INFLUENCE OF THE SCINTILLATION CRYSTAL OPTION ON THE DETECTOR RESPONSE OF PET DEVICES

by

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Positron emission tomography is a technology that provides unique and exquisite possibilities in functional diagnostics, in the sense that it is the most efficient and most reliable method for obtaining information about biochemical activity and cellular metabolism in the body, by determining exact localization and performing semi-quantitative assessment of the distribution of a radioactive tracer. This paper compares the characteristics of recently introduced lutetium based crystals to those of conventionally used bismuth-ortho-germanate scintillators; both options are used as scintillation detectors within the positron emission tomography systems. Energy resolution and scintillation decay time of lutetium based crystals and bismuth-ortho-germanate crystals was experimentally tested. Main scintillation detector parameters which affect the resulting detector response are considered and analyzed, since they serve as the basis for a positron emission tomography medical image.

Key words: positron emission tomography, scintillation crystal, detector response

INTRODUCTION

Positron emission tomography (PET) is a technology that provides unique and exquisite possibilities in functional diagnostics, primarily of oncological, but also neurological, coronary, infective and inflammatory diseases, in the sense that it is the most efficient and most reliable method for obtaining information about biochemical activity, cellular metabolism, and potential physiopatological processes in the body, by determining exact localization and performing semi-quantitative assessment of the distribution of a radioactive tracer. PET tracers contain short-lived β^+ emitters, the energy of which gets degraded through Coulombic interactions, from initial energies (in the range from 0.63 MeV (^{18}F) to 3.35 MeV (^{82}Rb)) to 511 keV, when positrons "capture" electrons and undergo positron-electron annihilation, which produces pairs of gamma photons. The two photons in an annihilation gamma pair are emitted at an angle of ~180° (180 0.5°, due to residual positron momentum). Coincident detection of these photons, within a limited time window, by scintillation detectors placed opposite one another, along the line of response (LOR), represents the basic diagnostic information. Modern detection systems are conceptually built up of 140-300 blocks, with over 30000 individual crystals, in a ring-shaped gantry. During a diagnostic procedure, the patient is positioned on the patient's table, which is moved in steps of certain length (the so called bed positions), and data is acquired during a preset time interval, from the region of interest (ROI) which is within detector's field of view (FOV). Distribution of radioactivity is reconstructed from several hundred million basic data. In addition to true coincidental events, there are the undesired events that need to be suppressed: scatter coincidences (when one or both annihilation γ-photons are scattered before detection) and random coincidences (when two photons from two different events are detected within the coincidence resolving time). At the macro level, consequences of detector intrinsic and geometrical imperfections need to be reduced (*e. g.*, intercrystal scattering, parallax effect, *etc.*), along with the impact of any patient movement during the imaging procedure.

The main drawback of a basic NaI(Tl) scintillation detector for use in PET diagnostics is the low detection efficiency of γ-rays with energies higher than 200 keV, due mainly to low density and low effective atomic number *Z*. For this reason, bismuth ortho-germanate – $Bi_4Ge_3O_{12}$ crystal (BGO) was initially chosen for PET detectors, back in the early 1970. BGO has a good response in the part of the energy spectrum of PET emitters. The importance of this di-

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agnostic method drives the technological progress forward, which results in new detector crystals being developed and implemented. Commercial successes on the road to an "ideal" scintillator have been based on lutetium, with cerium (Ce) as an activator. There are presently two commercial versions – LSO (lutetium oxyothosilicate, Lu_2SiO_5:Ce) and LYSO (lutetium-yttrium oxyorthosilicate, $Lu_{0.6}Y_{1.4}Si_{0.5}$:Ce) [1, 2].

This paper compares the characteristics of LYSO crystals to those of BGO scintillators, focusing on particular physical and operating conditions that exist during imaging procedures, as well as on the explicit effect on the resulting detector response, which serves as the basis for a PET medical image. The accent is on the roles and importance of all relevant crystal features, which provides guidance to the possibilities of further development of PET scintillation crystals and consequently PET detection.

PET SCINTILLATOR PARAMETERS

A scintillator crystal has to be mechanically hard and robust, non-hygroscopic, thermally stable, stable in color and transparency and radiation hard. From the perspective of detection, its most significant properties are sensitivity, energy resolution, light yield, scintillation decay time and melting point [2].

Sensitivity is expressed as the fraction of incident photons that cause scintillations. Density ρ [gcm^{-3}] and effective atomic number Z_{eff} of the scintillator are responsible for radiation absorption and photoelectric effect probability, which results in better attenuation and energy discrimination.

Higher energy resolution (ER) means more refined differentiation height of registered pulses, *i. e*., better function of the response of the detector at particular radiation energy. This means that the system can discriminate better between true and scatter events, and that narrower energy windows can be set, which results in lower noise in the image. ER mostly depends on light yield and intrinsic energy resolution.

Light yield is the conversion efficiency of radiation energy into light (photon/MeV). Higher light yield makes the value of energy deduced from the measured signal more precise (*i. e*., measurement uncertainty of incident PET photon energy is lower [3, 4]).

Intrinsic energy resolution of the crystal is related to crystal's dimensions and how the crystal has been cut, how this has been performed, and how light guidance has been implemented for each detector unit. Smaller detector crystals enable a larger acquisition matrix to be used, and therefore provide a better spectral resolution. On the other hand, if detectors are too small, inter-crystal scattering decreases the precision of determining event positions or it could lead to parallax effect: when an incident photon interact with detector element at the angle which is not perpendicular, the photon can penetrate and it could be detected by the another detector element [5]. Intrinsic energy resolution also depends both on the inhomogeneity presence or impurities in the crystal structures.

Scintillation decay time t_D (DT) is defined as the time needed for the pulse amplitude to decrease to the fraction of $1/e$ of its maximum initial value. DT corresponds to the interaction of a quantum of radiation with an atom in the detector material, which brings the atom into an excited energy state, from which it subsequently relaxes to the ground state by emitting visible light. A shorter current pulse is, naturally, desired, because it permits shorter coincidence timing (mitigates the influence of random effects, which makes the noise lower), enhances detector efficiency at higher photon fluence [6].

At high photon energies, electromagnetic showers appear when radiation goes through matter. The Moliere radius is a radius of a cylinder that on average contains at least 90% of electromagnetic shower's energy deposition. It depends linearly on radiation wavelength, as well as on the atomic number [7]. A smaller Moliere radius means better shower resolution and better shower separation, due to less overlapping.

Physical axial FOV represents the size of the body segment scanned within one time frame. With a larger FOV, a region being scanned can be segmented into smaller parts, which then makes the scan time shorter.

METHODS AND MATERIALS

Comparison of BGO and LYSO crystal parameters has been performed through the experimental analysis of the crystal energy resolution and scintillation decay time.

Energy spectrum (pulse height spectrum) has been obtained by the single light pulses integration so the energy resolution is obtained according to the well known equation

$$ER \quad \frac{\Delta E_{\text{FWHM}}}{E_{\max}} \; 100\% \qquad (1)$$

where ΔE_{FWHM} is full width at half maximum of the energy $E_{\max}$ [1, 2].

In order to measure scintillation decay time the methodology based on the principle presented by Bollinger and Thomas [8] has been adopted, since the method (also known as delayed coincidence method) estimates the real shape of the light pulse, without the signal distortion caused by the signal traversing the photomultiplier tube.

The measuring equipment consisted of:

(1) 10 BGO and 8 LYSO scintillation crystals (all with same dimensions of 5 mm 5 mm 20 mm),
(2) photodetectors, with an internal high voltage circuit,

(3) external voltage supply of 15 V,
(4) avalanche photodiodes,
(5) radioactive source ^{137}Cs with activity of 160 kBq, and
(6) radioactive source ^{22}Na with activity of 40 kBq.

In order to detect as much scintillation light as possible the crystals were wrapped in a teflon tape on all sides except the one coupled with the photodetectors. Optical grease (with refractive index of 1.465) was used in order to couple the crystal to photodetector.

Radioactive source ^{137}Cs was used since it emits gamma photons with an energy of 662 keV which is close to the photon energy of 511 keV detected in PET systems. Radioactive source ^{22}Na undergoes the beta plus decay, so it emits photons of 511 keV from the positron-electron annihilation. This source also emits photons of 1275 keV. It was used for the measurements of energy spectrum linearity.

RESULTS

Experimental results obtained from measurements of BGO crystals energy resolution performed without optical grease are shown in tab. 1. Energy spectrum of BGO crystals is shown in fig. 1. Without optical grease, the energy resolution was in the range of 20.1-22.1%, with mean value and corresponding standard deviation of 21.2 0.6%. Energy resolution measured for LYSO crystals was in the range of 12.9-15.3% with mean value and corresponding standard deviation of 13.6 0.1%, as it is shown in tab. 2.

Table 1. Energy resolution of BGO crystals obtained without optical grease

BGO crystal no.	*ER* [%]	BGO crystal no.	*ER* [%]
1	21.5 0.1	6	21.3 0.1
2	21.8 0.1	7	22.1 0.1
3	21.7 0.1	8	21.1 0.1
4	20.9 0.1	9	21.1 0.1
5	20.1 0.1	10	20.4 0.1

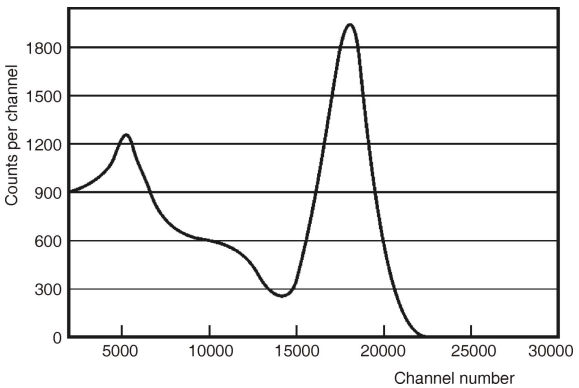

Figure 1. Energy spectrum of BGO crystals obtained without optical grease

Two randomly chosen BGO crystals were optically coupled to photodetector and measurements showed that the energy resolution was 16.5% and 16.3%. In case of LYSO crystals, the 4-th crystal shown in tab. 2 was chosen for energy resolution measurements with optical grease and the obtained value was 11.4%. Figure 2 shows the energy spectrum of LYSO crystals both with and without optical grease. All measurements were based on the channels which correspond to the energy of 662 keV (^{137}Cs). Standard deviation was less than 1% for BGO and less then 2% for LYSO crystals, for the energy peaks of 511 keV and 1275 keV, in respect to energy of 662 keV.

Experimental results of scintillation decay time obtained for BGO crystals are shown in tab. 3. Obtained mean value and corresponding standard deviation was 301 3 ns. Figure 3 shows scintillation decay time obtained by the experiment along with the exponential fit of the obtained results. Starting point for the results fitting was 100 ns while the ending point was 600 ns. Scintillation decay time of BGO crystal has biexponential behavior, *i. e.*, there is one fast compo-

Table 2. Energy resolution of LYSO crystals obtained without optical grease

LYSO crystal no.	*ER* [%]	BGO crystal no.	*ER* [%]
1	13.2 0.1	5	13.1 0.1
2	15.0 0.1	6	13.3 0.1
3	15.3 0.1	7	12.8 0.1
4	12.9 0.1	8	13.2 0.1

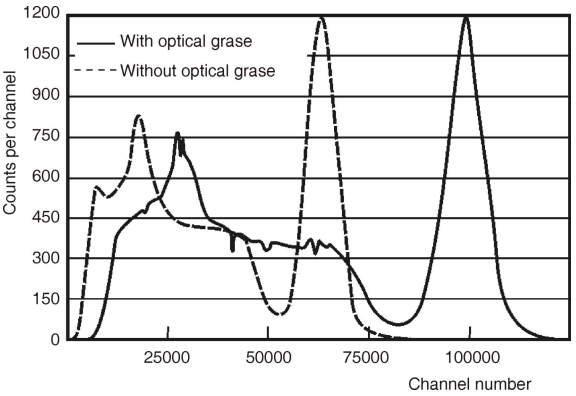

Figure 2. Energy spectrum of LYSO crystals, obtained without optical grease and with optical grease

Table 3. Scintillation decay time for BGO crystals

BGO crystal no.	*DT* [ns]	BGO crystal no.	*DT* [ns]
1	303 3	6	301 3
2	307 3	7	298 3
3	303 3	8	318 3
4	300 3	9	309 3
5	301 3	10	311 3

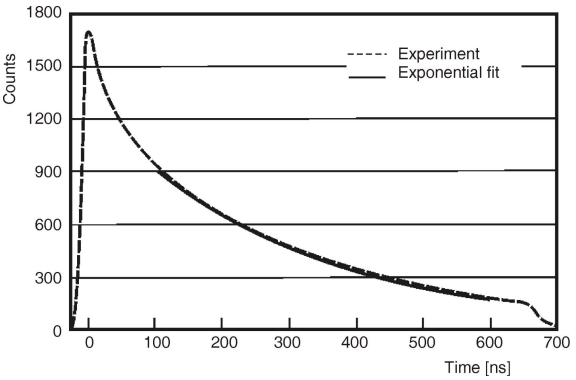

Figure 3. Scintillation decay time curves for BGO crystal

Table 4. Scintillation decay time for LYSO crystals

LYSO crystal no.	*DT* [ns]	BGO crystal no.	*DT* [ns]
1	48.0 0.2	5	47.1 0.2
2	47.7 0.2	6	49.7 0.2
3	49.5 0.2	7	47.4 0.2
4	46.4 0.2	8	48.2 0.2

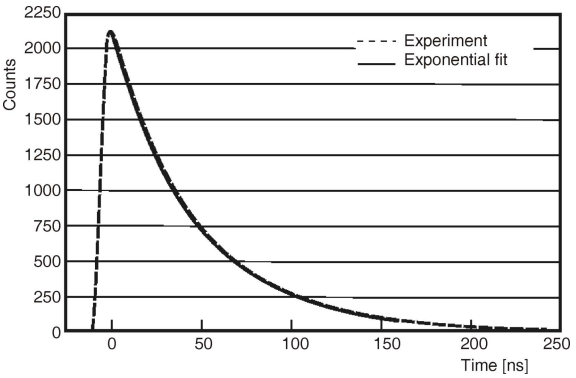

Figure 4. Scintillation decay time curves for LYSO crystals

nent of the decay time with duration of about 60 ns and its value is reduced $e^{-5/3}$ times after 100 ns, so its influence can be neglected. This component contains 10% of overall light yield.

Obtained mean value with corresponding standard deviation for LYSO crystals was 48.0 0.2 ns, according to the results shown in tab. 4. Scintillation decay time curve is shown in fig. 4. Since the LYSO crystal has the only one component of decay time the starting point for exponential fit is not of great significance. Time used for the ending point during the results fitting was 160 ns.

DISCUSSIONS

Obtained results of measuring the energy resolution and scintillation decay time (considering measurement uncertainty [9-11]), showed that LYSO crystals took advantage over BGO crystals in both investigated domains. It is important to notice that in both investigated types of scintillation crystals the better energy resolution is achieved by using the optical grease. Experiments also showed a higher fluctuation of the energy resolution with LYSO crystals. This could be explained as a consequence of the optical transport [12, 13]. It is proved that BGO crystals have longer scintillation decay time.

In addition to experimentally tested parameters of scintillation crystals there are more important parameters to compare which are being considered and currently investigated in order to find the optimal scintillation crystal for PET applications.

BGO crystal is a pure inorganic crystal that needs no activators. Its most significant advantage over lutetium crystals is its much higher efficiency for 511 keV photons [14]. Detector efficiency depends not only on material properties, but also on the width of the crystal. Currently employed BGO crystals are typically 3 cm wide, while LSO alternatives are 2 cm in width. Based on the ratio of corresponding efficiencies (90% : 65%), obtained at 511 keV photon energy and for specific activity of 37 MBq per 10 kg of body mass, which is an average activity in today's practice, the efficiency is found to be 38% higher in the first variant [15].

According to a Monte Carlo study [5], BGO detector has a better detection precision and lower inter-crystal scatter compared to lutetium alternatives. This is in accordance with the fact that its effective atomic number is almost 14% higher than that of LSO, which makes a higher probability for the photoelectric effect. As a result, the photoelectric (total absorption) cross-section at 511 keV is almost 60% higher in BGO crystals than in LSO alternatives [14].

Shortcomings of BGO crystals also include low light yield (about four times lower than that of lutetium crystals), larger physical axial FOV and more stringent environmental conditions (fluorescent intensity). The Moliere radius is also lower for both lutetium detector variants than for BGO [16].

Uneven sensitivity along detector's profile, caused by lower sensitivity around crystal edges is also considered. Regarding this issue, a minimum axial overlap of FOV with the next bed position is defined, with the effective axial FOV determined as the physical axial FOV minus the overlap. Smaller overlap is presently achieved in BGO detector systems [17].

Investigations are also directed to the finding the optimal scintillation detector size as well as its technical processing so that a smaller cross-sectional area is achieved, and thereby a larger acquisition matrix and better spatial resolution. Special attention is devoted to the manufacturing, growth, finishing and polishing of crystals and implementation of special reflective materials between crystal elements [2].

Alternative lutetium-based solutions have now reached experimental phase (as $Lu_{1.8}Gd_{0.2}SiO_5$:Ce, LGSO) as well as crystals based on other elements, especially $LaBr_3$ (with 5% Ce) [18] and Ce: $Gd_3Al_2Ga_3O_{12}$(Ce:GAGG) [19]. In addition to the development of detector block geometry (including investigations which combine various crystals within blocks, such as the GSO/LSO/BGO/CsI(Tl) structure with layers of scintillators with different decay times and dual-sided readout [19]), much work is dedicated to elimination of collection variability around detector edges, detector system design (including ways of enlarging the FOV size), implementation and improvement of silicon photomultipliers [18, 20] and other technical enhancements that contribute to the development of PET.

In comparison to alternative commercially available variants with lutetium crystals, BGO PET devices are much cheaper, especially when all the optionally offered assets of lutetium crystals are implemented, such as the hardware upgrade of the basic version which enhances the FOV and shortens scan time per bed position.

Lutetium carries the issue of natural radioactivity, with ^{176}Lu isotope (2.6% natural abundance) being radioactive (half-life of (3.56 0.07) 10^{10} years). The decay of ^{176}Lu emits β^- particles and γ-rays in the energy range from 88 keV to 400 keV, which do not present a problem for standard PET imaging, but affect low count rate measurements conducted in specific QC procedures for attenuation correction, when Ge-68 phantoms are used, with activities of the order of 5 kBq [21-23].

CONCLUSIONS

Experiments showed that LYSO crystals took advantage over BGO crystals in both investigated parameters, especially in decay time, which is about 5 times less with LYSO, compared both with and without optical grease. Moreover, 4 times higher light yield of LYSO crystals over BGO provides much higher potential in rejecting the spurious events. Energy resolution and scintillator decay time directly affect the noise equivalent count rate and PET image noise.

From the other hand, detector sensitivity is one of the main detector characteristics, since good image quality requires as many photons to be detected as possible, and, furthermore, higher sensitivity makes the imaging procedure quicker and the administered activity lower. In commercially available devices, maximum PET system sensitivity of 10 cps/kBq has been achieved with BGO crystals (and through optimization of detectors and of the front-end electronic design).

Work with standard radionuclides (^{18}F or ^{11}C) and administered activities, with modern smart software solutions and smaller basic voxel cells for reconstruction make arguments in favor of lutetium crystals not convincing enough to discard cheaper and reliable BGO PET detector devices for routine PET imaging.

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AUTHOR CONTRIBUTIONS

Theoretical analysis was carried out by V. M. Antić and P. V. Osmokrović. All of the authors have performed the analyses and discussion. The manuscript was written by V. M. Antić and K. Dj. Stanković. The figures and tables were prepared by V. M. Antić.

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УТИЦАЈ ИЗБОРА СЦИНТИЛАЦИОНОГ КРИСТАЛА НА ОДЗИВ ДЕТЕКТОРА ПЕТ УРЕЂАЈА

Позитронска емисиона томографија је технологија која пружа јединствене и изузетне могућности у функционалној дијагностици. Наиме, помоћу до сада најефикасније и најпоузданије методе долази се до информације о биохемијској активности и ћелијском метаболизму у телу, одређивањем тачне локализације и вршењем семиквантитативне процене дистрибуције радиоактивне супстанције. У овом раду упоређене су карактеристике недавно уведених кристала на бази лутецијума са карактеристикама конвенционалних бизмут-орто-германат сцинтилатора; обе варијанте се користе као сцинтилациони детектори код ПЕТ уређаја. Експериментално је тестирана енергетска резолуција и време гашења светлосног импулса у кристалу за LYSO и BGO сцинтилаторе. Разматрани су и анализирани главни параметри сцинтилационих детектора који утичу на резултујући одзив детектора, који служе као основа за ПЕТ медицинску слику.

Кључне речи: позитронска емисиона томографија, сцинтилациони кристал, одзив детектора