Implementation of radionuclide scanning in medical science

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***Abstract*— A radionuclide scan is a way of imaging bones, organs and other parts of the body by using a small dose of a radioactive chemical. There are different types of radionuclide chemicals. The one used depends on which organ or part of the body is to be scanned. Gamma rays are similar to X-rays and are detected by a device called a gamma camera. The gamma rays which are emitted from inside the body are detected by the gamma camera, are converted into an electrical signal, and sent to a computer. A radionuclide scan is used for bone scan, tumor scan, cancer scan, kidney scan, lung perfusion scan, 'myocardial perfusion scan and thyroid scan etc. To detect the gamma photons usually a large crystal of sodium iodide is used in gamma camera. Different radionuclides used with their specific applications are summarized.**

I. INTRODUCTION

A radionuclide scan is a way of imaging bones, organs and other parts of the body by using a small dose of a radioactive chemical. There are different types of radionuclide chemical. The one used depends on which organ or part of the body is to be scanned.

A radionuclide (sometimes called a radioisotope or isotope) is a chemical which emits a type of radioactivity called gamma rays. A tiny amount of radionuclide is put into the body, usually by an injection into a vein. Sometimes it is breathed in, or swallowed, depending on the test [1]. Different radionuclides tend to concentrate in different organs or tissues. So, the radionuclide used depends on which part of the body is to be scanned. For example, if radioactive iodine is injected into a vein it is quickly taken up into the tissues of the thyroid gland. So, it is used to scan the thyroid gland. Cells which are most active in the target tissue or organ will take up more of the radionuclide [2]. So, active parts of the tissue will emit more gamma rays than less active or inactive parts.

Gamma rays are similar to X-rays and are detected by a device called a gamma camera. The gamma rays which are emitted from inside the body are detected by the gamma camera, are converted into an electrical signal, and sent to a computer. The computer builds a picture by converting the differing intensities of radioactivity emitted into different colours or shades of grey [3].

The discovery of natural radioactivity by Henry Bequerel in 1896 was the start that led to the development of nuclear medicine. The year before Wilhelm Rontgen discovered X-rays and as early as 1897, it was concluded that X-rays could be used for therapeutic as well as diagnostic purposes [4,5]. 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 [6]. 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. The first electronic gamma camera with multiple photomultiplier tubes was reported in 1957 and presented by Anger 1958 [7]. The development of the gamma camera made it possible to show in real-time, the blood flow in a patient, watch the kidney's function, examine the liver as it generates 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). Gamma imaging carried out by injecting patient with a tracer that emits gamma rays [8].

Collimator is a device that make photons of light accurately parallel (collimate) and hence direct the primary photons to form an image. Rays cannot be focused so "principle of absorptive collimation" is used. An absorptive collimator projects an image of the source distribution onto the detector by allowing only those rays traveling along certain direction to reach the detector.

This paper throws an insight on various types of gamma cameras used in radionuclide scanning and the effect of different collimators on the scanned image.

II. GAMMA CAMERA

Gamma cameras are made of a crystal which produces a burst of light when gamma rays hit it. Light is picked up by detectors (photomultiplier tubes) located behind the crystal. Electrical output from detectors is fed to computer to produce image.

A. Rectilinear scanner

Rectilinear scanner was developed by Benedict cassen-1950s. A single moving radiation detector sampled the photon influence at a small region of the image plane at a time the detector was scanned mechanically in a raster-like pattern over the area of interest. The image was a pattern of dots imprinted on a sheet of paper by a mechanical printer that followed the scanning motion of the detector.

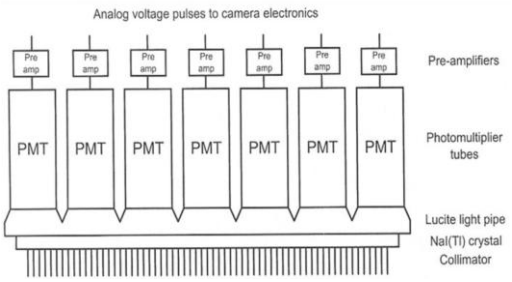

Figure.1 Anger scintillation camera stages

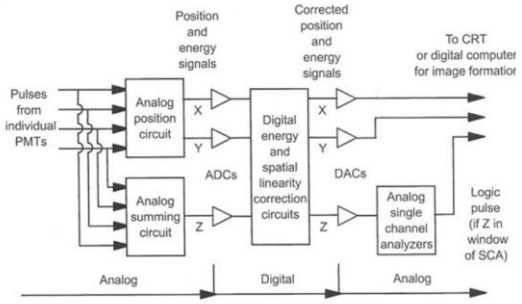

Figure.3 Hybrid camera stages

B. Anger scintillation camera

Anger scintillation camera developed by Hal Anger at Berkeley from 1952 to 1958 Anger Scintillation Camera (Fig. 1) replaced rectilinear scanners in the late 1960s [7]. It permits more rapid acquisition of images and enables dynamic studies. Being more flexible in its positioning, it permits images to be obtained from almost any angle.

Anger scintillation camera system components include collimator, detector, and crystal -NaI (Tl), light pipe photomultiplier tube (PMT), preamplifier, amplifier, pulse height analyzer (PHA) & image formation display.

They are specially designed for sick patients/breast imaging. having smaller detectors [9]. Detector size ~10×10 to 20×20square cm. Detector is NaI(Tl)/PMT or CsI(Tl) with silicon photodiode arrays and some systems employed cadmium zinc telluride (CZT) for direct detection of gamma rays.

C. Analog camera

Analog camera (Fig. 2) position circuit received pulses from individual preamps and by determining the centroid of these pulses, produced an X-position pulse and a Y-position pulse. Summing circuit added the pulses from individual preamps to produce an energy (Z) pulse proportional to total energy deposited in the crystal Z pulse sent to a single channel analyzer (SCA), produced a logic pulse only if Z pulse within a preset range of energies.

X and Y position pulses and the logic pulse from each interaction sent to a cathode ray tube (CRT) produced a momentary dot of light on its screen at a position determined by the X and Y pulses if a logic pulse from a SCA received simultaneously. Photographic camera aimed at CRT recorded the flashes of light, forming an image on film, dot by dot [6].

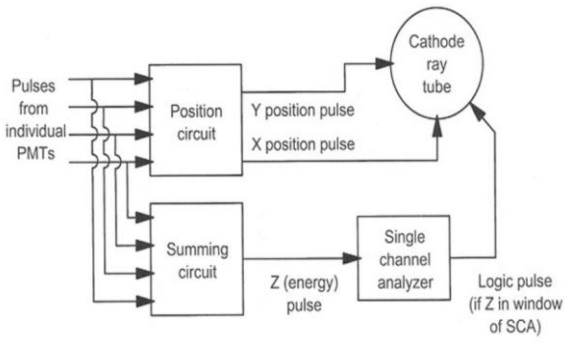

Figure.2 Analog camera stages

D. Hybrid camera

Position and summing circuits are analog X, Y, and Z pulses converted to digital signals by analog-to-digital converters (ADCs). Digital signals sent to digital correction circuits corrected digital X, Y, and Z signals converted back to analog voltage pulses [10].

Energy discrimination done in the analog domain by SCAs output to CRT as with analog camera or digitized again for display on a computer monitor (Fig. 3).

E. Digital camera

Digital camera pulses from individual PMTs are digitized position signals and energy signal formed using digital circuitry. Digital X, Y, and Z signals are corrected, using digital correction circuits, and energy discrimination applied in digital domain. Signals formed into digital images within a computer (Fig. 4).

F. Multiple crystal camera

Multiple crystal camera is developed by Bender and Bala. Small physically separate crystals were arranged in a matrix. Each Crystal was coupled to two PMTs. One PMT determine X position and the other determine Y position directly. At present, a single block of NaI(TI) divided into 400 detector element in a 20×20 matrix used. PMTs are coupled to block without any light pipe is a pixelated camera. In early state, CZT is used presently, CsI(Tl) and silicon photodiode is used. Efficiency is 80% has a size of 21×21cm and 4096 CsI 3×3mm [10]. It is good for small organs and cardiac imaging.

G. Solid State Camera

Solid state camera is a pixelated camera good for small organs and cardiac imaging [11]. In early state CZT is used presently CsI(TI) and silicon photodiode is used with efficiency of 80 % size of 21×21cm and 4096CsI 3×3mm.

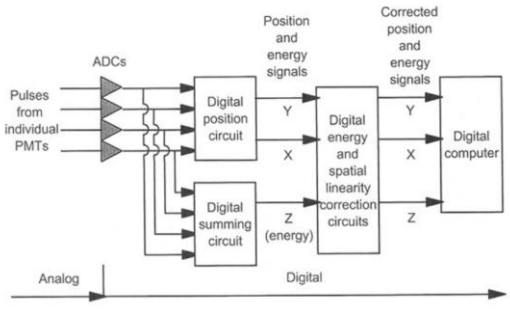

Figure.4 Digital camera stages

H. SPECT Gamma Camera

SPECT cameras have better contrast resolution and spatial localization and improved detection of abnormal function and quantification. But they have poor spatial resolution, lack of anatomical landmarks, longer scan times. The camera rotates 360 /180 degrees and collects 2D projection from different angles [12]. The acquisition 64×64 or 128×128 projections, time interval 15 to 20 sec per projection. Step increment 3 to 6 degrees. Photon attenuation and scattering affects the quality of images [13].

In hybrid SPECT/CT scanner, two approaches to clinical SPECT/CT are, low output slow acquisition CT scanner & Integrating commercially available CT scanner with gamma cameras slow CT rotation provides accurate registration with SPECT and allows for precise localization (Fig. 5).

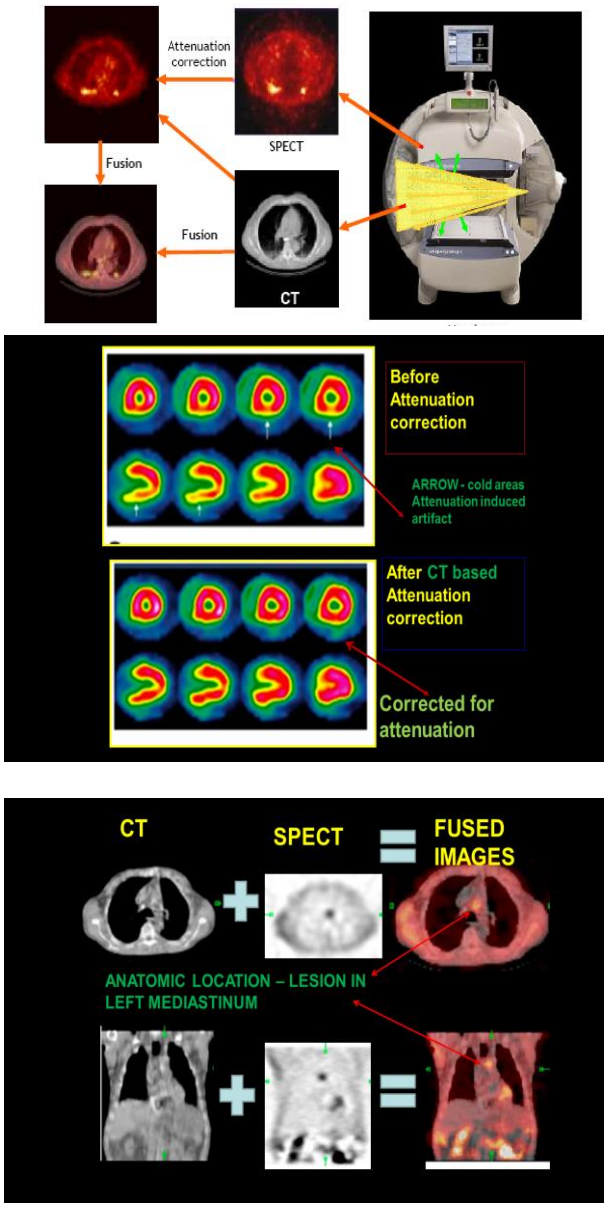

Figure.5 Implementation of SPECT camera

III. COLLIMATOR

It consists of hole/s (circular, square & hexagonal) in the absorptive medium (Lead) with walls/septa. Main interaction in collimating photons is photoelectric attenuation that depends on atomic mass (Z) of material used [11]. The need is to have highest density and largest Z as possible to keep septal thickness to a minimum. Lead (Z=82, d=11.3 g/cm^3) is cheap, machinable but fragile at 140 KeV, half-value layer (HVL) is 0.23mm. Tungsten (Z=74, d=19.3 g/cm^3) has useful properties but is harder to work with and is more expensive. At 140 KeV, HVL is 0.19mm. Gold & Uranium can also be employed but are not common (Au: Z=79, d=19.3 g/cm^3, HVL=0.14; U: Z=92, d=18.7 g/cm^3, HVL=0.14):

The efficiency of collimator is described by the ratio of number of photons that pass through the collimator to the number emitted. Sensitivity is dependent on area seen by hole. Larger hole space and thinner septa provide higher sensitivity. Choice of collimator is a tradeoff between spatial resolution and Sensitivity. Energy rating of the available collimators is given in Table 1.

A. Pinhole collimator

Pinhole collimator consists of small pinhole aperture in a piece of lead, tungsten or platinum [6]. The aperture is located at the end of a cone, typically 20-25cm from detector. Size of pinhole can be varied by using removable inserts. It is primarily used for magnification imaging of small objects inverted image (Fig. 6-a).

B. Parallel hole collimator

This is the most widely used collimator. It has an array of parallel holes perpendicular to crystal face. Resolution is worse with distance but sensitivity is independent with distance between source and detector [14]. Field of view increases with distance (Fig. 6-b).

Slant hole collimator (Fig. 6-c) is a parallel hole collimator in which all of holes are parallel to each other but angled by 25-30 degrees approximately. It is used for cardiac imaging.

C. Converging collimator

Converging collimator have an array of tapered holes that converge at a point at some distance in front of collimator (focal point). It produces magnified image so it is good for imaging smaller objects [15]. Resolution is high at surface whereas sensitivity slowly increases as source is moved from collimator face to focal plane and then decreases (Fig. 6-d).

D. Diverging collimator

Diverging collimator (Fig. 6-e) is upside down converging collimator having an array of tapered holes that diverge from hypothetical point behind crystal. It also gives magnified image and is used for imaging large objects on smaller field [15].

TABLE I.
ENERGY RATING OF AVAILABLE COLLIMATORS

Collimator Type	Max. Energy Rating	Radionuclides
Low Energy	140-200 KeV	99mTc,201TI,124I
Medium Energy	300 KeV	67Ga, 111 In
High Energy	360-560 KeV	131 I
Ultra High Energy	511 KeV	Positron Emitters

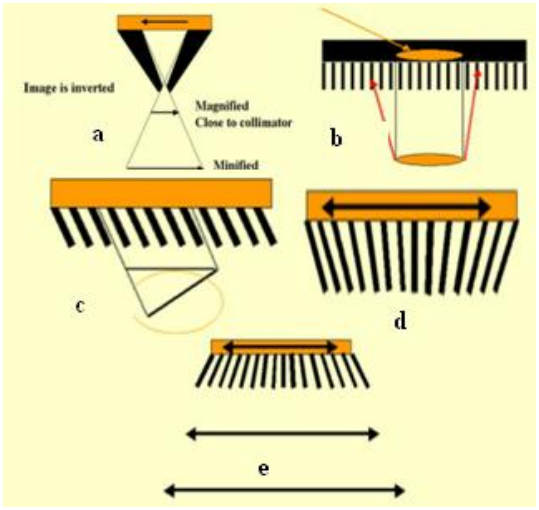

Figure.6 a: Pinhole collimator; b: Parallel hole collimator; c: Slant hole collimator; d: Converging collimator; e: Diverging collimator

E. Specialized collimator

Fan beam collimator is a combination of parallel hole collimator (along one axis) and a converging collimator (along other axis). It is used for tomography of small organs [16]. Single axis-diverging collimator has diverging holes in transverse direction but has parallel holes in axial direction used when transverse field of view of camera is smaller than patient's width.

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