Luminescence Efficiency of fast Yttrium Aluminum Garnet Phosphor Screens for use in Digital Breast Tomosynthesis.

Stratos L. David[1], Christos M. Michail[1], Ioannis G. Valais[2], Ioannis Seferis[2], George Varaboutis[2], Stauros Gatsos[2], Adrianos E. Toutountzis[1], George Fountos[2], Ioannis S. Kandarakis[2], George S. Panayiotakis[1]

[1] *Department of Medical Physics, Medical School University of Patras 265 00 Patras, Greece*
[2] *Department of Medical Instruments Technology Technological Educational Institution of Athens Ag. Spyridonos, Aigaleo, 122 10 Athens, Greece*

Corresponding author: kandarakis@teiath.gr

Abstract. *In the present study scintillating screens prepared from $Y_3Al_5O_{12}$:Ce (YAG:Ce) powder phosphor were evaluated for use in digital mammography. YAG:Ce has never previously been used in X-ray medical imaging, however since it emits green light (i.e peak at 550nm), it is expected to match well the spectral sensitivities of most photodetectors (photodiodes, CCDs and amorphous silicon sensors) incorporated in various digital breast tomosynthesis detectors. YAG:Ce was purchased in powder form and was used in order to prepare test screens in the laboratory. Screens were evaluated by determining the quantum detection efficiency (QDE), the energy absorption efficiency (EAE), the absolute luminescence efficiency (AE), the X-ray luminescence efficiency (XLE), the light emission spectrum, the X-ray to light intrinsic conversion efficiency and the spectral compatibility with various digital photodetectors. Results were compared with phosphor materials commercially employed in X-ray imaging such as CsI:Tl and Gd_2O_2S:Tb. Maximum YAG:Ce emission efficiency was observed for the 63 mg/cm^2 screen at 49 kVp. The spectral compatibility with amorphous silicon photodiodes (0.93) and CCDs (0.95) was found to be very high, better than the corresponding compatibility of the CsI:Tl screen, mostly used in current digital radiography detectors. Taking into account the YAG:Ce overall performance, its short decay time as well as its spectral compatibility with amorphous silicon detectors and CCDs, YAG:Ce could be of interest for further investigation for applications in digital breast tomosynthesis detectors.*

Keywords: Luminescence Efficiency, Matching Factor, Powder Phosphors, Mammography.

1 INTRODUCTION

Digital X-ray breast tomosynthesis is a technique for producing slice images using conventional X-ray systems. It is a refinement of conventional geometric tomography, which has been known since the 1930 [1]. In conventional geometric tomography, the X-ray tube and image receptor move in synchrony on opposite sides of the patient to produce a plane of structures in sharp focus at the plane containing the fulcrum of the motion; all other structures above and below the fulcrum plane are blurred and thus less visible in the resulting image. Tomosynthesis improves upon conventional geometric tomography in that it allows an arbitrary number of in-focus planes to be generated retrospectively from a sequence of projection radiographs that are acquired during a single motion of the X-ray tube. Tomosynthesis requires short scanning time. Furthermore, in order to improve the time resolution of a detecting element, fluorescence just after X-ray irradiation stops (afterglow), should be low. For the purpose, both decay time constant of the fluorescence, and the afterglow intensity, should be low. Therefore, in order to get a high performance X-ray scanner with weak afterglow, it is necessary to use a fluorescent material with both small decay time constant and weak afterglow intensity. Up to now, CsI/a-Si flat-panel detector with ~1000ns primary decay is used for breast tomosynthesis.

Commercial YAG:Ce powder phosphor (it is often referred to as P-46 phosphor) exhibits a fast decay time of approximately 70 ns with no afterglow emission. It has a density of 4.15 g/cm3 comparable to other scintillators already in use (e.g., CsI, NaI, etc.). The effective atomic number of YAG:Ce is Zeff = 23.8, which combined with its density gives a rather low radiation detection index of ρZ4eff =1.33x104. Moreover, YAG:Ce is not hygroscopic characterized by mechanical and chemical stability and high radiation hardness [2]-[5]. Taking into account all the aforementioned characteristics, in the present work, YAG:Ce was tested as to its suitability for use in digital breast tomosynthesis detectors.

Up to now, for X-ray imaging applications, a thin YAG scintillator screen with a high resolution imaging CCD camera, working at 40 kV, has been developed by Czech researchers [6]. Moreover, some previous studied focused on the X-ray luminescence efficiencies of YAG:Ce phosphor screens under vacuum ultraviolet excitation as well as under X-ray excitation in the range of X-ray tube voltages (80–120 kVp) employed in general medical radiography and fluoroscopy have been carried out [7]-[9].

To simulate in the present study, the intensifying screens used in digital breast tomosynthesis detectors, the phosphor material was used in the form of test screens of various thickness prepared in our laboratory [9]-[11]. For the evaluation of these screens, the following parameters were examined:

1. The X-ray detection, i.e. the quantum detection efficiency (QDE) and the energy absorption efficiency (EAE) as a function of X-ray tube voltage [7], [11]. QDE is the fraction of incident photons interacting within the scintillator mass. X-ray imaging detectors are energy integrating systems, i.e., their output signal is proportional to the X-ray energy imparted within the scintillator mass. Hence, when evaluating X-ray imaging systems, the calculation of the energy absorption efficiency (EAE) is also of significance. EAE is a measure of the locally absorbed energy, representing more accurately the efficiency of a detector to capture the useful X-ray imaging signal (i.e., the spatial distribution of primary X-ray photon interaction events) [11].
2. The absolute luminescence efficiency (AE). AE is defined by the ratio of the energy flux emitted by an excited phosphor screen over the incident X-ray exposure rate and expresses the ability of the phosphor to convert incident X-ray exposure into emitted light energy [7], [9]-[12].
3. The X-ray luminescence efficiency (XLE). XLE is a unit less quantity defined by the ratio of the energy flux emitted by an excited phosphor screen over the incident X-ray energy flux. XLE is suitable for evaluation of energy integrating X-ray imaging detectors, since it is directly related to the scintillator' energy conversion properties giving the fraction of incident radiation energy converted into emitted light. This quantity is independent from air absorption and air

ionization contribution, necessary to convert the exposure rate, into energy flux [11].

4. The spectral compatibility (a_s) of YAG:Ce to various photodetectors. It describes the degree of coincidence between the optical spectrum of the phosphor and the spectral sensitivity of the photodetector [13], [14].
5. The intrinsic conversion efficiency (η_c) expressing the fraction of absorbed X-ray energy converted into light within the mass of the phosphor [2].

2 MATERIALS AND METHODS

YAG:Ce was supplied in powder form (Phosphor technology, UK) with mean grain size of approximately 8μm. Five screens (thin layers) with coating thicknesses ranging from 13 to 107 mg/cm2 were prepared in laboratory by sedimentation of the powder on fused silica substrates. During the sedimentation process, sodium orthosilicate (Na2SiO3) was used as binding material between the powder grains [7], [10].

To determine AE and a_s, the phosphor screens were exposed to X-rays using a Senographe DMR G.E. X-ray mammography unit with molybdenum anode target and molybdenum filter. In this system the filter changes automatically from molybdenum to rhodium (Rh) and aluminum (Al) as tube voltage increases. The X-ray beam was filtered by an additional 30 mm thick block of Perspex in order to simulate beam hardening by human breast. The X-ray tube current was set at 10 mA. The molybdenum filter of the mammography unit was 30μm thick, the rhodium filter 25 μm and the aluminum filter 1mm. The distance between the X-ray tube and the YAG:Ce screens was 30 cm. X-ray tube voltage varied from 28 to 49 kVp. Tube voltage accuracy 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). AE is often expressed in units of $\mu Wm^{-2}/(mRs^{-1})$ thereafter referred to as the efficiency units (E.U.). Alternatively the emission efficiency of a scintillator may be expressed by employing the definition of the X-ray luminescence efficiency (XLE):

$$XLE(kVp, w_0) = \Psi_\lambda(kVp, w_0)/\Psi \qquad (1)$$

where Ψ is the incident X-ray energy flux, which may be determined from exposure measurements using the factor for converting exposure rate into energy flux [15], [16]. Both Ψ_λ and Ψ are expressed in μWm^{-2}.

The optical measurements required for the absolute efficiency and the spectral compatibility determination, were performed as follows:

The emitted light energy flux was measured by a photomultiplier (EMI 9798 B) coupled to a Cary 401 vibrating reed electrometer. Light was observed from the non-irradiated screen side (transmission mode technique or front screen configuration) as used in digital mammography detectors. To determine light energy flux accurately, the experimental data were corrected dividing by: (a) the spectral compatibility between the YAG:Ce phosphor light and the spectral sensitivity of the photomultiplier's photocathode (extended S-20); and (b) the geometric light collection efficiency of the phosphor-photomultiplier experimental arrangement. This efficiency depends on the angular distribution of the emitted light [17], and on the distance between the phosphor layer and the photomultiplier. The light emission spectrum of the YAG:Ce phosphor was measured by an Ocean Optics S2000 spectrometer. The spectrometer was optically coupled to the screen via a fiber optic (Avantes Inc. FCB-UV400-2).

The spectral compatibility, a_s, was then determined using the following relation [13]-[14]:

$$\alpha_s(\Delta\lambda) = \int_{\lambda_1}^{\lambda_2} \phi(\lambda) S_{OD}(\lambda) d\lambda \Bigg/ \int_{\lambda_1}^{\lambda_2} \phi(\lambda) d\lambda \qquad (2)$$

where $\phi(\lambda)$ denotes the emitted light spectrum and S_{OD} is the spectral sensitivity of the optical detectors. Equation (2) describes how well light may be captured by optical detectors.

In addition the above emission spectrum was employed to determine the intrinsic conversion efficiency η_c as follows [2]:

$$\eta_C = \frac{E_\lambda}{\beta E_g} Sq = \frac{2\pi\hbar c}{\lambda \beta E_g} Sq \qquad (3)$$

where $E_\lambda = 2\pi\hbar c / \lambda$ expresses the mean energy of the emitted light photon, c is the light velocity, $\hbar$ is the Planck's constant, λ is the mean light wavelength, determined by the light emission spectrum measurements. E_g denotes the energy gap separating the valence and conduction energy bands (for YAG, E_g=7,01 eV) [2], [7]. βE_g expresses the average energy required to create an electron-hole pair [2], which is then followed by light creation in the phosphor. This energy is often larger than E_g. For YAG:Ce it has been previously estimated that β=5.6 [2], [7], [9]. S is the probability of energy transfer from the electron-hole pairs to the centers of light creation (luminescent centers) within the phosphor. q is the quantum efficiency of these centers, e.g. the efficiency to absorb the energy transferred from the electron-hole pairs and convert it into light photons. It has previously been estimated that for Ce activated phosphors, the probability that electron-hole pairs are converted to light photons, Sq in relation (3), is close to unity [7], [9].

3 RESULTS AND DISCUSSION

In figure1 the response characteristic curve (output signal versus incident exposure) of the 63 mg/cm^2 YAG:Ce phosphor screen is plotted and shows a linear dependence between output signal and exposure rate in the 0.2 - 5 mR/s range. In figure 2 the variation of calculated QDE and EAE with X-ray tube voltage is shown for the three most efficient (33, 42 and 63 mg/cm^2) YAG:Ce screens in the energy range up to 40 kVp, which is mainly used in digital mammography. QDE values were found to be higher than the corresponding values of EAE. This is due to the emerging K-fluorescence photons or scattered X-rays that do not lead to energy deposition and thus are not included in the EAE calculation. For 28 kVp tube voltage, data from figure 2 show that only a fraction ~ 50% of detected photons (QDE) are locally imparted (EAE) meaning that the fraction contributing to accurate spatial registration of interacting photons is small. This is due to the low K-absorption energy edge of yttrium element at ~17 keV, which thus allows the generation of a relatively large amount of K-fluorescence photons within the mammography energy range.

Figure 3 shows the variation of the absolute luminescence efficiency of YAG:Ce with respect to the X-ray tube voltage for screens of various coating thickness. In the tube voltage range, from 28 to 49 kVp, the absolute luminescence efficiency increases significantly with increasing X-ray tube voltage. An interesting point to notice is that a sharp increase was observed at 42 kVp. This was attributed to the change of the mammography tube filter, Rh instead of Al. On the contrary, no significant variation was observed when the filter changed from Mo to Rh at lower energies (38 kVp).

Figure 4 shows the variation of the absolute luminescence efficiency of YAG:Ce with coating thickness for various X-ray tube voltages. For the scintillator screens of 33, 42, and 63 mg/cm2 coating thickness the absolute luminescence efficiency values are very similar at least in the range of X-ray tube voltages used in mammography daily practice (25-35 kVp).

Figure 5 shows the X-ray luminescence efficiency (XLE) of the 33, 42, and 63 mg/cm^2 coating thickness YAG:Ce powder phosphor screens with X-ray tube voltage in the range from 20 up to 40 kVp. Data shown were obtained using the definition of XLE in (1) and by converting exposure rate into energy fluence using the conversion factor:

$$X/\Psi(E) = (\mu_{en}(E)/\rho)_{air}/W_A/e \tag{4}$$

where $(\mu_{en}/\rho)_{air}$ is the X-ray mass energy absorption coefficient of air at energy E, and (W_A/e) is the average energy per unit of charge required to produce an electron - ion pair in air.

The values of (W_A/e) and $(\mu_{en}/\rho)_{air}$ were obtained from the literature [16]. In contrast to AE, defined in terms of incident exposure, which in turn depends on the energy dependent X-ray absorption and ionization properties of air, the shape of the XLE curves is strictly affected by the intrinsic properties of a screen, i.e. the energy absorption properties of the phosphor material and by the light transmission through the screens. Maximum XLE value (6.4x10-4) was observed for the screen of 33 mg/cm2 at 25 kVp. This result may be explained by considering that low energy X-rays do not penetrate deeply within the scintillator mass. Hence light photons, which are mainly created very close to the input screen surface, are forced to travel long trajectories to escape the rear surface of the screen. This increases the light attenuation (self absorption) within the scintillator mass in the case of the two thicker screens of 42 and 63 mg/cm2. On the other hand at higher voltages, X-rays penetrate deeper in the phosphor mass, light photons are thus created deeper, or more close to the screen output. Light is thus more easily transmitted through the phosphor grains and hence, at 30 kVp, the screen of 42 mg/cm2 showed higher XLE (6.1x10-4) than the other two screens.

Figure 6 shows a plot of the YAG:Ce emission spectrum measurements including seven spectral sensitivity curves corresponding to optical detectors currently used in a large variety of digital X-ray detectors. The optical detectors considered were the following: 1. Amorphous silicon photodiodes (a-Si) for layer thickness of 400nm (I04H) and 800nm (I08H), 2. Silicon photodiodes. The BR type uses a resin-coated window and the S1133 is a visible photodiode with a visual-compensated filter [18], 3. S100AF and S100AB Charge Coupled Devices (CCDs) and CCD with broadband anti-reflection (AR) and IR – AR coating which are employed in digital radiography, computed tomography, crystallography e.t.c. [SITe®], 4. High resolution CMOS digital detectors with hybrid and monolithic coating. As it is observed from figure 4, the YAG:Ce spectrum is lying well within the sensitivity limits of S100AB CCD, I08H a-Si:H, S1133 photodiode and CMOS hybrid with blue AR coating.

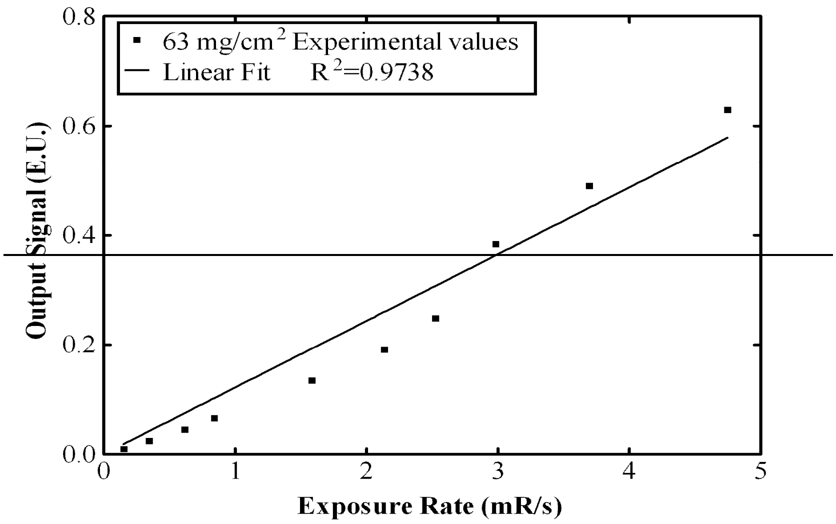

Figure 1 Characteristic curve of the 63 mg/cm2 YAG:Ce phosphor screen in the mammography range of exposures. Dots correspond to the measured values; solid line is a linear no threshold fit to the measured data.

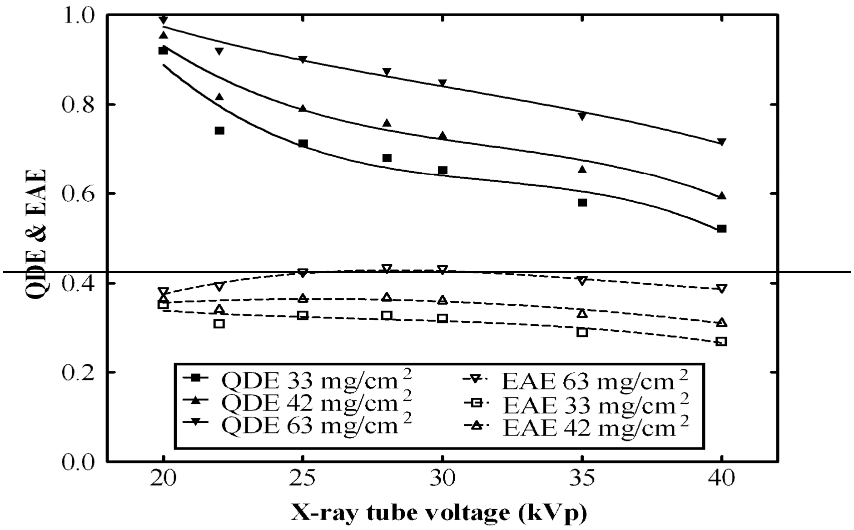

Figure 2 Variation of the quantum detection efficiency (QDE) and the energy absorption efficiency (EAE) of YAG 33, 42, and 63 mg/cm2 screens with tube voltage (in the range 20-40 kVp).

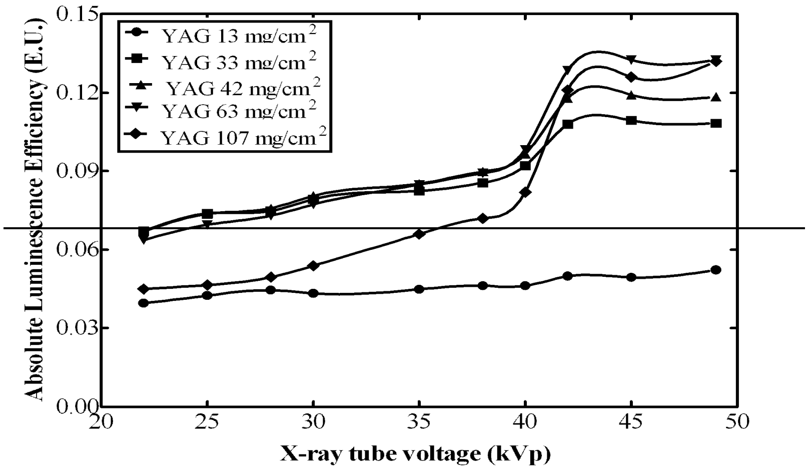

Figure 3 Variation of the absolute luminescence efficiency (AE) of YAG:Ce screens with tube voltage.

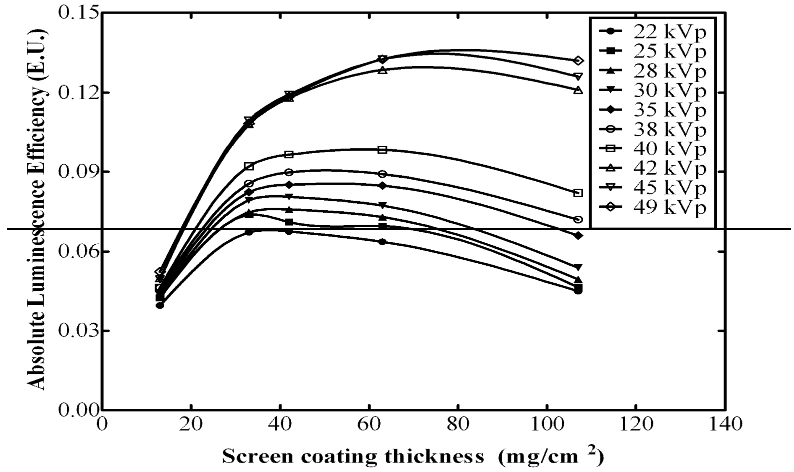

Figure 4 Variation of the absolute luminescence efficiency (AE) of YAG:Ce screens with coating thickness.

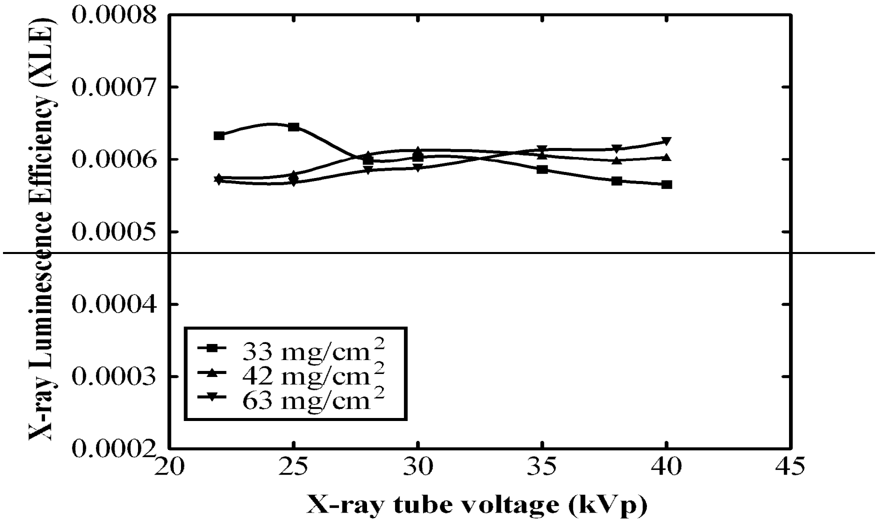

Figure 5 The X-ray luminescence efficiency (XLE) of 33, 42 and 63 mg/cm2 coating thickness YAG:Ce screens as determined from experimental AE data, for X-ray tube voltages between 20 - 40 kVp.

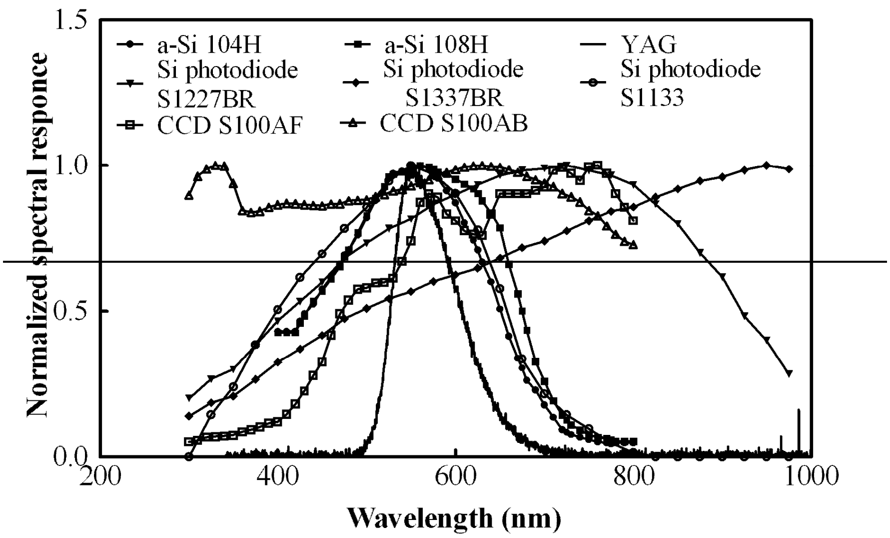

Figure 6 Normalized light emission spectrum of YAG:Ce phosphor and spectral sensitivities of various optical detectors.

The value of the intrinsic conversion efficiency (η_c), as estimated by the equation (3), was found η_c=0.05. This value is higher than the corresponding value of CaWO4 (η_c = 0.04), which is used in conventional radiographic cassettes, but lower than that of rare earth phosphors (η_c = 0.15) [19].

Optical detectors	Phosphors		
	YAG:Ce	CsI:Tl	Gd_2O_2S:Tb
a-Si:H I04H	0.88	0.77	0.91
a-Si:H I08H	0.93	0.84	0.93
S1227-BR HAMAMATSU	0.85	0.81	0.79
S1337-BR HAMAMATSU	0.60	0.56	0.55
S1133 HAMAMATSU	0.88	0.81	0.91
CCD S100AF SITe®	0.78	0.69	0.67
CCD S100AB SITe®	0.95	0.93	0.92
CCD (broadband AR coating)	0.79	0.81	0.80
CCD (IR AR coating)	0.71	0.68	0.67
CMOS (hybrid with NIR AR coating)	0.79	0.77	0.74
CMOS (monolithic 0.25μ)	0.61	0.63	0.60
CMOS (hybrid with Blue AR coating)	0.85	0.85	0.83

Table 1: Spectral compatibility of YAG:Ce, CsI:Tl and Gd_2O_2S:Tb with optical detectors

Table 1 shows the values of the spectral compatibilities, corresponding to the aforementioned detectors, calculated using (2). In the same table for comparison reasons, values related to the very often-employed CsI:Tl and the rare earth phosphor (Gd_2O_2S:Tb) are also shown. YAG:Ce shows very high spectral compatibility with CCD S100AB, which is also higher than CsI:Tl and Gd_2O_2S:Tb. Additionally, YAG:Ce is very well compatible with the a-Si:H I08H. Although the emission efficiency, and in particular the intrinsic conversion efficiency of this scintillator is not better than that of currently employed materials, its high spectral compatibility with amorphous silicon photodiodes, CMOS and CCD sensors, as well as its very short response time, may raise the interest to further investigate YAG:Ce for possible incorporation into some novel type of digital mammography detectors, such as scanning slot devices or devices dedicated to breast tomosynthesis techniques.

4 CONCLUSIONS

In the present study, YAG:Ce powder scintillator screens of 13, 33, 42, 63 and 107 mg/cm^2 coating thickness were prepared and examined under X-ray mammographic conditions. The intrinsic conversion efficiency – as estimated by (3) and the literature data - was found close to that of $CaWO_4$ phosphor but lower than that of CsI:Tl and GOS:Tb. The experimentally measured emission spectrum of YAG:Ce powder scintillator was found to peak at approximately 550-555 nm and showed very good compatibility with current electronic optical sensors (CCDs, CMOS and photodiodes). Maximum experimental absolute luminescence efficiency values were obtained for the screen of 63 mg/cm^2 using tube voltages varying from 43 to 49 kVp. In the range of X-ray tube voltages usually employed in mammography (25-35 kVp) the two thinner YAG:Ce of 33 and 42 mg/cm^2 showed the highest luminescence efficiency. Assuming that the thinner the phosphor coating the better the resolution properties of a screen, we could select the 33 mg/cm^2 screen as more appropriate for mammography applications. Our data on YAG:Ce screens indicate that further investigations, regarding the suitability of this scintillator for use in various digital mammography detectors, especially those requiring fast response, could be of interest.

Acknowledgments

The authors gratefully acknowledge Dr. Dimitropoulos from the department of Medical Imaging, “Euromedica” Medical center for providing medical equipment.

REFERENCES

[1] J. T. Dobbins, D. J. Godfrey, (2003), “Digital X-ray tomosynthesis: current state of the art and clinical potential” Phys. Med. Biol. vol. 48, pp.65-106.
[2] G. Blasse and B.C. Grabmaier, (1994), Luminescent Materials. Berlin, Germany: Springer.
[3] W. E. Van Eijk, (2002) “Inorganic scintillators in medical imaging,” Phys.Med.Biol., vol. 47, pp. R85-R106.
[4] M. J. Yaffe and J. A. Rowlands, (1997), “X-ray detectors for digital radiography,” Phys. Med. Biol., vol.42, pp. Rl.

[5] M. Nikl, (2006), "Scintillation detectors for X-rays," Meas. Sci. Technol., vol. 17, pp. 37-54.
[6] J. Touš, K. Blažek, L. Pína, B. Sopko, (2007), "High-resolution X-ray imaging CCD camera based on a thin scintillator screen," Rad. Meas., vol.42, pp. 925-928.
[7] D. Cavouras, I. Kandarakis, D. Nikolopoulos, I. Kalatzis, G. Kagadis, N. Kalivas, A. Episkopakis, D. Linardatos, M. Roussou, E. Nirgianaki, D. Margetis, I. Valais, I. Sianoudis, K. Kourkoutas, N. Dimitropoulos, A. Louizi, C. Nomikos, G. Panayiotakis, (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.
[8] A. Baciero, L. Placentino, K. J. Mc Carthy, L. R. Barquero, A. Ibarra, B. Zurro, (1999), "Vacuum ultraviolet and X-ray luminescence efficiencies of $Y_3Al_5O_{12}$:Ce phosphor screens," J. Appl. Phys., vol. 85, pp.6790-6795.
[9] I. Kandarakis, D. Cavouras, I. Sianoudis, D. Nikolopoulos, A. Episkopakis, D. Linardatos, D. Margetis, E. Nirgianaki, M. Roussou, P. Melissaropoulos, N. Kalivas, I. Kalatzis, K. Kourkoutas,N. Dimitropoulos, A. Louizi, C. Nomicos, G. Panayiotakis, (2005), "On the response of $Y_3Al_5O_{12}$: Ce (YAG: Ce) powder scintillating screens to medical imaging X-rays," Nucl. Instr. Meth. Phys. Res A, vol.538, pp. 615–630.
[10] N. Kalivas, I. Valais, D. Nikolopoulos, A. konstantinidis, A. Gaitanis, D. Cavouras, C. Nomicos, G. Panayiotakis, I. Kandarakis, (2007), "Light emission efficiency and imaging properties of YAP:Ce granular phosphor screens," Appl. Phys. A., vol. 89, pp. 443-457.
[11] S. David, C. Michail, I. Valais, A. Toutountzis, D. Cavouras, I. Kandarakis, G. Panayiotakis, (2008) "Investigation of luminescence properties of Lu_2SiO_5:Ce (LSO) powder scintillator in the X-ray radiography energy range," IEEE Trans. Nucl. Sci., Vol.55, No. 6, pp.3684-3691.
[12] S. David, C. Michail, I. Valais, D. Nikolopoulos, P. Liaparinos, N. Kalivas, I. Kalatzis, A. Toutountzis, N. Efthimiou, G. Loudos, I. Sianoudis, D. Cavouras, N. Dimitropoulos, C. D. Nomicos, I. Kandarakis and G. S. Panayiotakis, (2007) "Efficiency of Lu_2SiO_5:Ce (LSO) powder phosphor as X-ray to light converter under mammographic imaging conditions," Nucl. Instrum. Methods Phys. Res. A. vol. 571, pp. 346-349.
[13] G.E. Giakoumakis, (1991) "Matching factors for various light-source Photodetector combinations," Appl. Phys., vol.52, pp.7-9.
[14] I. Kandarakis, D. Cavouras, C. D. Nomicos. and G. S. Panayiotakis, (2001), "Measurement of the X-ray luminescence and spectral compatibility of the $CdPO_3CI$:Mn phosphor," Rad. Meas. vol.33, pp. 217-224.
[15] J. M. Boone, (2000), "X-ray production, interaction, and detection in diagnostic imaging," in Handbook of Medical Imaging, vol. 1, J. Beutel, H. L. Kundel, R. L. Van Metter, Ed. Bellingham, WA: SPIE Press Physics and Psycophysics, , pp. 40-67.
[16] J. R. Greening, Fundamentals of Radiation Dosimetry. London, U.K.: Instit. Phys., 1985.
[17] G. E. Giakoumakis, D. M. Miliotis, (1985), "Light angular distribution of fluorescent screens excited by X-rays," Phys. Med. Biol., vol. 30, pp. 21-25.
[18] Hamamatsu Corporation-photonics [Online]. Available: http://sales.hamamatsu.com/assets/pdf/parts_S/
[19] R. M. Nishikawa., M. J. Yaffe, (1990), "Model of the spatial-frequency

dependent detective quantum efficiency of phosphor screens," Med. Phys., vol. 17, pp. 894-904.