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Preliminary study for pixel identification on a modular gamma camera

A. Soluri *, G. Atzeni, A. Ucci, F. Cusanno, R. Massari

Institute of Biomedical Engineering, CNR, Rome, Italy

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ABSTRACT

Our group has recently investigated and produced new scintigraphic prototypes based on advanced scintillation structure. The aim of this study is to demonstrate the use of scintillation matrices with size equal to the overall area of the Position Sensitive Photomultiplier Tube (PSPMT), to design a modular gamma camera and study the solution of the dead area problems optimizing the overall pixel identification.

In this paper we investigate the response of different combinations with crystals integrated within tungsten structure, coupled with H8500, R8900-C12 and R11265-M64 Hamamatsu PSPMTs. Several scintillation matrices, whose dimensions match to the physical area of the PSPMT, have been analysed so that we have also studied limits of detection for the elements of the matrix in the critical zones of the PSPMT, i.e. corners and borders.

In order to enhance the detectability of scintillation elements we improved the light collection by depositing metallic layers or treating the tungsten structure with different coating materials, and shaping the external elements of the scintillation matrices.

The results have shown good energy resolution and the proposed method can be applied in medical imaging for obtaining high efficiency scintillation devices.

1. Introduction

The last fifteen years have seen a large application of PSPMTs (Position Sensitive Photomultiplier Tube) in medical imaging [1]. The use of different PSPMT models in the last twenty years has opened the way to the expression “High Spatial Resolution”, that simply refers to a gamma camera having an intrinsic resolution better than commercial systems (typically 3–4 mm). To design high resolution devices, exploiting the PSPMT excellent intrinsic spatial resolution, scintillation crystal arrays have replaced planar crystals [1]. To this purpose, our group has introduced an array based upon the use of CsI(Tl) crystals integrated into a square hole tungsten structure [2]. Such solution allows a square hole tungsten collimator to be used [3–5]. This ensures the matching of the collimator hole with the crystal, bringing a better signal-to-noise ratio due to the suppression of the pattern mismatching. This technology has been applied on several experimental prototypes used in medical field [6–8]. Moreover this alignment provides a better detection efficiency with respect to the use of a standard hexagonal hole collimator.

Many devices use advanced solutions to obtain a better intrinsic spatial resolution, although they are almost always limited to the use of a single PSPMT. The main problems that we have faced up are related to the construction of large devices for medical applications based upon an array of PSPMTs (multi-PSPMT gamma camera) [9]. One of the major limitations is surely constituted by the incapability to collect proper signal in the dead areas between adjacent photosensors.

When a crystal overlaps the dead area between two PSPMTs, each photosensor sees only a portion of the scintillation light emitted, part of the light falls in the dead area and cannot be collected. In this case there is ambiguity in the event location since the calculation of the scintillation position is not accurate, consequently there is a limitation, in terms of spatial resolution for the pixel identification within the area between photomultipliers. In this sense, the worst situation is at the junction of four PSPMTs [10] where their rounded corners greatly reduce the active area available. The presence of dead zones is related to the limits imposed by the construction techniques of the PSPMTs, in fact, all models of photomultiplier have overall dimensions that exceed the active area. The dead zone between PSPMTs (typically 2–4 mm) determines the minimum crystal size that can be used. Therefore it restricts the intrinsic spatial resolution.

Many authors [11–21] propose methods to spread the scintillation light to the active areas of the PSPMT next to dead zones. For this purpose a light guide is often placed between the scintillator

* Correspondence to: Institute of Biomedical Engineering, CNR, Via Salaria Km 29.300, C.P. 00015 Monterotondo (Roma), Italy. Tel.: +39 6 90672867; fax: +39 6 90672692.
E-mail address: soluri@isib.cnr.it (A. Soluri).

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and PSPMT, but the use of light guides introduces a degradation of the pixels resolution and the energy resolution.

The more is widened the spatial distribution of light in the light guide, the more the charge distribution involves a larger number of anodes; consequently the signal collected on each anode is lower as it is more extended the Scintillation Light Distribution (SLD). If the scintillation light is weak, the charge signals collected on some anodes may be under the limit of detection. So, when a crystal overlaps a dead zone, the absence of charge signal may strongly affect the position reconstruction.

Consequently, there are some limitations on the technical solutions for identifying all pixels of a scintillation matrix.

Approximately, in the area between PSPMTs the value of the intrinsic resolution obtainable cannot be less than the size of the dead zone. So, if the crystal size is much smaller than the dead area, the intrinsic resolution can vary dramatically between the centers and the peripheral zones of the PSPMT. Many works highlight their performances only in the active area of the PSPMTs, while the overall performances of a device should be evaluated.

For clinical use, the gamma camera intrinsic resolution is assumed to be constant across the field of view (FOV). The variation of the intrinsic spatial resolution could lead to a degradation of the scintigraphic images and might be needed appropriate correction algorithms in order to account for this effect [14–16]. These correction algorithms may require multi-anode readout electronics [13,17–20] that is more complex with respect to the resistive chain proposed. Therefore, in the design of a gamma camera, useful in clinical use, technical solutions that ensure small variations in the intrinsic spatial resolution should be preferred.

Although the PSPMTs have been used in the last years to assemble several devices, in order to achieve large area gamma cameras, few of such devices have been adopted into clinical applications due to the problems stated above. One of these is the Dilon 6800 Gamma Camera, that contains an array of NaI(Tl) crystals (3×3 mm pixel size) coupled to an array of 1 in.^{2} PSPMTs and has an intrinsic resolution of 3.3–4.7 mm [22]. The intrinsic resolution of this system is absolutely comparable with the value of 3.3 mm claimed by the Philips Brightview [23], a commercial gamma camera that consists of a large planar NaI(Tl) crystal optically coupled to an array of standard photomultiplier tubes (PMT) [24,25]. Therefore the development of large area gamma cameras with an array of PSPMTs and a light sharing, does not show clear advantages, also considering the higher cost per area required with respect to the standard technology.

Our aim is to develop a gamma camera by decoupling the PSPMTs, i.e. each photomultiplier becomes a detector module, so a large area can be produced connecting several independent modules obtaining a scalable system. In this modular approach each component is designed to optimize the performances in terms of intrinsic spatial resolution and its uniformity. This, evidently, has a direct bearing on the overall system. In details the size of each module corresponds to the PSPMT overall size, while the size of the crystals is chosen to make possible the identification of all the pixels. Consequently the choice of components greatly affects the performances.

Moreover each module can have its own dedicated electronics to calculate the position of the detected event. This last solution represents an innovation [26] with respect to those with single channel acquisition and those in which the channels are connected together by a resistive chain.

The FOV of the detector is so divided into independent zones that can acquire events in parallel mode, in that way the counting rate capability is greatly enhanced. This represents an innovation with respect to the current technology where the single event acquisition involves the full detector area.

In this work we have used and compared several scintillator arrays in order to select the one with the best pixel identification. These arrays have tungsten septa that separates two adjacent crystals [2]. This solution reduces the inter-crystal Compton scatter effect and eliminates the optical crosstalk, the image contrast is consequently enhanced [27–30].

Furthermore, the suppression of crosstalk and inter-crystal scattering is a necessary condition to make the detection modules independent, otherwise the light produced in the peripheral crystals would interfere in the adjacent modules.

The performances of the system, including the pixel identification, are strictly related to the light collection efficiency. Usually to improve the light collection efficiency the crystal surfaces that are not coupled with the PSPMT are coated with a white epoxy diffuse light reflector. In our case, as previously mentioned, the scintillation element is bordered by tungsten sheets. This allows us to use the surfaces of the septa to act as reflector in order to enhance the light collection. To accomplish this result we studied the deposition of several reflective materials on the tungsten sheets, even using different deposition techniques in order to have the best result in terms of reflection and stability.

Further, to improve the detection efficiency by increasing the active area of scintillator, we also have applied an innovative method that consists into the use of uncoated crystals within a tungsten structure whose septa are treated with a diffusive white deposit.

Moreover the light collection efficiency issue is particularly crucial on the borders of the PSPMT due to the presence of the dead areas. In order to improve the optical coupling, we have properly shaped the outlying matrix crystals so that their output face matches with the active area of the PSPMT (i.e. does not overlap the dead areas). Consequently the focusing of the scintillation light to the active area is definitely enhanced.

The proposed methods was employed to build several experimental setups in order to determine the best arrangements that optimize the module overall performances.

2. Equipment and method

Our work is based on the use of several Hamamatsu PSPMTs coupled to different scintillation assemblies. In particular we have built many matrices of crystals integrated in a tungsten grid. The mechanical accuracy required in the arrangement of the experimental devices is high: the positioning of the scintillation structures with respect to the surface of the PSPMTs critically affect the results. We solved this problem by using mechanical guides designed with the purpose of align the different PSPMTs to the scintillation arrays.

2.1. The PSPMTs

We have used three types of Hamamatsu PSPMTs (multi-strip PMTs as well as multi-anode PMTs), in combination with dedicated scintillation structures.

H8500 (Flat Panel) PSPMT has an external size of $52 \times 52 \times 28$ mm^3 with an active area of 49×49 mm^2. Its glass window thickness is 1.5 mm, the photocathode is made of bialkali and its multiplication system consists in 12 metal channel dynodes, so the PSPMT gain is about 3×10^6 at -1000 V. The multiplied charge is collected by an array of 8×8 anodes and the size of each anode is 6.08×6.08 mm^2 except the external ones with size of 6.28×6.28 mm^2.

Other measurements have been performed combining the dedicated scintillation structures to R11265-00-M64 Hamamatsu, which has an external size of $26.2 \times 26.2 \times 20.25$ mm^3 with an active area of 23.04×23.04 mm^2, its glass window thickness is