scintillation crystal, photomultiplier tubes and light guides, and a collimator, shielded with lead to reduce the detection of background radiation of the crystal.

2.1.1 The Scintillation Crystal

Most common is to use a thin and large circular crystal made of sodium iodide (NaI), activated with a trace of thallium (Tl) as scintillation. It has a diameter of 30-50 cm and its thickness is a trade-off between intrinsic resolution (the spatial resolution of the system without collimator) and detection efficiency of the incident photons. Due to the fact that most gamma cameras are designed for imaging $^{99m}$Tc-labeled pharmaceuticals with 140 keV low-energy $\gamma$ photon emissions, a crystal thickness of 9-12 mm provides the best compromise between resolution and efficiency [4, 12, 13]. The intrinsic efficiency for this thickness is high, as can be seen in Figure 2-1.

![Figure 2-1 The intrinsic total detection efficiency of gamma rays with perpendicular incidence for various thickness of a NaI(Tl) scintillator](image)

The energy resolution is typically 15-20 keV at 150 keV (see Figure 2-2) but a crystal of this thickness becomes rapidly less sensitive above 300 keV. Due to the crystals high atomic number (Z=53) and high density ($\rho$=3.69 g/cm$^3$) it absorbs above 90% of the 140 keV $^{99m}$Tc gamma rays, principally by the photoelectric process.
The Na(Tl) scintillator crystal emits blue-green light (means a wavelength $\lambda \approx 415$ nm) and this spectral output matches well the light-photon response characteristics of standard bialkali (SbK$_2$Cs) photomultipliers (see Figure 2-3) [4].

The crystal is fragile, very sensitive against temperature changes and easy to damage. In addition it is hygroscopic, therefore it is encapsulated in an aluminium cylinder with one flat window made of Pyrex glass facing the photomultiplier tubes.
2.1.2 The Photomultiplier Tubes

A photomultiplier tube consists of an evacuated glass envelope, containing a photocathode, an anode and several (about 10) intermediate electrodes (so called dynodes). The photocathode is made of a photoemissive material that emits electrons (in the order of 1 electron per 5 or 10 light photons for standard photomultiplier tubes) when stricken by light photons in proportion to the intensity of the light. The emitted electrons will then, since the photocathode is maintained mostly at ground potential, be accelerated toward the first dynode having a higher potential. The dynodes are covered by a material that emits secondary electrons (normally 3-4) when stricken by an electron. Those will then be accelerated to strike the next dynode, which have higher voltage than the previous one, and so on. Finally after about ten dynode steps the number of electrons have been multiplied by a factor $\sim 10^6$. Therefore, the current output is proportional to the number of light photons and each initial flash of light produces a pulse of charge or voltage large enough to be measured [4, 7, 12, 13]. The output of each photomultiplier tubes is fed to a preamplifier for further amplification and pulse shaping.

![Figure 2-4 The principle of a photomultiplier tube](image)

2.1.3 Light Guides

It is usually necessary to provide a light guide to interface the scintillation crystal to the photomultipliers. This light guides are made of transparent plastic with a refractive index close to 1.85, the refractive index of NaI(Tl), and they are carefully shaped to match the shape of the photomultiplier entrance window [13]. The light guide helps to minimise the light losses in the transfer of light from the scintillation crystal to the photomultiplier. It will also minimise the variation of light collection efficiency across the face of the crystal.

2.1.4 The Detector Assembly

The photomultiplier is mounted on the crystal as mentioned above mostly via light guides, and the optimal arrangement of photomultiplier tubes (with circular or hexagonal cross sections) closely packed onto the surface of a circular scintillation crystal is a hexagonal array (see Figure 2-5) containing 7, 19, 37, 61, 91 etc photomultipliers [4]. In a modern circular gamma camera 37-91 photomultipliers are used. The number of photomultiplier tubes affects the intrinsic resolution of the camera.
Figure 2-5 The optimal arrangement of photomultiplier tubes onto the surface of a circular scintillation crystal

2.1.5 Collimators

A collimator is used to select the direction of the photons incident to the camera and will determine the geometrical field of view (FOW). It also essentially determines the spatial resolution and the sensitivity of the system [4, 7, 11, 13]. A range of collimators is required to image different photon energies and to achieve sufficient compromise between spatial resolution and sensitivity. The most commonly used are the parallel-hole, converging, diverging and pinhole collimators. These types exist as low- or middle-energy collimators depending on the required thickness of absorber.

If looking at the photon energy dependence we can see that a general-purpose low-energy multihole collimator consist of a lead disk, typically 25 mm thick, and are equipped with ~25 holes/cm$^2$ closely packed circular or hexagonal holes. The holes have a diameter of 2.5 mm and the lead between them, called septa is 0.3 mm thick. This type of collimator can be used for energies up to 150 keV, e.g. $^{99m}$Tc.

A high-resolution collimator would have more and smaller holes per cm$^2$ and therefore lower sensitivity. This kind of collimator is often used when high resolution is required and the total number of detected photons is acceptable. This depends of course on the amount of radioactivity and the imaging time.

A “high-sensitivity” collimator has fewer and larger holes and consequently poorer resolution. This type is used for dynamic imaging where short exposure times are necessary and poorer resolution must be accepted.

For energies up to 400 keV, e.g. $^{111}$In, $^{67}$Ga, and $^{131}$I, one will use a medium-energy collimator. It has thicker septa (1.4 mm) and consequently fewer holes and much lower sensitivity.

The spatial resolution is strongly dependent on the distance between the source and the collimator [4]. In Figure 2-6 one can see that the spatial resolution ($R_s$) vary with the source-to-collimator distance for given values of intrinsic resolutions ($R_i$).
2.1.5.1 The Parallel Multihole Collimator

The parallel-hole type of collimator is the most commonly used in radionuclide imaging [4, 7, 11, 13]. It consists, as for all multihole collimators of an array of holes separated by lead septa. The holes are orientated perpendicular to the surface of the crystal, and therefore the size of the image of the radionuclide distribution has a one-to-one ratio. As mentioned above the collimator has a major effect on the spatial resolution and determines the detection efficiency of the gamma camera. The spatial resolution FWHM (Full Width Half Maximum) is related to the width of the average intensity distribution from a point source placed at a distance $z$ from the collimator face (see Figure 2-7).
The spatial resolution can be expressed as:

\[ R_s = \frac{2a(d + z + b)}{d} \]

where \( R_s \) is the spatial resolution, \( 2a \) is the diameter of the collimator hole, \( d \) is the thickness of the collimator and \( b \) is the gap between the back of the collimator and the image plane inside the crystal (see figure 2-7) [11].

### 2.1.5.2 The Diverging Multihole Collimator

The diverging multihole collimator [4, 7, 11, 13], were used earlier for imaging large objects, e.g. lungs, with a small camera, due to its reduction effects. Nowadays the small cameras have almost disappeared and have been taken over by the large field of view cameras.
2.1.5.3 The Pinhole Collimator

A pinhole collimator consists of a single small aperture 3-5 mm diameter at the end of a conical lead shield containing sufficient attenuating material to minimize photon penetration for energies up to 500 keV [4, 7, 11, 13]. It is used primarily for high resolution imaging of small organs. The spatial resolution can be expressed as:

\[ R_s = \frac{2a(d + z + b)}{d} \]

with similar notations as earlier (see Figure 2-9).

A significant magnification can also be achieved, although at the expense of some distortion particularly at the image edge. The magnification factor depends on the ratio z/d (see figure 2-9) [11]. The image becomes due to the geometry inverted.

![Figure 2-9 The principle of a pinhole collimator](image)

2.2 The Electronic Processing Unit

The two properties that are necessary to deduce for each scintillation event, is their coordinates in the crystal and their amplitude, to avoid detection of incompletely absorbed or Compton scattered photons or other distorted signals.

2.2.1 Position Determination “the Anger Logic” and Signal Discrimination

To provide positional information from the analogue outputs of the photomultiplier tubes, a resistive-coupled network is used. The output signals are attenuated by a resistor matrix, where the values of the resistors correspond to the positions of the photomultiplier tubes. An example of a simple resistor network containing outputs from seven photomultiplier tubes is shown in Figure 2-10.
Figure 2-10 An example of a resistor network used to provide positional information from seven analogue photomultiplier outputs. The scheme is from an early paper published by Hal O. Anger, the inventor of the gamma camera.

In Figure 2-10 the outputs of the seven photomultiplier tubes (PMTs) are each attenuated by four resistors. The output of the PMT 1 in the centre is of course attenuated by four resistors with the same value and the others with resistor values depending on their positions. The corresponding attenuated PMT outputs are added to provide four signals ($X^+, X^-, Y^+, Y^-$). The relative intensity of these signals will determine the x-y position, necessary for the image reproduction. The total intensity of the signal, $\Sigma$, is given by:

$$\Sigma = x^+ + x^- + y^+ + y^-$$

and the x and y positions by:

$$x = k \left( \frac{x^+ - x^-}{\Sigma} \right)$$

$$y = k \left( \frac{y^+ - y^-}{\Sigma} \right)$$

where $k$ is a constant. The above formulas in the literature are called as “the Anger logic” [4, 14].

This, the total energy, $\Sigma$-signal is analysed by a discriminator, (a Single Channel Analyser) with discrimination levels set to determine whether the $\Sigma$-signal corresponds to the full energy of the gamma photon, i.e. the photopeak [4, 15, 16,
The discriminator is used to choose the photopeak of the radionuclide that are used.

The lower level discriminator is set a few keV under the photopeak to avoid Compton scattered events to be registered and the upper level is likewise set to a few keV over the photopeak to avoid pile-up and other high-energetic events. If the $\Sigma$-signal fits the energy window of the discriminator a gate will open and the four analogue position signals will be digitised by the ADCs (Analogue-to-Digital Converters) of the data acquisition system and stored in a computer.

2.3 Application of Computers in the Radioisotope Imaging

Prior to the late 1970s, due to obvious reason gamma cameras were generally not associated with computers. Today, the applications of computers in radioisotope imaging include data acquisition, data processing, image display and manipulation, data storage and system control. These applications will briefly be discussed below.

To obtain static or dynamical 3-D images with a gamma camera SPECT (Single Photon Emission Computed Tomography) technique is used. The SPECT imaging modality is also discussed in this chapter.

2.3.1 Data Acquisition

Data from a gamma camera can be acquired and stored in a computer either in list mode or frame mode. In list mode the converted signals are stored as they arrive, as numbers corresponding to the x-y position of the registered photons [4, 7].

In the more commonly used frame mode, the acquired x-y data directly increment a two-dimensional memory matrix in the computer. The analogue pulses from the gamma camera are digitised in such a way that the storage location for the detected event is determined. If the energy of the pulse fits a preset energy window corresponding to the photopeak, the event is allowed to update the memory matrix.

There are different types of frame mode of data acquisition like the static, the dynamic and the gated [4, 7].

In static acquisition, a single image is acquired for a preset time interval or until the number of counts in the image reaches a preset number.

In dynamic acquisition, a series of images is acquired one after another, for a preset time per image. This type of acquisition is used to study dynamic physiological processes, such as the transport of a radiopharmaceutical in the human body.

Finally, gated acquisition is used to study dynamic processes that occur too rapidly to be effectively portrayed by dynamic acquisition e.g. if each image in the sequence contains too little data to be statically valid. If the dynamic process is repetitive, gated acquisition may permit the acquisition of an image sequence that accurately depicts the dynamic process. Gated acquisition is frequently used for evaluating cardiac mechanical performance and requires a physiologic monitor that provides a trigger pulse to the computer at the beginning of each cycle of the process being studied.

When using gated acquisition, first of all, space for a desired number of images is reserved in the computer memory. Next, several cycles are timed and the average time per cycle is determined. The time per cycle is divided by the number of images in the sequence to obtain the time per image $T$. Then the acquisition starts. When the first trigger pulse is received, the data from the acquisition interface is sent to the
first image in the sequence for a time $T_i$. Then it is stored in the second image in the sequence for a time $T_i$, after which it is stored in the third image and so on until the next trigger pulse is received. The process then starts again with the data being added to the first image matrix for a time $T_i$ and then to the second and so on. This loop continues until a preset acquisition time has elapsed, enabling sufficient counts to be collected for each image in the sequence to form a statically valid depiction of an average cycle.

In the case of cardiac studies an EKG monitor provides the trigger pulse to the computer when it detects a QRS complex (a peak in the EKG diagram depending on depolarisation during the contraction of the auricle). A typical value for the preset time interval is ten minutes. The EKG cycle is typically divided in 16-24 images.

![Figure 2-11 The principle of gated acquisition of a cardiac image sequence](image)

### 2.3.2 3-D Imaging (SPECT-Technique)

In nuclear medicine, tomography using single photon emitters is advantageous compared to planar imaging in many applications because of improved diagnostic accuracy. Currently, the most widespread method in use is Single Photon Emission Computed Tomography (SPECT). A SPECT system is characterized by one or more gamma cameras with collimators, mounted on a gantry, which allows rotation around the long axis of a supine patient [4, 11, 19].
During the rotation procedure, multiple two-dimensional projections of the three-dimensional radiopharmaceutical distribution can be acquired and stored on the computer. If a large number of linear and angular data samples are taken, it is possible to reconstruct the cross-sectional images that represent the radiopharmaceutical distributions in the body. A standard SPECT-examination involves 60 to 128 images taken from different angles around the object. The images are processed and combined in a computer to produce a section image. The image reconstruction, it will be discussed below, is normally performed with filtered backprojection algorithms.

When the projection images are acquired, they are usually corrected for axis-of-rotation (AOR) misalignments. With AOR-misalignments means that the distance, from the centre of the detector to the centre of the rotation, varies for different projection angles. This misalignment is often mechanical, due to inexact positioning of the camera head on the gantry. If no correction is made, a loss of spatial resolution in the image will occur.

Another error that one has to cope with is absorption. Gamma rays emitted from activity deep inside the patient produce fewer counts due to attenuation, than those emitted close to the surface of the body. This attenuation-effect needs to be corrected for and the most common procedure is to presume a constant attenuation coefficient throughout the body. Attenuation corrections can however overcompensate for the true absorption or cause streak artefacts, so its proper functioning should be verified using phantoms before it is used in clinical studies. The completed section image can visualize any slice through the patient within the projected volume and appropriate computer codes are used to present a 3-dimensional volume image of the organ that has been studied.
2.3.2.1 3-D Data Processing

To obtain a 3-D image with the SPECT method one has to apply more or less complex data reconstruction techniques such as back projection.

The straightforward back projection technique includes both back projection of the projection data and superposition of all projections at angles $0-\pi$ around the object [18]. Mathematically back projection can be expressed as:

$$f_{bp}(x, y) = \int_0^\pi p(x \cos \theta + y \sin \theta, \theta) d\theta$$

where $f_{bp}(x, y)$ is the picture value obtained after reconstruction at the position $(x, y)$ of the tomographic plane and $p(x \cos \theta + y \sin \theta, \theta) = p(r, \theta)$ is projection data at the point $(r, \theta)$ along the projection axis (see Figure 2-13).

![Figure 2-13 Back projection. A stationary x-y coordinate system defines the tomographic plane with reference to the body. A rotating r-s system defines the same plane with reference to the detector positioning. A coordinate transformation from the r-s to the x-y system is performed when back projecting the number of recorded events, $(p(r, \theta))$, into the tomographic x-y plane.](image)

The back projection has a serious drawback. The images, which are obtained, are blurred. To get a better image quality frequency filtering is used. In the spatial domain it is analogue with spatial filtering, which implies a convolution operation between the projection data and a spatial filter function before back projection.
2.3.3 Image Display and Manipulation

After the data acquisition, the correction and the processing have to take place to create an image. It is advantageous, to highlight the diagnostic information of interest. Modern gamma cameras provide the possibility to alter the dynamic range to optimise the information of the displayed image.

If one wants to look on high- and low contrasts regions separately, it is possible to enhance the displayed image by gray-level windowing to pick out different intensity regions. To get a quick overview of an image it is preferable to use high-contrast pseudo color presentation, which gives image with contours [4].

The number of pixels used for displaying is important. To few pixels cause pixel edge distortions. Other important features are the ability to rotate the image, to provide multi-image display and also to add textual information.

Nowadays, when looking at dynamic or tomographic images ciné displays are often used to show the images sequentially. The varying display rates might be used to simulate temporal changes or present spatial rotation.

2.3.4 Data Storage

Gamma cameras generate large volumes of data produced during dynamic and tomographic acquisitions. The acquisition requires modest short-term memory and large hard-disc space for semi permanent storage.

2.3.5 System Control

The role of computers in different types of system control has also increased since low-cost and powerful microprocessors became available.

The control of the gamma camera gantry is necessary for provision of multiview data acquisition [4]. It provides the motion that is needed for a lot of imaging methods, such as rotation in SPECT imaging, transitional in planar scanning and longitudinal tomography. In all cases the computer must store the position information of the gantry. This information is an essential part of the data acquisition and necessary to provide correct spatial and angular information required for data processing.

Another type of movement that has to be controlled for some types of scanning camera systems is the patient couch [4]. For example when using circular gamma camera orbits in SPECT, the accurate positioning of the couch is necessary for optimising the multiview information and to ensure that no points of interest in the ROI (Region Of Interest) is emitted outside of the field of view. Another possibility is to use elliptical gamma camera orbits, which allow the camera head to be placed optimally close to the patient at all angles and consequently maximise the spatial resolution. This process is best done by computer control of the patient couch during acquisition, so that the vertical and lateral motion is closely synchronised with the rotation of the gantry.
3 The Rebuilt GE Starcam 300A/M

The Physics Dept. of KTH received an old gamma camera manufactured by General Electric Medical Systems (GEMS) of the type GE Starcam 300A/M from the Nuclear Medicine Dept. of Danderyd Hospital. The goal of this Master Thesis was to modernise and convert this gamma camera so that it can be used as a demonstrator/laboratory exercise instrument at KTH. This conversion consisted mainly of the replacement of its digital and data acquisition part with dedicated hardware as input and a new computer with data acquisition software. The design and production of this hardware and software is made by the expertise at ATOMKI, Debrecen, Hungary.

The remaining parts of the gamma camera were either preserved in its original shape like the camera-head and gantry or been adopted to fit the new digital system. The analogue part was modified, also with help of ATOMKI, to suite the inputs of the new digital part of the camera.

3.1 The Imaging Detector

As mentioned above the imaging detector in the camera-head, was kept in its original shape. The imaging detector of the gamma camera consists of a 300 mm diameter NaI crystal housed in an aluminum and lead shield casing.

Mounted on one side of the crystal are 37 hexagonal photomultiplier tubes with associated preamplifier assemblies with an optical couplant.
The four \((X^+, X^-, Y^+, Y^-)\) analogue signals output

High Voltage Power Supply

PM Socket Connector, Amplifier and Resistors

Figure 3-1 The arrangement of photomultiplier tubes with associated preamplifier assemblies inside the detector head

On the other side is the exchangeable lead collimator mounted in order to condition the incoming gamma rays. The camera-head also contains the integrated detector power supply that delivers +22.5 V, -7.5 V to the preamplifiers and the negative high voltage (-HV) to the photomultiplier tubes.

Gamma rays that manage to pass the collimator are converted to light quanta by the scintillation crystal. The photocathode of the photomultiplier tubes convert the light to electrical signals. The signals are then amplified in the ten-step dynode ladder and the resulting charge is collected by the anode of the tube [20].
The 37 anode pulses are further amplified by an operational amplifier, placed on PM Tube socket printed-circuit board (the lower “pizza board”) and the output signal is fed through a ribbon cable to the upper “pizza board” (pin 8 of the interconnecting DIL socket) [20], (see Figure 3-1).

The amplitudes of the 37 PM tube signals depend on the place where the gamma photon is converted into scintillating light in the NaI crystal and of course on the total energy deposited. The sum of the 37 signals will give the energy and their relative amplitudes will determine the position in the x-y plane. The closest tube to the scintillation spot will deliver the largest amplitude, while the most distant ones will deliver the smallest.

To determine the position in the x-y plane the anode signal from each PM tube is individually attenuated by four different resistors according to the position of the tube.
Figure 3-3 The signal from PM tube nr. $i$ is attenuated by four resistors according to the position of the tube. The sum ($\Sigma$) is also detected to analyse the energy of the pulse.

The resistors with values ($R_i^{x+}$, $R_i^{x-}$, $R_i^{y+}$, $R_i^{y-}$) for each of the 37 anode outputs will create the resistor matrix with the values that are presented in Appendix p. V.

The respective coordinate signals ($X^+, X^-, Y^+, Y^-$ resp. $Y^-$) from the PM tubes are interconnected and will form an analogue signal bus for determining the position. A fifth signal in this bus is the sum of all amplitudes delivered by the 37 PM tubes. These five signals will then be sent to the analogue control board, (see Figure 3-3). (Electric schemes see appendix.)

To control the gain of the 37 photomultiplier tubes the system is equipped with an automatic gain control. To each photomultiplier tube a reference LED (Light Emitting Diode) is attached. A common oscillator turns these on every millisecond for 1-2 microseconds [20].
The amplitudes of the anode pulses are compared with a reference voltage setting and the gain of the tube is controlled by adjusting some of its dynode voltages (see Figure 3-4). However, if the system is unable to make the appropriate adjustments, a yellow warning LED on the preamplifier assembly (situated on the upper pizzaboard) will illuminate to indicate that a specific photomultiplier tube has an auto-tune failure [20], (see Figure (3-5).
3.2 The Collimators

The gamma camera is equipped with low energy collimators suitable for 140 keV (\(^{99m}\)Tc) studies. There are 3 multihole collimators (a general purpose parallel hole, a high-sensitive parallel hole and a diverging) and 1 pinhole collimator with three possible hole sizes, (see Figure 3-6 and 3-7).
3.3 Analogue Control Board

The analogue control board is the electronic link between the camera head and the input of the new data acquisition computer. It is located in the rack of the gamma camera gantry. It receives output signals from the imaging detector head and executes the following:

- Amplification
- Signal amplitude discrimination and pile-up event rejection
- Stretching of the 4 position-signals with stretchers
- Test signal generation

3.3.1 Amplification

The output $X^+$, $X^-$, $Y^+$ and $Y^-$-signals from the detector head are feed into four OP-amplifiers via input switches. These are used to select between the true signals from the detector head or test signals. The signals from the four OP-amplifiers are amplified 250 times before letting them go into a resistor network and then divide by 10. This technique is used in order to reduce the clamp effect, which is caused by the AC-coupled amplifier stage. Finally the signals are buffered by four OP-amplifiers [20].
3.3.2 Discrimination of Signals and Pile-up Events

3.3.2.1 Pile-up

Because of the random nature of nuclear events, there is a finite possibility that more than one event occur in a rapid succession [15, 20, 21]. Since the output pulses of a processing system are a few µs, there is a possibility that two or more of such pulses will be superimposed (pile-up). This will result in a false (an elevated) amplitude. The output pulse should therefore return to zero volts as soon as possible to minimise pile-up. If for example the output pulse has slow decaying tails (as in Figure 3-9), the effect on the pulse height spectrum will be an increased FWHM (Full Width Half Maximum) by distorted high-energy side of the peak. In a similar way output pulses having long undershoots will cause distortion of the low energy side of the energy peak.

Figure 3-9 (a.) A good exponential tail pulse from a preamplifier (b.) Pulse pile-up. A second pulse rides on the tail.

The pulse rate affects the fraction of piled-up pulses. At low pulse rates, the mean spacing between pulses is large and no pulse pile-up will occur.

At high pulse rates many pulses will overlap. Therefore it is preferable to choose a pulse shape that has a rapid return to the zero volts even if this not generally is the optimum for the noise and the ballistic deficit contributions to avoid the distorting pile-up effects. The time constant of the pulse shaping is a trade off between the noise contribution and the resolution.
3.3.2.2 Discrimination and Stretching

The four position signals from the camera-head are fed after amplification through 4 newly built stretchers. The position signals will be stretched (kept constant for a while) during the time the sum-signal ($\Sigma$) is analysed in a discriminator on the modified Analogue Control Board.

The discriminator contains three discrimination-levels: Sub-Lower Level (SLL), Lower Level (LL) and Upper Level (UL) [20]. If the $\Sigma$-signal fit the condition set by the discriminator (i.e. has a level between UL and LL), a circuit will open a gate in the data acquisition system and the four analogue position signals will be converted to digital signals in Analogue to Digital Converters (ADC’s).

The Sub Lower Level is always below the Lower Level and set just over the noise level (in this case about 80 mV), and indicates that an event take place.

The Lower Level is set by the LL potentiometer on the control panel. This discrimination level will reject Compton scattered events to be registered and should be set a few keV lower than the lowest energy of interest. In case of $^{99m}$Tc and $^{57}$Co, the energies are 140 keV and 122 keV, respectively.

The Upper Level is set by the UL potentiometer also on the control panel. This discrimination level is for avoiding pile-up events and LED pulses and should be set a few keV higher than the highest energy of interest.

3.3.2.3 Test signal generation

To test the data acquisition system a signal generator is used. The test signal generator creates a pulse of constant amplitude and shape when enabled by a control signal [20].

The generator output is feed into a resistor circuit, which divides the pulse into four signals corresponding to the $X^+$, $X^-$, $Y^+$ and $Y^-$-position.

The signals are then buffered by four OP-amplifiers and the resulting test signals ($TX^+$, $TX^-$, $TY^+$ and $TY^-$) are feed into the input switches if the control signal is in test mode. Since the amplitude of the test signal are constant it will appear as a point on the image display. By changing the resistor values it is possible to orientate the test point to a desired place on the image display. The preinstalled value with 1K on every resistor gives equal amplitudes for the four position signals, meaning that the test point is situated in the middle of the image.
The test signal frequency is set by means of three countrate select lines CR Ø, CR 1 and CR 2, which can be varied between 4.88 kHz and 625 kHz in powers of two (see Table 1).

**Table 1 The eligible frequencies**

<table>
<thead>
<tr>
<th>CR 2 (*16)</th>
<th>CR1 (*4)</th>
<th>CRØ (*2)</th>
<th>Frequency (kHz)</th>
</tr>
</thead>
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<td>0</td>
<td>4.88</td>
</tr>
<tr>
<td>0</td>
<td>0</td>
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<td>9.76</td>
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<td>0</td>
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</tr>
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<td>1</td>
<td>0</td>
<td>312</td>
</tr>
<tr>
<td>1</td>
<td>1</td>
<td>1</td>
<td>625</td>
</tr>
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</table>
4 Measurements

Measurements have been performed with the rebuilt gamma camera using the improved Analogue Control Board connected to the new data acquisition system. The spatial resolution for different collimators of the GE Gamma Camera with varying source to collimator distances was measured. Also the homogeneity of the scintillation crystal, i.e. differences in spatial resolution for different x-y positions, has been investigated.

4.1 The measurement method

To obtain the spatial resolution for different camera parameters like distance and type of collimator, the spatial FWHM (Full Width Half Maximum) in the x-y plane was measured. The events were collected with help of the acquisition PC as 2-dimensional images and the data could be presented as x-, and y-projections (see Figure 4-1).

![Figure 4-1 The presentation of a projection of detected events](image)

To obtain the spatial resolution expressed in measurable values one had to convert the two channel numbers (x-y projections of the 2-dimensional images) to distances measured in centimeters.

By placing a record player with a rotating $^{57}$Co point source in front of the camera-head, an image of a circular orbit with a fixed diameter was created.

The source rotated with a diameter of 13 cm and at a distance of 5.5 cm from the parallel hole collimator.
From the x-y projections of the 2-dimensional image of the circle, the distance expressed in channel numbers between the two outermost peaks could be determined (see Figure 4-3).

The projections gave the diameter of the circle in channel numbers, 85 and 82, for the x- and y- projections respectively. The average value of 83.5 channels corresponding to 13 cm was adopted.

### 4.2 Results

The spatial resolution for the three multihole collimators were measured for ten source-to-collimator distances ranging from 2.3 cm to 111 cm. The specification of these three collimators are presented in Table 2.
<table>
<thead>
<tr>
<th>Collimator type/ Data</th>
<th>General purpose parallel hole collimator</th>
<th>High-sensitivity parallel collimator</th>
<th>Diverging collimator</th>
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</thead>
<tbody>
<tr>
<td>Diameter (cm)</td>
<td>30.6</td>
<td>30.6</td>
<td>30.6</td>
</tr>
<tr>
<td>Thickness (cm)</td>
<td>3.3</td>
<td>3.3</td>
<td>3.3</td>
</tr>
<tr>
<td># holes per cm⁻²</td>
<td>~25.7</td>
<td>~14.6</td>
<td>~26.1</td>
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</tbody>
</table>

Table 2 Specification over the multihole collimators

The spatial resolution of the GEMS pinhole collimator were also measured for different hole diameters for five different source to collimator distances ranging from 3.4 cm to 58.6 cm. Data for the pinhole collimator is collected in Table 3.

<table>
<thead>
<tr>
<th>Collimator type/ Data</th>
<th>Pinhole collimator</th>
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<tr>
<td>Diameter (cm)</td>
<td>30.6</td>
</tr>
<tr>
<td>Length (cm)</td>
<td>24.3</td>
</tr>
<tr>
<td>Hole diameter (mm)</td>
<td>3, 4, and 6</td>
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</table>

Table 3 Data for the pinhole collimator

To check the homogeneity of the camera head assembly, the spatial resolution at 37 different positions, corresponding to the positions of the PM tubes, were measured for each collimator. The source positions were selected according to recommendation of the Camera-head manufacturer.

The differences in spatial resolution are used in the diagrams of FWHM variation with source to collimator distance to show the uncertainty for the measured values. The error bars in Figure 4-6 – 4-8 and 4-10 are deduced from the possible variation in FWHM if the chosen measuring point is in front of an area including the 7 innermost photomultiplier tubes.
The results were as followed:

FWHM: 1.6 cm +/- <5%

FWHM: 1.6 cm +/- <10%

FWHM: 1.6 cm +/- >10%

Figure 4-4 The spatial resolution at 37 position for the general purpose parallel hole collimator at a source to collimator distance of about 20 cm
Figure 4-5 The spatial resolution at 37 positions for the high sensitive parallel hole collimator at a source to collimator of about 13 cm
Figure 4-6 The variation of FWHM when using the general purpose parallel hole collimator for different source to collimator distances.
Figure 4-7 7 The variation of FWHM when using the high sensitive parallel hole collimator for different source to collimator distances
Figure 4-8 The variation of FWHM when using the diverging collimator for different source to collimator distances
Figure 4-9 Comparison of the variation of FWHM for varying source to collimator distances between the multihole collimators.
Figure 4-10 The FWHM for the pinhole collimator with three different hole diameters for different source to collimator distances
4.3 Conclusions

The relationship between the spatial resolution and source-collimator distance for different collimators was as expected. The theoretical values from the formulas of the FWHM, which are given in the chapters 2.1.5.1 and 2.1.5.3 are not included due to the lack of information about the distance between the rear side of the collimator and the image plane inside the crystal. In addition the source that has been used in the measurement was not a point source. It had a diameter of 3-5 mm and therefore a quantitative comparison with the predictions of the above mentioned formulas was not meaningful. However one can conclude that the diverging collimator provides the best and the high-sensitivity parallel hole collimator the worst spatial resolutions among the multihole collimators. The pinhole collimator also gave an expected result. The FWHM was better for smaller hole diameters and for longer source-collimator distances. Of course the real spatial resolution is better for shorter source-collimator distances, but the value of FWHM should be corrected for the collimator magnification so it refers to distances in the object instead of distances in the camera crystal. The homogeneity control showed that some inhomogeneities in the camera-head assembly is present. It can depend on crystal artefacts, caused by earlier temperature changes or other external damages. If looking on the two plots over the spatial resolution for different x-y positions one can see that the differences in spatial resolution vary for several positions. This depends probably on small but visible damages of the collimators. Another possibility is a non-uniform response of the photomultiplier tubes. It is possible to compensate for inhomogenieties, using a flood field source (a uniform source that covers the hole scintillation crystal) and dedicated software.
References

20. General Electric Medical Systems. Service Manuals for the Starcam Mobile 300 A/M Gamma Camera, Physics Dept. KTH Royal Institute of Technology
Acknowledgements

This is a Master of Science work at the Department of Physics of the Royal Institute of Technology (KTH), Stockholm. It was performed at the Department of Physics of KTH and at the Department of Electronics of ATOMKI in Debrecen, Hungary.

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Finally, this Master of Science Work could not have been realised without the generous donation of the gamma camera and other valuable equipments by The Nuclear Medicine Department of Danderyd Hospital.
Appendix
Block Scheme

Gantry p. VIII

Camera-Head p. III-VI

Distribution Box (Front)

Distribution Box (Back)

X+ Y+

Detector-Cable p. VII

D G M

X- Y-

Analogue Control Board p. XII-XVIII

Power Supplies p. IX-XI

+ 5V Regulated

+/- 15V

+ 24V Raw

+ 24V Regulated

AC Transformer

240V

240V

240V AC Transformator
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SOCKET LEMO 46-406706 REAR VIEW

NOTE
CAMERA STAND 300, BRAKE CABLE PIN 5 CONNECTED TO PIN 5 IN CONNECTOR PLUG.
MOBILE CAMERA BRAKE CABLE PIN 1 CONNECTED TO PIN 5 IN CONNECTOR PLUG.