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Evaluating scintillators used in radiation detectors of medical imaging systems by the effective fidelity index method

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Abstract

Objective: The performance of medical X-ray image receptors depends: (1) on the scintillator light emission efficiency; and (2) on the compatibility of the scintillator light spectrum with the spectral sensitivity of the light detector (film, photocathode, or photodiode), employed in conjunction with the scintillator. In this study, a scintillator performance measure, the effective fidelity index (EFI), is defined as function of both the scintillator light emission efficiency and spectral compatibility. *Materials and Method:* CsI:Na, Gd_2O_2S:Tb and La_2O_2S:Tb scintillators were employed in the form of phosphor screens prepared in our laboratory with various coating thicknesses. The screens were irradiated with X-rays employing tube voltages ranging between 50–120 kVp. *Results:* The EFI performance of CsI:Na was found to increase with screen coating thickness and it was best when combined with the orthochromatic film or the ES/20 photocathode. Gd_2O_2S:Tb showed peak EFI performance at 70 mg/cm^2 coating thickness and it was well combined with the light detectors considered. *Conclusion:* In accordance with our results, CsI:Na may be employed in radiography when adequately protected against humidity. Gd_2O_2S:Tb suitability for conventional imaging was verified and it was found that it may be useful in all types of digital imaging. La_2O_2S:Tb could also be used in digital detectors of imaging applications demanding medium X-ray tube voltages.

Keywords: Scintillators; Phosphor screens; X-ray luminescence

1. Introduction

X-ray or γ-ray medical image receptors comprise a scintillator (phosphor) coupled to a light detector. The latter may be a film in a radiographic cassette, the photocathode of fluoroscopic image intensifiers or nuclear medicine photomultipliers, the photodiodes of digital radiography or of computed tomography detectors. The performance of a scintillator is usually assessed by its efficiency to emit light of high intensity and of suitable spectrum to match the sensitivity of light detectors.

In the present study, the effective fidelity index (EFI) of scintillators is defined, based on Linfoot's definition [1] of image fidelity (see Appendix A). EFI is an index that compares the light emitted by a scintillator and matched to the sensitivity of a light detector against the light of an ideal scintillator—light detector system. The latter may be defined as a detector absorbing all incident X-rays and converting all their energy into emitted light, which is perfectly matched to the light detector. EFI measurements were performed at various X-ray

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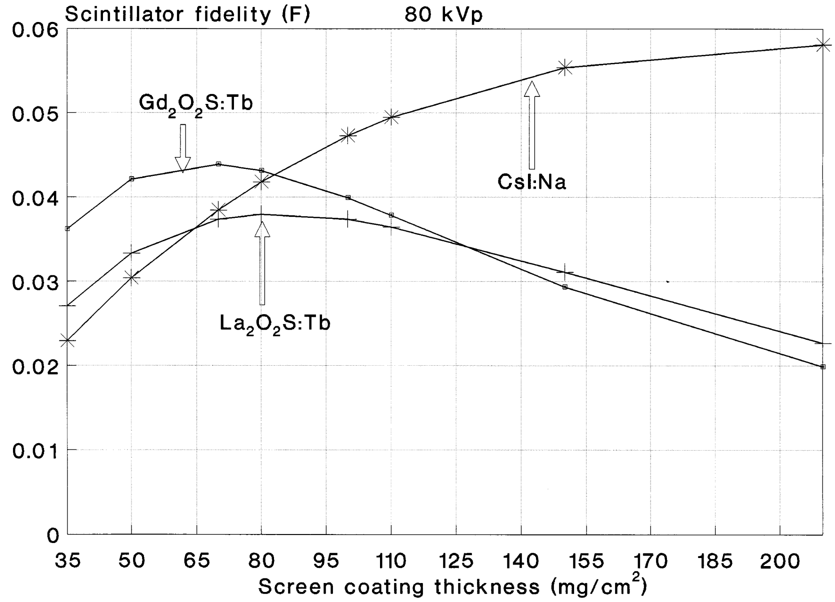

Fig. 1. Variation of the scintillator fidelity of CsI:Na, Gd_2O_2S:Tb and La_2O_2S:Tb scintillators with screen coating thickness. Measurements were obtained at 80 kVp.

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tube voltages and using laboratory prepared test scintillating screens of various thicknesses in order to compare three of the most efficient scintillator materials used in X-ray image receptors, namely CsI:Na, Gd_2O_2S:Tb and La_2O_2S:Tb. CsI:Na is usually employed in image intensifiers and Gd_2O_2S:Tb and La_2O_2S:Tb in radiographic cassettes. It was also investigated the usefulness of these materials in other medical imaging systems.

2. Materials and methods

Scintillators are employed in X-ray imaging in the form of fluorescent layers often called phosphor screens. The fidelity of an imaging scintillator (see Appendix A) can be defined as:

$$F = 1 - \frac{[J_{\mathrm{ideal}} - J_{\mathrm{L}}]^2}{J_{\mathrm{ideal}}^2} \quad (1)$$

where J_{L} is the light intensity emitted by the scintillator for a given level of incident X-ray intensity and J_{ideal} is the corresponding intensity emitted by an ideal scintillator. According to the definition of an ideal scintillator (see Appendix A), J_{ideal} must be numerically equal to the intensity of the X-rays falling on the scintillator surface.

The quantity of light utilized to form the final image depends on the compatibility between the spectrum of the emitted light and the spectral sensitivity of the light detector. To account for the spectral compatibility effects, J_{L} must be multiplied by the spectral matching factor a_{SP}. The latter expresses the fraction of light spectrum that can sensitize the light detector (see Appendix B). Hence, the effective fidelity index of a scintillator can be defined as follows:

$$\mathrm{EFI} = 1 - \frac{[J_{\mathrm{ideal}} - J_{\mathrm{eff}}]^2}{J_{\mathrm{ideal}}^2} \quad (2)$$

where,

$$J_{\mathrm{eff}} = J_{\mathrm{L}} \times a_{\mathrm{SP}} \quad (3)$$

EFI was experimentally determined for the CsI:Na, Gd_2O_2S:Tb and La_2O_2S:Tb scintillators. A total of eight test screens from each material were prepared in our laboratory with coating thicknesses ranging from 35 to 210 mg/cm^2. Gd_2O_2S:Tb and La_2O_2S:Tb materials were in powder form with mean grain size of about 7 μm. The screens were prepared by sedimentation using a previously described technique [2–5]. CsI:Na screens were prepared by evaporation of pure CsI with a quantity of NaI in order to activate the pure material with Na. These screens were almost compact with $\approx 90\%$ packing density and their internal structure consisted of needle like columns. Light photons were channeled out along these columns, and thus, light intensity attenuation was minimized. However, CsI:Na is very hygroscopic demanding appropriate protection from humidity. On the other hand, Gd_2O_2S:Tb and La_2O_2S:Tb screens being of granular structure show increased light scattering and attenuation phenomena but they are not affected by humidity.

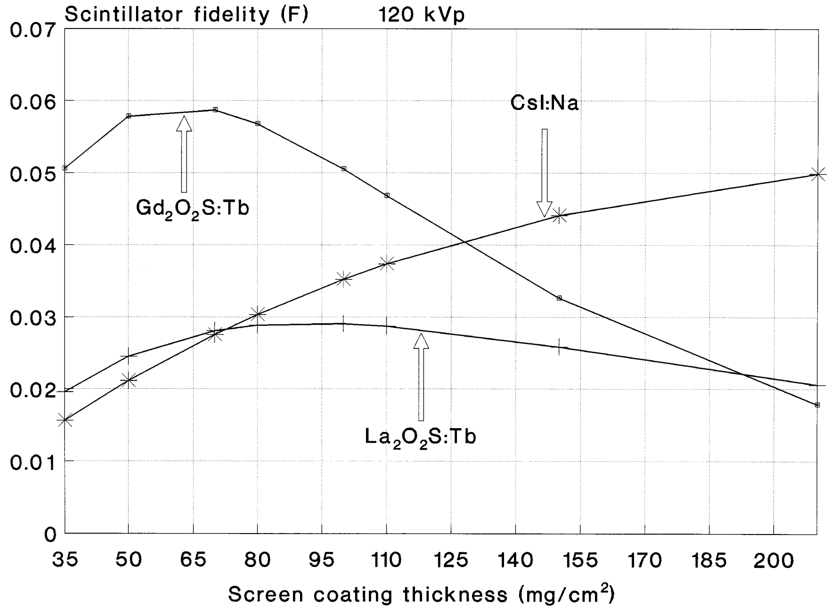

Fig. 2. Variation of the scintillator fidelity of CsI:Na, Gd_2O_2S:Tb and La_2O_2S:Tb scintillators with screen coating thickness. Measurements were obtained at 120 kVp.

The screens were exposed to X-rays employing tube voltages from 50 to 120 kVp. The emitted light intensity J_L was measured by a photomultiplier (EMI 9558 QB) interfaced to an electrometer (Cary 401). J_{ideal} was determined from exposure measurements, since it was considered numerically equal to the incident X-ray intensity. Exposure rate $\dot{X}$ was measured by a PTW dosemeter and was converted into X-ray intensity (energy flux) [6,7] using the following relation:

$$J_{ideal} = \dot{X} \left\{ \left[\frac{\mu_{en}}{\rho} \right]_{air} \frac{e}{W_{air}} \right\}^{-1} \tag{4}$$

where, $[\mu_{en}/\rho]_{air}$ is the mass energy absorption coefficient of air, e is the electron charge and W_{air} is the average energy required to produce an ion pair in air. To simulate clinical conditions, the incident X-ray beam was filtered by an additional 20 mm aluminum filter to approximate X-ray attenuation by the patient's body.

J_{eff} was calculated according to Eq. (3). To determine the spectral matching factor a_{SP}, the spectrum of the emitted light and the spectral sensitivity of the light detectors had to be known (see Appendix B). Light spectra were measured in our laboratory by an Oriel 7240 grating monochromator and spectral sensitivities were obtained from manufacturers' data. The light detectors considered were the Agfa Curix Ortho GS orthochromatic film, often used in radiographic cassettes, the ES/20 photocathode, often used in image intensifiers and photomultipliers, and the silicon (Si) photodiode used in CCD arrays of digital radiography or in solid state detectors of CT systems.

3. Results and discussion

Fig. 1 shows the variation of the scintillator fidelity (***F***) of CsI:Na, Gd_2O_2S:Tb and La_2O_2S:Tb screens with coating thickness. Measurements were performed at 80 kVp. For screens up to 80 mg/cm^2, the performance of Gd_2O_2S:Tb was found better than CsI:Na or La_2O_2S:Tb. For screens thicker than 70 mg/cm^2, the fidelity of Gd_2O_2S:Tb decreased continuously due to the increased light attenuation effects within the material. The shape of the La_2O_2S:Tb *F*-curve was found similar to that of Gd_2O_2S:Tb. However, the La_2O_2S:Tb peak fidelity was reached at 80 mg/cm^2 while, for thicker screens, its rate of decrease was slightly lower than that of Gd_2O_2S:Tb. The fidelity of CsI:Na was found to increase continuously with screen thickness. For screens thicker than 100 mg/cm^2 the fidelity of CsI:Na was significantly higher than the fidelities of Gd_2O_2S:Tb or La_2O_2S:Tb. Fig. 2 shows the *F*-curves of the same screens at 120 kVp. *F*-curves were found similar in shape to those obtained at 80 kVp. However, at 120 kVp, Gd_2O_2S:Tb was much more efficient than La_2O_2S:Tb or CsI:Na for screens up to 110 mg/cm^2. The shape of the *F*-curves in both Figs. 1 and 2 may be explained by considering that in Gd_2O_2S:Tb and La_2O_2S:Tb, the fidelity initially increases due to an increase in X-ray absorption with thickness. However, after a maximum value is attained the fidelity starts to decrease because light attenuation becomes more important at thick screens, mainly due to light scattering effects on phosphor grains. On the other hand, light scattering is minimized in CsI:Na due to the almost compact, non-granular, structure of this material,

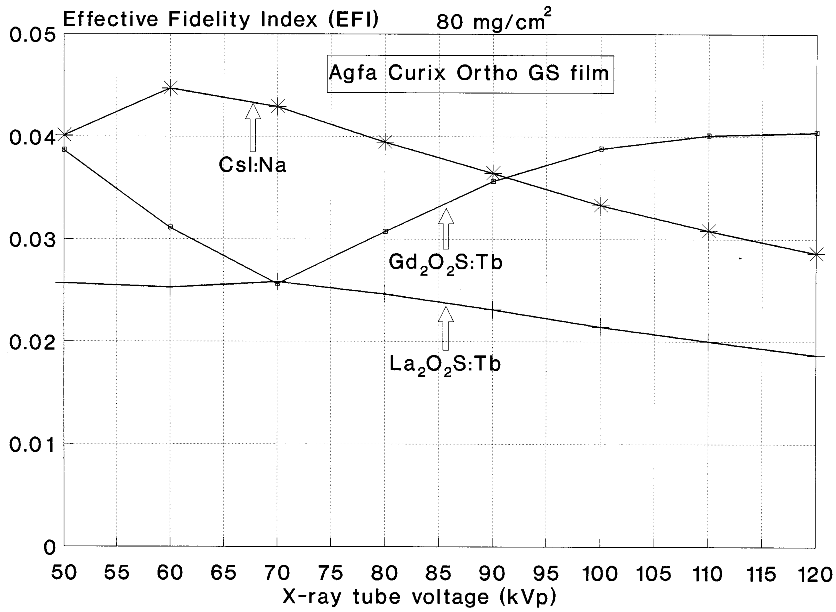

Fig. 3. Variation of the effective fidelity index (EFI) with X-ray tube voltage for the CsI:Na, Gd_2O_2S:Tb and La_2O_2S:Tb 80 mg/cm^2 scintillators combined with the Agfa Curix Ortho GS radiographic film.

which channels the light out of the screen with minimal losses. Thus, the CsI:Na fidelity is continuously increasing with screen coating thickness.

Fig. 3 shows the variation of the effective fidelity index with X-ray tube voltage for the three 80 mg/cm^2 scintillator screens combined with the Agfa Curix Ortho GS radiographic film. CsI:Na was higher in the 50–90 kVp range, which is very often used in many radiographic techniques. For tube voltages higher than 90 kVp, the EFI of Gd_2O_2S:Tb was found superior. The low performance of Gd_2O_2S:Tb at 70 kVp must be attributed to its very low X-ray absorption efficiency for energies just below 50.2 keV, which is the energy of the *K*-absorption edge of Gd [8]. These energies approximately correspond to the mean energy of a 70 kVp X-ray spectrum. EFI-curves shown in Fig. 3 indicate that CsI:Na screens could be employed in radiography, provided they are adequately protected against humidity.

Fig. 4 shows the variation of EFI with X-ray tube voltage for the three 80 mg/cm^2 scintillator screens combined with the extended sensitivity S-20 photocathode. Gd_2O_2S:Tb performed better for X-ray voltages higher than 90 kVp. This result may be of value in designing image intensifiers for fluoroscopy or digital imaging. Fig. 5 similarly presents the variation of EFI with X-ray tube voltage for scintillators combined with the Si photodiode. It is interesting to note that: (1) within a short kVp region (63–77 kVp), the La_2O_2S:Tb-Si combination performed best; (2) the Gd_2O_2S:Tb-Si combination outperformed the others within the 50–63 kVp and 77–120 kVp ranges; and (3) the CsI:Na-Si combination remained lowest in the whole kVp range. Therefore, Gd_2O_2S:Tb could be the scintillator of choice in many digital imaging systems employing Si or CCD arrays, such as CT detectors or digital X-ray image receptors.

Acknowledgements

This study is dedicated to the memory of Professor G.E. Giakoumakis, leading member of our team, whose work on phosphor materials has inspired us to continue.

Appendix A. Fidelity of imaging scintillators

Image fidelity has been defined [1] as:

$$F = 1 - \frac{\int_x \int_y [J_{\mathrm{ideal}}(x, y) - J_{\mathrm{L}}(x, y)]^2 \, \mathrm{d}x \, \mathrm{d}y}{\int_x \int_y [J_{\mathrm{ideal}}(x, y)]^2 \, \mathrm{d}x \, \mathrm{d}y} \quad \text{(A1)}$$

where, $J_{\mathrm{L}}(x, y)$ is the real image intensity and $J_{\mathrm{ideal}}(x, y)$ is the ideal image intensity at a point (x, y) on the image area. The integration in Eq. (A1) is extended over the whole image area. If the image receptor is a scintillator detector in the form of a screen, J_{L} and J_{ideal} may be considered as the emitted light intensities that form the image on the screen's surface. J_{L} and J_{ideal} are expressed in J/m^{2}s.

In a scintillator evaluation experiment, the screen may be uniformly illuminated by X-rays. $J_{\mathrm{L}}(x, y)$ has constant mean value over the screen output surface.

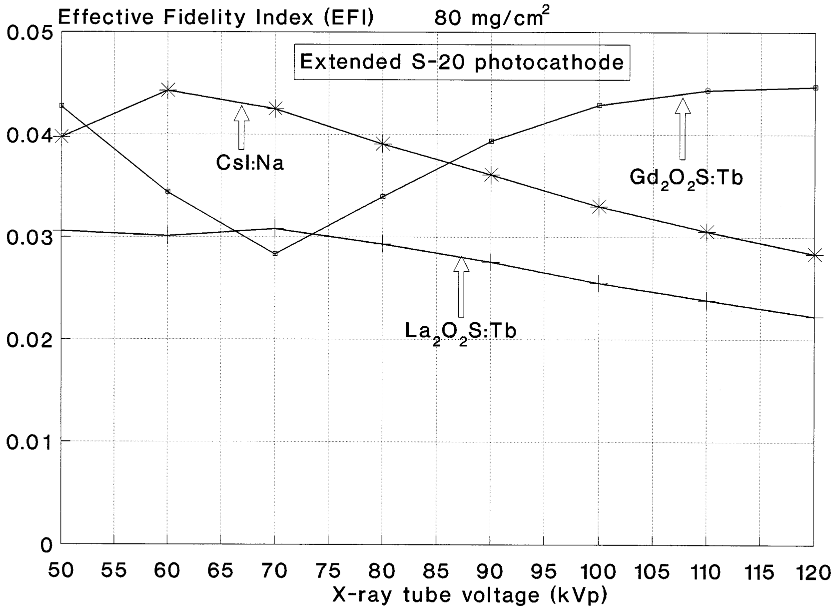

Fig. 4. Variation of the effective fidelity index (EFI) with X-ray tube voltage for the CsI:Na, Gd_2O_2S:Tb and La_2O_2S:Tb 80 mg/cm^2 scintillators combined with the extended S-20 photocathode.

This value expresses the screen emission efficiency and depends on the intensity of the incident X-ray beam and on the intrinsic physical properties of the scintillator material as indicated by the relation:

$$J_L(x, y) = J_Q(x, y)\bar{\eta}_Q\bar{\eta}_C\bar{G}_L(\mu_{OP}) \qquad (A2)$$

where, J_Q denotes the mean intensity of the incident X-ray beam, $\bar{\eta}_Q$ is the mean value of the X-ray detection efficiency averaged over the screen area, $\bar{\eta}_C$ is the mean intrinsic X-ray to light conversion efficiency, expressing the fraction of absorbed X-ray energy that is converted into light within the screen material, $\bar{G}_L$ is the mean light transmission efficiency expressing the fraction of light that is emitted after transmission through the screen material [9].

The ideal image intensity J_{ideal} may be defined as the output light intensity emitted by a screen when only signal conversion processes take place without signal losses ($\bar{\eta}_Q = 1$, $\bar{\eta}_C = 1$, $\bar{G}_L = 1$). Thus, from Eq. (A2), $J_{ideal} = J_Q$. Eq. (A1) may then be written as:

$$F = 1 - \frac{[J_Q - J_L]^2}{J_Q^2} \qquad (A3)$$

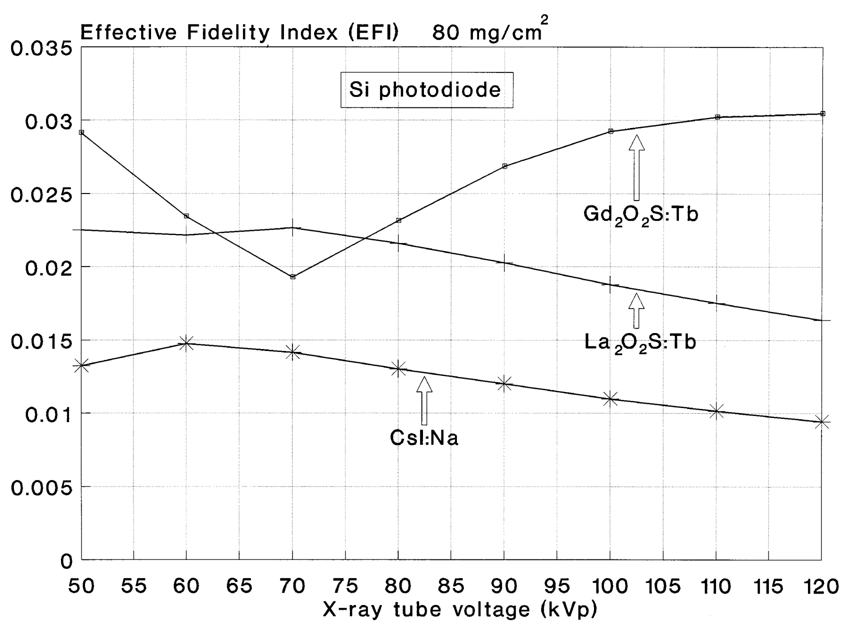

Fig. 5. Variation of the effective fidelity index (EFI) with X-ray tube voltage for the CsI:Na, Gd_2O_2S:Tb and La_2O_2S:Tb 80 mg/cm^2 scintillators combined with the Si photodiode.

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which defines the fidelity of a scintillator. F is a parameter taking values between 1 and 0.

Appendix B. Spectral matching factor

The spectral matching factor was calculated by the formula [2,10]

$$a_{SP} = \frac{\int_{\lambda 1}^{\lambda 2} S_P(\lambda) S_D(\lambda)\, d\lambda}{\int_{\lambda 1}^{\lambda 2} S_D(\lambda)\, d\lambda} \quad \text{(B1)}$$

where λ_1 and λ_2 are the lower and upper wavelength limits of the spectrum, $S_P(\lambda)$ is the normalized spectrum of the scintillator, experimentally determined with an Oriel 7240 grating monochromator, $S_D(\lambda)$ is the normalized spectral sensitivity distribution of the light detector, obtained from manufacturers' data. For an ideal scintillator–light detector combination $a_{SP} = 1$.

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