Physica Medica 48 (2018) 84–90

Contents lists available at ScienceDirect

Physica Medica

journal homepage: www.elsevier.com/locate/ejmp

Original paper

Novel light-guide-PMT geometries to reduce dead edges of a scintillation camera

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ARTICLE INFO

Keywords:
Gamma-ray detectors
Maximum likelihood estimation
Photomultiplier tube
Dead edge

ABSTRACT

Anger cameras based on monolithic NaI scintillators read out by an array of PMTs are predominant in planar gamma imaging and SPECT. However, position estimation of gamma interactions is usually severely degraded near the edges of the scintillator which can be extremely undesirable for applications like breast imaging. Here we propose a relatively cost-effective solution based on the use of scintillators with absorptive edges with an unconventional light-guide-PMT layout employing a maximum likelihood positioning algorithm. The basic design on which we aim to improve consists of a monolithic NaI(Tl) scintillator read out by 3×5 square PMTs (conventional layout, CL) that could be suitable for molecular breast imaging. To better detect gamma interactions near the crystal's critical edge, we tried different set-ups: we replaced the 5 large PMTs near the edge by 11 smaller PMTs (small-sensor layout, SSL); we emulated rectangular PMTs along the critical edge by inserting a row of 5 rectangular light-guides that direct the light toward square PMTs placed behind (shifted layout, SL); we inserted rectangular light-guides alternatingly, such that the PMTs are in an interlocking pattern (alternating shifted layout, ASL). The performance of our designs was tested with Monte Carlo simulations. Results showed that SSL, SL, and ASL gave better spatial resolution near the critical edge than CL (3.4, 3.6, and 4.1 mm near the edge compared with 5.3 mm for CL), and thus resulted in a larger usable detector area. To conclude, for applications where small dead edges are crucial, our designs may be cost-effective solutions.

1. Introduction

Gamma detectors that deliver information on the interaction position and energy of incoming gamma photons are key elements in nuclear medicine scanners. Both in planar scintigraphy and in SPECT, gamma detectors based on continuous NaI(Tl) scintillators that are read out by an array of photomultiplier tubes (PMTs) – usually referred to as the Anger camera – have been predominant for decades. In Anger cameras, the gamma photon's interaction position and its energy are conventionally estimated using Anger logic [1], which is based on calculating the centroid of the PMT outputs. Anger logic has become popular because it can be simply implemented with a resistor/capacitor network and Anger logic combined with heuristic linearity and non-uniformity corrections provides satisfactory position and energy estimation results in most applications. Unfortunately, the positioning linearity and spatial resolution are usually poor near the scintillator's edges, a situation often referred to as the dead edge effect. This effect has implications for the usable field-of-view of a gamma camera which is smaller than the scintillator's surface.

Although reducing dead edges is almost always profitable to enhance the usable detector surface and thus the system's sensitivity, in whole-body SPECT the presence of dead edges is usually accepted because with the large-area detectors that are commonly applied, the size of the dead edges is relatively small and because not using the detector's edges does not have to lead to image artefacts. However, in other applications, the use of the detector's edges can be absolutely necessary in order to arrive at useful images. Examples of this include planar breast imaging [2,3] and a dedicated multi-pinhole molecular breast tomosynthesis (MP-MBT) technique proposed in our group [4,5]. In the proposed MP-MBT scanner, a woman is lying prone on a patient bed with her breast pendant in a hole in the bed. The breast is mildly compressed and two gamma cameras are placed on either side of the breast close to the chest wall. In simulations, such a design resulted in a tumour-to-background contrast-to-noise ratio 2 – 3 times higher than commercial planar scanners. The edge area of the detector in this design is used to image the part of the breast close to the chest wall. However, in conventional Anger cameras, the dead edge roughly equals the PMT radius and as most common PMTs are two or three inches in diameter,

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https://doi.org/10.1016/j.ejmp.2018.04.004
Received 17 February 2018; Accepted 4 April 2018

about 25 mm or 40 mm at the edges would be unusable if we would employ a standard Anger camera in MP-MBT. Therefore, a detector with small dead edge is essential for MP-MBT.

To improve positioning linearity near the edges, several solutions have been proposed over the years. In some cases, PMTs were extended over the edges of the scintillator both for continuous crystals [6] and pixelated or semi-pixelated scintillators [7–9]. However, in MP-MBT there is no room for such a placement of PMTs since the scintillator extends till the patient bed. Another option is to read out the continuous crystals with smaller light sensors, including position-sensitive PMTs [10–12], avalanche photodiodes [13,14], silicon photomultipliers [15,16], charge-coupled devices [17], or to use a combination of pixelated scintillators and these small light sensors [18–22]. However, using small light sensors instead of PMTs for large surface gamma detectors (such as in MP-MBT, $240 \times 140\,\text{mm}^2$ area) leads to enormously increased costs. A third option is to use semiconductor gamma detectors instead of scintillation-based detectors. These detectors transfer gamma energy directly into an electrical signal and are already applied in several dedicated breast scanners [23,24]. Besides being able to reduce dead edges, semiconductor detectors improve energy resolution over scintillator detectors, although several studies have shown that the benefit of this in dedicated breast scanners is limited [25–28]. However, like small light sensors, the use of semiconductor detectors significantly increases the gamma camera's costs over those of the Anger camera.

Besides using new detector materials or advanced light sensors, several algorithms to better decode the scintillation position from the light distribution in PMT-read out scintillators have been proposed, e.g. maximum likelihood estimation [29,30], chi-squared error estimation [31], the *k*-nearest-neighbour method [16], a Gaussian filter algorithm [17], advanced light model fitting [11,14], and different machine learning algorithms [13]. These decoding processes are more sophisticated and also more computationally demanding than weighted averaging, as is done in Anger logic, but they have been proven to be more effective in resolving scintillations near the edges. These algorithms are often used together with black-edge detectors which use absorbing material at the sides of the scintillator [6,7,10,11,29]. Such absorptive edges increase the position dependence of the light spread near the edges, and thus improve position estimation in these areas.

Inspired by several of the above-mentioned elements, the aim of this paper is to propose a novel gamma scintillation detector design that has a cost comparable to that of the Anger camera but has improved spatial resolution and positioning linearity near the edges. This is achieved by using smart light-guide-PMT geometries to emulate smaller light sensors near the edges and by using a black-edge scintillator combined with a maximum likelihood (ML) positioning algorithm. PMTs used have a square shape in order to optimally cover the rectangular scintillator. Different designs are evaluated using Monte Carlo simulations.

2. Methods

2.1. Gamma detector designs

Detector dimensions are chosen such that they are suitable for the MP-MBT scanner proposed in our group [4,5] which has a minimum requirement for the active detector area of $240 \times 140\,\text{mm}^2$ and the scintillator thickness is 9.5 mm. We test four different designs in a simulation study which all fulfil the minimum dimension requirement.

The first design (Fig. 1(a)), which is the most basic (therefore dubbed 'conventional layout', CL) comprises a $240 \times 180 \times 9.5\,\text{mm}^3$ NaI(Tl) scintillator, a 14 mm thick glass light-guide, and 15 Hamamatsu R6236 PMTs ($60 \times 60\,\text{mm}^2$ square PMTs with $54 \times 54\,\text{mm}^2$ photocathodes) [32]. The entrance surface of the scintillator is painted white (reflective) while the edges are black (absorptive). As a comparison, in Section 3.1, we will also show some results for the same design but with a white-edge scintillator. Note that in our design, PMTs placed at the right and left sides of the gamma detector partly extend over the edges. In this way, the left and right edges are effectively read out by half-sized PMTs which is expected to improve resolution and linearity in these edge areas [6]. However, at the upper edge which is assumed to be the critical edge of the detector, such an approach is not feasible as there is no space to allow for this (this is the edge placed close to the patient's chest wall).

An alternative to CL could be the use of smaller PMTs, e.g. Hamamatsu R1548-07 ($24 \times 24\,\text{mm}^2$ square PMTs with $(2\times)$ $8 \times 18\,\text{mm}^2$ photocathodes [32]), which is the second design tested ('small-sensor layout', SSL; Fig. 1(b)). In that case, 21 PMTs would be needed to cover the 240 mm long upper edge. As the price per PMT is approximately constant, the costs for PMTs would increase by 40% while the scintillator size would be reduced to $240 \times 144\,\text{mm}^2$. In principle, smaller PMT sizes are only required in the direction perpendicular to the edge and one would thus like to use rectangular PMTs if these would be commercially available for the same price. As this is not the case, we propose an alternative design: the 'shifted layout' (SL, Fig. 1(c)). In this layout, an additional light-guide, with a cross-section that is half the PMT area, is inserted in between the original light-guide and each of the upper row PMTs. The additional light-guides are covered by Lambertian reflectors like Polytetrafluoroethylene (PTFE) with 98% reflectivity [33]. The length of the additional light-guide is assumed to be 160 mm, longer than the length of the PMTs (123 mm). The light-guide is assumed to be borosilicate crown glass. A variant on SL is the 'alternating shifted layout' (ASL, Fig. 1(d)), in which the additional light-guides still exist, but the PMTs are placed in an interlocking layout instead of in a conventional grid. Because the second light-guide is half as wide as the PMT front face, the scintillator sizes for SL and ASL are both $240 \times 150\,\text{mm}^2$. We come back to this reduced area in the discussion section.

2.2. Simulations

The performance of our gamma detector designs is assessed by the well-validated Monte Carlo simulation software GEANT4 Application for Tomographic Emission (GATE) [34–36]. The optical surface parameters in GATE are tuned in such a way that the simulator gives the best agreement with our available clinical Anger camera with 3 inch round PMTs. Here we simulate square PMTs, and we assume the light propagation in the new setups remains valid. In Table I. the relevant parameters used in the GATE simulations are listed.

The refractive index of the white reflector was set to 1.0 which reflects the presence of an air gap between the white reflector and the scintillator/light-guide. Furthermore, low reflectivity as reported in [39,40] is assumed which is representative for high-quality black edges because it has been reported that the quality of the black absorber is crucial in the black-edge scintillation camera performance.

To assess spatial resolution and positioning linearity, NEMA suggests to put lead masks with thin parallel slits on the gamma detector and irradiate them with gamma rays from a source placed at a relatively large distance above the detector to approximate parallel rays perpendicularly directed towards the detector surface [41]. In this way, the line response function (LRF) from each slit is obtained, and from these LRFs, positioning linearity and spatial resolution in horizontal and vertical directions are measured. In GATE this measurement is simulated by irradiating the gamma detector with vertical and horizontal line sources of 140 keV gamma photons (energy of ^{99m}Tc gamma emission). Gamma emitters are evenly distributed in the infinitely thin lines (as is shown in the solid black lines in Fig. 2) and all gamma photons are emitted perpendicular to the detectors. The interval between two neighbouring lines is 10 mm, and the outer horizontal and vertical lines are all 2 mm from the edges of the scintillator.

To obtain the light collection map and linearity correction map for Anger logic estimation (further discussed in Sections 2.3 and 2.4), the gamma detectors are also irradiated by point sources of 140 keV gamma photons, and the point response functions are determined. From each