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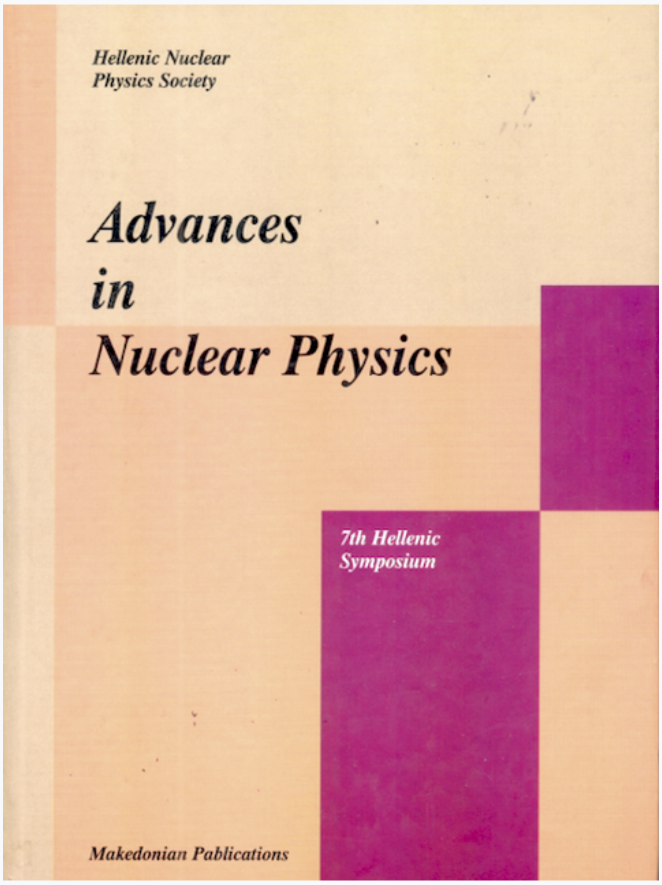

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Behaviour of Y_2O_3:Eu^{3+} Scintillator under Radiation used in Medical Applications

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Abstract

The Y_2O_3:Eu^{3+} scintillator was studied for use in radiation detectors of medical imaging systems. Y_2O_3:Eu^{3+} was used in the form of laboratory prepared test screens. The x-ray luminescence efficiency of the screens was measured for tube voltages up 250 kVp. The intrinsic x-ray to light conversion efficiency (nc) and other optical parameters of the Y_2O_3:Eu^{3+} scintillator related to optical scattering, absorption, and reflection were determined. The light emission spectrum of Y_2O_3:Eu^{3+} was measured (λ=613 nm). The x-ray luminescence efficiency peaked at 180 mg/cm2 screen coating weight. The intrinsic x-ray to light conversion efficiency was found to be n_c=0.095. Results indicated that Y_2O_3:Eu^{3+} is a medium to high overall performance material that could be used in medical imaging systems.

1 Introduction

Europium activated yttrium oxide Y_2O_3:Eu^{3+} is a scintillator material that has been employed in radiation detectors in various applications but not in detectors of medical imaging systems. The latter use scintillators coupled to photosensitive detectors (film, photodiodes, photocathode, CCD arrays) to capture ionizing radiation emerging from the patient's body.

In this work a detailed study of the x-ray luminescence efficiency of Y_2O_3:Eu^{3+} under the conditions used in diagnostic radiology is presented. The scintillator had the form of screens consisting of phosphor grains in a binding material and x-ray energy varied in the range from 40-250 kVp.

2 Material and Methods

2.1 Theory

The x-ray luminescence efficiency η_Φ of a scintillator is defined as the ratio of light energy flux Ψ_L emitted when an x-ray energy flux Ψ_X incidents on the surface of the scintillator detector.

$$\eta_\Phi(E_o,t) = \Psi_L(E_o,t)/\Psi_x(E_o) \qquad (1a)$$

where

$$\Psi_L(E_o,t) = N_L\overline{E_\lambda} \qquad and \qquad \Psi_x(E_o) = N_x(E_o)\overline{E} \qquad (1b)$$

and

$$N_L = N_x(E_o)\eta_Q(E_o,t)\eta_c[\overline{E}/\overline{E_\lambda}]G(s,a,r,E_o,t) = N_x(E_o)\eta_\Phi(E_o,t)[\overline{E}/\overline{E_\lambda}] \quad (2)$$

which is in accordance with the expression [1] of the efficiency of a phosphor as the product of quantum detection efficiency (η_Q), intrinsic conversion efficiency (ηC) and light transmission efficiency (G(s,a,r,E_0,t))

$$\eta_\Phi(E_o,t) = \eta_Q(E_o,t)\eta_c G(s,a,r,E_o,t) \qquad (3)$$

N_L is the number of light photons emitted by N_X x-ray photons incident per unit of area and time, η_Q (E_0,t) denotes the x-ray quantum detection efficiency of the fluorescent layer of thickness t at x-ray energy E_0. G(a,s,r,E_0, t) is the light transmission efficiency giving the fraction of produced light photons that are transmitted through the material and are emitted from the surface of the fluorescent layer. s, a, r are coefficients of optical scattering (s), optical absorption (a) within the scintillator material, and of optical reflection (r) at the boundaries of the layer [1-4]. $\overline{E}, \overline{E}_\lambda$are the corresponding mean energies of the incident x-ray quanta and emitted light quanta. E_0 is the maximum energy in the spectrum of the incident x-ray quanta. E_0 is numerically equal to the x-ray tube voltage.

The luminescence efficiency is theoretically calculated [1-3,5] considering that the x-ray absorption, light generation, and light transmission within the screen phosphor material is described by the differential equations:

$$\frac{dI_F}{dt} = -(a+s)I_F + sI_B + \frac{1}{2}\eta_c\mu(E)N_x(E)exp(-\mu(E)t) \qquad (4)$$

where, I_F is the forward directed light intensity relative to the x-ray beam direction. μ(E) is the x-ray mass attenuation coefficient [6], t is the penetration depth of x-rays. The solution of these differential equations gives:

$$\eta_\Phi(E,t) = \frac{\eta_c\gamma T\mu(E)(1+\rho)exp(-\mu(E)t)}{2(\mu(E)^2-\sigma^2)} \times$$

$$\frac{(\mu(E)-\sigma)(1-\beta)exp(-\sigma t)+2(\sigma+\mu(E)\beta)exp(\mu(E)t)-(\mu(E)+\sigma)(1+\beta)exp(\sigma t)}{(1+\beta)(\rho+\beta)exp(\sigma t)-(1-\beta)(\rho-\beta)exp(-\sigma t)} \qquad (5)$$

where :

T: transparency of the screen's substrate

ρ: ρ = (1-r)/(1+r) where r is the reflectivity of the screen's substrate

γ: conversion factor converting energy fluence (W/m2) into exposure rate (mR/s).

σ, β: coefficients directly related to the absorption (a) and scattering (s) coefficients of optical photons within the screen, by σ=[a(a+2s)]$^{1/2}$ and β=[a/(a+2s)]$^{1/2}$

2.2 Experimental Methods

Y_2O_3:Eu^{3+} was used in the form of scintillating screens of various coating weights. The screens were prepared in laboratory by sedimentation of the scintillator in powder form on fused silica substrates [2,3,5]. They were excited to luminescence in a Siemens Stabilipan x-ray unit with tube voltages up to 250 kVp.

The light flux emitted was measured by an EMI 9558 QB photomultiplier with an extended S-20 photocathode connected to a Cary 401 electrometer. The x-ray energy flux was determined by measuring the x-ray exposure using a PTW Simplex dosemeter and an appropriate conversion factor [7]. Optical reflectivity measurements were also performed according to [1] to determine parameters ρ and β.

3 Results and Discussion

Table 1 shows results on x-ray luminescence efficiency at various x-ray tube voltages for a scintillating screen of 80 mg/cm^2 coating thickness. The efficiency is expressed in number of emitted optical quanta per incident x-ray quantum.

Results concerning other screens with higher or lower coating weight had similar behaviour. The most efficient screen was the one having 180 mg/cm^2 coating thickness. The luminescence efficiency continuously decreases with increasing x-ray tube voltage following a similar variation of the quantum detection efficiency. The latter continuously decreases for x-ray energies higher than 17 keV where the K- absorption edge of yttrium appears.

Table 2 shows the values of optical parameters η_C, σ, β in comparison with corresponding parameters of other scintillator materials.

Table 1: Experimental values of x-ray luminescence efficiency for an 80 mg/cm^2 scintillator screen.

x-ray luminescence efficiency (light quanta per incident x-ray quantum)	x-ray tube voltage (kVp)
245	40
230	50
210	60
195	70
170	80
170	90
165	100
145	110
140	120
135	140
118	160
115	180
110	200
100	250

The values of η_C and σ were found by fitting equation (5) to experimental data employing the Levenberg-Marquard method [8]. The intrinsic x-ray to light conversion efficiency of Y_2O_3:Eu^{3+} (0.095) was found higher than the corresponding efficiency of CaWO4 (0.05), which is conventionally used in medical radiography, and approximately equal to NaI:Tl (0.10) used in nuclear medicine. However, ηC was considerably lower than that of Y_2O_2S:Tb, La_2O_2S:Tb, and Gd_2O_2S:Tb (0.18-0.20) employed in some modern either conventional or digital x-ray imaging systems.

As shown in Table 2 the main advantage of the Y_2O_3:Eu^{3+} scintillator is the low value of light attenuation parameter σ denoting lower optical scattering, which is due to the longer wavelength (613 nm) of the emitted light. The latter was measured with an 7240 Oriel monochromator.

Table 2: Intrinsic efficiency and optical parameters of scintillators

Scintillator	η_C	σ (cm^2/g)	β
Y_2O_3:Eu	0.095	25	0.03
$CaWO_4$	0.05	30	0.04
ZnSCdS:Ag	0.207	34	0.04
NaI:Tl	0.10	-	-
Gd_2O_22S:Tb	0.20	30	0.03
La_2O_2S:Tb	0.18	30	0.03
Y_2O_2S:Tb	0.18	30	0.03
CsI:Na	0.10	-	-

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