Radiol Oncol 2000; 34(4): 381-5.

Current trends in diagnostic nuclear medicine instrumentation

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Introduction. *The basic principles of nuclear medicine imaging instruments are described with the emphasis on multi crystal scintillation and semiconductor gamma cameras, positron emission scanners without detector inter-ring attenuation septa, new detector materials and semiconductor CZT surgical probes for localization of metastases. For the estimation of minimal useful detector diameter the theoretical equation was derived for the 20% loss of absorbed gamma rays at the edge of the detector.*

Conclusions. *Nuclear medicine instrumentation has been passing throgh vigorous development in the last years and will be most likely also in the near future. New detection materials with much better physical characteristics than the standard NaI as regards the stopping power, energy resolution, fragility, decay time, light output, and density will most likely replace the NaI. It is expected that new imaging devices with several thousands of tiny crystals or semiconductor array of small position sensitive areas will improve the sensitivity and specificity of clinical studies. At the same time the small surgical probes made from these materials are also becoming very popular in surgery tracing the regional metastases.*

Key words: gamma cameras; tomography, emission-computed-instrumentation; multi-crystal scintillation gamma camera, semiconductor gamma camera, 2-D and 3-D PET scanner

Introduction

In nuclear medicine imaging the use of multi-detector systems for total-body, brain and heart scanning have recently gained increasing popularity. The systems with 2 or 3 Anger type gamma cameras 180°, 120° or 90° apart with small or large field of view are most popular. These systems have considerably improved the sensitivity, resolution and scanning time compared to the single camera systems. PET imaging with rings of small detectors in multi-slice configuration with lead or tungsten septa between slices (2-D) or without septa (3-D) has been successfully implemented using [F^{18}]FDG in a limited axial region and most recently also for total-body tomography. The biggest improvement in spatial resolution of PET was gained in the past mainly by the reducing the size of the crystals which currently resolve the structures to 5 mm in size. An intense development of the imaging systems is under way with a large number of tiny crystals made of the newly developed high dense and fast responding materials.

Received 23 May 2000
Accepted 14 June 2000

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Methods

Principles of image detection

Generally, three basic methods of nuclear medicine image formation are being used: 1. a single large scintillation crystal with a large number of photo multiplier tubes (PMT), 2. a large number of tiny scintillation crystals with position sensitive photo-multiplier tube (PSPMT) and 3. a large semiconductor crystal with an array of tiny "n-p" sensitive areas.

In the first method, the gamma ray hits a large circular or rectangle thin crystal and the induced scintillation light is distributed between the PMTs (Figure 1a) according to their viewing spatial angles. The x and y positions of PMTs are weighted by their electric signal responses from all PMTs and the X and Y coordinates of the scintillation cloud striking the array of PMT. The corresponding energy is computed (Anger type of gamma camera). The intrinsic spatial resolution of the imaging device strongly depends on the crystal thickness, slightly less on the size (number) shape of PMT and the position circuit. In contrast to the sensitivity of the system, the spatial resolution increases by the crystal thickness. The energy and time resolution depend mainly on the crystal material. Nowadays nearly all planar gamma cameras are of this kind.

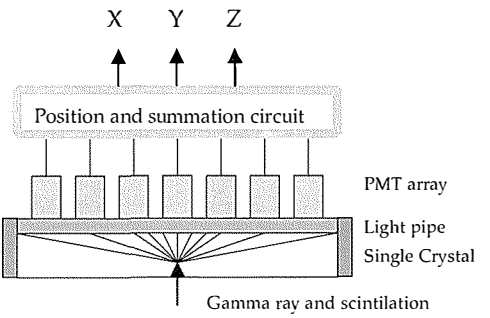

Figure 1a. Imaging detector with a single large scintillation crystal with set of PMTs.

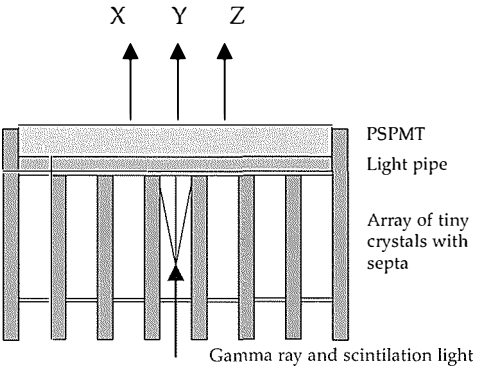

Figure 1b. Imaging detector with an array of tiny scintillation crystals with PSPMT.

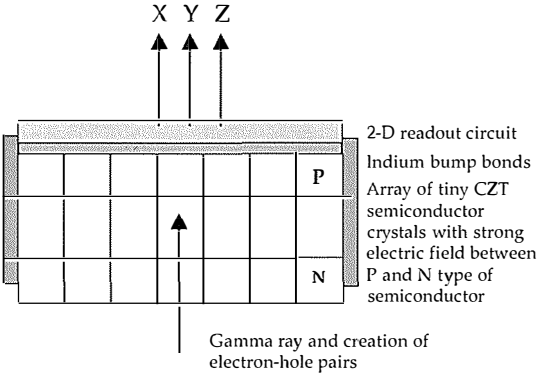

Figure 1c. Imaging semiconductor array detector.

In the second method (Figure 1b), the gamma ray is absorbed in a tiny crystal and all the induced scintillation light is collected by a small area of position sensitive photo multiplier tube which converts the incident light in a very thin layer into a charge or current which is then converted to digital E (energy) signal. Each small sensitive area of PSPMT provides also corresponding spatial coordinates X and Y for the particular exposed crystal. The spatial resolution strongly depends on the size of the crystals and on the thickness and material of the septa. Each crystal represents a pixel in the final digital image. The thinner and longer the crystal and

the thinner the septa, the better is the resolving power of the imaging system. Still, there is a limitation in the cross sectional size of the crystal (Figure 2).

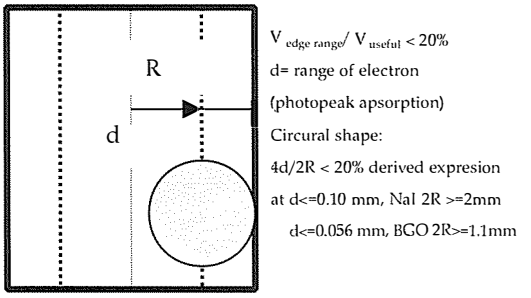

Figure 2. An estimate of the limit for the crystal cross sectional size. The shadow region represents the volume of induced ionization and the place from which the scintillation light contributes to the energy signal. The data for stopping power of electrons were taken from.[1]

If gamma ray hits the crystal too close to the reflector cover or to the optical fiber (crystal is inside optical fiber) then some of the ionization does not contribute to the scintillation light; therefore, the energy signal is weaker. Consequently, a definite volume close to the edge is not useful and is treated as scattered. An estimate is given for the circular shape of the crystal and for two popular scintillation materials (NaI and BGO). The scintillation light will be reduced in approximately 20 % of absorbed gamma rays. Some of the signals coming from these 20 % will still be included in the lower part of the photopeak but some will be lost. Therefore, there is no meaning of using thinner sized crystals than 1-2 mm depending on the material (for NaI crystal this size is limited to 2 mm and for BGO crystal 1 mm). The sensitivity increases with the crystal length, density and cross sectional size. The energy resolution is improved by more efficient PSPMT and more effective collecting of scintillation light in crystal.

In the third method (Figure 1c) the incident gamma ray is absorbed in the region of "p-n junction" region of the semiconductor crystal and a large number of electron-hole pairs is created. Their number (approximately 3 - 5 eV/electron-hole pair is spent on average) is proportional to the energy of the gamma ray and is nearly ten times greater than the quantity of the scintillation light (approximately 30 eV per ionization). For the same reason, the energy resolution is better. Because of the improved energy resolution the quantity of the Compton scattered gamma rays in the energy peak window is considerably reduced, whereas the contrast of the structures in the scan is much better (Figure 3).

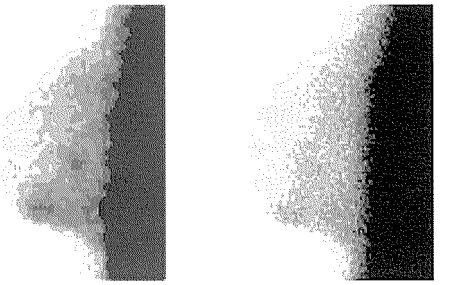

Figure 3. Comparison of breast scans from CZT and NaI Anger gamma camera.

The efficiency of the lately developed CZT crystal is even better than for NaI. The intrinsic spatial resolution of the CZT gamma camera is considerably better and is about 2 - 3 mm for 140 keV compared to 3 - 4 mm for NaI gamma camera. On the other hand, the collection time for electron-hole pairs is at least 100 times shorter than is the decay time for scintillation light in NaI crystal; therefore, the increased count rate can be achieved (250.000 counts/s). The weight of such imaging system is 100 times lower and the CZT gamma camera is easily portable to emergency departments or elsewhere. It is expected that the price for such gamma camera will also be much lower because of the less complicated production.

Table 1. Physical properties for some interesting scintillation materials. NaI-Thalium-doped sodium iodide, BGO-Bismuth germanate (Bi4Ge3O12), LSO-Lutetium oxyorthosilicate, YSO-Yttrium oxyorthosilicate

	NaI	BGO	LSO	YSO
Density (g/cm3)	3.67	7.13	7.40	4.54
Effective Z	51	74	66	34
Decay time (ns)	230	300	40	70
Relative light output	100	15	75	120
Energy resolution	7.8 %	10 %	<10 %	<7.5 %
1/μ for 140 keV	4.2 mm	0.82	1.0	7.7
1/μ for 511 keV	30 mm	11	12	26

New materials for detection crystals

Some new detector materials were developed recently which promise a considerable improvement of nuclear medicine imaging devices. These materials are presented in Table 1.[2]

The LSO is intrinsically radioactive and is not useful for SPET but can be used for PET (coincidence measurement excludes the single decay and absorption of gamma ray inside LSO crystal). The use of LSO and YSO is very promising in the so-called phoswhich detector where the YSO crystal (1 - 2 cm) is in front and the LSO crystal (1 - 2 cm) is optically coupled to the YSO. The YSO is used for attenuation of low energy (in the range of 100 keV) and LSO serves as the light pipe and for attenuation of high energy gamma rays. The induced scintillation signals from both crystals can be separated because of their different decay times.

The BGO is nearly exclusively used for PET, but will probably be replaced by this phoswhich detector.

One of the most interesting detectors is semiconductor CZT (cadmium zinc telluride) which has even better stopping power for 140 keV gamma rays than NaI (Tl) and much better energy resolution (for factor of 10). An array with a large number of very small sensitive areas can be formed so that each of these areas is a separate pixel in the digital image.

SPET

The biggest improvement in the SPET was the development of several detector heads which drastically improved the system sensitivity. In cardiac SPET, the use of two heads at 90° and the whole-body bone SPET or scanning at 180° shortens the acquisition time or doubles the acquisition counts. The improvement of the gamma camera features was mainly due to the development of the so-called digital head electronics which replaced the old analog position circuit by the digital one. The output from each PMT is digitized and the spatial coordinates are then computed. All corrections for non-linearity, spatial and energy non-uniformity can be performed on-line by the use of special fast processors.

In the future, the probable development of SPET will involve the building the tomographic system of several modular multi-crystal detectors which will introduce even greater flexibility than that with two or three big planar gamma cameras at different angular setup. Each module will probably be an array of very tiny crystals from YSO and LSO or semiconductor CZT detector.

Another possible approach in modular design of SPET will be the development of special models for each organ (*i.e.* thyroid and cardiac tomograph needs relatively small sized detector's modules) and, possibly, much lower prices for small SPET systems.

PET

The latest improvements are mainly due to the development of the so-called 3-D tomographs which do not use septa between planes with rings of crystals.[3] By omitting the septa the sensitivity is increased by ten times and the amount of scattered photons for approximately 30 %. In such configuration of several thousands tiny crystals all possible coincidence events between any two crystals are used in the reconstruction algorithm. The system works in a true 3-D mode. Another possibility of PET is the use of double-head SPET system with or without collimator (high-speed electronics is essential) in a coincidence mode. This type of PET is considerably less sensitive but is interesting to perform both PET and SPET studies.

A much better sensitivity of the system is expected in future from the new generation of the PET. It will be of extreme importance in the imaging of specific biochemical bindings, such as receptor binding. In this applications a small amount of the injected radioactivity is collected by a target organ (usually less than 1 %).

Surgical gamma probe

This application of radioactivity tracing becomes very important in surgery for identifying the regional metastases. Currently interesting clinical field where the small detector probe is of great importance is lymph node dissection of the axilla or regional nodes in the breast cancer patients and in some melanoma patients. The role of the surgical gamma probe is to localize the sentinel node transcutaneously and intra-operatively. To meet a high sensitivity, good spatial and spectral resolution and appropriate ergonomic characteristics several of different commercially available probes were evaluated.[4] It was found that CZT probe was the most appropriate for low energies (140 keV from ^{99m}Tc) and the NaI probe for high energy (364 keV ^{131}I).

Conclusions

Nuclear Medicine instrumentation has been passing through vigorous development in the last years and will be most likely also in the near future. New detection materials with much better physical characteristics than the standard NaI as regards the stopping power, energy resolution, fragility, decay time, light output, and density will most likely replace the NaI. It is expected that new imaging devices with several thousands of tiny crystals or semiconductor array of small position sensitive areas will improve the sensitivity and specificity of clinical studies. At the same time, the small surgical probes made of these materials are also becoming very popular in surgery tracing the regional metastases.

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