Luminescence Efficiency of Gd_2O_2S:Eu Powder Phosphors as X-ray to Light Converter

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Abstract. *The aim of the present study was to examine the performance of the Gd_2O_2S:Eu powder scintillator for use in medical imaging applications. For this purpose various scintillator screens with coating thicknesses 33.1, 46.4, 63.1, 78.3 and 139.8 mg/cm^2 were prepared in our laboratory by sedimentation of Gd_2O_2S:Eu powder on silica substrates. Quantum Detection Efficiency (QDE) and Energy Absorption Efficiency (EAE) were evaluated. Light emission efficiency and optical emission spectra of the screens were measured under X-ray excitation using X-ray tube voltages (50–140 kVp) employed in X-ray radiography. Spectral compatibility with various optical photon detectors (photodiodes, photocathodes, charge coupled devices, films) were determined using emission spectrum data. Gd_2O_2S:Eu showed peak emission in the wavelength range 620–630 nm. The 139.8 mg/cm^2 phosphor screen appeared with the maximum light emission efficiency. Due to its reddish emission spectrum, Gd_2O_2S:Eu showed excellent compatibility with the sensitivity of many currently used photodetectors and could be considered for application in X-ray imaging especially in various digital detectors.*

Keywords: Luminescence Efficiency, Matching Factor, Powder Phosphors, Gd_2O_2S:Eu.

1 INTRODUCTION

Most radiation detectors, employed in x-ray radiography, consist of a scintillator/phosphor screen coupled to an optical detector (photographic emulsion film, photocathode, photodiode, etc.) [1]–[5]. Currently the most preferred phosphors are Gd_2O_2S:Tb and CsI:Tl. Gd_2O_2S:Tb has been proven very useful in conventional radiography screen-film systems. However, in digital radiography systems, based on crystalline silicon (Si) optical detectors (CCDs, photodiodes) the green light emitted by terbium-activated phosphors is not very efficiently detected [3], [6], [7]. This is because a large number of Si based devices, incorporated in X-ray imaging systems are not adequately sensitive to these wavelengths (500-550 nm) [3]. Since most Si based photodetectors are more sensitive in the red wavelength range, it would be of interest to investigate the emission efficiency of red emitting phosphors [6], [7].

For this purpose, europium (Eu)-activated phosphors, emitting at wavelengths towards the red region of the light spectrum could be used instead of green emitting,

Tb-activated phosphors. These red emitting phosphors show adequate matching with some films exhibiting high sensitivity to red light such as those used in laser imagers. Furthermore many europium doped scintillators, and particularly Gd_2O_2S:Eu, have been previously found comparable to terbium-activated phosphors [8]. In addition europium-activated phosphors have a decay time of the order of one millisecond (slightly higher than Gd_2O_2S:Tb) which is acceptable for applications that do not involve high framing rates [9], [10].

The aim of the present study was to investigate the response of Gd_2O_2S:Eu scintillator to X-rays employed in general X-ray radiography [11]. The principal criteria taken into account in the evaluation of Gd_2O_2S:Eu phosphor were: the X-ray quantum detection (QDE) and the X-ray energy absorption efficiency (EAE), the light emission efficiency, the spectrum of the emitted light and the spectral compatibility to optical detectors incorporated in medical imaging systems. To our knowledge the light emission properties of the Gd_2O_2S:Eu have not been previously systematically investigated under X-ray Radiography conditions.

2 MATERIALS AND METHODS

2.1 Experimental procedure

The screens, necessary for the experiments, were purchased in powder form (Phosphors Technology Ltd, England, code: UKL63/N-R1, with mean grain size of approximately 8 μm and density of 7.3 g/cm^3). The phosphors were used in the form of thin layers (test screens) to simulate the intensifying screens employed in X-ray Radiography. Five Gd_2O_2S:Eu thick scintillating screens, with coating thicknesses of 33.1, 46.4, 63.1, 78.3 and 139.8 mg/cm^2 were prepared by sedimentation of the powder phosphors on fused silica substrates (spectrosil B) [12], [13].

Experiments were performed on a Philips Optimus X-ray unit. Appropriate beam filtering (20mm Al) was applied to simulate X-ray beam hardening by human body [14].

The fluorescence light flux emitted by the X-ray excited screens was measured using an experimental setup comprising a light integration sphere (Oriel 70451) coupled to a photomultiplier (EMI 9798B) connected to a Cary 401 vibrating reed electrometer [14]. Tube voltage (ranging from 50 to 140 kVp) was checked using an RMI model 240 multifunction meter. Incident exposure rate measurements were performed using a Radcal 2026C ionization chamber dosimeter (Radcal Corp. USA).

2.2.1 Calculation of Radiation Detection parameters

2.2.1 Quantum Detection (QDE) and Energy Absorption Efficiency (EAE)

The efficiency of a scintillator to detect photons is described by the quantum detection efficiency (QDE). QDE is the fraction of incident photons interacting with the scintillator mass described previously [15].

For polyenergetic X-rays the QDE of a scintillator layer of coating thickness w is written as:

$$QDE(E) = \frac{\int_0^{E_0} \Phi_0(E)(1-e^{-(\mu_{tot,t}(E)/\rho)W})dE}{\int_0^{E_0} \Phi_0(E)dE} \quad (1)$$

where E_0 is the maximum energy of X-ray spectrum and $\mu_{tot,t}(E)/\rho$ is the X-ray total mass attenuation coefficient of the scintillator. $\Phi_0(E)$ is the X-ray photon fluence (photons per unit of area) incident on the scintillator.

X-ray imaging detectors are energy integrating systems, i.e., their output signal is proportional to the X-ray energy absorbed within the scintillator. Hence, when evaluating X-ray imaging systems, the calculation of the energy absorption efficiency (EAE) is also of importance. EAE may be calculated by the relation:

$$EAE(E) = \frac{\int_0^{E_0} \Phi_0(E)E\left(\frac{\mu_{tot,en}(E)}{\mu_{tot,t}(E)}\right)(1-e^{-(\mu_{tot,t}(E)/\rho)W})dE}{\int_0^{E_0} \Phi_0(E)EdE} \quad (2)$$

Ψ_0 is the incident X-ray energy fluence and $\mu_{tot,en}$ is the total mass energy absorption coefficient of the scintillator. $\mu_{tot,en}$ includes all mechanisms of energy deposition locally at the point of X-ray interaction within the scintillator's mass.

All secondary photons, e.g., K-characteristic fluorescence X-rays, created just after the primary interaction effect, are assumed to be lost. Thus EAE, being a measure of the locally absorbed energy, represents more accurately the efficiency of a detector to capture the useful X-ray imaging signal (i.e., the spatial distribution of primary X-ray absorption events). Attenuation and absorption coefficients were calculated using tabulated data [15].
In the present study both efficiencies were calculated as described in previous studies [16]. The required values of the total attenuation coefficients and the total energy absorption coefficient of Gd_2O_2S:Eu scintillator were calculated from tabulated data on energy absorption and attenuation coefficients of gadolinium, sulphur and oxygen [17].

2.2.2 Experimentally determined quantities

2.3.1 Absolute Efficiency

The light emission efficiency, of a phosphor, may be experimentally estimated under X-ray imaging conditions, by determining the absolute luminescence efficiency (AE) defined by equation (3):

$$\eta_A = \dot{\Psi}_\lambda / \dot{X} \quad (3)$$

where $\dot{\Psi}_\lambda$ is the emitted light energy flux (energy of light per unit of area and time), $\dot{X}$ is the incident exposure rate that excites the phosphor to luminescence. AE, is traditionally expressed, in units of $\mu W \times m^{-2} / (mR \times s^{-1})$, thereafter referred to as

efficiency units (E.U.). Light flux measurements were corrected for the spectral mismatches between the emitted light and the spectral sensitivity of the photocathode (extended S20) of the photomultiplier [14].

2.3.2 Emitted Light Spectrum & Spectral Compatibility

The emitted light spectra of the Gd_2O_2S:Eu powder phosphors were measured (under X-ray excitation) by an optical spectrometer (Ocean Optics Inc., HR2000). The light emitted by the irradiated powder Gd_2O_2S:Eu phosphor screens were transferred to the spectrometer through a 2.0 m long, 400 μm fiber optic, (Avantes Inc. FCB-UV400-2, Colorado, USA).

Since scintillators and phosphor screens are always used in combination with photodetectors (radiographic films, photodiodes, photocathodes etc), an estimation of the emitted light spectrum compatibility with the spectral sensitivity of photodetectors is required. This compatibility is often expressed by the spectral matching factor [13], αs, which can be calculated as in equation (4):

$$\alpha s = \int \phi\lambda(\lambda) SD(\lambda) d\lambda \Big/ \int \phi\lambda(\lambda) d\lambda \qquad (4)$$

where $SD(\lambda)$ is the normalized spectral sensitivity distribution of the photodetector used with the phosphor and $\phi\lambda$ is the emitted light spectrum.

3 RESULTS AND DISCUSSION

Calculated results based on tabulated data for the X-ray energy absorption and quantum detection efficiency for the Gd_2O_2S:Eu phosphor screens are shown in figures 1 and 2. Both EAE and DQE decreased with increasing energy and increased with increasing coating thickness. At low X-ray tube voltages, the thicker screen (139.8 mg/cm^2) absorbs relatively larger fractions of incident X-ray energy (e.g. 0.47 at 50 kVp). At higher voltages X-ray photons are more penetrating and X-ray energy absorption is lower (e.g. 0.30 at 90 kVp, 139.8 mg/cm^2 screen). QDE values (figure 2) are always higher than the corresponding values of EAE. This is due to the emerging K-fluorescence photons or scattered x-rays, which are not included in the EAE calculation [15], [17]. As it may be seen, for the 63.1 mg/cm^2 thick screen at 40 kVp, EAE (0.37) is approximately 12% lower than QDE (0.42). This difference increases with increasing X-ray tube voltage, being approximately 55% at 80 kVp (Figures 1 and 2) [15].

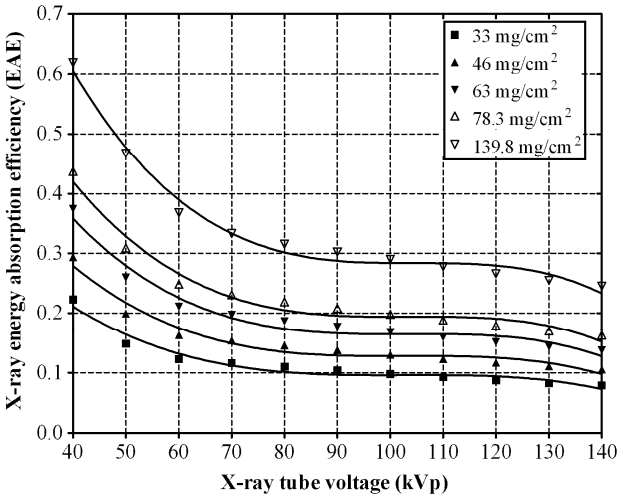

Figure 1 X-ray energy absorption efficiency (EAE) for the Gd_2O_2S:Eu screens in the radiographic energy range.

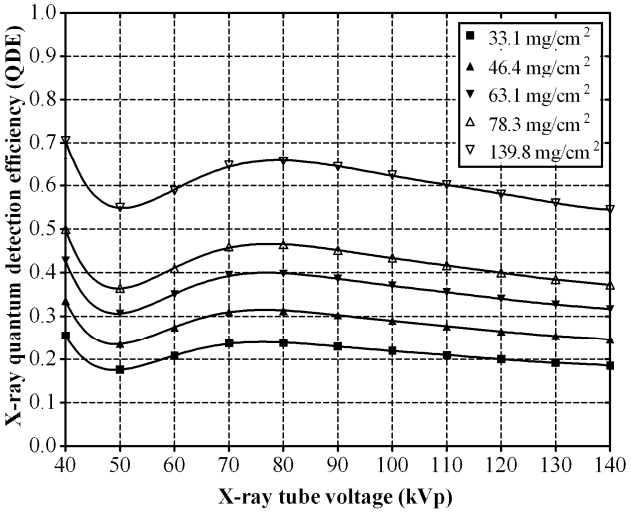

Figure 2 X-ray quantum detection efficiency (QDE) for the Gd_2O_2S:Eu screens in the radiographic energy range.

Figure 3 shows the variation of the absolute luminescence efficiency of Gd_2O_2S:Eu screens with X-ray tube voltage in the range from 50 to 140kVp. The fitted curve shown in figure 3 was obtained by a polynomial second order curve fitting. As it may be seen, absolute luminescence efficiency increases with increasing X-ray tube voltage until a maximum AE value (at 90 kVp) and decrease thereafter. This is due to the fact that as X-rays penetrate deeper in the phosphor mass, at higher voltages, light photons are created closer to the screen output and thus more easily transmitted through the phosphor grains. The 139.8 mg/cm^2 Gd_2O_2S:Eu screen showed the highest absolute luminescence efficiency values (e.g. 20.66 E.U. at 90kVp).

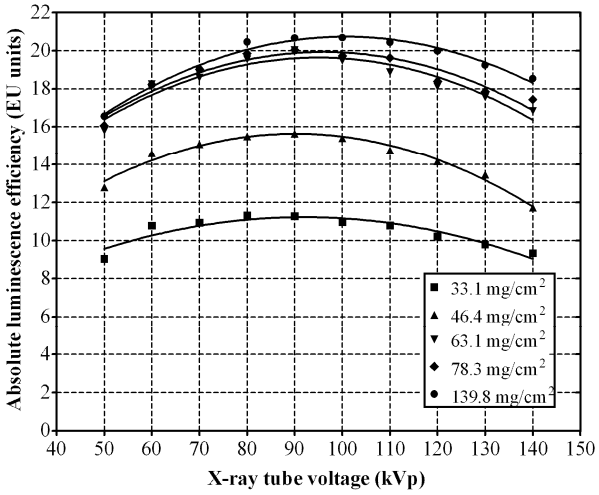

Figure 3 Variation of the absolute luminescence efficiency (AE) of the Gd_2O_2S:Eu powder phosphor screens with X-ray tube voltage. Points correspond to experimental values and solid line represent fitting curve. Efficiency units (E.U.): ($\mu W.s/mR.m^2$).

Figure 4 shows the measured light emission spectra of the 33.1, 46.4, 63.1, 78.3 and 139.8 mg/cm^2 Gd_2O_2S:Eu phosphor screens under X-ray excitation (100 kVp). The peak values of the light spectra of all the phosphor screens were found at 626 nm, which is attributed to the Eu^{3+} activator.

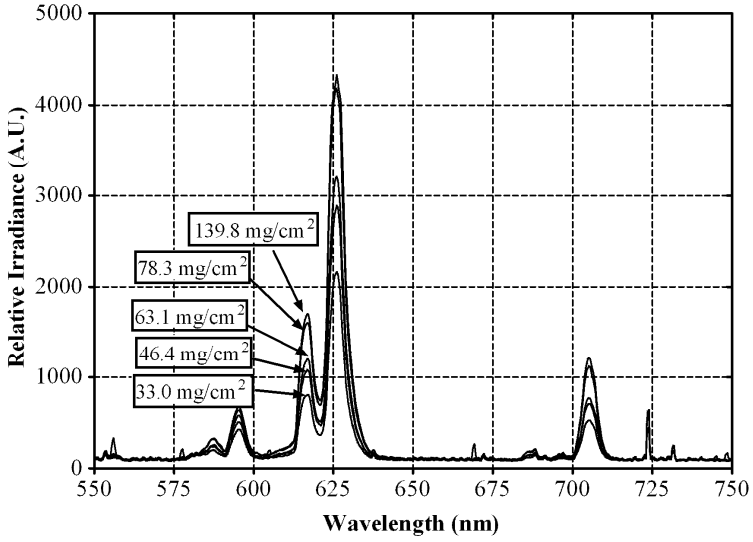

Figure 4 Optical emission spectra of Gd_2O_2S:Eu phosphor measured under X-ray excitation for the 33.1, 46.4, 63.1, 78.3 and 139.8 mg/cm^2 coating thicknesses scintillators.

Table 1 shows spectral matching factors data for the Gd_2O_2S:Eu scintillator combined with various red sensitive photodetectors. Gd_2O_2S:Eu has better spectral matching with CCD arrays than Gd_2O_2S:Tb. This result indicates that the red emitting Gd_2O_2S:Eu is more suitable for CCD based digital detectors. These data show that the

scintillator may be efficiently coupled to all practical types of red sensitive optical photodetectors both conventional and electronic. Gd_2O_2S:Eu could thus be considered for use in a variety of X-ray radiography detectors.

Optical Detectors	Gd_2O_2S:Eu	Gd_2O_2S:Tb
CCD S100AB SITe®	0.97	0.54
Agfa Scopix LT 2B	0.98	0.52
APD Hamamatsu S5343 M=50	1.00	
c-Si Photodiode	0.68	0.54
a-Si Photodiode	0.83	0.92
a-Si:H 104H Photodiode	0.85	0.90
a-Si:H 108H Photodiode	0.89	0.87
GaAsP Hamamatsu Photocathode	0.96	0.94
Extended S20 Photocathode	0.61	0.78

Table 1: Spectral matching factors.

4 CONCLUSIONS

In this study we examined the light emission performance of five Gd_2O_2S:Eu phosphors screens under X-ray radiography conditions. The luminescence efficiency, quantum detection efficiency and the energy absorption efficiencies of these screens were found adequately high to compete terbium and cerium doped scintillators in X-ray imaging. The measured light spectra showed an excellent compatibility to various electronic photodetectors used in digital imaging devices and to red sensitive films. These findings show that Gd_2O_2S:Eu could be of interest for applications in X-ray radiography detectors.

Acknowledgments

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REFERENCES

[1] Arnold, B.A., (1979), “Physical characteristics of screen-film combinations. In: Haus, A.G. (Ed.), The Physics of Medical Imaging: Recording System, Measurements and Techniques. American Association of Physicists in Medicine, New York, pp. 30-33.

[2] Liaparinos P, Kandarakis I, Cavouras D, Delis H, and Panayiotakis G. S. (2007), **"**Modelling granular phosphor screens by Monte Carlo methods**",** Med. Phys. Vol. 33 pp. 4502-4514.
[3] Gurwich, A.M., (1995), "Luminescent screens for mammography". Radiat. Meas. Vol.24, pp.325-328.
[4] Kandarakis I, Cavouras D, Panayiotakis G S, Agelis T, Nomicos C D and Giakoumakis G S (1996) **"**X-ray induced luminescence and spatial resolution of La_2O_2S:Tb phosphor screens**"**. Phys. Med. Biol. Vol. 41 pp. 297-307.
[5] Hendee W. R., (1970) *Medical radiation physics*, Medical Publishers, Chicago.
[6] Van Den Bergh R and Leblans P, (2005) **"**Rare earth activated lutetium oxysulfide phosphor for direct x-ray detection**"**, EP14 937 98.
[7] Rodnyi P A, Mikhrin S B, Mishin A N and Sidorenko A V, (2001) **"**Small-size pulsed x-ray source for measurements of scintillator decay time constants**"**, IEEE Trans. Nucl. Sci. Vol. 48 pp. 2340-2343.
[8] Okumura M, Tamatani M, Matsuda N, Takahara T and Fukuta Y, (2002) **"**Ceramic scintillator, method for producing same, and X-ray detector and X-ray CT imaging equipment using same**"**. U.S. Patent 6 384 417
[9] Nagarkar V V, Tipnis S V, Miller S R, Lempicki A, Brecher C, Szupryczynski P and Lingertat H, (2003) **"**A New X-Ray Scintillator for Digital Radiography**"**, IEEE Trans. Nucl. Sci. Vol. 50 pp. 297-300.
[10] Lempicki A, Brecher C, Szupryczynski P, Lingertat H, Nagarkar VV, Tipnis S V, Miller S R (2002) **"**A new lutetia-based ceramic scintillator for X-ray imaging**"**, Nucl. Instr. and Meth. A. Vol. 488 pp.579–590.
[11] Gambaccini M, Taibi A, Del Guerra A, Marziani M and Tuffanelli A, (1996) **"**MTF evaluation of a phosphor-coated CCD for x-ray imaging**"**, Phys. Med. Biol. Vol. 41 pp. 2799–2806.
[12] G E Giakoumakis, C D Nomicos, E N Yiakoumakis and E K Evangelou, (1990) "Absolute efficiency of rare earth oxysulphide screens in reflection mode observation", Phys. Med. Biol., , Vol. 35, No 7, pp.1017-1023.
[13] Kandarakis I, Cavouras D, Panayiotakis G S and Nomicos C D, (1997) **"**Evaluating x-ray detectors for radiographic applications: A comparison of ZnSCdS:Ag with Gd_2O_2S:Tb and Y_2O_2S:Tb screens**"**, Phys. Med. Biol. Vol. 42 pp. 1351-1373.
[14] Valais IG et al. (2005) **"**Luminescence Efficiency of Gd_2SiO_5:Ce Scintillator Under X-Ray Excitation**"**, IEEE Trans. Nucl. Sci. Vol. 52 pp. 1830-1835.
[15] Boone JM, (2000) **"***X-ray production, interaction, and detection in diagnostic imaging in Handbook of Medical Imaging***"**. Beutel J, Kundel H L and Van Metter R L, SPIE Press Bellingham.
[16] Cavouras D et al., (2005) **"**Light emission efficiency and imaging performance of $Y_3Al_5O_{12}$: Ce (YAG: Ce) powder screens under diagnostic radiology conditions**"**, Appl. Phys. B Vol. 80 pp. 923-933.
[17] Hubbel J H and Seltzer S M, (1995) **"**Tables of X-ray mass attenuation coefficients and mass energy absorption coefficients 1 keV to 20MeV for elements Z=1 to 92 and 48 additional substances of dosimetric interest**"**. US Department of commerce NISTIR 5632.