BASIC PRINCIPLES OF SCINTILLATION DETECTORS AND GAMMA CAMERA

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OSNOVNI PRINCIPI SCINTILACIONIH DETEKTORA I GAMA KAMERE

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ABSTRACT

General principles of construction and the use of the scintillation detectors and gamma camera are presented in this article. Various types of the scintillators and their usage in these devices are also described. The given information is oriented to those medical personnel and medical students, who are neither familiar with physics nor with physical terminology. We hope that our explanations will make them easier to understand aforementioned principles.

Key words: *Nuclear Medicine, Imaging, Scintillation detector, Gamma camera*

SAŽETAK

U ovom radu su prikazani generalni principi konstrukcije i korišćenja scintilacionih detektora i gama kamera. Takođe su prikazani različiti tipovi scintilatora i njihovog korišćenja u takvim uređajima. Ove informacije su orijentisane ka medicinskom personalu i studentima, koji nisu dovoljno familijarni sa fizikom i sa fizičkom terminologijom. Nadamo se da će naša objašnjenja njima olakšati razumevanje ovih principa.

Ključne reči: *Nuclerna medicina, Snimanje, Scintilacioni detektor, Gama kamera*

INTRODUCTION

The discovery of artificial radioactivity in 1934 by the Curies, the development of the scintillation detector in 1948 by Hofstadter, and the construction of a gamma scintillation camera in 1957 by Anger, were key events in the last century in regards to nuclear medicine instrumentation. We intend to describe the principles of construction and operation of scintillation detectors and gamma cameras. This paper is primarily written to help medical personnel and medical students, who are neither familiar with the physics nor with the physical terminology to be proficient in detector fundamentals.

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SCINTILLATION DETECTORS

Scintillators

Scintillating materials, which can be organic or inorganic, have the property that when excited by ionizing radiation, a gamma photon for instance, will reemit the absorbed energy. The phenomenon is presented as a visible light photons emission following the absorption of gamma photons in the crystal lattice of the scintillator. The mechanism of scintillation is based on the fact that some impurities or imperfections in the crystal lattice can create energy states in the forbidden band, known as activator sites.

What is a scintillation detector?

The scintillation detector constitutes one of the most useful tools available for the detection of a wide range of radiation. The principle of scintillation detector is the interaction of the incident radiation with a scintillating material that releases the energy deposited in the form of light photons. These light photons are subsequently detected by the photomultiplier tubes (PMT), which convert the light photons to electrons. These electrons are multiplied within the PMT to produce a measurable current pulse that is proportional to the energy of the incident radiation.

In nuclear medicine, most detectors are based on inorganic scintillators. These types of the scintillators could be divided into three different groups. The first class of scintillators are those activated by doped impurities; activator sites are produced by adding impurities to the crystal lattice, such as thallium activated sodium iodide (NaI(Tl)), thallium activated cesium iodide (CsI(Tl)), and cerium activated gadolinium orthosilicate (Gd_2SiO_5(Ce)). The second type of detectors are made of self-activated scintillating materials, in which case the activator sites are produced by a stochiometric excess of one of the constituents of the solid. such as cadmium sulphide (CdS) with excess Cd and bismuth germanate (BGO). Finally, the third group of scintillators are those scintillators comprised of pure crystals, in which case activator sites are produced by imperfections in the crystal lattice, such as in diamond.

Properties required for good scintillator

- The detection efficiency should be high for the incident radiation; this corresponds to the material having a high atomic number and high density.
- The crystal itself must be transparent to the scintillation light.
- The conversion of radiation into light should be linear over a wide range of energies. The light yield should be proportional to the deposited energy.
- The crystal should have high stopping power, which depends on the effective Z (or high linear attenuation coefficient).
- The crystal should convert the energy of the radiation absorbed into detectable light with high scintillation efficiency, i.e., high light output (LO). It is the number of visible photons produced in the scintillator under gamma radiation. This is usually is expressed in terms of photons/MeV.
- The decay time of the produced scintillation light should be short.
- The index of refraction of a scintillator should be similar to that of glass (1.5), and the wavelength of scintillation light should be similar to the maximum PMT sensitivity.
- The scintillator should be both available and inexpensive.

These properties influence energy resolution, which defines the ability of the detector to determine the incident energy of the radiation. High resolution is necessary in order to distinguish gamma-sources of slightly different energies.

Thallium doped sodium iodide NaI(Tl) is the most widely used scintillation material. Its light output is greater than that of other scintillators, and it has a convenient emission range (in coincidence with maximum efficiency region of photomultiplier with bialkali photocatodes). The main disadvantage to Nal(TI) is hygroscopy.

Cesium iodide doped with sodium CsI(Na) is currently a widely used material. High light output (85% of that of NaI(TI)), emission in the blue spectral region (coincident with the maximum sensitivity range of the most popular PMT with bialkali photocatodes), and substantially lower hygroscopicity in comparison with that of NaI(TI), makes this material a good alternative for NaI(TI) in many standard applications.

Complex oxide crystals, such as gadolinium silicate doped with cerium (Gd_2SiO_5(Ce) or GSO), BGO, CWO, PWO, and NBWO, have a number of advantages over alkali halide crystals: high effective atomic number, high density, good energy resolution, low afterglow, and non-hygroscopy. Due to these features, detectors with oxide crystals are fail-safe, and there is no need for hermetisation.

Zinc selenide ZnSe(Te) scintillation material was created especially for matching with the photodiode, with an emission maximum at 640 nm. However, ZnSe(Te) crystals possess poor transparency.

Organic scintillators can be either liquid or solid and are characterised by their high response speed and relatively low light output.

Construction and Function of Scintillation Detectors

The basic scintillation detector consists of

- Scintillator (usually crystal NaI(Tl))
- Light guide, which is usually a thin layer of some transparent material with appropriate index of refraction, similar to the of refraction index of the crystal and photocathode
- Photo-detector (usually photomultiplier tube-PMT)

The Photomultiplier tube (PMT) consists of

- Vacuum glass envelope with a transparent window to couple the scintillator with a light guide
- Photocathode to absorb scintillation photons and to emit photoelectrons

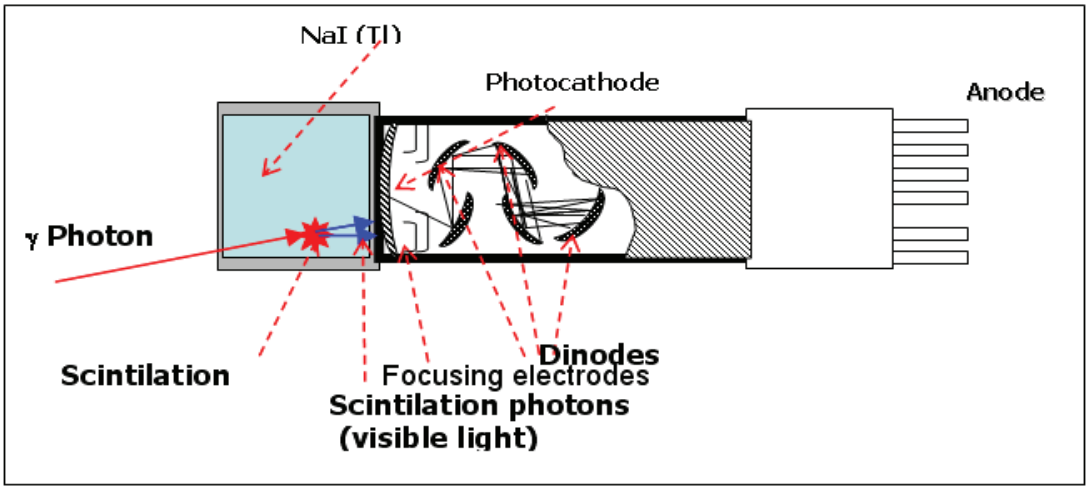

Figure 1. Schematic of a scintillator detector.

- Focusing electrodes to guide photoelectrons to the first dynode
- Dynode chain - there is a voltage drop between each dynode (100-300 V)
- Anode - the output stage of the PMT

* Amplitude of PMT anode output pulse is proportional to the energy E_γ of the incident photon. This is a very important characteristic of scintillation detector.

How does it work?

A fraction of the light photons produced in the scintillator will hit the PMT glass window. The light that passes through the glass envelope will hit the photocathode. Commonly, a photocathode consists of a mixture of potassium, cesium, sodium, and antimony. The photocathode must be thick enough to absorb the light photons, yet thin enough to prevent absorption of the photoelectron. Its purpose is to absorb the photon and emit an electron. Focusing electrodes guide the electrons, or the photoelectrons, to the first dynode. When the photoelectron strikes the dynode after being accelerated through 100-300 V, it will cause an emission of 2-5 secondary electrons, which in turn will be attracted to the next dynode and produce a multiplication of the 2-5/dynode stage.

The multiplication factor for a dynode is given by

$$\delta = \frac{N_s}{N_p}$$

where N_s is the number of secondary photoelectrons and N_p is the number of incident primary electrons. Typical values of δ are approximately between 4 and 5, and the overall gain in the PMT is given by

$$\text{gain} = \alpha\delta^N$$

In this relationship, α is the collection efficiency (~1) of the photoelectrons and N is the number of dynodes.

The gain of a 10-stage PMT tube is ~ 5^{10} or 10^7.

GAMMA SCINTILLATION CAMERA

While the shape and design of the detector and electronic processing have been altered since its inception, the basic components and principles of the gamma camera have changed little over that time.

Hal Anger 1958 developed a visualisation device with 10-cm diameter circular NaI(Tl) crystal, pinhole collimator, and 7 PMTs. The author named this device "scintillation camera", although it is now usually referred as an "Anger camera".

What is Nuclear Imaging?

Nuclear imaging involves injecting a small amount of a chemical substance, tagged with a short-lived radioactive tracer, into a patient. Depending on the chemical substance used, the radiopharmaceutical concentrates in the part of the body being investigated and emits gamma rays. A gamma camera then detects the source of the radiation to build a picture. These are called scans.

Nuclear medical imaging may be divided into three categories:

- Conventional or planar medical imaging
- Single photon emission computed tomography (SPECT)
- Positron emission tomography (PET)

Conventional and SPECT imaging use gamma cameras of the Anger type, which are more or less modified. Cameras of this general type have a single crystal viewed by arrays of detectors (Figure 2).

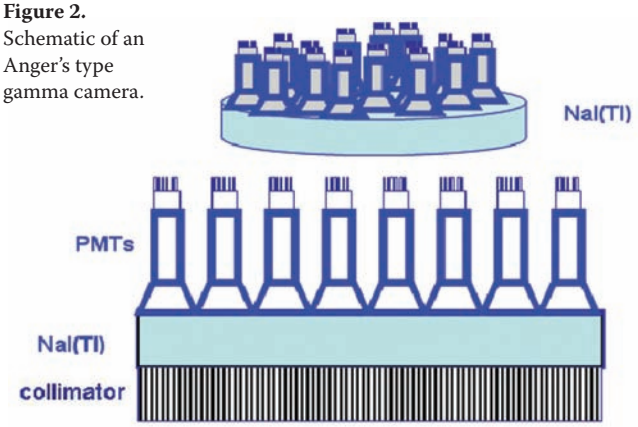

Figure 2. Schematic of an Anger's type gamma camera.

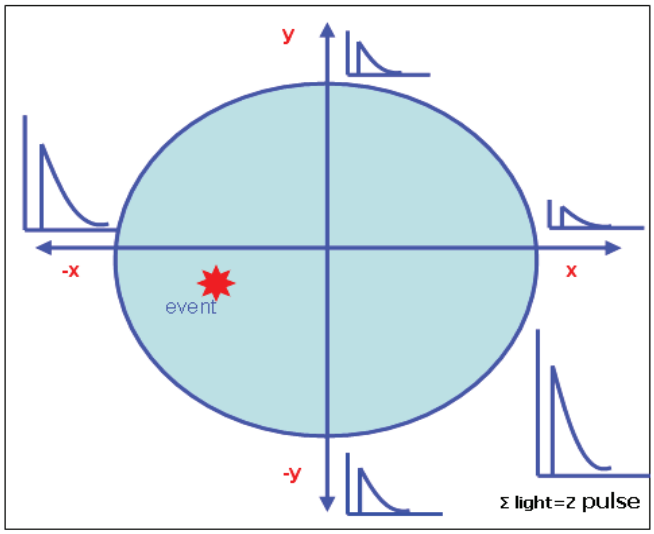

Figure 3. Schematic of detection and position of the event in the crystal of Anger's gamma camera.

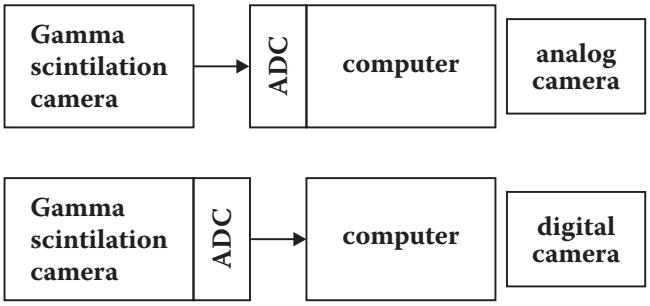

FIGURE 4. Analog and digital type GSC.

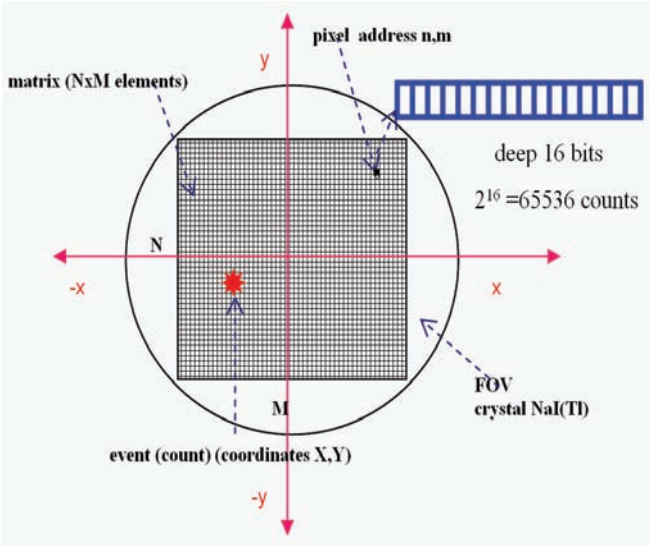

Figure 5. Computer memory and principle of data storage.

How do Gamma Scintillation Cameras work?

The emitted photons are first discretised by direction using a collimator. The collimator ensures that each small part of the crystal views only a small area of the organ to be imaged. Each interactions of a gamma ray with the detector crystal is called an 'event'. Scintillation imaging produces many single events in the crystal (Figure 3). The scintillation, which is produced in the crystal, originates very close to the point of interaction (event).

Each photomultiplier tube in the array detects the scintillation light of each single event. The PMT converts the detected light into an electronic pulse. The amplitude pulse is proportional to the intensity of the detected light; i.e., it is related to the proximity of the PMT to the point of interaction (Figure 5). Each event is located using information collected from each PMT. From all detected outputs, a position-specific logic circuit (Anger logic) estimates the position of each event. Because of this processing, the vertical and axial positions of PMTs are weighted by their electric signal responses from all PMTs. That circuit then gets X and Y coordinates of each single event. A single image is comprised of many events, represented as dots on the analog image.

Gamma Camera Computer System

For optimal performance, the gamma scintillation camera is often coupled with a dedicated computer. Analog-to-digital conversion of the signal from gamma camera is provided by an analog-digital converter (ADC) module. This module is part of dedicated computer in older models of analog gamma cameras, or part of gamma camera itself in newer digital models (Figure 4).

Signals for the definition of horizontal and vertical positions (X and Y coordinates, respectively) of the events will then be converted to digital form. The ADC conversion is usually performed with ADC converters for each channel (separately for the X and Y coordinate). As trigger for AD conversion process of each single event uses corresponding Z pulse. The ADC module and dedicated software developed in our institution (9) could be an example how it can be realised.

The digital image in the computer's video memory matrix is presented as a rectangular object in (or around) the field of view of detector crystal (FOV), which consists of NxM elements (Figure 5). These elements are usually called pixels. In most cases, N is equal to M and the shape of the matrix is a rectangular quadrilateral. Each pixel of the matrix has its own unique address (n,m), which corresponds to the X and Y coordinates of the event in the crystal. According to the given digitalised X and Y coordinates of the actual event, an appropriate pixel is increased by 1 for each registered event. In this case, an event is commonly referred to as a count. Many single events will be represented in video memory as a digital image of the spa-

tial distribution of the radiopharmaceutical. The number of events that can be stored in each pixel depends on the third dimension (depth) of the used matrix. A matrix of n bits deep can store 2^n events. It is common practice in nuclear medical imaging to use matrices 8 to 64 bits deep and with up to 512x512 pixels. Figure 6 presents one common nuclear medicine image, a scintigram of the thyroid.

Computer systems coupled with gamma camera enable data processing, safe long-term data storage, provide dynamic, gated and ECT studies. If an image in computer memory is digital, we could further process using, for example, digital filtration, edge detection and enhancement, shape recognition, etc.

Newer trends in construction

Over the past decade, great progress has been made in combining anatomical imaging, such as CT or MRI, with modes of functional imaging, such as SPECT and PET. The first advancement was the development of software for the fusion of images generated from those different "stand alone" machines, followed by the development of hybrid devices such as SPECT/CT and PET/CT. There are currently many models of hybrid imaging devices, with SPECT or PET as functional imaging devices that are often coupled to a 64-slice CT for use as anatomical imaging devices. More powerful hybrid imaging machines such as PET/MRI are being developed as the modality of choice in clinical imaging, and such machines will likely be the ultimate future imaging modality.

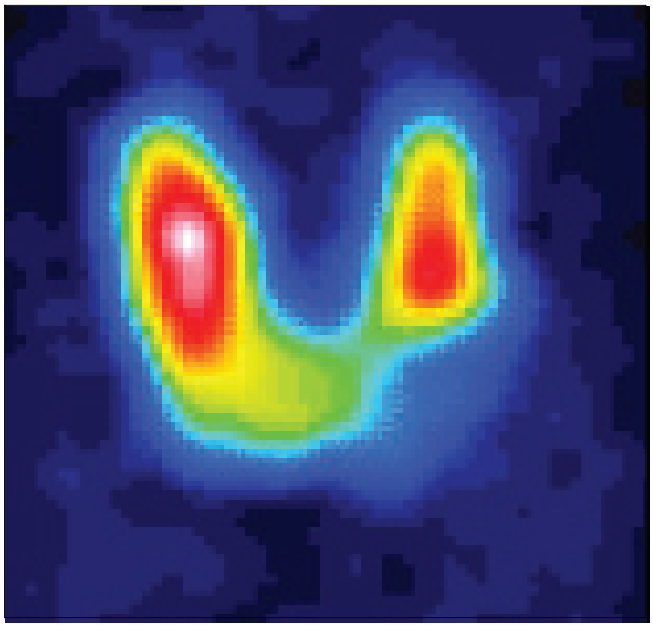

Figure 6. Image of the thyroid ("cold node" in left lobe).

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