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

CURRENT TRENDS IN NUCLEAR INSTRUMENTATION IN DIAGNOSTIC NUCLEAR MEDICINE

Valentin Fidler

University Medical Center Ljubljana, Nuclear Medicine Department , Slovenia

ABSTRACT

An overview of physical principles of imaging techniques in Nuclear Medicine is presented with emphasis on latest trends in the development of multi-crystal scintillation and semiconductor gamma cameras, increased sensitivity of PET scanners with omitting the septa between the rings of crystals, development of miniature CZT probe system for radioactivity tracing in surgery for identifying the regional metastasis and development of new detecting materials. An estimate for minimal crystal diameter for NaI and BGO is given for approximately 20% loss of absorbed gamma rays in the crystal edge region.

Key words: 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 filed of view are most popular. These systems have considerably improved the sensitivity, resolution and the scanning time comparing 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 in the past was gained mainly by the reducing the size of the crystals which currently resolve the structures to 5 mm in size. There is a strong development of the imaging systems with a large number of tiny crystals made by the newly developed high dense and fast responding materials.

METHODS

Principles of image detection

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

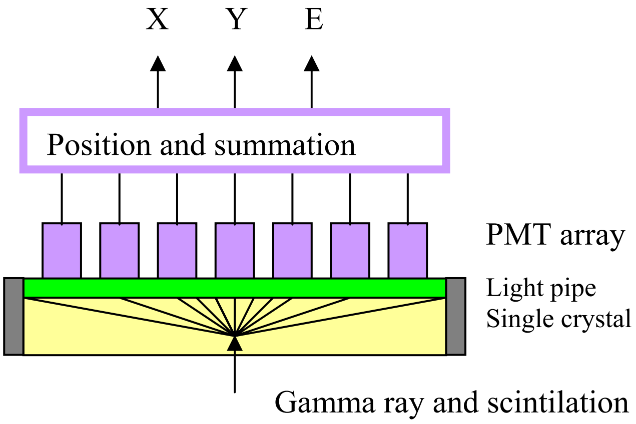


Fig. 1 A. Imaging detector with single large scintillation crystal with set of PMTs.

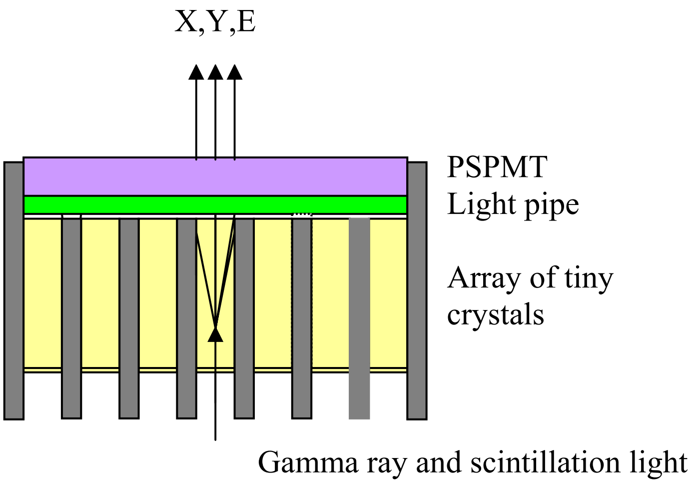


Fig. 1B. Imaging detector with array of tiny scintillation crystals with PSPMT.

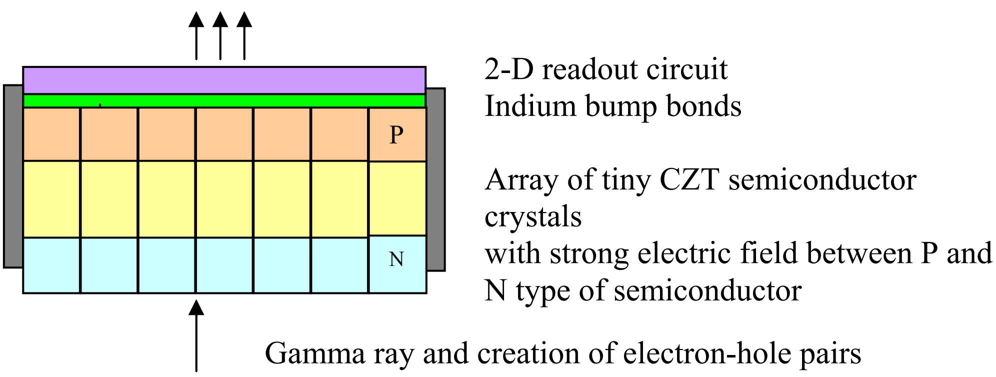
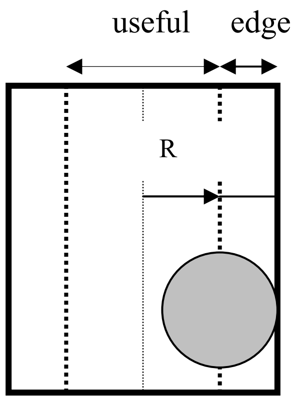


Fig. 1C. Imaging semiconductor array detector.

In the first method the gamma ray strikes a large circular or rectangle thin crystal and the induced scintillation light is distributed between the PMTs (Fig. 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 and 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. The sensitivity of the system oppositely to 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.
In the second method (Fig. 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. Thinner and longer the crystal and thinner the septa better is the resolving power of the imaging system. Still, there is a limitation in the cross sectional size of the crystal (Fig. 2). If gamma ray strikes the crystal too close to the reflector cover or to the optical fiber (crystal is inside optical fiber) then some of the ionization doesn't contribute to the scintillation light and therefore the energy signal is smaller. Because of this, 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). For approximately 20 % of absorbed gamma rays, the reduction of the scintillation light will be present. Some of the signals coming from these 20 % will still be included in the lower part of the photopeak but some will be lost. The conclusion is that 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 by 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.



$V_{edge\ range} / V_{useful} < 20\ \%$
d = range of electron (photopeak absorption)
Circular shape :
$4\,d / 2\,R < 20\ \%$ derived expression
at $d <= 0.10$ mm, NaI $2R >= 2$ mm
$d <= 0.056$ mm, BGO $2R >= 1.1$ mm

Fig. 2. 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).

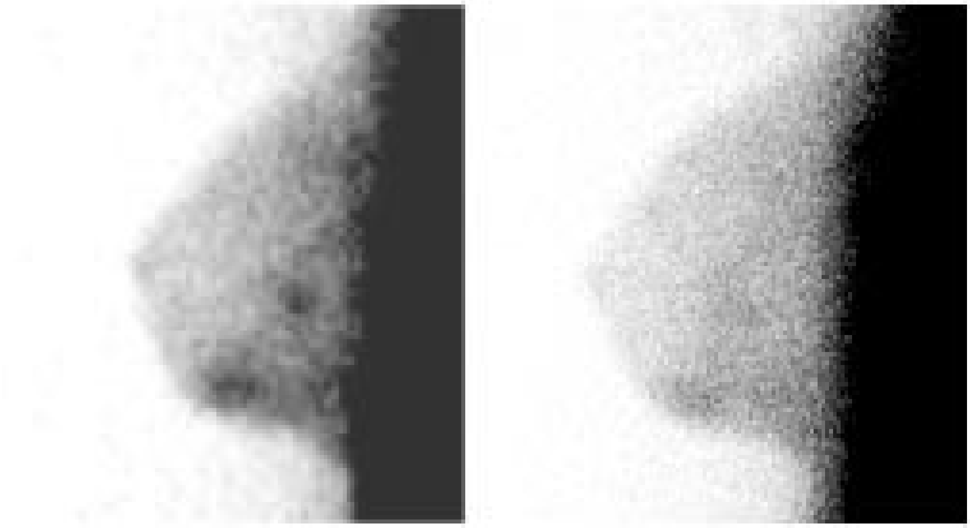
In the third method (Fig. 1C) the incident gamma ray is absorbed in the region of “p-n junction” region of the semiconductor crystal and a large number of electron-hole pairs are created. Their number (approximately 3 - 5 eV/electron-hole pair is spent on average) is proportional to the energy of the gamma ray and is nearly ten times greater than the quantity of the scintillation light (approximately 30 eV per ionization). For the same factor the energy resolution is better. The efficiency of the late 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 and 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.

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).

	NaI	**BGO**	**LSO**	**YSO**
Density (g/cm^3)	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

Table 1. Physical properties for some interesting scintillation materials. NaI – Thalium-doped sodium iodide, BGO – Bismuth germanate ($Bi_4Ge_3O_{12}$), LSO – Lutetium oxyorthosilicate, YSO – Yttrium oxyorthosilicate.



Picture 1. Comparison between CZT camera and NaI Anger camera breast scans.

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). Very promising is the use of LSO and YSO in 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 the difference in the 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 becomes semiconductor CZT (cadmium zinc telluride) which has even better stopping power for 140 keV gamma rays than NaI (Tl) but much better energy resolution (for factor of 10). The array with large number of very small sensitive areas can be build so that each of these areas is a pixel in the digital image.

SPET

The biggest advance in the SPET was the implementation of several detector heads which drastically improved the system sensitivity. For cardiac SPET the use of two heads at 90^o and the whole-body bone SPET or scanning at 180^o shortens the acquisition time or increases the acquisition counts two times. The improvement of the gamma camera features was mainly due to the development of so-called digital head electronics which had replaced the old analog position circuit by the digital one. The output from each PMT is digitized and then the spatial coordinates are 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 be building the tomographic system from the arrangement of several modular multi-crystal detectors which will introduce even greater flexibility than 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 in a specialization for some organs (i.e. thyroid and cardiac tomograph needs relatively small sized detector's modules) and possible 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 don't 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 only approximately 3o %. 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 dual-head SPET system with or without collimator (high-speed electronics is essential) in a coincidence mode. But this type of PET is considerably less sensitive but is interesting for performing 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 and in some melanoma patients. The role of the surgical gamma probe is to localise the sentinel node transcutaneously and intra-operatively. To meet a high sensitivity, good spatial and spectral resolution and appropriate ergonomic characteristics several 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 is passing vigorous development last years and most likely in the near future as well. New detection materials with much better physical characteristics than the standard NaI concerning 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.

REFERENCES

(1) Mladjenovic M. Radioisotope and Radiation Physics. Academic Press, NJ, 1983: 148 - 9.

(2) Links JM (1998). Advances in nuclear medicine instrumentation: consideration in the design and selection of an imaging system. Eur J of Nucl Med 1998; 25/10: 1453-66.

(3) Meikle SR, Dahlbom M (). Positron emission tomography. In: Murray IPC and Ell. P Nuclear Medicine in Clinical Diagnosis and Treatment. Churchill Livingstone Press 1994; 1327 - 37.

(4) Tiourina T, Arends B, Huysmans D, Rutten H, Lemaire B and Muller S. Evaluation of surgical gamma probes for radiological sentinel node localization. Eur J Nucl Med 1998; 25/9: 1224 - 31.