Comparison of two position sensitive gamma-ray detectors based on continuous YAP and pixellated NaI(TI) for nuclear medical imaging applications*

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Abstract Dedicated position sensitive gamma-ray detectors based on position sensitive photomultiplier tubes (PSPMTs) coupled to scintillation crystals, have been used for the construction of compact gamma-ray imaging systems, suitable for nuclear medical imaging applications such as small animal imaging and single organ imaging and scintimammography. In this work, the performance of two gamma-ray detectors：a continuous YAP scintillation crystal coupled to a Hamamastu R2486 PSPMT and a pixellated NaI(TI) scintillation array crystal coupled to the same PSPMT, is compared. The results show that the gamma-ray detector based on a pixellated NaI(TI) scintillation array crystal is a promising candidate for nuclear medical imaging applications, since their performance in terms of position linearity, spatial resolution and effective field of view (FOV) is superior than that of the gamma-ray detector based on a continuous YAP scintillation crystal. However, a better photodetector (Hamamatau H8500 Flat Panel PMT, for example) coupled to the continuous crystal is also likely a good selection for nuclear medicine imaging applications.

Key words gamma-ray detector, PSPMT, continuous YAP scintillation crystal, NaI(TI) scintillation array crystal

PACS 87.58.Pm, 29.40.Mc, 29.40.Gx

1 Introduction

In recent years great efforts have been made in developing compact high-resolution gamma-ray imaging systems to improve nuclear medical imaging[1, 2]. Small and flexible gamma-ray imaging systems (gamma camera and SPECT and PET, for example) with good spatial resolution have been used widely for nuclear medical imaging applications. The applications include small animal imaging and single organ imaging (thyroid gland and cardiac and brain, for example) and scintimammography[3], etc. The nuclear medical imaging system in general consists of collimator and position sensitive gamma-ray detector with image reconstruction software and display unit. The vital component of such imaging system is the gamma-ray detector and the performance determines its main parameters, which are the spatial resolution and the energy resolution, the position linearity, the effective field of view (FOV), the size and cost. The key for the gamma-ray imaging system to improve the nuclear medical imaging qualities is to optimize the performance of the gamma-ray detector. To obtain clearer nuclear medical image and decrease the size of gamma-ray imaging system, the gamma-ray detector should be small and flexible with good spatial resolution. In an attempt to optimize the gamma-ray detector, we compared the performances of two position sensitive gamma-ray detectors by using experiment methods. The two gamma-ray detectors were based on position sensitive photomultiplier tube (PSPMT[4]): first, a continuous YAP scintillation crystal coupled to a Hamamastu R2486 PSPMT and second, a pixellated NaI(TI) scintillation array

Received 17 March 2008, Revised 4 June 2008
* Supported by National Nature Science Foundation of China (10275063)
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crystal coupled to a Hamamastu R2486 PSPMT. In this paper, the first is named as continuous-crystal gamma-ray detector (CCD); and the second is named as array-crystal gamma-ray detector (ACD). In this work, the performances of two gamma-ray detectors are assessed in terms of spatial resolution and position linearity and energy resolution.

2 Equipment and methods

We have built up the continuous-crystal gamma-ray detector (CCD) and the array-crystal gamma-ray detector (ACD), respectively.

YAP scintillation crystal was already used in high-resolution gamma-ray imaging applications[5]. It has an emission spectrum peaking at 370 nm and its scintillation decay time is 30 ns. The light output is about 17000 photons/MeV. The properties of this scintillator allow it to be used in gamma-ray imaging systems with very high spatial and time resolution. The size of YAP scintillation crystal is 20 mm×20 mm×4 mm in continuous-crystal gamma-ray detector.

The NaI(TI) scintillation array crystal[6] we are reporting on here is composed of 2 mm×2 mm elements separated by a 0.2 mm thick white reflecting powder which also covers the back side of the array. The square area of Nal(Tl) scintillating array is 48.2 mm×48.2 mm. There are in total 484 pixels in a 22×22 matrix. The crystal thickness is 5 mm.

The main advantage of PSPMT is that the anode provides position information, thus only a PMT is necessary in order to obtain the X-Y coordinates for each detected photon. The Hamamatsu PSPMT R2486 has a bialkali photocathode, a12-stage coarse mesh dynode structure, and multiple anode wires crossing each other in the X and Y directions. The PSPMT R2486 has $16(x)+16(y)$ anodes. The output signals from each dynode can be divided through external resistive chains. The readout of the signals from PSPMT is in this way: An incident photon interacts with the crystal and deposits some of the energy and produces scintillation light. The scintillation light strikes the photocathode and liberates photoelectrons, which are multiplied at a 12-stage coarse mesh dynode system by an electric potential of typically 950 V for ^{241}Am. Thus, an electron cloud reaches the crossed wire anode stage. The readout of the 16 anodes signals can obtain the exact position of the incident photon in the XY plane. In this case, the electronic readout is based on conventional resistive divider technique. The four preamplifiers read out the four ends of the two resistive chains connected to the X and Y wires, respectively. The R2486 PSPMT resistive chain readout features easy signal processing. However, the resistive chain readout decreases the spatial resolution of system. In the future, we may be able to use a better photodetector based on multi-anode readout (Hamamatau H8500 Flat Panel PMT, for example) to improve the intrinsic performance of the detector.

To evaluate the detection performance of the CCD and ACD, a set of measure instrument is utilized. Fig. 1 shows the block diagram of the measure instrument. The ^{241}Am source is collimated with ϕ1 mm copper collimator, the position of which can be changed in X and Y directions by step motor, so as to scan the surface of crystal (YAP and NaI(TI)). The entire system, including data acquisition and real time analysis, is controlled by PC with LabVIEW software.

The energy spectrum for a given position is obtained, and the full width at half maximum (FWHM) of the 59.5 keV full energy peak is taken as a proper window width. In data acquisition, only the events inside the energy window are recorded. With the energy and X and Y position information, the center of gravity of scintillation light can be calculated. The data are processed with PAW[7] software. In the experiment, the energy window width, the image position, the count number for each irradiation spot and the other data are stored as independent files.

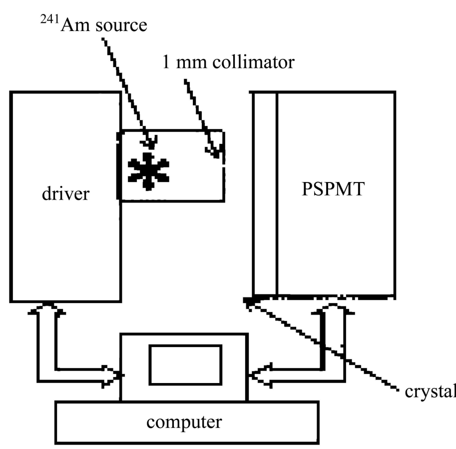

Fig. 1. The block diagram of the measure instrument.

3 Results and discussion

3.1 Position linearity

To study the position linearity of the two gamma-ray detectors, the surfaces of YAP continuous crystal and NaI(TI) array crystal have been scanned by the collimated ^{241}Am source, respectively. The scan range of YAP is from −10 mm to +10 mm with a step of 1 mm along one crystal axis. The data of 19 points are collected. The scan range of NaI(TI) is from −24 mm to +24 mm with a step of 2.2 mm along one crystal axis. Each pixel of a row is irradiated. The data of 22 points are collected.

Figure 2 shows the position linearity response of CCD (a) and ACD (b), respectively. The data are fit with a straight line.

From Fig. 2(a), the following conclusions can be obtained: (1) The position linearity covers the same crystal dimension (12 mm). The slope of position linearity is constant within 15% in a range of 12 mm. A change of slope of about 60% occurs at the PSPMT boundary (4 mm). The results show that the position linearity in a range of 12 mm is good. So the effective field of view (FOV) is only 60%. (2) Position linearity response can be described by differential non-linearity and absolute non-linearity[8]. For a uniform distribution of mechanical position of impact point, the differential non-linearity is defined as the standard error of distance between borders upon two image points. Absolute non-linearity is defined as the maximum deviation between image position and straight line fit. The average values of differential non-linearity and absolute non-linearity are 0.566 mm and 3.866 mm, respectively.

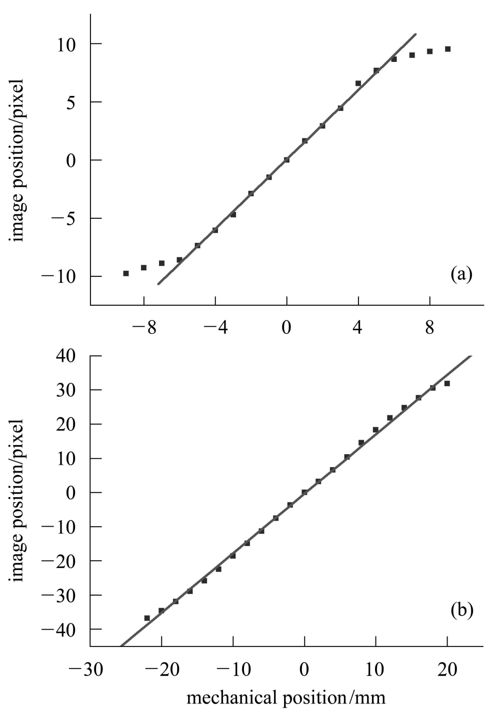

Fig. 2. The position linearity response of two detectors: (a) CCD and (b) ACD.

From Fig. 2(b), the following conclusions can be obtained: (1) The position linearity covers the whole crystal dimension (48 mm). The slope of position linearity is constant within 10% in a range of 40 mm. A change of slope of about 30% occurs at the PSPMT boundary (4 mm). The results show that the position linearity in a range of 40 mm is good. The effective field of view (FOV) is 83%. (2) The average values of differential non-linearity and absolute non-linearity are 0.375 mm and 1.426 mm, respectively.

Comparing the position linearity responses of CCD and ACD, the latter shows a larger effective field of view (FOV) and a better position linearity range.

3.2 Energy resolution

Figure 3 shows the typical energy response of the CCD (a) and ACD (b), respectively. At 59.5 keV, the CCD shows an energy resolution of 21% with respect to the 20% of the ACD. The difference between these two detectors is very small. The energy resolution seems to have relatively little impact on the effective field of view (FOV) of the detectors.

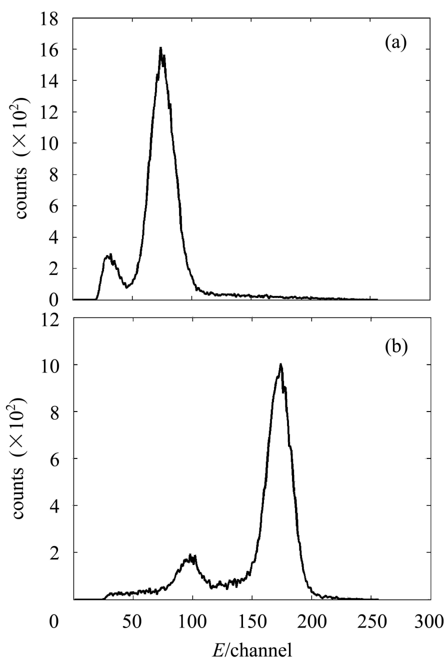

Fig. 3. The typical energy response of two detectors: (a) CCD and (b) ACD.

3.3 Spatial resolution

In the experiment, each irradiation spot is stored as an independent file for a point-by-point analysis of spatial resolution. Fig. 4 and 5 show a typical Point Spread Function (PSF) of CCD and ACD, respectively. The irradiation spot is located in the middle of the crystal. By fitting a Gaussian curve to the Point Spread Functions, spatial resolutions of 4.8 mm FWHM and 2.3 mm FWHM are obtained, respectively. By using the same analysis process, the spatial resolutions of all the irradiation spots are obtained. Fig. 6 shows the spatial resolution response as a function of position for CCD (a) and ACD (b), respectively. Fig. 6(a) shows a mean value of spatial resolution of 5.7 mm FWHM and a best value of 4.3 mm FWHM. Fig. 6(b) shows a mean value of spatial resolution of 2.5 mm FWHM and a best value of 2.2 mm FWHM.

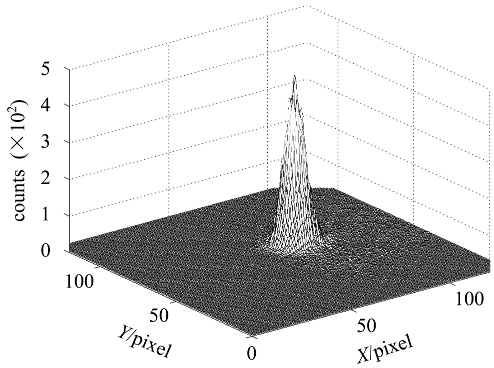

Fig. 4. A typical point spread function (PSF) of CCD.

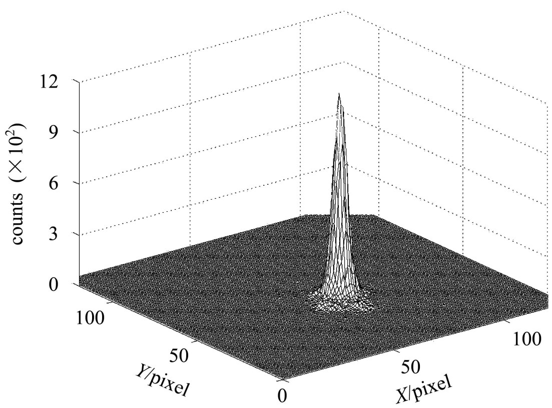

Fig. 5. A typical point spread unction (PSF) of the ACD.

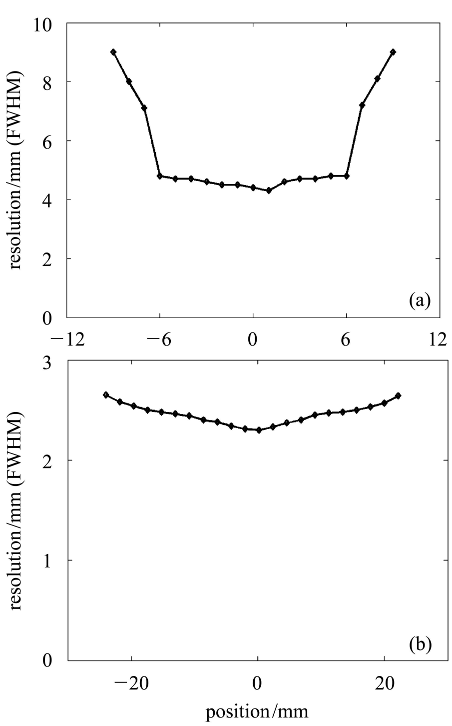

Fig. 6. Spatial resolution response as a function of position: (a) CCD and (b) ACD.

Comparing the spatial resolutions of CCD and ACD, the following conclusions can be obtained: (1) At the edges of the crystal, it is clear that the spatial resolution values are poor as a result of the shrinkage effect[9]. However, the shrinkage effect of CCD is larger than ACD. It demonstrates that the large spread of light distribution on the boundaries of the continuous crystal aggravates the shrinkage effect and the position distortions near the edges of the crystal. On the contrary, the scintillation array crystal can reduce the shrinkage effect and decrease the position distortions greatly via producing very focused light spot[10]. (2) ACD shows a larger effective field of view. It is in agreement with the foregoing results. (3) The change of spatial resolution is not evident in the effective field of view. It shows that the detection performances of CCD and ACD are steady in the effective field of view. (4) ACD shows a significantly better value of spatial resolution. The results demonstrate that the scintillation array crystal could improve the spatial resolution greatly via restricting the spread of photons. Each pixel of the scintillation array crystal can produce a very focused light spot, so it improves the spatial resolution of ACD when the position of the gamma ray interaction is estimated by a centroid calculation of light distribution. On the contrary, the use of a continuous crystal involves a large spread of light distribution and as a consequence a spatial resolution worse than the scintillation array crystal and a small position linearity range are what results.

4 Conclusions

In this study, the tests of position linearity, energy response and spatial resolution are carried out by coupling the PSPMT R2486 to a continuous YAP scintillation crystal and to a pixellated NaI(TI) scintillation array crystal. Compared with the continuous-crystal gamma-ray detector (CCD), the results indicate that the array-crystal gamma-ray detector (ACD) has good position linearity, good spatial resolution and larger effective field of view. It has demonstrated that the array-crystal gamma-ray detector is a practical technique for nuclear medicine imaging applications. Thus the array-crystal gamma-ray detector is a promising candidate for nuclear medical imaging applications.

Nevertheless further researches are needed. A better photodetector can be employed to improve the intrinsic performances of the detector. In the future, we may be able to use the Hamamatau H8500 Flat Panel PMT coupling with the continuous crystal (continuous Lanthanum tri-Bromide Cerium-doped ($LaBr_3$:Ce) crystal, for example) to achieve higher spatial resolution[11]. The H8500 Flat Panel PMT is based on metal channel dynode for charge multiplication and 8×8 anodes for charge collection and position calculation. Compared with the R2486 PSPMT resistive chain readout, the H8500 Flat Panel PMT multi-anode readout has a better spatial resolution. So the continuous-crystal gamma-ray detector based on Hamamatau H8500 Flat Panel PMT coupled to the continuous crystal is also likely a good selection for nuclear medicine imaging applications.

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