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Comparative evaluation of single crystal scintillators under x-ray imaging conditions

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ABSTRACT: The present study is a comparative investigation of the luminescence properties of (Lu,Y)₂SiO₅: Ce (LYSO: Ce), YAlO₃: Ce (YAP: Ce), Gd₂SiO₅: Ce (GSO: Ce) and (Bi₄Ge₃O₁₂) BGO single crystal scintillators under x-ray excitation. Results will be of value in designing dual modality tomographic systems (PET/CT, SPECT/CT) based on a common scintillator crystal. All scintillating crystals have dimensions of 10 × 10 × 10 cm³ are non-hygroscopic exhibiting high radiation absorption efficiency in the energy range used in medical imaging applications. The comparative investigation was performed by determining the x-ray luminescence efficiency (emitted light flux over incident x-ray energy flux) in the range of x-ray energies employed in: (i) general x-ray imaging (40–140 kV, using a W/Al x-ray spectrum) and (ii) x-ray mammography imaging (22–49 kV, using a Mo/Mo x-ray spectrum). Additionally, light emission spectra of crystals at various x-ray energies were measured, in order to determine the intrinsic conversion efficiency and the spectral compatibility to optical photon detectors incorporated in medical imaging systems. The light emission performance of LYSO:Ce scintillator studied was found very high for x-ray imaging.

KEYWORDS: X-ray detectors; Gamma camera, SPECT, PET PET/CT, coronary CT angiography (CTA); X-ray radiography and digital radiography (DR); X-ray mammography and scinto- and MRI-mammography

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1 Introduction

BGO combines good scintillation characteristics however, its decay time (~ 300 ns) limits its applications in fast imaging, i.e. Spiral Computed Tomography (SCT) and Positron Emission Tomography (PET) [1].

Cerium (Ce^{3+}) doped Gadolinium (Gd_2SiO_5 or GSO) oxyorthosilicate and Yttrium orthoaluminate (YAlO_3 : Ce or YAP: Ce) are fast emitting scintillators employed mainly in PET and animal PET detectors [2]. However GSO and YAP exhibit relatively lower light yield (≥ 8000 ph/MeV for GSO and 21000 ph/MeV for YAP) than Lutetium Yttrium Oxyorthosilicate, $(\text{Lu}, \text{Y})_2\text{SiO}_5$:Ce (LYSO: Ce), which is a promising scintillation crystal [3], with similar scintillation properties as LSO:Ce [4].

The present study is a comparative investigation of GSO:Ce, YAP:Ce and LYSO: Ce single crystal scintillators with the traditional BGO scintillator, under mammographic and general x-ray medical imaging conditions for potential use in dual modality tomographic systems (PET/CT, SPECT/CT) based on a common scintillator crystal.

2 Materials and methods

The emission efficiency (light yield) of a scintillator may be evaluated by determining the x-ray to light conversion efficiency or luminescence efficiency (LE) [5] (emitted light energy flux over incident x-ray energy flux).

An additional important parameter to be examined in medical imaging detectors is the Spectral Matching Factor (SMF), i.e. the compatibility of the scintillator's light emission spectrum to the spectral sensitivity of various optical photon detectors [5].

Another critical parameter is the intrinsic conversion efficiency η_C , i.e. the percent of the absorbed energy converted into light within the scintillator [6].

All crystals used in this study had dimensions of $10\text{ mm} \times 10\text{ mm} \times 10\text{ mm}$. Cerium activated crystals were doped with 0.5% mol of cerium (Ce^{+3}). The crystals were irradiated by X-rays using a Philips Optimus x-ray unit and a General Electric Senographe DMR x-ray mammography unit. Appropriate beam filtering was applied to simulate x-ray beam hardening by human body [7].

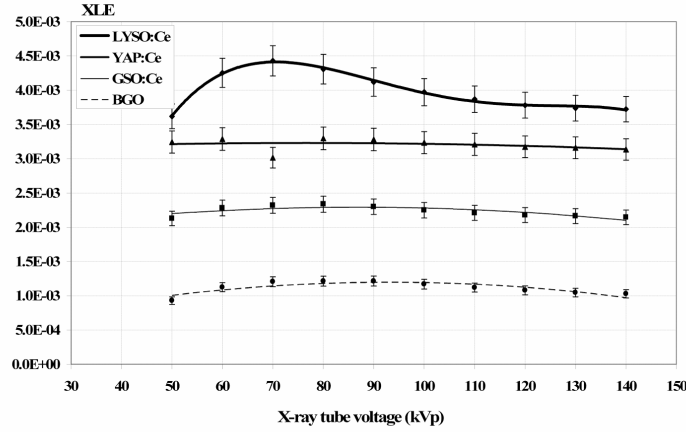


Figure 1. The x-ray luminescence efficiency (XLE) of LYSO:Ce, YAP:Ce, GSO:Ce and BGO as determined by the experimental data for x-ray tube voltages between 50–140 kVp (general radiography). Points: measured data, line: fitted curve.

The x-ray luminescence efficiency was determined by performing x-ray energy and light flux measurements, previously described by Valais et al. [5]. The intrinsic conversion efficiency, η_C , was calculated as follows [6]:

$$\eta_C = \frac{\bar{E}_\lambda}{E_g} \cdot \left(\frac{S \cdot Q}{\beta} \right) \quad (2.1)$$

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 absorption efficiency of the activator, 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 was obtained from light emission spectrum measurements. The energy gap, E_g , for each material was obtained from published data [6, 8, 9]. The intrinsic conversion efficiency values are reported in table 1.

The SMF was examined for five optical photon detectors currently used in digital radiography, computed tomography and nuclear medicine (table 2).

3 Results and discussion

The variation of the [5] x-ray luminescence efficiency is shown in figures 1 and 2. Figure 1 shows the luminescence efficiency curves of BGO, GSO:Ce, YAP:Ce and LYSO:Ce scintillators, for energies between 22–45 kVp, used in x-ray mammography (Mo spectrum), and figure 2, for energies between 40–140 kVp, used in general x-ray imaging (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 [10].

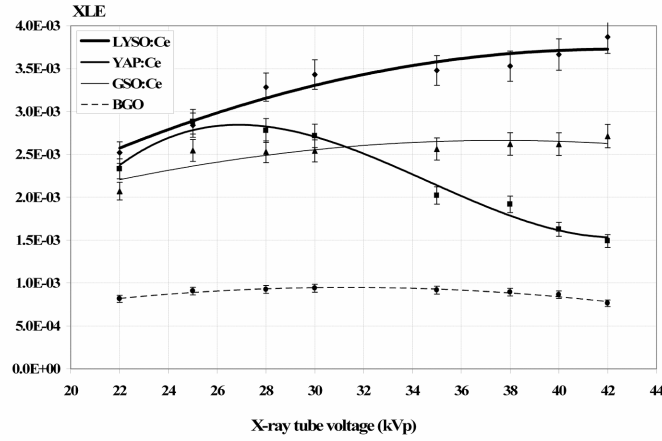


Figure 2. The x-ray luminescence efficiency (XLE) of LYSO:Ce, YAP:Ce, GSO:Ce and BGO as determined by the experimental data for x-ray tube voltages between 22–42 kVp (mammography). Points: measured data, line: fitted curve.

Table 1. Theoretical maximum intrinsic conversion efficiency of LYSO:Ce, YAP:Ce, GSO:Ce and GO scintillators.

| Parameter | LYSO:Ce | YAP:Ce | GSO:Ce | BGO |
|------------------|--------------------|--------------------|--------------------|--------------------|
| $E_g(\text{eV})$ | 6.4 ^a | 7.7 ^a | 6.2 ^c | 5.0 ^a |
| β | 4.4 ^d | 5.6 ^c | 5.8 ^b | 8.6 ^d |
| η_C | 0.105 ^c | 0.077 ^d | 0.081 ^d | 0.060 ^c |

^a Data are from [8].

^b Data are from [9].

^c Data are from [11].

^d Calculated data.

LE data are presented in two separate figures (figures 1 and 2) 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), 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 figures 1 and 2, between XLE values obtained under similar x-ray tube voltages but at different x-ray units (W anode or Mo anode).

In figure 1 LYSO:Ce exhibits a noticeable increase in XLE at the x-ray tube voltages between 60–70 kVp. This non-proportionality effect may be attributed to the absorption in the K-edge of Lu ions (63.3 keV) [10]. Similarly in figure 2, YAP:Ce exhibits a non-linear response in the x-ray tube voltage range between 22 to 28 kVp mainly due to the absorption of the K-edge of Y ions (20 keV) [10]. Table 1 shows maximum values of the intrinsic conversion efficiency, i.e. $Q = 1$ and $S = 1$ in (2.1).

The mean energy of light photons \bar{E}_λ was calculated from light emission spectrum measurements as shown in figure 3.

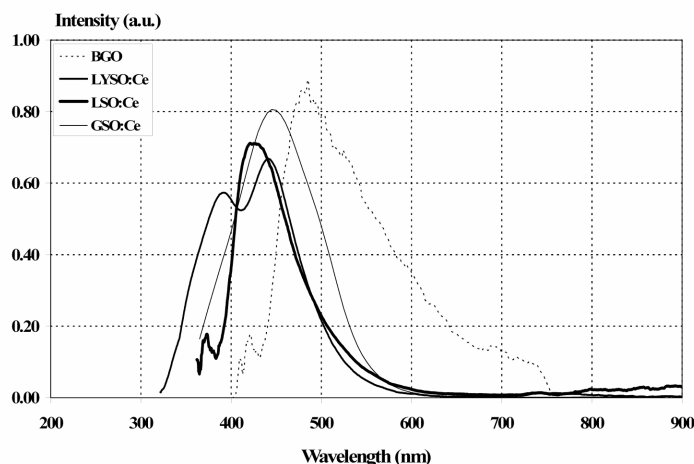


Figure 3. Normalized spectral response of BGO, LYSO:Ce, YAP:Ce and GSO:Ce single crystal scintillators.

Table 2. Spectral Matching Factors of BGO, GSO:Ce, YAP:Ce and LYSO:Ce with optical detectors.

| Optical Detectors | BGO | GSO:Ce | YAP:Ce | LYSO:Ce |
|----------------------------|------|--------|--------|---------|
| Extended S-20 Photocathode | 0.89 | 0.90 | 0.93 | 0.94 |
| APD Hamamatsu S5343 | 0.56 | 0.76 | 0.39 | 0.60 |
| a-Si:H 108H Photodiode | 0.74 | 0.70 | 0.07 | 0.57 |
| PSPMT Hamamatsu H8500 | 0.80 | 0.71 | 0.95 | 0.86 |
| CCD S100AB SITe® | 0.87 | 0.88 | 0.85 | 0.88 |

These spectra were found to be well matched with the spectral sensitivity curves of most optical detectors (table 2).

In table 2, BGO, GSO:Ce and LYSO:Ce exhibit highest compatibility when combined to EMI S-20 photocathode, whereas YAP:Ce when combined to PSPMT H8500.

In the present investigation we observed a clear superiority in the LYSO:Ce light output under medical x-ray excitation over BGO, GSO:Ce and YAP:Ce scintillators. Similarly, although GSO:Ce and BGO have been reported to exhibit similar light yields, our results (figures 1 and 2) demonstrate a well defined superiority of GSO:Ce, as far as the XLE is concerned in both radiographic and mammographic x-ray tube voltages. The above considerations may be reasoned considering the differences in non-proportionality curves at low energies reported by Balcerzyk et al. [12] and Dorenbos et al. [10].

4 Conclusions

The light emission performance of LYSO:Ce was found higher than YAP:Ce, BGO and GSO:Ce. The emission spectra of all four scintillators examined in our study are well matched with the spectral sensitivities of the optical photon detectors often employed in radiation detectors. The intrinsic conversion efficiency of LYSO:Ce was the highest of all scintillators examined and this in turn may explain the superiority of LYSO:Ce XLE. Taking into account the luminescence efficiency and the short decay time of LYSO:Ce scintillator, it can be considered as potential detector in modern fast single detector multimodality imaging systems.

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