

Luminescence Properties of $(\text{Lu}, \text{Y})_2\text{SiO}_5 : \text{Ce}$ and $\text{Gd}_2\text{SiO}_5 : \text{Ce}$ Single Crystal Scintillators Under X-Ray Excitation for Use in Medical Imaging Systems

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Abstract—The luminescence of $(\text{Lu}, \text{Y})_2\text{SiO}_5 : \text{Ce}$ (LYSO:Ce) and $\text{Gd}_2\text{SiO}_5 : \text{Ce}$ (GSO:Ce) crystals was studied for use in tomographic medical x-ray imaging. LYSO:Ce and GSO:Ce are high density (7.1 g/cm^3 and 6.71 g/cm^3 respectively), high atomic number (71 for Lu and 64 for Gd), non-hygroscopic, and short decay time (40 ns and 60 ns respectively) scintillators. Evaluation was performed by determining: 1) the luminescence efficiency (LE) (emitted light energy flux over incident x-ray energy flux) in x-ray energies employed in general x-ray imaging (40–140 kVp) and in mammographic x-ray imaging (22–49 kVp), 2) the light emission spectrum, determined at various x-ray energies (22–140 kVp), and 3) the spectral compatibility to optical photon detectors, incorporated in medical imaging systems. Both scintillation materials exhibited adequately high LE in the x-ray diagnostic energy range, with LYSO:Ce's LE being distinctively higher. LYSO:Ce and GSO:Ce were found most compatible with the S-20 photocathode (0.9 for both materials) and adequately compatible to the amorphous silicon photodiode (0.74 for both materials), incorporated in many digital x-ray detectors.

Index Terms—Inorganic scintillators, luminescence efficiency, LYSO:Ce, matching factor, radiation detectors.

I. INTRODUCTION

INORGANIC scintillators coupled to optical photon detectors (photocathodes, photodiodes etc.) are used as detectors in most currently employed medical imaging systems [1]. The sensitivity and performance of imaging systems, such as Positron Emission Tomography (PET), Single Photon Emission

Computed Tomography (SPECT), or X-ray Computed Tomography (CT), increases when more efficient and faster scintillating crystals are used [2]. Scintillators are usually employed in the form of single-crystals, granular scintillating screens or ceramic blocks.

Cesium Iodide crystals doped with Thallium (CsI:Tl) are widely used in many medical imaging applications (e.g., gamma-ray imaging, x-ray CT, digital radiography, mammography, dentistry (intra-oral radiography), fluoroscopy, etc) [1], since they exhibit high gamma ray detection efficiency and light yield (66000 photons/MeV) [3], [4]. However, CsI:Tl crystal's high afterglow together with its relatively long decay time (800 ns), as compared to decay times met in cerium (Ce^{3+}) doped scintillators, prohibits its application in fast medical imaging modalities, such as spiral CT and PET. As a result, there is a growing interest in introducing new scintillator materials (Table I).

On the other hand, cerium-based (Ce^{3+}) scintillators exhibit faster scintillation response. The Ce^{3+} ion has one electron in the 4f state, which is lifted to the empty 5d shell upon excitation. As a result, an electric dipole $5d \rightarrow 4f$ transition will occur during de-excitation with a decay time in the order of a few nanoseconds (≥ 30 ns) [3]. The Cerium (Ce^{3+}) doped Lutetium oxyorthosilicate ($\text{Lu}_2\text{SiO}_5 : \text{Ce}$ or LSO:Ce) scintillator [5] could be considered as a good compromise between fast response (40 ns) and high light yield (≥ 26000 photons/MeV) scintillators. However, one factor worth considering is its high price in relation to some commonly employed scintillators.

Another Ce^{3+} doped scintillator of commercial use is the Gadolinium oxyorthosilicate (Gd_2SiO_5 or GSO:Ce) crystal. GSO:Ce is characterized by its high density ($\rho = 6.71 \text{ g/cm}^3$), high effective atomic number ($Z = 59$), and high radiation detection efficiency ($\rho Z_{eff}^4 = 84 \cdot 10^6$). Additionally, due to the presence of Ce^{3+} ion activator, GSO:Ce exhibits fast response (60 ns). Moreover, its low cost, as compared to LSO, led GSO:Ce to be used in commercial positron emission tomography detectors [6]. However, GSO:Ce shows a comparatively low light yield (≥ 8000 photons/MeV) and is yet to be employed in a wide range of medical imaging modalities.

A promising next generation scintillation crystal is Cerium doped Lutetium Yttrium Oxyorthosilicate, $(\text{Lu}, \text{Y})_2\text{SiO}_5 : \text{Ce}$ (LYSO:Ce). LYSO:Ce is a mixed LSO/YSO (5–10%, Y) non-hygroscopic crystal that offers high density (7.1 g/cm^3), high

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TABLE I
INORGANIC SCINTILLATORS FOR MEDICAL IMAGING

Scintillator Name	Density (g·cm ⁻³)	ρZ_{eff}^4 (10 ⁶)	Hygroscopicity	Light yield (Photons/MeV)	Decay time (ns)	Emission maximum (nm)
CsI:Tl	4.51	38	Slightly	66 000	800-900	550
Gd ₂ O ₂ S:Pr	7.3	103	No	50 000 ^a	3·10 ³	510
YAlO ₃ :Ce	5.5	7	No	21 000	30	350
Y ₃ Al ₅ O ₁₂ :Ce	4.6	39	No	16 700	80	530
Gd ₂ SiO ₅ :Ce	6.7	84	No	8 000	60	440
Gd ₃ Ga ₅ O ₁₂ :Ce	7.1	58	No	40 000 ^a	1.4·10 ⁵	730
Lu ₂ SiO ₅ :Ce	7.4	143	No	26 000	40	420
Lu _{2(1-x)Y_{2x}SiO₅:Ce}	7.1	63	No	<30000	40	420 ^b

^aMeasured at ~60-80 keV; all others at 662 keV.

^bData are from [7].

Data are from [1].

light output (≤ 30000 ph/MeV), good energy resolution ($\sim 10\%$) and short decay time (40 ns). These properties make LYSO:Ce suitable for a wide range of γ -ray detection applications [7].

In the present study, the light emission characteristics of LYSO:Ce and GSO:Ce single crystal scintillators were investigated under x-ray excitation. Findings could be of interest in tomographic x-ray imaging, such as general purpose CT, breast CT, combined PET/CT. To this aim, five parameters related to the light emission properties of the scintillator were studied, using experimental methods and calculations, namely: i) the luminescence efficiency (LE), i.e. the efficiency to convert the incident x-ray energy flux into emitted light energy flux, ii) the intrinsic conversion efficiency, iii) the spectral matching factor (SMF), and iv) the effective efficiency (EE). A similar approach, with LYSO and LSO crystal scintillators has been reported by Pepin *et al.*, 2004, measuring the performance of the two materials under VUV and γ -ray excitation for phoswich PET detectors [8]. To our knowledge LYSO:Ce has never been previously investigated under medical x-ray imaging conditions.

II. MATERIALS AND METHODS

A. Definitions

The emission efficiency (light yield) of a scintillator may be evaluated by determining the x-ray to light conversion efficiency or luminescence efficiency (LE). This is a dimensionless parameter defined by the ratio:

$$\eta_{\Phi} = \dot{\Psi}_{\lambda} / \dot{\Psi}_X \quad (1)$$

where $\dot{\Psi}_{\lambda}$ is the light energy flux, emitted by a scintillator excited by an incident x-ray energy flux $\dot{\Psi}_X$. LE describes the radiation detection sensitivity of energy integrating detectors, i.e. when the output signal is proportional to the total energy absorbed within the scintillator's mass. LE depends strictly on the intrinsic energy conversion properties of the material, namely (i) the total absorption efficiency, i.e. the fraction of incident energy absorbed within the scintillator mass, including secondary x-ray photon absorption (K-fluorescence and scattered x-ray photons) (ii) the intrinsic radiation-to-light energy conversion efficiency, being the fraction of absorbed energy converted into light within

the crystal, and (iii) the light energy transmission efficiency, expressing the fraction of light escaping the crystal [9], [10]. Following the previous considerations, LE may be defined as the fraction of incident energy converted into output light, i.e. light emitted by the crystal surface, facing the optical detector. To be more consistent with x-ray imaging experimental conditions, LE may be determined by measurements of emitted light energy flux and incident exposure rate. The ratio of these quantities defines the absolute luminescence efficiency (AE) [9]–[12], given in relation (2)

$$\eta_A = \dot{\Psi}_{\lambda} / \dot{X} \quad (2)$$

where the x-ray energy flux, $\dot{\Psi}_X$ in (1), has been replaced by the incident exposure rate, \dot{X} . The latter can be easily experimentally determined through ionization chamber measurements. Absolute efficiency is thus expressed in units of [$\mu\text{W} \cdot \text{m}^{-2} / \text{mR} \cdot \text{s}^{-1}$ or Efficiency Units (E.U.)].

Since LE depends on the intrinsic energy conversion properties of the scintillator, it may be of interest to obtain an estimation of their values.

The absorption efficiency $\eta_a(E)$ may be approximated by an exponential function [14] of the x-ray mass total absorption coefficient ($\mu_{tot,a}(E)/\rho$) of the scintillator material. This coefficient can be calculated by employing tabulated data [15], [16]. However, in the present study, the total absorption efficiency was 100%, due to the large size of the crystals employed.

The intrinsic x-ray energy to light conversion efficiency (η_C) may be estimated [17] by the expression

$$\eta_C = \frac{\bar{E}_{\lambda}}{E_g} \cdot \left(\frac{S \cdot Q}{\beta} \right) \quad (3)$$

where \bar{E}_{λ} is the mean energy of the emitted light photons, E_g is the forbidden energy gap between the valence and the conduction energy bands, S is the transfer efficiency of the electron-hole pair, expressing the fraction of electron-hole energy transferred to the site of the activator (Ce^{3+}), Q is the quantum efficiency of the luminescent centre i.e. the activator Ce^{3+} , expressing the fraction of transferred electron-hole pair energy absorbed at the activator site, and β is a parameter characterizing the excess energy, above E_g , required to be absorbed so as to

allow for an electron-hole pair generation. The mean energy of light photons \bar{E}_λ may be determined from the spectrum of light $S_P(\lambda)$ emitted by the scintillator crystal.

The light transmission efficiency was experimentally determined as described in later a section.

In addition, the effect of the light spectrum on the overall detector (scintillator-optical detector) performance may be estimated by determining the spectral compatibility of the scintillator's emitted light with the spectral sensitivity of the optical photon detector. Spectral compatibility may be estimated by the spectral matching factor (SMF), which is defined by the ratio [18]

$$\alpha_S = \frac{\int S_P(\lambda)S_D(\lambda)d\lambda}{\int S_P(\lambda)d\lambda} \quad (4)$$

where S_D is the spectral sensitivity of the optical photon detector. To express the overall sensitivity of a specific scintillator-optical detector combination, the effective efficiency (EE) has been defined [11], [19] as:

$$\eta_{eff} = \eta_\Phi \alpha_S. \quad (5a)$$

B. Experiments and Calculations

The luminescence efficiency and related properties, i.e. total absorption efficiency, intrinsic energy conversion efficiency, light transmission, spectral matching factor, and effective efficiency of the two scintillators, were determined employing the following experiments and calculations.

The LYSO:Ce and GSO:Ce crystals examined in this study were supplied by Photonic Materials Ltd., Scotland, U.K., and by Hitachi Chemicals Co., Ltd, Japan, respectively. Both crystals had dimensions of 10 mm × 10 mm × 10 mm and they were doped with 0.5% mole of Cerium (Ce⁺³). The crystals were irradiated by X-rays using the following imaging units: (i) a General Electric Senographe DMR x-ray mammography unit, equipped with a molybdenum (Mo) anode target and molybdenum filter (to produce x-ray photon energies for tomographic breast imaging), and (ii) a Philips Optimus x-ray unit with a tungsten (W) anode target and 2 mm Al filter (to produce x-ray energies used in general purpose computed tomography). In the x-ray mammography unit the filter changed automatically from molybdenum to rhodium (Rh) and aluminum (Al) filters, as tube voltage increased from medium to higher mammographic voltages. The whole range of available x-ray tube voltages varied from 22 to 49 kVp in the mammography unit and from 40 to 140 kVp in the general x-ray unit. 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 by human breast [20]. Similarly, a 20 mm thick Al was employed in the general radiography unit to simulate beam hardening by the human body [21].

The absolute luminescence efficiency was determined, according to relation (2), by performing x-ray exposure and light flux measurements. The exposure rate, \dot{X} in (2), was measured at the crystal's site, using a Radcal 2026C dosimeter (Radcal

Corp., USA). Light energy flux, $\dot{\Psi}_\lambda$, measurements were performed using an experimental setup, comprising a light integration sphere (Oriol 70451) coupled to a photomultiplier (EMI 9798B) [14]. The luminescence efficiency was determined using the experimental absolute efficiency data and by converting the incident x-ray exposure rate into x-ray energy fluence according to the conversion formula (6) [13], [22]:

$$\dot{\Psi}_X = \dot{X} \left[\frac{(W_{air}/e)}{\langle \mu_{en}/\rho \rangle_E} \right] \quad (6)$$

where $\langle \mu_{en}/\rho \rangle_E$ is the x-ray mass energy absorption coefficient of air, averaged over the x-ray spectrum and $[W_{air}/e]$ is the average energy per unit of charge required to produce an electron-ion pair in the air. Hence, using relations (1), (2) and (6), LE may be expressed as follows [18]:

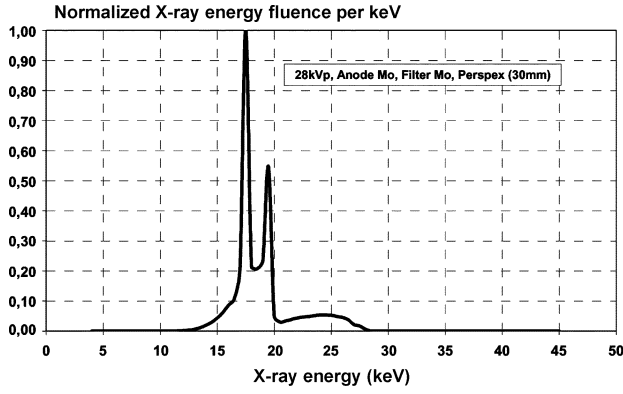
$$\eta_\Phi = \frac{\langle \mu_{en}/\rho \rangle_E}{[W_{air}/e]} \cdot \eta_A. \quad (7)$$

The value of the ratio (W_{air}/e) was obtained from literature [23]. The energy averaged value of the x-ray mass energy absorption coefficient of air $(\bar{\mu}_{en}/\rho)$ was determined using relation (8):

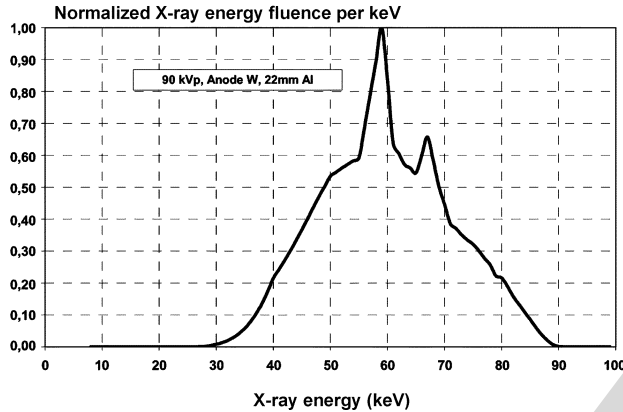
$$\langle \mu_{en}/\rho \rangle_E = \left(\int \Psi_X(E) [\mu_{en}(E)/\rho] dE \right) / \int \Psi_X(E) dE. \quad (8)$$

Function $\Psi_X(E)$, denotes the spectral distribution of the x-ray energy fluence incident on the scintillator crystal. This function was determined according to previously published models for computer generating tungsten and molybdenum x-ray spectra [24], shown in Fig. 1. $\mu_{en}(E)/\rho$ in relation (8) is the monoenergetic value of the mass energy absorption coefficient of air at energy level E . The total absorption efficiency, $\eta_a(E)$ in relation (2), was determined [14] by calculating the mass total absorption coefficient of the scintillator under investigation. Data for the chemical elements Lu, Y, Si, O, Gd and for the photon energies employed in our study were obtained from literature [15], [16]. The intrinsic energy conversion efficiency was determined according to relation (3). The values of the parameters Q and S were assumed to be equal to 1. This is a reasonable assumption [3] and leads to the determination of the maximum value of η_C . Values for the parameter β , for inorganic scintillators, range between 2 and 3 [17], but in some cases of crystalline scintillators can exceed the value of 5 (see Table II) [1], [17]. The mean optical photon energy \bar{E}_λ was obtained from light emission spectrum measurements. The energy gap, E_g , for each material was obtained from the literature [1], [25]–[27]. The light transmission efficiency, $\eta_\lambda(\lambda)$, was determined from light transmission measurements obtained by a Perkin-Elmer UV/Visible lamda 15 Spectrophotometer (Perkin-Elmer Life And Analytical Sciences, Inc., USA).

To determine both the mean light photon energy \bar{E}_λ and the spectral matching factor α_S , the emitted light spectrum $S_P(\lambda)$ of the LYSO:Ce and GSO:Ce crystals was measured using a grating optical spectrometer (Ocean Optics Inc., HR2000). Spectrometer light measurements were performed under x-ray



(a)



(b)

Fig. 1. X-ray spectra for a) 28 kVp and b) 90 kVp calculated according to [24].

TABLE II
THEORETICAL MAXIMUM INTRINSIC CONVERSION EFFICIENCY OF LYSO:Ce
AND GSO:Ce SCINTILLATORS

Parameter	LYSO:Ce	GSO:Ce
E_g (eV)	6.40 ^c	6.20 ^d
β	2.5 ^b	5.78 ^a
η_c	0.18 ^e	0.08 ^a

^aData are from [17]
^bData are from [25]
^cData are from [26]
^dData are from [27]
^eCalculated data

excitation. The light emitted by the irradiated scintillator was transferred to the spectrometer through a 2.0 m long, 400 μm fiber optic, (Avantes Inc. FCB-UV400-2, Colorado, USA). X-ray measurements were performed at various x-ray tube voltages up to 140 kVp. To improve measurements sensitivity with minimum tube load, the X-ray tube was positioned very close to the scintillation crystal surface for acquiring most of the emitted light. Corrections for light signal degradation due to fiber optic light losses were taken into account. Spectral sensitivity ($S_D(\lambda)$) data were obtained from corresponding manufacturer's (Hamamatsu, EMI, etc.) datasheets. Six optical photon detectors, currently used in a large variety of imaging detectors (digital and conventional radiography, fluoroscopy,

TABLE III
SPECTRAL MATCHING FACTORS OF LYSO:Ce AND GSO:Ce WITH SOME
OPTICAL DETECTORS

Optical Detectors	LYSO:Ce	GSO:Ce
GaAs Photocathode	0.93	0.93
Extended S-20 Photocathode	0.90	0.90
GaAsP Hamamatsu Photocathode	0.67	0.66
a-Si:H 108H Photodiode	0.74	0.74
Si/S1133 Hamamatsu Photodiode	0.44	0.45
CCD S100AB SiTe®	0.88	0.88

computed tomography, nuclear medicine, etc.) and their spectral matching factor with the LYSO:Ce and GSO:Ce spectra were examined (Table III). These optical detectors were: (i) GaAs photocathode, (ii) extended S20 EMI photocathode with quartz window, (iii) GaAsP Hamamatsu photocathode, (iv) a-Si:H/108H amorphous silicon photodiode, corresponding to intrinsic layer thickness of 800 nm (108H), (v) Si/S1133 Hamamatsu crystalline silicon photodiode, and (vi) CCD S100AB SiTe.

III. RESULTS AND DISCUSSION

Figs. 2 and 3 show the luminescence efficiency curves of LYSO:Ce and GSO:Ce scintillators, determined by experiments and (7), for energies between 22–45 kVp, used in mammographic applications (Mo spectrum), and for energies between 40–140 kVp, used in general x-ray imaging applications (W spectrum). Under both mammographic and general x-ray imaging conditions, the luminescence efficiency curves showed a nonlinear response with increasing x-ray tube voltage. This finding seems to be consistent with previously reported data that light yield varies in both LYSO:Ce and GSO:Ce due to non-proportionality effects [14], [28]–[30]. LYSO:Ce was found clearly higher than GSO:Ce, in the whole energy range up to 140 kVp (see Fig. 3). However, the difference in efficiency between the two materials is smaller than that depicted on Table I. This may be due to differences in the corresponding light yields under different x-ray energies (22–140 kVp in our experiments against 662 keV in Table I). LE data are presented in two separate figures (Figs. 2 and 3) to clearly indicate the differences between the corresponding experimental conditions. The shape of the x-ray energy spectra (and the corresponding mean x-ray photon energy), shown in Fig. 1, are strongly affected by the x-ray tube anode material, the anode filter, and the total filtration, employed to simulate the patient's body. This may explain the differences, observed in Figs. 2 and 3, between LE values obtained under equal x-ray tube voltages but at different x-ray units (W anode or Mo anode).

In Table II the intrinsic conversion efficiency (η_c) of LYSO:Ce and GSO:Ce is shown, calculated using relation (3) and published data [17], [25]–[27]. The difference between the two scintillators was due to the value of parameter β (expressing the fraction of energy transferred to phonons) and the Gd-to-Ce⁺³ transfer in GSO [31], which depends on the Ce⁺³ concentration affecting also the Gd⁺³ sub-lattice [32].

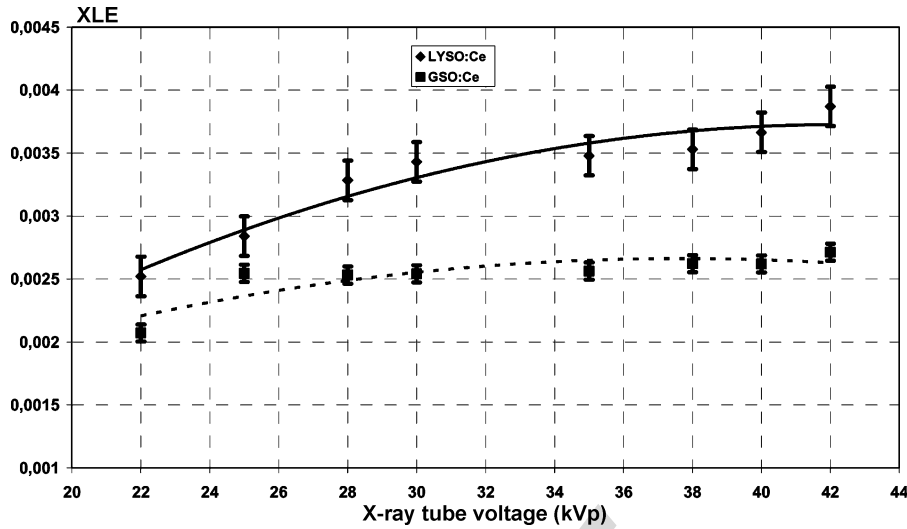


Fig. 2. The x-ray luminescence efficiency (XLE) of LYSO:Ce and GSO:Ce as determined by the experimental data for x-ray tube voltages between 22–42 kVp (mammography). Points: measured data, line: fitted curve.

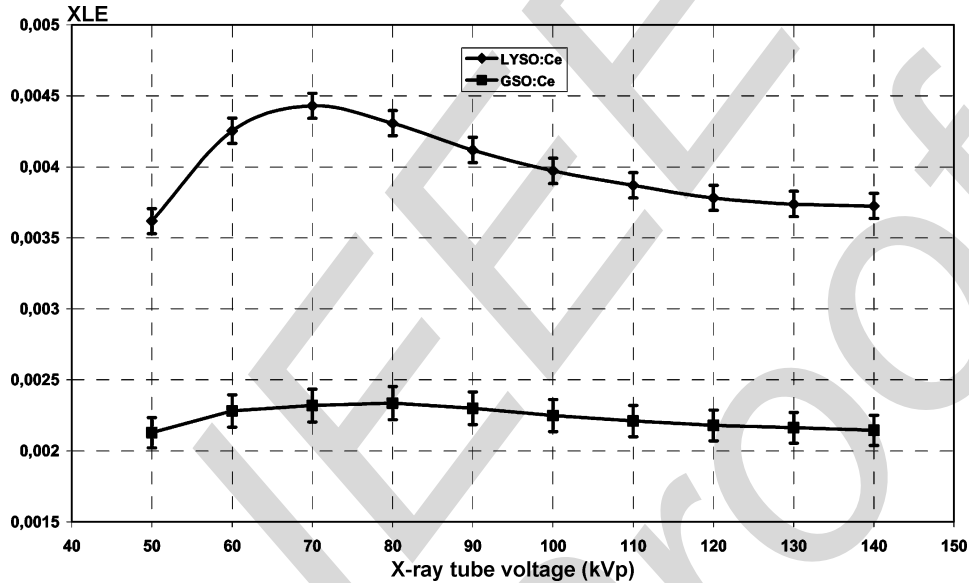


Fig. 3. The x-ray luminescence efficiency (XLE) of LYSO:Ce and GSO:Ce as determined by the experimental data for x-ray tube voltages between 40–140 kVp (general radiography). Points: measured data, line: fitted curve.

Parameter β is 5.78 for GSO:Ce instead of 2.5 for LYSO:Ce [25]. It must be noted that the intrinsic efficiency of LYSO:Ce ($\eta_C = 0.185$) was estimated to be higher than many scintillators (CsI, BGO, Gd₂O₂S) currently used in various medical imaging applications [1], [9], [17]

Figs. 4 and 5 show the light transmission and absorption measurements obtained using the Perkin-Elmer UV/Visible lamda 15 Spectrophotometer. In Fig. 4, transmittance expresses the fraction (%) of light passing through the crystals and reaching the spectrophotometer's optical detector. Both crystals showed high transmittance (84.0% for LYSO:Ce and 82.5% for GSO:Ce) in the visible region above 450 nm. These values depend on the dimensions and on the Ce concentration of the specific scintillator sample. However, it seems to be higher than the corresponding light transmission values previously published for LSO:Ce [33] and in consistency with values

recently published for LYSO:Ce scintillators under high energy physics experimental conditions [34]. The relatively high light transmittance is an additional factor ameliorating the luminescence efficiency of LYSO:Ce. In the region from 450 nm down to 412 nm, LYSO transmittance showed a tendency to decrease very slightly. For lower wavelengths, transmittance was found to drop rapidly down to values close to zero. As one should expect, by considering the energy gap of the undoped crystals (6.4 eV, for LYSO, and 6.2 eV, for GSO), the absorption for both crystals should be limited to values close to 200 nm. This may be estimated using the relation ($\lambda_{cf} = hc/E_g$) giving the cut-off wavelength (λ_{cf}) in terms of the energy gap (E_g) [35]. However, the absorption cut-off shifts close to 380 nm and this may be regarded to be mainly due to the cerium dopand [33], to its specific concentration, and the specific crystal sample

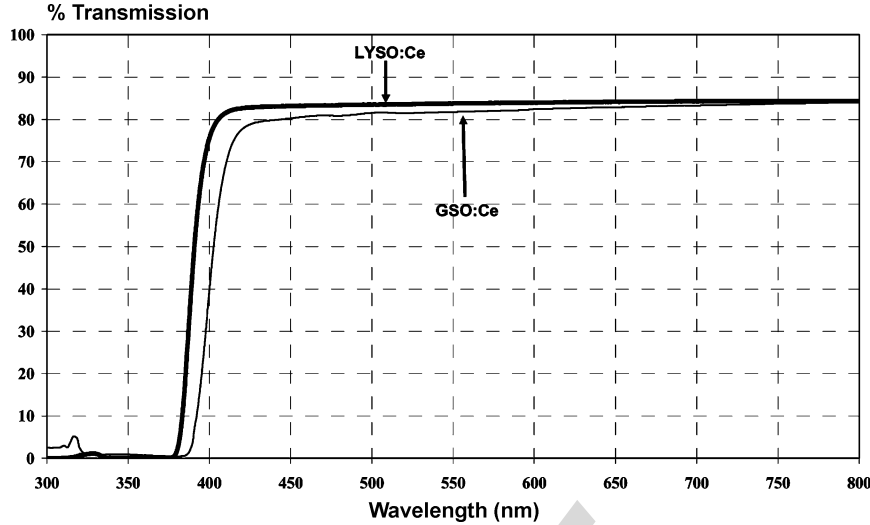


Fig. 4. Light transmission measurements using a spectrophotometer Perkin Elmer Lamda 15. For best resolution the scale of measurement was set from 300 nm to 500 nm. Above 500 nm or below 300 nm the transmittance of LYSO:Ce and GSO:Ce was constant.

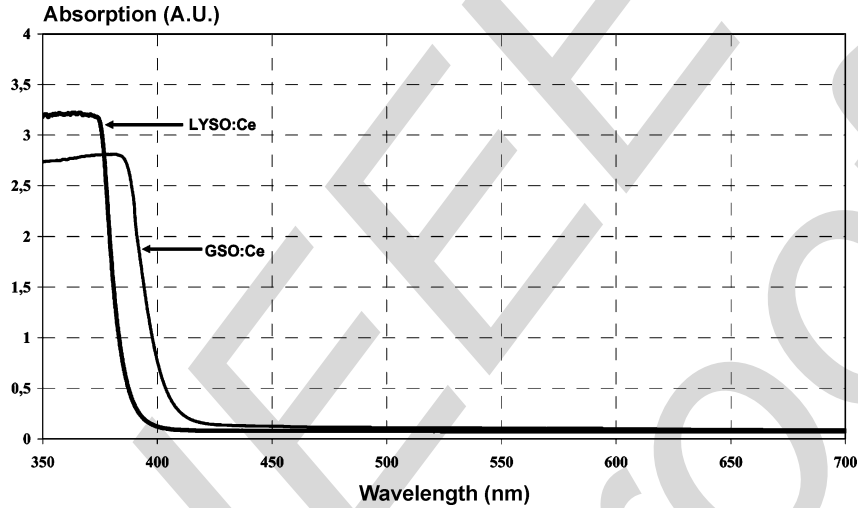


Fig. 5. Light absorption measurements using a spectrophotometer Perkin Elmer Lamda 15. Above 450 nm the absorption of LYSO:Ce was constant. (A.U.: Arbitrary Units).

thickness. In Fig. 5, absorption is expressed as the inverse transmittance.

Fig. 6(a) and (b) shows the normalized spectral response of LYSO:Ce and GSO:Ce crystals, respectively, compared to the spectral sensitivities of six optical photon detectors. X-ray excited LYSO:Ce spectrum, extending from 390 to over 470 nm and peaking at about 430 nm, was found very similar to the shape of the spectrum of x-ray excited LSO:Ce crystal (426 nm) reported by others [33].

Table III summarizes the calculated spectral matching factors of LYSO:Ce and GSO:Ce scintillators with some optical photon detectors, using relation (4). The spectral sensitivity of the optical photon detectors (S_D), in Table III, were taken from manufacturer's datasheets. The highest spectral compatibility was found for the GaAs photocathode (matching factor: 0.93). High compatibility was also exhibit with two optical detectors of interest in x-ray imaging, namely the amorphous silicon photodiode and the CCD. Differences in spectral compatibility are

reflected in Fig. 7, which shows the effective luminescence efficiency of the LYSO:Ce scintillator with S-20, a-Si, and CCD optical detectors.

IV. CONCLUSIONS

In conclusion, our measurements showed that the luminescence efficiency of LYSO:Ce and GSO:Ce scintillators was found adequately high for applications in the energy range employed in x-ray imaging. The luminescence efficiency of LYSO:Ce seems to be affected by the effects of non-proportionality of light yield reported in previous studies [29], [30]. The luminescence efficiency of LYSO:Ce was found higher than that of GSO:Ce crystal. The intrinsic conversion efficiency of LYSO:Ce was estimated to be significantly higher than many of the currently employed scintillators. The light emission spectrum, extending from 390 nm to over 470 nm

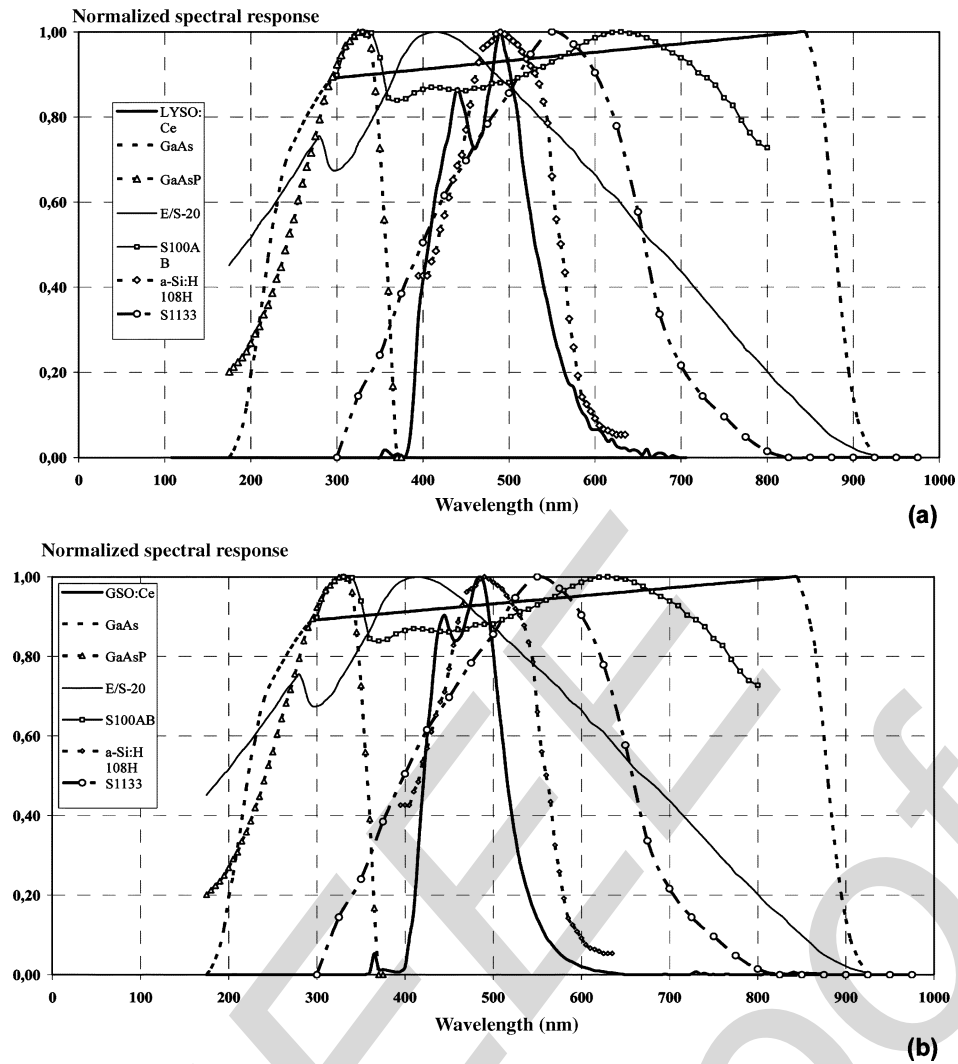


Fig. 6. Normalized spectral response of (a) LYSO:Ce and (b) GSO:Ce, compared to the spectral sensitivities of GaAs, ES-20, GaAsP, CCD S100AB, a-Si:H 108H and S1133 photodetectors.

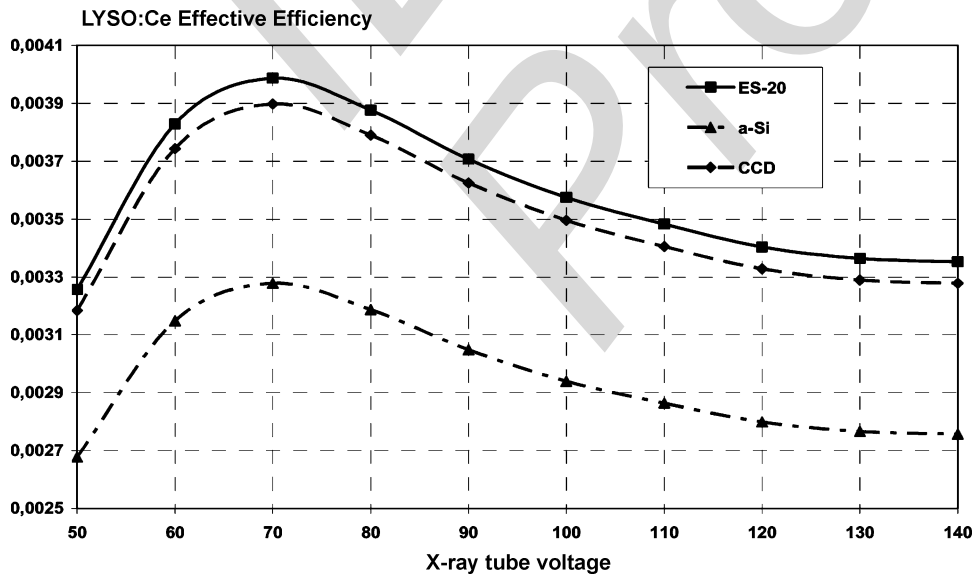


Fig. 7. Effective efficiency of LYSO:Ce crystal with S-20 EMI photocathode, a-Si photodiode, and CCD.

and peaking at about 430 nm for LYSO:Ce, was found compatible (70%–95%) to many currently employed optical photon

detectors (amorphous and crystalline silicon photodiodes etc). Taking into account the very short scintillation decay time of

both scintillators, they can be considered as potential substitute of currently employed scintillators (e.g. CsI:Tl) in breast computed tomography, in modern fast image producing x-ray computed tomography systems, especially those employed in combined PET/CT devices, as well as in imaging applications requiring relatively high x-ray tube voltages (e.g. computed tomography detectors, pixelated scintillation digital detectors, nuclear medicine imaging detectors etc).

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