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High count rate gamma camera with independent modules

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ABSTRACT

Advances in nuclear medical imaging are based on the improvements of the detector's performance. Generally the research is focussed on the spatial resolution improvement. However, another important parameter is the acquisition time that can significantly affect performance in some clinical investigation (e.g. first-pass cardiac studies). At present, there are several clinical imaging systems which are able to solve these diagnostic requirements, such as the D-SPECT Cardiac Imaging System (Spectrum Dynamics) or the Nucline Cardiodesk Medical Imaging System (Mediso). Actually, these solutions are organ-specific dedicated systems, while it would be preferable having general purpose planar detectors with high counting rate.

Our group has recently introduced the use of scintillation matrices whose size is equal to the overall area of a position sensitive photomultiplier tube (PSPMT) in order to design a modular gamma camera. This study allowed optimising the overall pixel identification by improving and controlling the light collection efficiency of each PSPMT. Although we achieved a solution for the problems about the dead area at the junction of the PSPMTs when they are set side by side. In this paper, we propose a modular gamma camera design as the basis to build large area detectors. The modular detector design allows us to achieve better counting performance. In this approach, each module that is made of one or more PSPMTs, can actually acquire data independently and simultaneously, increasing the overall detection efficiency. To verify the improvement in count rate capability we have built two detectors with a field of view of $\sim 5 \times 5\text{cm}^2$, by using four R8900-C12 PSPMTs (Hamamatsu Photonics K.K.). Each PSPMT was coupled to a dedicated discrete scintillation structure designed to obtain a good homogeneity, high imaging performance and high efficiency. One of the detectors was designed as a standard gamma camera, while the other was composed by four independent modules. Then, we compared the counting rate measurements demonstrating the validity of the modular approach and opening the way to build larger area devices.

1. Introduction

A standard gamma camera, as is well known, contains a NaI(Tl) crystal coupled to a large number of photomultiplier tubes (PMTs). The glass window of the scintillator acts as a light guide that optimises the optical coupling.

The γ-rays' interactions in the scintillator will create a light that is detected in principle by all the PMTs. The signal amplitudes from the PMTs decrease with the distance from the original scintillation point. Therefore the readout electronics allows us to detect position and energy information for each γ-ray interaction. Electronics consists of a resistive network or individual analogue to digital converter (ADC) for each PMT [1].

The detection of a scintillation event involves all the PMT; following an interaction there is a time interval, known as dead time, in which the entire area of detection is blind to other events, also for the PMTs not directly affected from the scintillation light.

Dead time causes counting losses of the detection system that will lead inaccurate counting rates especially at high count rates. Dead time corrections can be applied to estimate the true count rate; these corrections, however, become increasingly inaccurate when counting losses increase. The inaccuracies due to dead time losses can significantly affect performance in certain high counting rate applications such as in first-pass cardiac studies [2].

Furthermore, at high counting rates, there is a high likelihood that two or more pulses are sufficiently close together to be treated as a single event. This effect is known as pile-up. When it happens the pulses sum together and produce an amplitude that is not representative of each event. Hence, the pile-up effect distorts energy information, but moreover contributes to counting losses.

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Detectors may exhibit two dead time behaviours: in the first, called the non-paralyzable model, events occurring during the dead time go undetected and ignored. In the paralyzable model, instead, events occurring during the dead time act as new detected events and extend the dead time itself. So, due to the dead time τ, the detector perceives counts at a lower rate m with respect to the true events rate n.

For the non-paralyzable model, the observed count rate is [3]

$$m=\frac{n}{1+n\tau}.$$

While for the paralyzable model the measured count rate is [3]

$$m=ne^{-n\tau}.$$

In a modular device the active area is divided into several sub-areas which act as independent detectors. The γ-ray flux is divided among the modules mitigating the dead time effects. Considering the whole system as a single detector, an overall or "effective" dead time can be introduced. In this regard, the improvement in counting rate capability could be interpreted as a reduction of the effective dead time.

Considering a device consisting of $\mathcal{N}$ modules, and denoting n_i as the true events rate for the ith module, for the non-paralyzable model the measured count rate will be

$$m=\sum_{i=1}^{\mathcal{N}}\frac{n_i}{1+n_i\tau}.$$

While for the paralyzable model the recorded count rate will be

$$m=\sum_{i=1}^{\mathcal{N}}n_i e^{-n_i\tau}.$$

Obviously, the best results are obtained when the workload is divided evenly among the modules, which is the case of identical modules subjected to a uniform flux of radiation.

In this situation the γ-rays detection events are distributed with equal probability among the detectors that have the same detection efficiency. Denoting as n the true interaction rate of the overall system, the true events rate for the ith module can be expressed as $n_i=n/\mathcal{N}$, so the recorded count rate for non-paralyzable model simplifies to:

$$m=\frac{n}{1+n\tau/\mathcal{N}}.$$

While for paralyzable model is

$$m=ne^{-n\tau/\mathcal{N}}.$$

The space division into $\mathcal{N}$ independent sub-areas of the field of view (FOV) reduces the effective dead time by a factor up to $1/\mathcal{N}$. Clearly, the improvement amount depends on the distribution of the activity and on the source–detector geometry. However, in principle, this approach allows faster counting rates for a device with given active area.

The spatial resolution is an important feature used to evaluate imaging performance of gamma camera systems. Acquired images have limited spatial resolution because of the physical characteristics of the detector and the associated electronics. In order to improve the intrinsic spatial resolution may be increased the efficiency of the light collection, for example, using more efficient PMTs or an enhanced optical coupling. The intrinsic spatial resolution depends, also, on the crystal thickness. Moreover, the intrinsic spatial resolution could be improved by corrections for detector non-linearity and non-uniformity. However, the different approaches used to improve the spatial resolution have led to large FOV gamma camera whose best intrinsic spatial resolution is at least 3 mm full width at half maximum (FWHM) at 140 keV [2]. This is mainly due to technological limitations such as the presence of dead zones on the PMTs.

In a gamma camera a critical factor that limits the imaging performance is the scintillation light distribution (SLD). A light guide must be used between the crystal and the system of PMTs, in order to optimise the optical coupling. Using a thin light guide, the spatial resolution will be non-uniform along the active area, with the worst resolution at the PMTs centre. Using a thick light guide, instead, the spatial resolution will be sufficiently uniform, but the amount of light reaching a single PMT will be weak, which worsens the measuring accuracy and therefore also the resolution. There is, therefore, an optimal thickness that allows a sufficient uniformity of spatial resolution simultaneously to a sufficient measuring accuracy [4].

To improve the intrinsic spatial resolution of a gamma camera, a choice widely used is to employ discrete scintillator arrays. This solution allows focussing the scintillation light onto a small area of the photodetector, then the FWHM of the SLD is reduced. Indeed, the scintillation light from γ-ray absorption within a given crystal will be confined to that crystal. Thus the intrinsic spatial resolution is set by the crystal size [5]. The narrow SLD emerging from the scintillator crystals must be read out by high resolution photo-sensors, such as are the PSPMTs. The detectors that exploited this solution have produced promising results, although they are usually limited to the use of a single PSPMT.

The main problems we have faced up are related to the construction of large area devices for medical applications based on an array of PSPMTs (multi-PSPMT gamma camera) [6]. One of the major limitations of PSPMTs is their inability to properly collect the light in correspondence of the dead area at the junction of the PSPMTs when they are put beside. Practically, to obtain large area detectors these dead areas should be smaller than 3–5 mm.

Our group has introduced a solution based on CsI(Tl) crystals integrated into a square hole tungsten structure [7], and has recently investigated to optimise their pixel identification in order to design a modular gamma camera [8]. The use of tungsten structure allows us to reduce the inter-crystal Compton scatter and eliminates the optical crosstalk, achieving an enhancement of the image contrast [9,10]. This solution also allows the confinement of the scintillation light in the crystal in which the γ-ray has interacted, this condition ensures that the position of interaction can be calculated independently for each module. The modular design [11] constitutes a substantial innovation with respect to the single charge collection surface proposed by Soluri et al. [6] in the past, in which the anode wires of each photomultiplier were connected with the contiguous ones. In the present work we have build a modular detector in which each PSPMT represents an independent module. Since the PSPMTs are mutually independent, each module can be exploit their excellent spatial performance to optimise the intrinsic spatial resolution.

2. Equipment and method

In order to verify the rate counting capability of a modular gamma camera we have built two detectors. Both devices consist of a 2×2 array of R8900-200-C12 PSPMTs (Hamamatsu Photonics K. K.), optically coupled to opportune scintillation assemblies, as shown in Fig. 1.

The R8900-200-C12 has an external size of $26.2\times26.2\times27\ \text{mm}^3$ with an active area of $23\times23\ \text{mm}^2$. This PSPMT features a 10 metal channel dynodes charge multiplication system, which provide a gain of about $1\cdot10^6$ at -800 V. The multiplied charge is collected by 6(X)+6(Y) strip anodes. An additional reflective dynode positioned after the anode structure assures a better charge collection. The R8900-200-C12 exploits the "ultra bialkali"