nuclear medicine

Detector Applications Information Note

Nuclear Medicine

Radiation detectors play a major role in non-invasive clinical investigations.

In *nuclear imaging*, imaging systems called gamma cameras determine the radiation distribution, transit time or uptake of a pharmaceutical labeled with a suitable, short-lived radioactive nuclide administered to the patient. The cameras' externally placed radiation detectors measure the radiation, and special imaging devices display the radioactivity distribution. The images yield information about the function of an organ or part of the body. SPECT and PET are two of the imaging techniques used.

In *X-ray diagnostics*, the transmitted intensity of an X-ray beam is measured. The standard technique of irradiating the patient with an X-ray beam and producing an image on photographic film is still widely used. However, the availability of advanced computers and sophisticated radiation detection and image processing techniques has led to the development of digital radiographic imaging sytems - CT scanners. These systems measure the transmitted radiation with an array of radiation detectors. The signals from the arrays are used to construct a density image that can be analyzed using computer techniques.

In *Radioimmunoassay (RIA) measurements*, isotopes of Iodine (I-125, I-132) or other radionuclides are used to label organic molecules in samples taken from the patient. Analyzers incorporating single or multiple small NaI(Tl) well detectors count the activity in the samples to determine the presence of certain agents. This information can help in diagnosing certain conditions.

Single Photon Computerized Tomography (SPECT)

In nuclear medicine, it is often necessary to view the image of the distribution of γ-ray-emitting isotopes throughout the patient. The *gamma camera* is a device that senses the two dimensional coordinates of a γ-ray photon as it interacts in a large-area detector. It forms an image through the accumulation of many such events over the exposure time.

A pin-hole or parallel hole lead collimator restricts the γ-rays that strike the detector so that an image of the two dimensional radioactivity distribution is projected on the scintillation crystal. A matrix of photomultiplier tubes mounted on the back of the scintillation crystal detects the scintillation light. By using two or more large area detectors and by rotating them around the patient, a three dimensional image of the radioactivity distribution can be obtained. This technique is called SPECT.

Photon statistics play an important role in determining the position of an interaction event accurately. Because of the rather low radiation energies detected, only NaI(Tl) scintillation crystals with their large light output can be used in detectors for these applications at present.

Current gamma cameras use NaI(Tl) scintillation crystal detector plates in sizes up to 80 cm (31.5"), in cylindrical or rectangular geometries. Larger sizes can be made. Since Tc-99m, which emits 140 keV γ-rays, is the nuclide most often used, a crystal thickness of 1cm is sufficient. However, using thicker plates allows 511 keV studies to be done as well.

The production of large area gamma camera scintillation crystal plates requires special techniques to ensure a homogeneous response over the entire surface as well as to maintain long term, hermetic seal integrity.

Positron Emission Tomography (PET)

The PET technique evolved in the mid-1970's and uses positron emitting isotopes (e.g. C-11, O-15, F-18) to study chemical and biological processes in the body. The primary $\beta+$ radiation that these nuclides emit cannot penetrate far in tissue - only a few millimeters. When a positron has slowed down in the surrounding material to nearly rest, it will annihilate with an electron, creating two nearly collinear 511 keV gamma photons.

By setting a coincidence requirement on two opposing scintillation detectors, measuring only 511 keV quanta, and by accumulating several events, information about the position of the nuclide in the object an be obtained. A full tri-dimensional image of the radioactivity distribution can be obtained using advanced imaging techniques. With PET, images of biologically active compounds can be made for studying metabolic processes.

PET systems require no collimators like those used in SPECT. The half life of most positron emitters is small so that the count rate in detectors for PET can be high (up to a MHz).

PET detectors require that the detector material be of high enough Z to maximize the photoelectric cross sections for 511 keV, be fast enough to handle the count rates involved, and be able to distinguish coincident events. Furthermore, detectors for PET should show a reasonable pulse height resolution for 511 keV so that scattered events in the patient can be rejected.

Early PET scanners used NaI(Tl) scintillation crystal detectors but most current devices are equipped with BGO crystal detectors. The trend is towards higher count rates which makes scintillators like CsF, BaF_2 and undoped CsI (all relatively fast but with low light output) more attractive.

When using ultra-fast scintillators, *Time of Flight Positron Emission Tomography (TOFPET)* becomes possible. In this technique, the difference between the moments of interaction of the two annihilation quanta provides information on the position of the annihilation. Time-of-flight information also allows the rejection of annihilation events outside a selected area. This improves the signal-to-noise ratio.

Computerized Tomography (CT)

A CT scanner is a sophisticated X-ray machine that produces a three dimensional density image of all or part of the body. This is accomplished by measuring the transmitted intensity of X-rays in a small collimated beam at different angles. Often, a CT scanner consists of an X-ray source and an array of small scintillation crystals coupled to silicon photodiodes that rotates around the patient. The detectors operate in DC mode and measure the current from the diodes.

Since photodiode arrays are used, the scintillator should show sufficient light output in the long wavelength part of the spectrum. Furthermore, a high density to stop X-rays up to 150 keV in a thin crystal is desired. Scintillators like CsI(Tl), $CdWO_4$ and BGO can be used.

A very important parameter for a CT scanner is a constant detector response, which is necessary to obtain a reliable, high contrast image. Since a detector element is irradiated every several milliseconds, the afterglow of the scintillator becomes a factor in this. $CdWO_4$ has a low afterglow, and is a high density material well suited for use in CT scanners. CsI(Tl) shows a larger light output but it also has a longer afterglow than $CdWO_4$. BGO can only be used for special applications due to its relatively low light output.

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