Available online at www.sciencedirect.com
ScienceDirect
Nuclear Instruments and Methods in Physics Research A 569 (2006) 201–204
NUCLEAR INSTRUMENTS & METHODS IN PHYSICS RESEARCH Section A
www.elsevier.com/locate/nima

Evaluation of the light emission efficiency of LYSO:Ce scintillator under X-ray excitation for possible applications in medical imaging

Ioannis G. Valais[a,b], Ioannis S. Kandarakis[b], Anastasios Konstantinidis[b], Dimitrios N. Nikolopoulos[b], Ioannis Sianoudis[c], Dionisis A. Cavouras[b], Nikolaos Dimitropoulos[d], Constantinos D. Nomicos[e], George S. Panayiotakis[a,*]

[a]*Department of Medical Physics, Medical School, University of Patras, 265 00 Patras, Greece*
[b]*Department of Medical Instruments Technology, Technological Educational Institution of Athens, Ag. Spyridonos, Aigaleo, 122 10 Athens, Greece*
[c]*Department of Physics, Chemistry and Materials Technology, Technological Educational Institution of Athens, Ag. Spyridonos, Aigaleo, 122 10 Athens, Greece*
[d]*Department of Medical Imaging, "Euromedica" Medical Center, 2 Mesogeion Avenue, Athens, Greece*
[e]*Department of Electronics, Technological Educational Institution of Athens, Ag. Spyridonos, Aigaleo, 122 10 Athens, Greece*

Available online 5 September 2006

Abstract

Lutetium–yttrium-based scintillators, such as LYSO:Ce, have a high effective atomic number, are non-hygroscopic, fast emitting materials, and promising candidates for use in positron emission imagers. The present study investigates the light emission characteristics of (Lu, Y)$_2$SiO$_5$:Ce (LYSO:Ce) single-crystal scintillator under X-ray imaging conditions. Also, the parameters related to the luminescence emission spectrum and emission efficiency were studied using experimental methods. Various X-ray tube voltages currently employed in X-ray imaging techniques were used. Measurements were performed using an experimental set-up based on a photomultiplier coupled to an integration sphere. In addition, the emission spectrum under UV and X-ray excitation was measured using an optical grating monochromator to determine the spectral compatibility of optical photon detectors incorporated in medical imaging systems. The absolute efficiency of LYSO:Ce was found to increase with increasing X-ray tube voltage (from 2.2 EU at 22 kVp to 22.4 EU at 140 kVp), while its spectrum, peaking at about 430 nm, was found compatible with most optical detectors (photodiodes, photocathodes, charge coupled devices, etc.). The matching factor was estimated to range from 0.76 to 0.92 (for a silicon photodiode and for a GaAsP photocathode, respectively).

PACS: 07.85.Fv; 29.40.Mc

Keywords: Inorganic scintillators; LYSO:Ce; Luminescence efficiency; Radiation detectors; Matching factor

1. Introduction

In most X- and γ-ray medical imaging modalities, inorganic scintillators are employed as radiation in light converters [1–3]. Cerium-doped (Ce^{3+}) scintillators have a much faster response than most currently employed materials (e.g. thallium-activated sodium iodide and cesium iodide scintillators, terbium- or praseodymium-activated gadolinium oxysulfide rare-earth materials, etc.—Table 1) [4,5].

Cerium-doped lutetium yttrium oxyorthosilicate, (Lu, Y)$_2$SiO$_5$:Ce or LYSO:Ce, is a promising next generation scintillation crystal (Table 1). LYSO:Ce is a mixed LSO/YSO (5–10% Y) non-hygroscopic crystal that offers high density (7.1 g/cm^3), high light output ($\geqslant$30000 ph/MeV), good energy resolution (~10%) and short decay time (40 ns)[6]. Although LYSO:Ce and LSO:Ce demonstrate similar behavior, as far as the decay scheme is concerned [7], LYSO:Ce appears to be of approximately 20% higher light yield than LSO under low-energy (35 kV) X-ray excitation [8].

In the present study, we investigate the light emission characteristics of LYSO:Ce single-crystal scintillator under

*Corresponding author. Tel.: +30 2610 996113; fax: +30 2610 996113.
E-mail address: panayiot@upatras.gr (G.S. Panayiotakis).

0168-9002/$ - see front matter © 2006 Elsevier B.V. All rights reserved.
doi:10.1016/j.nima.2006.08.018

Table 1
Scintillators for medical imaging radiation detectors

Name	Density ($g\,cm^{-3}$)	ρZ_{eff}^4 (10^6)	Hygroscopicity	Light yield (photons/MeV)	Decay time (ns)	Emission maximum (nm)
NaI:Tl	3.67	24.5	Yes	41 000	230	410
CsI:Tl	4.51	38	Slight	66 000	800–6000	550
Gd_2O_2S:Tb	7.3	103	No	**60 000**[a]	1 × 106	545
Gd_2O_2S:Pr, Ce, F	7.3	103	No	35 000[a]	4 × 103	510
$YAlO_3$:Ce	5.5	7	No	21 000	30	350
$Y_3Al_5O_{12}$:Ce	4.6	39	No	16 700	80	530
Gd_2SiO_5:Ce	6.7	84	No	8 000	60	440
$Gd_3Ga_5O_{12}$:Ce	7.1	58	No	40 000[a]	1.4 × 105	730
Lu_2SiO_5:Ce	7.4	143	No	26 000	40	420
$Lu_{2(1-x)}Y_{2x}SiO_5$:Ce	7.1	63	No	< 30 000	40	420[b]

[a]Measured at ~60–80 keV; all others at 662 keV.
[b]Data are from Ref. [6]. Data are from Ref. [1].

X-ray irradiation. In addition, parameters related to the light emission efficiency (light yield) and the luminescence emission spectra were studied using experimental methods. The luminescence spectra and light yield of LYSO:Ce has been previously studied under low-energy X-rays (35 kV), γ rays and VUV excitation [7,8]. To our knowledge, LYSO:Ce has never been previously investigated under medical X-ray imaging conditions.

2. Materials and methods

2.1. Definitions

The emission efficiency (light yield) of scintillators may be evaluated by measuring the absolute luminescence efficiency (AE) [9–12]. AE has been defined as the ratio of the light energy flux ($\dot{\Psi}_\lambda$), emitted by an excited scintillator, over the incident exposure rate ($\dot{X}$):

$$\eta_A = \dot{\Psi}_\lambda / \dot{X} \tag{1}$$

where η_A is the absolute efficiency expressed in AE-units ($\mu W\,m^{-2}/mR\,s$). AE expresses light yield and scintillator sensitivity under X-ray medical imaging conditions, where detectors are operated in energy integration mode.

A major consideration is the spectrum of the emitted light and its compatibility to the spectral sensitivity of various optical photon detectors. Spectral compatibility may be estimated by the spectral matching factor (SMF), which has been defined by the ratio [13]

$$\alpha_S = \int S_P(\lambda) S_D(\lambda) / \int S_P(\lambda)\,d\lambda \tag{2}$$

where $S_P(\lambda)$ is the spectrum of the light emitted by the scintillator, $S_D(\lambda)$ is the spectral sensitivity of the optical photon detector and λ denotes the wavelength of the emitted light.

The efficiency corresponding to a specific scintillator–optical photon detector combination has been expressed by the effective efficiency (EE), given as the AE multiplied by the corresponding SMF [10,14].

2.2. Experiments

The LYSO:Ce crystal (chemical formula: $Lu_{2(1-x)}Y_{2x}SiO_5$, with 5–10% Y) used in this study was supplied by Photonic Materials Ltd., Scotland, UK, with dimensions of $10 \times 10 \times 10$ mm, doped with 0.5% mol of cerium (Ce^{+3}). The crystal was irradiated by X-rays using two X-ray machines: (i) A Philips Optimus general radiography X-ray unit (tungsten anode, 2 mm Al filter), operated at various X-ray tube voltages from 40 to 140 kV. This unit was used to simulate irradiation conditions employed in general-purpose computed tomography. (ii) A General Electric Senographe DMR X-ray mammographic unit with molybdenum anode and molybdenum filter (Mo/Mo). The filter changed automatically to rhodium (Rh) and aluminum (Al) filters as we moved from medium to higher mammographic voltages. X-ray tube voltage varied from 22 to 49 kV. This unit was used to simulate irradiation conditions for tomographic breast imaging. For measurements performed under X-ray mammographic conditions, the X-ray beam was filtered by a 30-mm-thick block of Perspex to simulate beam hardening in the human breast [15]. Similarly, a 20-mm-thick Al was employed in the general radiography unit to simulate beam hardening by the human body [16].

The AE was determined according to Eq. (1), by performing X-ray exposure and light flux measurements. The exposure rate was measured at the crystal's position using a Radcal 2026C dosimeter (Radcal Corp., USA). Light energy flux measurements were performed using a light integration sphere (Oriel 70451 [17]) coupled to a photomultiplier (EMI 9798B), equipped with an extended sensitivity S-20 photocathode and connected to a Cary 401 vibrating reed electrometer.

The SMF was determined using Eq. (2). The emitted light spectrum $S_P(\lambda)$ of LYSO:Ce crystal was measured using the Ocean Optics optical grating spectrometer (Ocean Optics Inc., S2000). Spectral sensitivity ($S_D(\lambda)$) data for various optical photon detectors were obtained from corresponding manufacturer's (Hamamatsu, EMI, etc.)

datasheets. Seven optical photon detectors currently used in a large variety of imaging detectors (digital and conventional radiography, fluoroscopy, computed tomography, nuclear medicine, etc.) and their SMF with LYSO:Ce, GSO:Ce, and YAP:Ce crystal spectra were examined (Table 2). These optical detectors were the following: (i) GaAs photocathode, (ii) extended S20 EMI photocathode with quartz window, (iii) GaAsP Hamamatsu photocathode, iv) a-Si:H/108H amorphous silicone photodiode corresponding to intrinsic layer thickness of 800 nm (108H), (v) Si/S1133 Hamamatsu crystalline silicone photodiode, (vi) Agfa Curix Ortho GS film, and (vii) CCD S100AB SITe®.

3. Results and discussion

Fig. 1 shows the variation of the AW of the LYSO:Ce crystal with X-ray tube voltage, under general X-ray imaging conditions (40–140 kV). The fitted curve shown in the figure was obtained by a logarithmic fitting. Fig. 2 shows the variation of the AE of LYSO:Ce in the mammographic X-ray tube voltage range (22–49 kV, Mo/Mo spectrum). As it may be seen, AE increases, in a non-proportional way [19], with increasing X-ray tube voltage. The fitted curve shown in the figure was obtained by a fifth-order polynomial fitting. An interesting point to notice in Fig. 2 is that, at 42 kV, a sharp increase was observed. This was attributed to the change of the mammography tube filter from Rh to Al. On the contrary, no significant variation was observed when the filter changed from Mo to Rh at lower energies (38 kV).

The shape of the absolute efficiency curves is significantly affected by the light transmission through the scintillator crystal. As the X-ray tube voltage increases, the X-ray beam penetrates deeper within the scintillator block and on average X-ray photons are absorbed at deeper points within the scintillator crystal. Thus, light is generated at points closer to the scintillator–optical detector interface, causing minimal light attenuation within the crystal. The same behavior was observed in a previous study of ours that involved the GSO:Ce crystal [18].

Fig. 3 shows the normalized spectral response of LYSO:Ce crystal compared to the spectral sensitivities of various optical detectors.

Fig. 4 shows the luminescence EE of LYSO:Ce single-crystal scintillator with some optical detectors. EE curves are quite close although the optical detectors used in our graph were of a different composition.

Highest EE was found for GaAs photocathode (SMF: 0.92635). High values were also found for most optical detectors examined in this study.

The AE value of LYSO:Ce (27 EU) was found high enough compared to that of GSO:Ce value (20 EU) [18,20],

Table 2
Spectral matching factor of LYSO:Ce, YAP:Ce and GSO:Ce scintillators with optical detectors

Optical detectors	LYSO:Ce	YAP:Ce	GSO:Ce
GaAs photocathode	0.92635	0.91141	0.92707
Extended S-20 Photocathode	0.90185	0.97971	0.90294
GaAsP Hamamatsu photocathode	0.66715	0.39677	0.65991
a-Si:H 108H photodiode	0.70491	0.44754	0.68984
Si/S1133 Hamamatsu photodiode	0.76025	0.50809	0.77425
CCD S100AB SITe®	0.88449	0.85977	0.87917
Agfa Curix Ortho GS film	0.85778	0.99280	0.62837

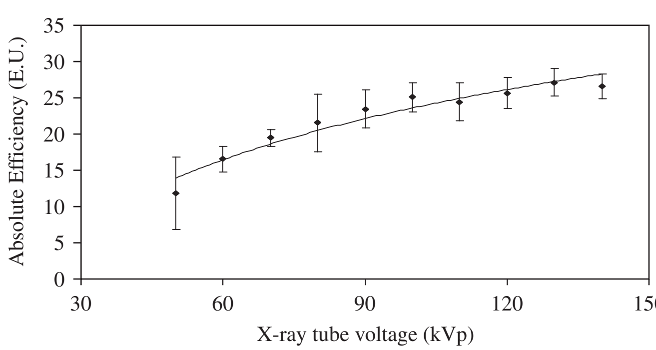

Fig. 1. Variation of absolute luminescence efficiency (AE) of LYSO:Ce crystal for radiographic X-ray tube voltages between 40 and 140 kV. AE units: $\mu W\,s/mR\,m^2$. Points: measured data, line: fitted curve.

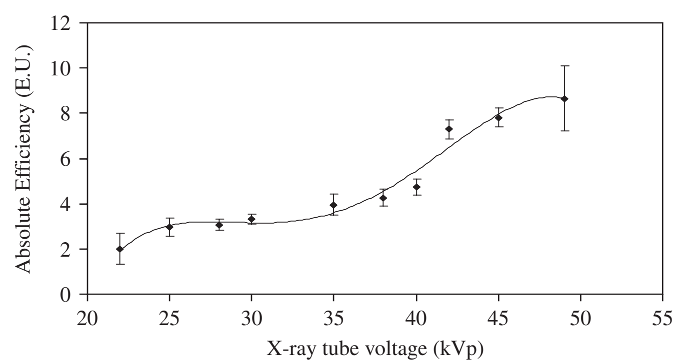

Fig. 2. Variation of absolute luminescence efficiency (AE) of LYSO:Ce crystal for mammography X-ray tube voltages between 22 and 49 kV. AE units: $\mu W\,s/mR\,m^2$. Points: measured data, line: fitted curve.

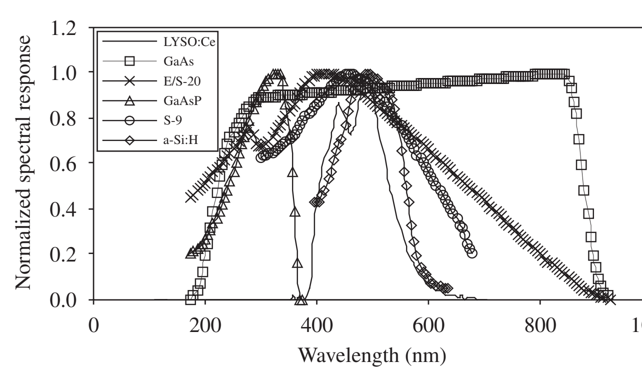

Fig. 3. Normalized spectral response of LYSO:Ce compared to the spectral sensitivities of GaAs, GaAsP, ES-20, S-9, and a-Si:H 108H photodetectors.

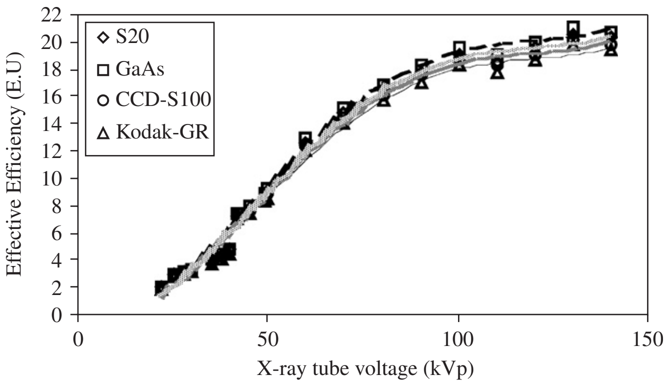

Fig. 4. Effective efficiency of LYSO:Ce crystal using S-20 EMI photocathode, GaAs photocathode, CCD S100AB photodiode, and Kodak X-omatic GR film.

which suggests that the former scintillator crystal could be useful in designing detectors for X-ray computed tomography as well as for the recently proposed breast computed tomography systems [21]. In addition, these data could be useful to obtain an estimation of the efficiency of the LYSO:Ce crystal under mono-energetic radiation used in ordinary nuclear medicine.

4. Conclusions

In conclusion, our measurements showed that the absolute efficiency of LYSO:Ce scintillator crystal increased with X-ray tube voltage, being adequately high for X-ray imaging applications. In addition, the LYSO:Ce's emission spectrum was found compatible to many currently employed optical photon detectors. LYSO:Ce may be considered for use in breast computed tomography and modern fast-image-producing X-ray computed tomography systems and imaging applications requiring relatively high X-ray tube voltages (e.g. computed tomography detectors, pixelated scintillation digital detectors, nuclear medicine imaging detectors, etc.).

Acknowledgments

This work was financially supported by the EPEAEK program ''Archimidis''. The authors gratefully acknowledge Photonic Materials Ltd., Scotland, UK, for supplying the LYSO crystal.

References

[1] C.W.E. van Eijk, Phys. Med. Biol. 47 (2002) R85.
[2] J.A. Mares, A. Beitlerova, M. Nikl, N. Solovieva, C. D'Ambrosio, K. Blazek, P. Maly, K. Nejezchleb, F. de Notaristefani, Radiat. Meas. 38 (2004) 353.
[3] V.V. Nagarkar, T.K. Gupta, S.R. Miller, Y. Klugerman, M.R. Squillante, G. Entine, IEEE Trans. Nucl. Sci. NS-45 (1998) 492.
[4] C.W.E. van Eijk, Nucl. Instr. and Meth. A. Phys. Res. 460 (2001) 1.
[5] P. Dorenbos, J.T.M. de Haas, C.W.E. van Eijk, C.L. Melcher, J.S. Schweitzer, IEEE Trans. Nucl. Sci. NS-41 (1995) 735.
[6] LYSO Product Specifications, Photonic Materials Ltd., Scotland, UK, Internet site: ⟨www.photonicmaterials.com⟩.
[7] L. Qin, H. Li, S. Lu, D. Ding, G. Ren, J. Crystal Growth 2–4 (2005) 518.
[8] A.J. Wojtowicz, W. Drozdowski, D. Wisniewski, J.L. Lefaucheur, Z. Galazka, Z. Gou, T. Lukasiewicz, J. Kisielewski, Opt. Mater. 28 (2006) 85.
[9] I. Kandarakis, D. Cavouras, E. Kanellopoulos, G.S. Panayiotakis, C.D. Nomicos, Nucl. Instr. and Meth. A. Phys. Res. 417 (1998) 86.
[10] I. Kandarakis, D. Cavouras, G.S. Panayiotakis, C.D. Nomicos, Phys. Med. Biol. 42 (1997) 1351.
[11] D. Cavouras, I. Kandarakis, P. Prassopoulos, E. Kanellopoulos, C.D. Nomicos, G.S. Panayiotakis, Technol. Health Care 7 (1999) 53.
[12] G.W. Ludwig, J. Electrochem. Soc. 118 (1971) 1152.
[13] I. Kandarakis, D. Cavouras, C.D. Nomicos, G.S. Panayiotakis, Nucl. Instr. and Meth. B. Phys. Res. 179 (2001) 215.
[14] D. Cavouras, I. Kandarakis, G.S. Panayiotakis, A. Bakas, D. Triantis, C.D. Nomicos, Br. J. Radiol. 71 (1998) 766.
[15] J.M. Boone, Med. Phys. 29 (5) (2002) 869.
[16] T. Holm, R. Moseley, Radiology (1964) 898.
[17] Oriel 70451 Integrating Sphere Data Sheet, Oriel Instruments, Available: ⟨http://www.spectra-physics.com/com/cda/products/⟩.
[18] I. Valais, I. Kandarakis, D. Nikolopoulos, I. Sianoudis, N. Dimitropoulos, D. Cavouras, C.D. Nomicos, G.S. Panayiotakis, IEEE Trans. Nucl. Sci. NS-52 (2005) 1830.
[19] J.D. Valentine, B.D. Rooney, P. Dorenbos, IEEE Trans. Nucl. Sci. NS-45 (1998) 1750 (Correspondence).
[20] I. Valais, D. Nikolopoulos, I. Kandarakis, H. Delis, N. Dimitropoulos, D. Cavouras, C.D. Nomicos, G.S. Panayiotakis, in: Proceedings of the 10th Mediteranian Conference on Medical Biology and Engineering, Ischia, Naples, Italy, 31 July–5 August, 2004 P568, IFMBE Proceedings, vol. 6, 2004, p. 84.
[21] B. Chen, R. Ning, Med. Phys. 29 (5) (2002) 755.