Continuing Education

The Developing Technology of Imaging Detectors

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This is the first in a continuing education series of four articles related to scintillation camera technology. After reading this article, the reader should be able to: 1) understand the basic principles of present detector technology; and 2) discuss the advantages and disadvantages of gas, semiconductor, and inorganic crystalline detectors.

Detectors of ionizing radiation are constantly evolving. The technology of imaging detectors evolves along a parallel course but the demands of imaging impose severe constraints on the detectors. The physical principles of imaging detectors used in single photon imaging in nuclear medicine are reviewed in this article, and the factors that affect imaging system performance are identified. The key performance parameters are intrinsic efficiency, energy resolution, dead time, and spatial resolution. These parameters are compared for a variety of potential new detectors and for the Anger scintillation camera. Detectors included in the comparison are the multiwire proportional chamber, the gas scintillation proportional chamber, semiconductor detectors and crystal scintillation detectors. The impact of the requirements of various types of planar imaging and single photon emission tomographic imaging are discussed. Specific applications that have shown promise are discussed and compared. This paper will summarize the prospect for various detectors to make a future contribution to nuclear medicine imaging.

REVIEW OF PRINCIPLES

All detectors used for imaging in nuclear medicine accomplish their goal in four major steps. Those steps are the following:

1. Interaction with the detector medium.
2. Production of electrons.
3. Amplification.
4. Production of energy and position signals.

We will discuss in some detail these steps for each of the detectors to be considered.

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There are several measures of performance which are used to describe the qualities of gamma ray imaging detectors. These are:

a. intrinsic efficiency
b. energy resolution
c. dead time
d. position resolution.

Cost also has to be considered as it contributes to the decisions on applying the detector along with the specific application of the detector (e.g., is it for planar or SPECT imaging? Is a detector required for first-pass cardiac studies or for equilibrium studies? Is the organ to be imaged the brain or the liver?) The answers to these questions will often influence the decisions on what detector and, hence, imaging system is preferable.

Measures of Performance

Intrinsic efficiency. The efficiency of a detector is the probability of a gamma ray interaction—the first step of the detection process. We can define intrinsic efficiency of a detector by the following equation:

$$\text{intrinsic efficiency, } \text{Eff}_{\text{int}} = \frac{\text{\# of photons detected}}{\text{\# of photons emitted}}.$$

In nuclear medicine, detection will be either by a photoelectric event or a Compton event. If by a Compton event, there is the possibility of the scattered gamma ray also being absorbed in the detector. This is usually undesired, but we will discuss one detector in which this effect is exploited.

The detection process for the photoelectric effect increases as the fourth power of the atomic number of the detector and decreases as the third power of the gamma ray energy. This can be expressed in the following equation:

$$\mu \sim \frac{Z^4}{E^3},$$

where μ is the linear attenuation coefficient for the photoelectric effect. The photoelectric interaction results in a photoelectron that travels some distance in the detector. The length of the electron path depends on the density of the detector material.

Intrinsic efficiency is easily calculated from the familiar exponential relationship of attenuation of radiation,

$$Eff_{int} = 1 - e^{-\mu x},$$

where x is the thickness of the detector and μ is the linear attenuation coefficient composed of the sum of the linear attenuation coefficient for the photoelectric effect, and the linear attenuation coefficient for the Compton effect.

It is best if a detector has a high atomic number to enhance the photoelectric effect and a high density so that detector thickness is kept to a minimum. A thick detector, while it stops the gamma rays, may result in parallax errors or in unwanted detection of photons from Compton events in the detector.

Energy Resolution. Energy resolution is the ability of the detector to separate two nearly equal energy gamma rays. It is dependent on the gamma ray energy and the nature of detection or energy conversion process. Subsequent amplification either in the detector or in the preamplifier may also affect the final energy resolution.

The conventional measure of the energy resolution is the full width at the half maximum points of the photopeak expressed either in keV (FWHM) or as a percent of the photopeak energy (%FWHM).

The primary determinant of the width of the photopeak is the number of charge pairs produced by the detected event. In some detectors, the number of electrons produced follow a Poisson distribution (i.e., the variance is the number of electrons). Expressed symbolically it is:

$$\text{variance} \sim N_e,$$

where N_e is the number of charge pairs produced by the detector. In a scintillation detector the charge pairs are the electrons produced at the first dynode of the photomultiplier tube (PMT). In other detectors the variance is much smaller than that predicted by the Poisson distribution and is described by

$$\text{variance} \sim F \times N_e,$$

where F is called the Fano factor and always ≤ 1. This is one instance where we can say that mother nature helps us since the smaller the Fano factor the better the energy resolution will be. Semiconductor detectors and gas detectors have Fano factors $\sim$ 0.2 or less whereas solid scintillation detectors such as NaI(Tl) have Fano factors of 1.0.

Dead Time. The dead time is a measure of count rate capability and is measured in microseconds for nuclear medicine imaging detectors. Dead time is determined by the length of time the detector takes to process an event that has been absorbed and by the degree of parallelness (or ability to handle simultaneous pulses). Detectors that can have parallel channels include mosaic detectors, ring detectors, and multidetector imaging devices, all of which are composed of multiple detector units.

The dead time performance of detectors can be characterized by a pulse of charge that reaches a maximum and then falls to zero. The pulse can be characterized by the rise time and the fall time determined at the 10% and 90% points. The time constant is also used to describe the fall time. Fall time is usually $\sim$ 3–4 times the time constant and is dependent on the pulse shape. The pulse width is the time between the 10% points. It is often necessary to discuss "afterglow" or long-tailed pulses. These are undesirable and lead to undesirable pile up effects. Ideal detectors would be characterized by a box-like pulse with short rise and fall times.

Position Resolution. This is defined as the ability to separate two closely spaced radioactive sources. It is usually described as being extrinsic or intrinsic depending upon whether the collimator is present or not. The most meaningful is extrinsic spatial resolution since it is relevant for clinical use. However, we will discuss intrinsic resolution since it is not application dependent and is well characterized for different detectors. The collimator eventually used with any detector should be optimized for that detector and for the intended application.

CURRENT DETECTOR TECHNOLOGY

The current imaging detector of choice in nuclear medicine for most applications is the thallium activated sodium iodide (NaI[Tl]) single crystal Anger camera. It is available from a large number of manufacturers, is not very costly and does most imaging jobs well. It is the standard against which all other detector technologies must be compared.

Sodium Iodide Detectors

Sodium iodide has been with nuclear medicine almost since the start of the field. The first detectors used in the field we now call nuclear medicine were G-M counters in the early 1940s. Imaging was performed by scanning with a hand-held G-M tube (*1*). After World War II, there was a brief flurry of activity with calcium tungstate crystals. Sodium iodide crystals were first used in 1948 (*2*) in a physics setting and by the early 1950s had become the standard detector in mechanical scanning. Anger first reported on the camera that bears his name in 1958 (*3*), and a camera using a matrix of NaI(Tl) crystals (Autofluroscope) made its appearance in 1960 (*4*).

The factors that made NaI(Tl) so enduring are its high atomic number (Z=53), modestly high density (3.67 g/cm^3), and modest energy resolution. Sodium iodide detectors are reasonably rugged and reliable, have reasonable cost, and come in practically any size.

The Anger camera is so well known that only a brief discussion of its performance properties will be presented. The key points in its operation are that a gamma ray enters the crystal and undergoes either a photoelectric interaction or a Compton interaction. The photoelectron or Compton electron that is produced is stopped in $\sim$ 1 mm. Scintillation photons will be produced in direct proportion to the energy of the electron. There will be an iodine x-ray of $\sim$ 30 keV produced if there is a photoelectric interaction. This will be absorbed relatively close to the original interaction site, and the light produced will add to that produced during the absorption of the photoelectron and will contribute to the full energy peak or photopeak. If a Compton event occurred, the scattered photon may

interact in the crystal. If the scattered photon does interact in the crystal, that compound event will be indistinguishable from a photopeak event and it will be counted and imaged. The position information, however, will be in error since the scintillations came from both the interaction point of the original gamma ray and from the interaction point of the scattered gamma ray. The effect of this error is a loss in spatial resolution and is relatively minor at 140 keV and increases as the energy and detector thickness increase.

Intrinsic Efficiency. The camera has an efficiency of ~0.85 for 140 keV at 9 mm thick. This can be increased with a thicker crystal at some loss in spatial resolution.

Energy Resolution. The energy resolution is ~ 10–11% on the best cameras and varies inversely with the square root of energy. The energy resolution is limited by the amount of light picked up by the PMT array. Approximately, 300 eV of gamma ray energy are required to produce an electron at the PMT. This means that a limited number of electrons are produced in the PMT, thus limiting the statistical accuracy of both energy and spatial resolution. The energy resolution is not likely to be improved upon to any significant degree in the Anger camera.

Dead Time. Dead time is normally on the order of 2–3 μsec with peak counting rates of 150–200 kcps. By sacrificing some of the light output, peak counting rates of 500 kcps can be obtained. Spatial resolution degrades by a factor of 2–3 when this is done.

Position Resolution. State-of-the-art position resolution is 3–4 mm at 140 keV. This is a function of the photon energy. It increases (i.e., gets worse) as the square root of energy below 100 keV since it is limited by the number of light photons produced and ultimately captured by the PMTs. In fact the major limit on spatial resolution at energies below 100 keV is the amount of light captured by the PMTs. Spatial resolution also degrades as photon energy increases due to the increase in the proportion of scatter events that will be produced and detected in the crystal. This effect is enhanced with thicker crystals.

OTHER DETECTOR TECHNOLOGIES

For purposes of discussion the detectors will be divided into three major categories: gas detectors, semiconductors detectors, and inorganic crystalline scintillation detectors. Gas detectors will be discussed first since they were historically the first used in nuclear medicine.

Gas Detectors

It takes ~ 30 eV to produce a negative electron and positive ion or charge pair in most gasses. In addition, the Fano factor is ~ 0.2 so the actual number of electrons produced for each gamma ray of a given energy has a small variance compared to sodium iodide detectors. This leads to potentially very good energy resolution. This energy resolution may or may not be realized depending upon the subsequent steps in the signal forming process.

Today gas detectors as imaging detectors have a limited role to play in nuclear medicine, but there are several detectors that look promising for special applications and are well worth examining more closely. These are the multiwire proportional chamber (MWPC) and the gas scintillation proportional chamber (GSPC).

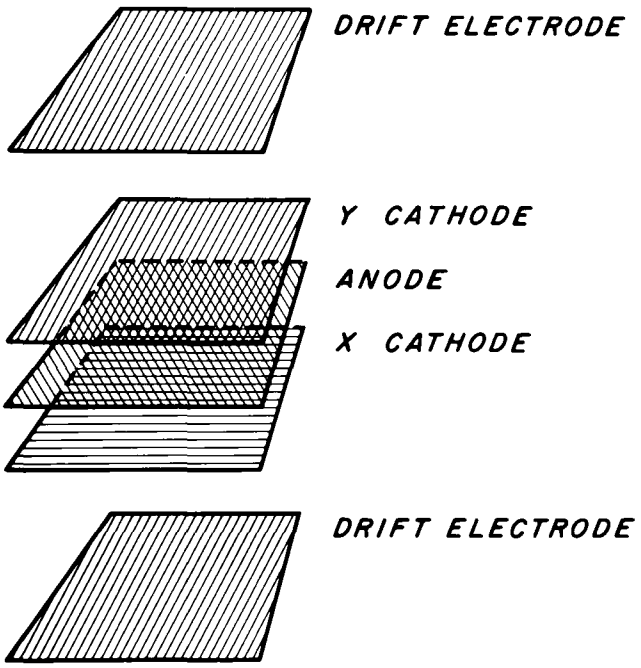

FIG. 1. Basic components of a multiwire proportional chamber. The cloud of electrons produced after the gamma ray interacts in the region between the drift planes moves under the influence of an applied electric field towards the anode wires. A high field around the anode wires causes the electrons to ionize the gas atoms and multiply the number of charge carriers in direct proportion to the original number of electrons. The two cathodes are at right angles to each other so X and Y coordinates can be determined.

Multiwire Proportional Chamber (MWPC)

Multiwire proportional chambers have been used in high energy physics since the mid 1960s and have been under investigation since the mid 1970s (*5–9*) for use as imaging detectors in nuclear medicine for both positron imaging and single photon imaging.

Figure 1 shows the main components of a MWPC used for imaging gamma rays. It consists of a drift region and interaction region of several centimeters thickness. The heart of the chamber is the much thinner cathode-anode region where amplification takes place and where the position signals are formed. The anode has a high positive voltage with reference to the cathode. The cathodes consist of wire planes. The wires are spaced ~ 2 mm apart. The upper and lower cathodes have the wires running in orthogonal directions. Gas pressures of from 1–10 atmospheres (atm) have been used.

The electrons produced in the initial absorption event are drifted under the influence of an applied electric field into the region of the anode wires. The anode has a high field especially close to the wires. The electrons are drawn into this field and are accelerated to high velocities by the field. When the electrons reach a certain critical velocity, they collide with

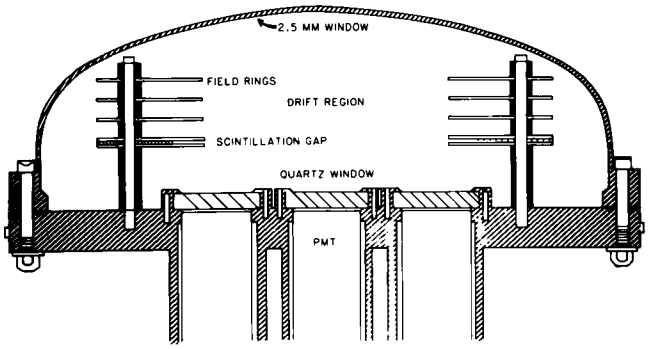

FIG. 2. Major components of a gas scintillation proportional chamber. Electrons interact in the drift region as in the MWPC and are drifted toward the scintillation gap. The electric field in the gap is just below the level that would cause further ionization, except that excitation of the gas takes place. As the gas relaxes, ultraviolet photons are emitted and pass through the windows and into the PMTs. The position and energy signals are formed just as in an Anger camera.

the gas atoms causing ionization of the gas and a proportional increase in the number of charge carriers. Multiplication factors of 10^5 to 10^6 are typical. Position readout is accomplished by taking the signals induced on the cathodes into either a delay line readout (*7, 8*) or having an amplifier on each wire (*5*).

Intrinsic Efficiency. The gas usually used is xenon mixed with a low percentage of an organic gas to aid in controlling the proportional amplification process. Xenon has a high atomic number of 54 but it has a low density of 5.89×10^{-3} g/cc at 1 atm of pressure which leads to a rather low intrinsic sensitivity unless quite high pressures are used. MWPCs operating at 4–10 atm of pressure have been designed for use with isotopes below 100 keV.

Energy Resolution. The energy resolution of gas detectors is relatively high since in xenon it takes 25 eV to produce a charge pair and the Fano factor is 0.19. However, this is degraded considerably during the proportional amplification process because of the difficulty in maintaining consistency in the electric field at the anode wires. Practically speaking, the best energy resolution seems to be approximately a factor of 2–3 better than that of sodium iodide detectors.

Dead Time. The time characteristics of MWPCs are good. Pulse width on the order of 1 μsec are quite feasible. This has been exploited in cameras made for imaging first-pass cardiac studies with short half-life isotopes where low dead time is a necessity (*9,10*). Peak counting rate of over one million counts per second have been reported (*9*).

Position Resolution. Position resolution is limited by the range of the photoelectron in the gas and more importantly by the range of the fluorescent x-ray emitted. This x-ray has an energy of ~ 30 keV and in low density gas has a range of several centimeters. This can lead to long tails in the line spread function limiting the low contrast performance of a camera.

This effect is minimal when the incoming photon energy is ~ 40 keV and gets worse with increasing energy. There are ways to minimize its effect by imaging not on the full energy peak but on the escape peak (*6,9*).

Gas Scintillation Proportional Chamber (GSPC)

The GSPC can be thought of as a gas version of the sodium iodide Anger camera (*10–13*). Figure 2 shows the major components. The gas is pure xenon. As in the MWPC, there is, a rather thick interaction region and drift space. A region composed of two transparent screen like electrodes is the heart of this device and is called the scintillation gap. An array of PMTs detects the light produced as in an Anger camera.

The initial gamma ray interaction occurs in the gas. The electron cloud produced by the photoelectron drifts toward the scintillation gap where the electrons undergo acceleration—enough acceleration so they excite the xenon gas, but not so much that ionization occurs. As the xenon relaxes back to the ground state, photons are emitted. These photons are in the ultraviolet part of the spectrum. The light gain or number of photons per electron entering the scintillation gap is several thousand. By careful selection of materials, the light photons can reach the PMTs and be converted to position signals and energy signals as in the Anger camera.

Intrinsic Efficiency. The intrinsic efficiency is the same as that for any gas detector filled with xenon gas. Prototypes have been built that operate at 4 atm pressure. Proposed designs are being constructed that operate at 40 atm to achieve a 70% intrinsic efficiency at 140 keV (Moore SC, *personal communication.*)

Energy Resolution. Energy resolution is excellent for this type of device since the amplification of the initial signal is dependent only on the electric field in the scintillation gap which can be well controlled. Degradation can occur when light is lost in transmission to the PMTs. Careful selection of materials can minimize this. Energy resolution of 5% at 80 keV has been achieved. Better than 3% energy resolution at 140 keV should be possible with optimum optical components (Lanza RC, *personal communication).*

Dead Time. The dead time is determined by the time it takes the electron cloud to traverse the scintillation gap. Minimum time is ~ 1 μsec, and the pulse is of the ideal box shape. This would permit excellent performance at high count rates, with peak rates of one million counts per second, while maintaining excellent energy resolution. This property of the chamber has not been exploited at this time.

Position Resolution. The position resolution is determined by the range of the photoelectron and the distance that the fluorescent x-ray travels, just as in the MWPC. In the Anger camera, a major limitation to the resolution is the number of light photons at the PMTs. In the GSPC, this is not the major limiting factor because there are several thousand more light photons available compared to an Anger camera. In a properly designed GSPC, a resolution of 3–5 mm at 4 atm has been obtained and resolution of 2 mm at 40 atm and 140 keV have been predicted (Lanza RC, *personal communication).*

Semiconductor Detectors

Semiconductor detectors have enticed more than a few peo-

ple with their superior energy resolution. Unfortunately, this energy resolution has been realized in only a few research imaging detectors (*14–16*). The cost is very high and the maximum size is ~5 cm square. Bigger detectors are made by putting the 5-cm modules together. More recent work has been done with intrinsic germanium planar detectors and with a cadmium telluride linear matrix detector. At this time, however, no other semiconductor detectors appear promising for future use.

All semiconductor detectors work essentially the same. Electron-hole pairs or charges are raised into a conduction band when a gamma ray is absorbed in the crystal. The charges are free to move under the influence of an applied electric field.

Intrinsic Germanium. Intrinsic germanium detectors are manufactured from highly purified germanium and have a maximum thickness of ~ 1 cm and a size of 7 cm diameter. The atomic number is a modestly high 32, which is acceptable for gamma ray imaging. The density is 5.33 g/cm^3 which is high in relation to other detector materials. The Fano factor, which is very difficult to measure, is ~ 0.05 so mother nature has been nice. These detectors must be cooled to liquid nitrogen temperatures when in use.

Several groups have worked on these detectors in the past, but currently the field seems to be at a standstill. The general approach to obtaining the energy and position signals has been similar for the various groups. The single crystal of germanium is prepared by depositing the contacts on both sides and then mechanical or chemical processing is performed to produce small groves on each side. The top groves would run in the X direction and the groves on the bottom side would run in the Y direction. This results in active regions ~ 2 mm square across the surface which are isolated from each other. X and Y information is obtained from the strips on either side using delay lines or current division techniques.

Intrinsic Efficiency. The intrinsic efficiency is relatively low at 140 keV because of the small thickness and the low atomic number.

Energy Resolution. Energy resolution is excellent although not near the maximum one can expect for nonimaging germanium detectors because of degradation due to the segmenting of the crystal to permit a row and column type of readout. Approximately 3% at 140 keV has been achieved.

Dead Time. Dead time is moderate on the order of 3–5 μsec and is determined by the need to wait for all the charge to be collected.

Position Resolution. Position resolution is mainly determined by the size of the segmentation. In the prototypes produced so far resolution was ~2 mm.

Cylindrical ingots of germanium have been explored for scanners (*17*). Excellent energy resolution and spatial resolution were achieved. Such an approach may yet have application in multidetector SPECT.

An interesting combination of a germanium detector and an Anger camera is being developed by Singh (*18*) and is described below.

Cadmium Telluride. With a density of 6.20 g/cm^3 and atomic numbers of 48 and 52 for cadmium and tellurium, respectively, CdTe appears to have some desirable characteristics for gamma ray detection. It operates at room temperature. The main problems arise because of the limited size and need for a matrix approach to achieve a working imaging detector. This has been done for a linear array which gets a two-dimensional image by rotating the detector array, collecting the data on a computer, and then reconstructing the image (*18*). The detector tends to be expensive and trouble prone but encouraging results have been obtained.

Inorganic Crystalline Scintillation Detectors

This category includes, of course, the familiar NaI(Tl) crystals that have been used for a long time and have been described previously. There are a number of other inorganic crystals that have found significant use in PET systems. These include bismuth germanate, $Bi_4Ge_3O_{12}$, barium fluoride, BaF_2, and cesium fluoride, CsF. All of these have less light output than NaI(Tl) and do not seriously challenge it for single photon imaging. They have excellent dead time characteristics. The major drawbacks are that they are expensive or impossible to obtain in large sizes. Low light output leads to inferior energy resolution when compared to sodium iodide. They seem to offer little use to a single photon imaging system.

As previously mentioned, the Anger camera is the standard by which all new detectors must be compared. It does all things well. It is not, however, perfect and there are a few areas (e.g., SPECT) where specialized detectors may be expected to improve on the Anger camera.

SPECT. The major limitations to SPECT today are spatial resolution, scatter radiation, and lack of full quantitative capability. The major route to improve spatial resolution is to optimize collimator design through special magnifying collimators. New detectors may be able to improve scatter rejection and thus help with quantitative ability.

Improvements in scatter rejection can best be realized by using detectors with improved energy resolution. Sodium iodide detectors will not get substantially better. We must look to either semiconductor detectors or gas detectors. Both can deliver better scatter rejection and excellent spatial resolution but at a reduced intrinsic efficiency. An overall system design indicates a factor of 3–4 in signal detection can be gained through a combined approach of collimator design and a gas scintillation proportional detectors.

Planar Imaging. If high count rate studies become desirable in nuclear medicine, gas detectors may have a contribution to make since they maintain energy resolution, spatial resolution, and count rate capability to higher limits than the NaI based cameras.

CURRENT RESEARCH AND DEVELOPMENT

Where are the research efforts being directed at the present time? Why is there any need for research effort on new detectors? What can be improved upon?

Imaging detectors that are presently in the research and developmental stages are the potential new detectors that may find their way into regular use in the nuclear medicine environment in the near future.

Crystalline Scintillation Detectors

There are not likely to be major changes in the technology of NaI detectors. There will be, however, changes in the electronic components of these detectors. We will see all-digital cameras, scatter rejection schemes, and other improvements in the handling of the PMT signals, but the basic detector will remain unchanged.

As examples of these types of changes, two new systems are discussed. A new detector system designed to perform SPECT has recently been introduced*. It represents one of the most highly developed Anger camera—analog camera systems. Three identical detectors are mounted on a common gantry. Previously published specifications for the detectors are intrinsic resolution of 4 mm, energy resolution of 10.2% and a detector efficiency of 90%.

A digital camera system† designed for brain SPECT is now under construction. This system uses a toroidal NaI crystal with digitization of the PMT signals and constitutes the first truly digital gamma camera. Estimated performance specifications are intrinsic resolution of 6 mm, energy resolution of 11%, and intrinsic efficiency of 90%.

Gas Detectors

A MWPC is under active development by a team from Baylor University and NASA in Houston. This effort has resulted in an attempt to bring MWPCs into the nuclear medicine community expressly designed to perform first-pass cardiac studies with 9.3 minute half life tantalum-178 (*9,10*). Performance parameters include dead time low enough to allow counting rates of 1,000,000 cps with a resolution of 9 mm.

A team from Worcester Polytechnic Institute, Massachusetts Institute of Technology and Harvard Medical School is designing and constructing a GSPC designed to peform SPECT imaging of the brain with ^{99m}Tc and ^{123}I. Intrinsic resolution is expected to be 2 mm with energy resolution of 2.2% and efficiency of 73% at 140 keV. The detector is designed to run with 40 atm of pure xenon as the scintillator.

Hybrid Detector

A detector under development at the University of Southern California does not fit neatly into any of the above catagories (*18*). The detector is actually a hybrid consisting of a germanium detector followed by a conventional Anger camera. No collimator is required. A gamma ray to be imaged is emitted from the source and undergoes Compton scattering in the germanium detector where the energy of the Compton electron can be measured precisely. The scattered photon must interact in the second detector, the Anger camera. By applying the known relationships between energy and angle for Compton scattering the three-dimensional source distribution can be reconstructed. The development of this novel system will be interesting to follow.

In summary, the Anger scintillation camera will remain the detector of choice in nuclear medicine in the foreseeable future. Furthermore, specialized Anger cameras for SPECT applications are appearing in clinical use and eventually will become a common feature in nuclear medicine departments. Gas detectors will make possible contributions in the early 1990s, and semiconductor detectors await further innovations before their role can be predicted.

NOTES

*Triad, Trionix Reserach Laboratory, Inc. Twinsburg, OH
†ASPECT, Digital Scintigraphics, Inc., Cambridge, MA

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