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ΤΜΗΜΑΤΑ ΙΑΤΡΙΚΗΣ - ΦΥΣΙΚΗΣ

ΔΙΑΤΜΗΜΑΤΙΚΟ ΠΡΟΓΡΑΜΜΑ ΜΕΤΑΠΤΥΧΙΑΚΩΝ
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ΔΙΠΛΩΜΑΤΙΚΗ ΕΡΓΑΣΙΑ

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ΕΝΕΡΓΟΠΟΙΗΤΗ ΙΟΝΤΩΝ ΔΗΜΗΤΡΙΟΥ (Ce³⁺) ΓΙΑ ΧΡΗΣΗ ΣΕ ΨΗΦΙΑΚΟΥΣ
ΑΝΙΧΝΕΥΤΕΣ ΙΑΤΡΙΚΗΣ ΑΠΕΙΚΟΝΙΣΗΣ

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Τεχνολογικού Εκπαιδευτικού Ιδρύματος Αθήνας

Πάτρα, Οκτώβριος 2006, Ελλάδα
MASTER THESIS

EXPERIMENTAL AND THEORETICAL DETERMINATION OF THE IMAGING CHARACTERISTICS IN NEW PHOSPHOR-SCINTILLATOR MATERIALS WITH CERIUM ($Ce^{3+}$) ACTIVATORS APPLIED IN MEDICAL DIGITAL DETECTORS

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Introduction

Image quality in general radiography and mammography is intimately linked to the precise and accurate acquisition of information from the x-ray beam transmitted by the patient, i.e. to the performance of the x-ray detector. Detectors for diagnostic applications must meet the needs of the specific radiological procedure which is to be used. Key parameters are spatial resolution, uniformity of response, contrast sensitivity, dynamic range and acquisition speed. These parameters are directly dependent to the light quanta produced as an intermediate stage on phosphor x-ray converters (Beutel et al, 2000).

Hence the present work sought, both to examine and evaluate the predictions given by a theoretical model for MTF calculation of phosphor screens. The transfer of energy through the phosphor screen is modelled as a series of cascaded stochastic processes assuming that the screen consists of many thin phosphor layers. Swank was the first that introduced a formula for the MTF calculation by modelling the escape of light in the screen as diffusion process (Swank R K, 1973). Others has continued this work and applied it into commercially used materials such as Gd$_2$O$_2$:Tb and CsI:Tl (Giakoumakis 1989; Nishikawa Yaffe, 1990, Kandarakis et al. 1997; Zhao and Rowlands 2004).

The predictions of the theoretical model were compared with the obtained experimental results and validated with data that already exist from previous works. The theoretical model, for calculation of various parameters applied on phosphor screens, was developed on the MATLAB 7.01 platform.

The phosphor screen where to this study focuses is Lutetium oxyorthosilicate Lu$_2$SiO$_5$:Ce (LSO). Since its discovery by Melcher and Schweitzer in 1992, Lutetium oxyorthosilicate (LSO) has attracted a great deal of attention and has been recognized as one of the best scintillating materials. In effect, this material is very good compromise among fast scintillators (e.g., BaF$_2$), high light output scintillators (e.g., Gd$_2$O$_2$: Tb) and dense scintillators (e.g., Bi$_4$Ge$_3$O$_{12}$). The great interest of LSO: Ce is due to many important advantages, such as high luminescence efficiency, high density of 7.4 g/cm$^3$, fast decay time of 40 ns, suitable emission wavelength (420 nm) and very good chemical stability compared to other scintillators. It is known that the scintillator LSO:Ce in crystal form has applications in Positron Emission Tomography, nuclear physics, high energy physics and environmental monitoring. According to our knowledge has not yet been such a study to this material under general radiographic and mammographic exposure conditions.
Chapter 1

Introduction

Theory

Nature of X-rays

X-rays are electromagnetic waves whose wavelengths range from about 0.1 to $100 \times 10^{-10}$ m (NASA).

![Fig. 1: The Electromagnetic Spectrum. The wavelength of radiation produced by an object is usually related to its temperature.](image)

They are produced when rapidly moving electrons strike a solid target and their kinetic energy is converted into radiation. The wavelength of the emitted radiation depends on the energy of the electrons. X-rays were first discovered in 1895 by Wilhelm Conrad Roentgen, who was awarded the 1901 Nobel Prize in physics for this achievement.

![Fig. 2: Wilhelm Conrad Roentgen.](image)

The discovery caused worldwide excitement, especially in the field of medicine; by 1900, there already were several medical radiological societies. Thus, the foundation was laid for a new branch of medicine devoted to imaging the structure and function of the body.

![Fig. 3: X-ray imaging.](image)
Production of X-rays

1. Ionization: A hole in an inner shell (here: K shell) is generated by an incident high-energy electron that loses the corresponding energy $E$ transferred to the ejected electron.

2. X-ray emission: The hole in the K shell is filled by an electron from an outer shell (here: L3). The superfluous energy is emitted as a characteristic X-ray quantum.

![Diagram of X-ray generation](image)

**Fig. 4:** Generation of X-rays.

The X-rays are produced from electrons that have been accelerated in vacuum from the cathode to the anode. Emission occurs when filament is heated by passing current through it. When the filament is hot enough, the electrons obtain thermal energy sufficient to overcome the energy binding the electron to the metal of the filament. After accelerated they will be stopped at a short distance. Most of the electron energy will produce heat at the anode. Some percentage will be converted to X-ray by two main methods. Deceleration of charged particle results in the emission of electromagnetic field called Bremsstrahlung radiation. These rays will have wide, continuous distribution of energies with the maximum being the total energy the electron had when reaching the anode. The number of X-rays will be small at lower energies and increased for higher energies.
The x-ray beam consists of a spectrum of energies over an energy range determined by the peak kilovoltage (kVp), the generator waveform, and the amount of inherent and added filtration. Lower energy photons in the spectrum are more readily attenuated, which results in a gradual shift to higher energies in the transmitted beam as it passes through an attenuating material. The gradually increasing average energy produces a corresponding decrease in the average attenuation coefficient of the transmitted beam. Directly measuring an x-ray spectrum in the field is impractical, but an estimate of the spectral distribution of x-rays and the relative penetrability or quality of the beam is useful. Transmission measurements of intensity through a series of attenuator thicknesses (e.g., aluminum) can characterize the beam quality in terms of the HVL. The HVL is the thickness of an attenuator required to reduce the initial beam intensity by one half, or by a factor of 2 (Seibert and Boone, 2005).
X-ray interactions with matter

X-ray interactions in general can result in energy deposition and, in many cases a secondary x-ray will be present after the initial interaction. Examples of this include scattered x-rays, characteristic x-rays, and annihilation radiation. Details of the 4 basic x-ray interactions pertinent to medical x-ray applications namely, photoelectric absorption, Rayleigh scattering, Compton scattering, and pair production are the focus of this section.

Photoelectric absorption

Photoelectric absorption involves the interaction of an incident x-ray photon with an inner shell electron in the absorbing atom that has a binding energy similar to but less than the energy of the incident photon. The incident x-ray photon transfers its energy to the electron and results in the ejection of the electron from its shell (usually the K shell) with a kinetic energy equal to the difference of the incident photon energy, $E_0$, and the electron shell binding energy, $E_{BE}$. The vacated electron shell is subsequently filled by an electron from an outer shell with less binding energy (e.g., from the L or M shell), producing a characteristic x-ray equal in energy to the difference in electron binding energies of the source electron shell and the final electron shell. If the incident photon energy is less than the binding energy of the electron, the photoelectric interaction cannot occur, but if the x-ray energy is equal to the electronic binding energy ($E_0=E_{BE}$), the photoelectric effect becomes energetically feasible and a large increase in attenuation occurs. As the incident photon energy increases above that of the electron shell binding energy, the likelihood of photoelectric absorption decreases at a rate proportional to $1/E^3$. The K absorption edge refers to the sudden jump in the probability of photoelectric absorption when the K-shell interaction is energetically possible. Similarly, the L absorption edge refers to the sudden jump in photoelectric absorption occurring at the L-shell electron binding energy (at much lower energy). After photoelectric interaction, ionization of the atom occurs and a free electron and a positively charged atom are produced. Kinetic energy (motion) of the ejected photoelectron can cause further electron–electron ionization, with most energy locally deposited. Also, a subsequent cascade of electron transitions to fill the vacated inner electron shell results in emission of characteristic radiation (Evans 1955; Seibert and Boone, 2005).
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Fig. 7: The free electron collides with the tungsten atom, knocking an electron out of a lower orbital. A higher orbital electron fills the empty position, releasing its excess energy as a photon.

**Compton scattering**

Together with the scattering of photons on free electrons, the photoelectric effect, and pair production, Compton scattering contributes to the attenuation of x-rays in matter. As the binding energy of electrons in atoms is low compared to that of passing near-relativistic particles, this is the relevant process in radiography. Closely related are Thompson scattering (classical treatment of photon scattering) and Rayleigh scattering (coherent scattering on atoms).

Compton Scattering, also known as incoherent scattering, occurs when the incident x-ray photon ejects an electron from an atom and an x-ray photon of lower energy is scattered from the atom. Relativistic energy and momentum are conserved in this process and the scattered x-ray photon has less energy and therefore a longer wavelength than the incident photon. Compton scattering is important for low atomic number specimens. The change in wavelength of the scattered photon is given by:

\[ \frac{c}{v'} - \frac{c}{v_0} = \lambda' - \lambda_0 = \frac{\hbar}{m_0 c} (1 - \cos \theta) \quad \text{Equation 1} \]

Theta is the scattering angle of the scattered photon. Note the fundamental constants for the speed of light, Planck constant, and electron mass (Evans 1955; Seibert and Boone, 2005).

Fig. 8: Compton Scattering. The free electron is attracted to the tungsten atom nucleus. As the electron speeds past, the nucleus alters its course. The electron loses energy, which it releases as an X-ray photon.
Rayleigh scattering

An incident x-ray photon can interact with an electron and be deflected (scattered) with no loss in energy. This process is also known as coherent or elastic scattering, and it occurs by temporarily raising the energy of the electron without removing it from the atom. The electron returns to its previous energy level by emitting an x-ray photon of equal energy but with a slightly different direction. Most x-rays are scattered forward by this mechanism, because the atom cannot experience significant recoil without otherwise removing the electron. There is no absorption of energy, and the majority of the x-ray photons are scattered with a small angle. The probability of Rayleigh scattering is given the symbol $\sigma_{coh}$ or $\sigma_R$. In soft tissue, probability of this event occurring is low, on the order of ~5% of all scattering events, because of the low effective atomic number of soft tissues ($Z=7.5$). The possibility of Rayleigh scattering increases with increasing $Z$ of the absorbed and decreasing x-ray energy (Evans 1955; Seibert and Boone, 2005).

Pair production

Pair production can occur when the incident x-ray photon has energy greater than 1.02 MeV, which represents the rest mass energy equivalent of 2 electrons (i.e., $E=2m_0c^2$, where $m_0$ is the rest mass of the electron [9.11*10^{-31} kg] and $c$ is the speed of light [3.0*10^8 m/s]). The interaction of the incident photon with the electric field of the nucleus results in the production of an electron ($e^{-}$) and a positron ($e^+$) pair, with any photon energy in excess of 1.02 MeV being transferred to the kinetic energy of the $e^-/e^+$ pair equally. Interestingly, ionization of the atom does not occur, although charged particles are formed and their kinetic energy can result in subsequent ionization within the local area. Once the positron expends its kinetic energy, it will combine with any available electron and produce annihilation radiation, resulting from the conversion of the rest mass energies of the $e^-/e^+$ pair into (nearly) oppositely directed 511 keV photons. The probability of pair production, commonly given the symbol $\pi$, increases with energy above 1.02 MeV. Though annihilation radiation is crucial for PET, pair production occurs at energies well above those used for diagnostic x-ray imaging (Evans 1955; Seibert and Boone, 2005).
Fig. 9: Illustrative summary of x-ray interactions. (A) Primary, unattenuated beam does not interact with material. (B) Photoelectric absorption results in total removal of incident x-ray photon with energy greater than binding energy of electron in its shell, with excess energy distributed to kinetic energy of photoelectron. (C) Rayleigh scattering is interaction with electron (or whole atom) in which no energy is exchanged and incident x-ray energy equals scattered x-ray energy with small angular change in direction. (D) Compton scattering interactions occur with essentially unbound electrons, with transfer of energy shared between recoil electron and scattered photon, with energy exchange described by Klein–Nishina formula.
Chapter 1    Introduction

Attenuation coefficients

Linear attenuation coefficient

The interaction mechanisms discussed here combine to attenuate the incident photon beam as it passes through matter, through the removal of x-ray photons from the x-ray beam either by absorption or scattering events. For a monoenergetic beam of \(N_0\) photons incident on a thin slab of material of thickness \(x\) with a probability of attenuation, \(\mu\), the fractional reduction of the number of photons from the beam is constant. An exponential relationship exists between the incident and transmitted photon fluence (number of photons/mm\(^2\) area) after passing through a total material thickness, \(x\), of an attenuating material, as:

\[
N_x = N_0 e^{-\mu x} \quad \text{Equation 2}
\]

The unit of thickness is commonly expressed in centimetres, so the corresponding unit of \(\mu\) is cm\(^{-1}\), where \(\mu\) is the linear attenuation coefficient, which represents the probability of attenuation per centimetre of a material. The total linear attenuation coefficient is the sum of the linear attenuation coefficients for the individual interaction mechanisms, as \(\mu = \tau + \sigma_r + \sigma + \pi\). Like the individual interaction coefficients, values of \(\mu\) are strongly dependent on incident x-ray energy and on the physical density of the interacting medium. The images that CT scanners produce are in fact maps of the spatially varying linear attenuation coefficients of the tissues being scanned (Evans 1955; Seibert and Boone, 2005).

Mass attenuation coefficient

At a given photon energy, the linear attenuation coefficient can vary significantly for the same material if it exhibits differences in physical density. A classic example is water, water vapour, and ice. The mass attenuation coefficient, \(\mu/\rho\), compensates for these variations by normalizing the linear attenuation by the density of the material. By doing so, the mass attenuation coefficients for water, water vapour, and ice are identical. For the mass attenuation coefficient, “thickness” becomes the product of the density and linear thickness of the material, or \(\rho x\). This is known as the mass thickness with units of g/cm\(^3\) x cm=g/cm\(^2\). The reciprocal of the mass thickness represents the units of mass attenuation coefficient, cm\(^2\)/g, and the corresponding Lambert–Beers equation is:

\[
N_x = N_0 e^{-\left(\frac{\mu}{\rho}\right)x} \quad \text{Equation 3}
\]

Just like the total linear attenuation coefficient, the mass attenuation coefficient for a specific material is a sum of the individual interaction probabilities (Evans 1955; Seibert and Boone, 2005):
Mass transfer and mass energy coefficients

Attenuation per se is not a direct measure of the amount of energy imparted to the medium, because in most instances some of the energy of the incident photon is transferred away from the site of interaction by scattered x-rays or other energy transfer mechanisms. Knowledge of the energy transfer is important for determination of radiation dose to the tissues as well as the signal captured by an x-ray detector. The mass energy transfer coefficient, $\mu_e/\rho$, describes the amount of energy transferred to charged particles within the medium at the site of interaction and takes into account radiative losses (if any) caused by characteristic radiation emitted after photoelectric absorption and for x-ray scatter emission after the Compton scatter interaction. The mass energy absorption coefficient, $\mu_m/\rho$, additionally considers the probability of energetic recoil electrons interacting within the medium and creating bremsstrahlung radiation (created in the same way as energetic electrons interact in the target of the x-ray tube to produce x-rays) that can radiate away from the site of interaction. Re-emission of bremsstrahlung energy mainly occurs with recoil electrons having high kinetic energies and with high-Z absorbers.

Within the diagnostic imaging energy range, the mass energy and mass transfer coefficients are approximately equal, $\mu_m/\rho \sim \mu_e/\rho$ (Evans 1955; Seibert and Boone, 2005).

Attenuation coefficients for compounds

Attenuation coefficients for compounds (materials comprised of $\geq 2$ elements) can be determined as the weighted average (by mass) of the individual mass attenuation coefficients of the compound’s constituent elements, as:

$$\frac{\mu}{\rho}_{\text{compound}} = \sum_{i=1}^{N} m_i \left( \frac{\mu}{\rho} \right)_i$$

where $m_i$ is the mass fraction (fraction of the element’s mass contribution to the total mass) and $\left( \frac{\mu}{\rho} \right)_i$ is the mass attenuation coefficient of element $i$ in the compound. This is important for estimating attenuation probabilities of compounds and materials that cannot be easily measured and particularly for computer simulations (Evans 1955; Seibert and Boone, 2005).
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Fig. 10: Mass Attenuation Coefficients for the Lu$_2$SiO$_5$:Ce phosphor calculated with the XMuDat photon attenuation data version 1.0.1 software, data source J H Hubbel, S M Seltzer, NISTIR 5632, 1995.

Absorbed dose

Absorbed dose is a non-stochastic quantity applicable to both indirectly and directly ionizing radiations. For indirectly ionizing radiations, energy is imparted to matter in a two step process. In the first step (resulting in KERMA) the indirectly ionizing radiation transfers energy as kinetic energy to secondary charged particles. In the second step these charged particles transfer some of their kinetic energy to the medium (resulting in absorbed dose) and lose some of their energy in the form of bremsstrahlung losses. The absorbed dose is related to the stochastic quantity energy imparted. The absorbed dose is defined as the mean energy imparted by ionizing radiation to matter of mass $m$ in a finite volume $V$ by:

$$D = \frac{\bar{d}E}{dm} \quad \text{Equation 6}$$

The energy imparted is the sum of all energy entering the volume of interest minus all energy leaving the volume, taking into account any mass-energy conversion within the volume. Pair production, for example, decreases the energy by 1.022 MeV, while electron-positron annihilation increases the energy by the same amount. Note that because the electrons travel in the medium and deposit energy along their tracks, this absorption of energy does not take place at the same location as the transfer of energy described by KERMA. The unit of absorbed dose is joule per kilogram (J·kg$^{-1}$). The special name for the unit of absorbed dose is the gray (Gy) (Podgorsak, 2003).
Exposure

Exposure $X$ is the quotient of $dQ$ by $dm$, where $dQ$ is the absolute value of the total charge of the ions of one sign produced in air when all the electrons and positrons liberated or created by photons in mass $dm$ of air are completely stopped in air:

$$ X = \frac{dQ}{dm} \quad \text{Equation 7} $$

The unit of exposure is coulomb per kilogram (C/kg). The special unit used for exposure is the roentgen $R$, where $1 \text{ R} = 2.58 \times 10^{-4} \text{ C/kg}$. In the SI system of units, roentgen is no longer used and the unit of exposure is simply $2.58 \times 10^{-4} \text{ C/kg}$ of air (Podgorsak, 2003).

KERMA

KERMA is an acronym for Kinetic Energy Released per unit MAss. It is a non-stochastic quantity applicable to indirectly ionizing radiations, such as photons and neutrons. It quantifies the average amount of energy transferred from the indirectly ionizing radiation to directly ionizing radiation without concerns to what happens after this transfer. Energy of photons is imparted to matter in a two-stage process. In the first stage, the photon radiation transfers energy to the secondary charged particles (electrons) through various photon interactions (photo-effect, Compton effect, pair production, etc). In the second stage, the charged particle transfers energy to the medium through atomic excitations and ionisations. KERMA is expressed in the units of J/kg which is also the radiation unit, the gray (Gy) In this context, the KERMA is defined as the mean energy transferred from the indirectly ionizing radiation to charged particles (electrons) in the medium $dE_{\nu}$ per unit mass $dm$:

$$ K = \frac{dE_{\nu}}{dm} \quad \text{Equation 8} $$

Air-KERMA in air

For a monoenergetic photon beam in air, the air-KERMA in air, $(K_{\text{air}})_{\text{air}}$, at a given point away from the source is proportional to the energy fluence or photon fluence as follows:

$$ (K_{\text{air}})_{\text{air}} = \Psi \left( \frac{\lambda_{\nu}}{\rho} \right)_{\text{air}} = \phi hv \left( \frac{\mu_{\nu}}{\rho} \right)_{\text{air}} \quad \text{Equation 9} $$

where $(\mu_{\nu}/\rho)_{\text{air}}$ is the mass energy transfer coefficient for air at photon energy $hv$.

KERMA $K$ consists of two components: the collision KERMA $K_{\text{col}}$ and the radiative KERMA $K_{\text{rad}}$, i.e. (Podgorsak, 2003):

$$ K = K_{\text{col}} + K_{\text{rad}} $$
Scintillators in medical imaging

Scintillators in x-ray imaging

Scintillators are employed in most of the current medical diagnostic imaging modalities using x-rays. This is explained by the comparatively good detection efficiency of scintillators for hard radiation. Yet, the various diagnostic methods differ considerably and consequently the radiation requirements also differ. These requirements are not always met by the scintillator specifications. New, digital diagnostic systems with an excellent image quality and a short image acquisition time and excellent quality real-time imaging systems for interventional radiology at a low radiation dose are two main reasons for continuous scintillator search (Johns and Cunningham, 1983).

X-ray imaging

In x-ray radiography (static imaging) an attenuation profile of a part of the human body is projected onto a two-dimensional position-sensitive radiation detector (PSD) using the focus of an x-ray tube as a point source. Thus, information on anatomical detail is obtained. A variety of radiation detectors are used. For example, at present, for chest radiography x-ray phosphor-screen-film cassettes, typically of 35x43 cm², are applied as the PSD on a large scale. In most cases Gd₂O₂S:Tb is used as the x-ray intensifying screen phosphor. Furthermore, a digital chest-radiography system is used based on the electrostatic read-out of a drum covered with an amorphous-selenium (a-Se) semiconductor layer as the PSD. Recently, new digital radiography (DR) systems have been introduced employing an a-Se photoconductor layer deposited on top of an amorphous-silicon (a-Si:H) thin-film transistor array or employing a Gd₂O₂S:Tb phosphor screen deposited on top of an array of a-Si:H photodiodes coupled to an array of thin-film transistors.

In mammography film-screen cassettes of 18x24 cm² are generally used. In this case a resolution of 0.1 mm is required to observe micro calcifications. Conventional film-screens systems usually consist of two equal phosphor-binder layers (thickness up to 0.3 mm per layer) and two equal emulsion layers, one on each side of the film base. More advanced systems for chest radiography are asymmetric in structure, i.e. the two screens have different thicknesses and the two emulsions are of different contrast. In mammography only one relatively thin screen (0.07 mm) and one emulsion are used to realize the required resolution (Van Eijk, 2002; Yaffe and Rowlands, 1997).
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Introduction

Fig. 11: Application of the scintillators in x-ray imaging.

Charge-coupled devices (CCDs)

The charge-coupled device was developed in 1970. Because of its compactness and dynamic range characteristics, it has virtually replaced the vacuum camera tube in commercial and home video and it has found many applications in digital imaging. This has largely come about due to the development of techniques for producing extremely pure crystalline silicon and for very large scale integration (VLSI). CCDs are particularly well suited to digital radiography because of their high spatial resolution capability, wide dynamic range and high degree of linearity with incident signal. They can be made sensitive to light or to direct electronic input. A CCD is an integrated circuit formed by depositing a series of electrodes, called ‘gates’ on a semiconductor substrate to form an array of metal-oxide-semiconductor (MOS) capacitors (figure 12).

By applying voltages to the gates, the material below is depleted to form charge storage ‘wells’. These store charge injected into the CCD or generated within the semiconductor by the photoelectric absorption of optical quanta. If the voltages over adjacent gates are varied appropriately, the charge can be transferred from well to well under the gates. In area CCDs, a ‘frame transfer’ system (figure 13) is employed to obtain rapid readout.
Chapter 1

Introduction

Fig. 13: (a) Structure of a CCD array, illustrating motion of stored charge in one direction as the potential wells are adjusted under control of the gate electrode voltages. (b) Typical readout configurations of CCDs showing (a) the frame transfer, (b) interline transfer and (c) time delay integration (TDI) devices. In (a) and (b) readout storage areas which are shielded from illumination are required.

Charge is initially accumulated on ‘detector’ pixels and then transferred to an array of ‘storage’ pixels from which the signal can be read line by line. Alternatively, ‘interline readout’ CCDs (figure 13 (b)) have a line of optically shielded storage and transfer pixels adjacent to each column of detector elements. The charge is rapidly unloaded into the storage column, freeing the detector elements to accumulate new signal, and transferred down the storage column elements to a master output register which sequentially receives signal from each storage column. These modes of operation are used for the small-format-area imagers which may be coupled to phosphors through fibre optics or lenses. Although they provide rapid readout, these systems require a storage area that is approximately equal to the active photo-detector area. In the case of interline devices, because the storage area is immediately adjacent to the detector columns, this may cause the effective fill factor of the detector to be reduced. Area format CCDs are available in sizes varying from 256X256 pixels or less to 2048X2048 or more. However, real-time readout (30 frames/s) is currently restricted to devices of 1000X1000 pixels or less. For scanning systems, as discussed above, it is usually more practical to operate the CCD in time delay integration (TDI) mode (figure 13(c)). Here, a storage section is not required as the charge is simultaneously integrated and shifted down the CCD detector columns toward the horizontal readout register. This type of analogue integration is desirable as it is relatively noise free. In addition, because all of the detector elements in a column contribute to each image pixel imaged by that column, the image produced by TDI is relatively insensitive to a few pixels in the column that may suffer from abnormally low or high sensitivity. In any CCD, the charge is transferred ‘bucket brigade’ style over many adjacent elements. It is, therefore, critical that the efficiency of each transfer is extremely high. Lack of transfer efficiency can
cause a serious loss of spatial resolution in the detector. If the signal must be shifted across $\eta$ elements and the efficiency per transfer is $\varepsilon$, then the overall charge transfer efficiency is $\varepsilon^\eta$. Even if $\varepsilon$ is 0.999, the efficiency falls to 90% over 100 transfers and 37% over 1000 transfers. The effect of less-than-perfect transfer efficiency is smearing of the image in the readout direction. In commercial CCDs, values of $\varepsilon$ as high as 0.999999 are achievable. Also of importance is the well storage capacity of the device. Depending on pixel size, capacities of 300 000 to several million electrons are possible. CCDs designed for video applications tend to be designed to have extremely small (15 $\mu m$) pixel dimensions. For medical applications, a larger size (25–100 $\mu m$) is generally desirable because this offers greater well capacity and better conforms with other constraints on spatial resolution. It is important that the CCD be designed with appropriate ‘anti-blooming’ protection to prevent degradation of the image if some of the charge wells are overfilled. This can occur in situations where the x-ray detector is exposed to the unattenuated x-ray beam, for example at the edges of the patient. When the CCD is used in TDI mode, the well capacity must be such that the integral charge over all stages of integration can be accommodated. For example, if each detector element accumulates 50 000 electrons per row and there are 64 rows in the CCD over which integration will take place, the well capacity must be 3.2 million electrons (Yaffe and Rowlands, 1997).

**Flat-panel systems**

A flat-panel digital detector, in principle, could perform all current radiological modalities: Radiography, fluoroscopy and fluorography. It could provide high image quality and an instant readout. Existing x-ray equipment could be easily adapted to employ such detectors. The technology of large-area active matrix arrays which would form the readout structure for a flat-panel system has been developed for liquid crystal displays (LCD) for over a decade.

![Fig. 14: (a) Vertical structure of a unit pixel and its equivalent circuit. (b) The vertical structure of a color TFT LCD panel.](image-url)
Active matrix LCDs (AMLCDs) have been made using amorphous (hydrogenated amorphous silicon (a-Si:H), polycrystalline (poly-Si) or cadmium selenide (CdSe) semiconductors.

Each display panel consists of two sheets of glass with a uniform layer of liquid crystal in between. One sheet is the active matrix itself (i.e. a large-area integrated circuit consisting of a large number of thin film field-effect transistors (TFTs) connected to individual pixel electrodes in a matrix). The other sheet has a uniform electrode layer. Two general approaches for flat-panel digital x-ray detectors are currently under investigation. In the first a phosphor layer is used to absorb x-rays and the resultant light photons are detected by a large-area photodiode array read out with active devices (for example, thin film transistors or diode switches) integrated onto the plate at each pixel. In the second approach (direct method), x-rays are detected in an amorphous selenium layer and the resulting charges released are collected on individual pixel electrodes. Finally, the readout occurs using the active matrix as in the indirect method. The potential advantages of such self-scanned, readout systems include their compactness permitting better access to patients than bulky devices such as conventional XRIIs. Since they are flat, they can be expected to be largely free from veiling glare, geometrically uniform. Unlike XRIIs they are immune to stray magnetic fields. These properties facilitate quantitative image analysis, registration and clinical comparison of images from other modalities, 3D reconstruction applications such as cone beam volume CT, and use in magnetic environments such as MRI rooms (Yaffe and Rowlands, 1997).

**Phosphor flat-panel detectors**

Several groups are developing large-area photo-detector arrays composed of individual photodiodes made with amorphous silicon, onto which a conventional x-ray absorbing phosphor, such as Gd₂O₂S, is placed or thallium-doped cesium iodide (CsI:Na) is grown. The principle of operation of an amorphous silicon detector is shown schematically in figure 15. The detector pixels are configured as photodiodes which convert the optical signal from the phosphor to charge and store that charge on the pixel
capacitance. Being low-noise devices, the photodiodes provide a very large dynamic range, of the order of 40 000. The signal is read out by activation of scanning control lines for each row of the device, connected to the gates of TFTs located on each detector pixel. An entire row of the detector array is activated simultaneously and the signal is read on lines for each column in the array which connects all the TFT sources in that column to a low-noise charge amplifier. The amplified signals from the columns are then multiplexed and digitized. This allows fast detector readout and requires a number of electronic channels equal to the number of columns of the array. Both radiographic and fluoroscopic systems have been described. Alternatively, instead of TFT readout various diode switching schemes can be used. The advantage of the diode approach is that since the photodiode has to be made anyway, the switching diode can be made at the same time without increase in the number of processing steps. The disadvantages of diode readout are a strong nonlinearity and large charge injection. The area allocated to each pixel of the array must contain the photodiode, switching device and control and signal lines so that the fill factor is less than 100%. This potential loss of x-ray utilization efficiency becomes proportionately greater as the pixel size is decreased and provides a challenge for the application of this technology to very high-resolution applications. The advantage of utilizing CsI as the x-ray absorber is that it can be grown in columnar crystals which act as fibre optics. When coupled to the photodiode pixels, there is little lateral spread of light and, therefore, high spatial resolution can be maintained. In addition, unlike conventional phosphors in which diffusion of light and loss of resolution become worse when the thickness is increased, CsI phosphors can be made thick enough to ensure a high value of $\eta$ while maintaining high spatial resolution (Yaffe and Rowlands, 1997).

The quest for the ideal inorganic scintillator

The past half century has witnessed the discovery of many new inorganic scintillator materials and numerous advances in our understanding of the basic physical processes governing the transformation of ionizing radiation into scintillation light. Whereas scintillators are available with a good combination of physical properties, none provides the desired combination of stopping power, light output and decay time. So accordingly to the use of a scintillator in medical diagnostic imaging system, scintillator should appear the desirable combination of fluorescent properties.

History

A history of the discovery of important inorganic scintillator materials important in the sense that they either became commercially available and widely used or triggered further developments or new research directions-is shown bellow.
The discovery of scintillator materials may be divided into three periods. The first period included the earliest scintillators: CaWO$_4$ first used in the year following Roentgen’s discovery of x-rays and ZnS used by Crooke’s to detect and count radioactivity and by Rutherford to study alpha particle scattering. The second period began with the development of the photomultiplier tube in the 1940s and the development of the thallium-activated NaI(Tl) by Hofstadter. In a burst of exploration during the following few years, the scintillation properties of most pure and activated alkali halide crystals were investigated. In the ensuing decades a steady precession of new scintillator materials appeared including the discovery of fast core-valence luminescence in BaF$_2$. A third period- the past two decades-has witnessed a veritable renaissance in research and development of scintillator materials, prompted in large part by the need for scintillators for precision calorimetry in high energy physics and for high light output scintillators for medical imaging (Derenzo et al., 2001; Tong Yu et al., 1997).

Among the various phosphors used in X-ray detectors of medical imaging systems, Gd$_2$O$_2$S:Tb and CsI:Na are widely accepted as the highest performing X-ray to light converters suitable for most X-ray imaging applications. Gd$_2$O$_2$S:Tb is a high density and effective atomic number material with K-absorption edge at 50.2 keV, which is well within the X-ray spectra often employed in medical imaging. Gd$_2$O$_2$S:Tb also exhibits one of the highest intrinsic X-ray to light conversion efficiencies (15-20%). These properties augment X-ray detection efficiency and light output both allowing for patient dose and image noise reduction. On the other hand, CsI:Na exhibits lower K-absorption energies (36 and 33.2 keV) and lower X-ray detection efficiency and X-ray to light conversion efficiency (10%). However, its
light output is high enough due to the CsI:Na non-granular intrinsic crystal structure that forms needle-like columns, giving highly directional light propagation and minimized optical scattering. These properties reduce light spread and optical loses within the phosphor material, resulting in increased light output and excellent spatial resolution. Nevertheless, a major drawback of CsI:Na is its hygroscopic properties demanding suitable light transparent protective covers to prevent damage from humidity (Kandarakis et al., 1998).

**Lutetium Oxyorthosilicate (Lu$_2$SiO$_5$:Ce)**

Cerium (Ce$^{3+}$) doped scintillators or phosphors are of particular interest for medical imaging, because of their very fast response. The latter is dominated by the very efficient $5d \to 4f$ electronic transitions of the Ce$^{3+}$ ion (Beutel et al., 2000; Van Eijk, 2002; Blasse, 1994). Since its discovery by Melcher and Schweitzer in 1992 (Melcher, 1992), Lutetium oxyorthosilicate (LSO) has attracted a great deal of attention and has been recognized as one of the best scintillating materials (Melcher and Schweitzer, 1992; Melcher, 1992). In effect, this material is very good compromise among fast scintillators (e.g., BaF$_2$), high light output scintillators (e.g., Gd$_2$O$_2$: Tb) and dense scintillators (e.g., Bi$_4$Ge$_3$O$_{12}$) (Melcher, 1992). The great interest of LSO: Ce is due to many important advantages, such as high luminescence efficiency, high density of 7.4 g/cm$^3$, fast decay time of 40 ns, suitable emission wavelength (420 nm) and very good chemical stability compared to other scintillators. It is known that the scintillator LSO: Ce in crystal form has applications in Positron Emission Tomography, nuclear physics, high energy physics and environmental monitoring (Melcher and Schweitzer, 1992).

**The luminescence efficiency of scintillators**

Scintillators as luminescent materials convert the energy of ionising radiation into emission of light. The physical processes occurring in scintillators are sometimes assumed to be incompletely known, but according to the state-of-the-art three different processes are distinguished:

- The absorption of the ionising radiation.
- The transfer of the absorbed energy to the luminescent centres.
- The emission process.

In solid materials where atoms are close together, the possible energy levels for the electrons are a continuous band called the valence band. Electrons in this band are confined to atoms or groups of atoms. If they are excited by x-rays, they can move up to the conduction band. Here the electrons are
free to roam and as they return to the valence band the excess energy is released as an optical photon (Blasse, 1994).

**The electron-hole pair mechanism**

The wavelength of photons that is able to cause electron excitation from valence band to conduction band must agree with the following condition:

\[
\frac{hc}{\lambda} \geq E_g \\
\text{or } \lambda \leq \frac{hc}{E_g}
\]

Equation 10

where \(E_g\) is the energy gap. Assuming that energy gap is expressed in eV and wavelength in \(\mu\)m the above condition is converted to:

\[
\lambda \leq 1.24E_g
\]

Equation 11

When the photon is absorbed in the lattice creates secondary electrons and holes by ionization. One initial particle with high energy may create many electron–hole pairs. However, the fast particle and the created charge carriers may lose energy to the lattice by exciting vibrations. In order to create an electron-hole pair of energy \(E_g\) a much larger amount of energy is needed. Generally the average energy required to create an electron-hole pair is equal to:

\[
E = \beta E_g
\]

Equation 12

Robbins showed the dependence of \(\beta\) on the so-called energy loss parameter \(K\) according to the following diagram:

![Fig. 17: The dependence of \(\beta\) on the loss parameter \(K\).](image)

The parameter \(K\) is given by,
Here $\varepsilon_1$ is the high-frequency dielectric constant, $\varepsilon_2$ the static dielectric constant and $\nu_{LO}$ the frequency of the longitudinal optical vibration mode. The values of $\beta$ range from about 3 (GaP, ZnS, CsI, NaI), 4 (La$_2$O$_2$S), 5, 6 (Y$_3$Al$_5$O$_{12}$) up to 7 (CaWO$_4$, YVO$_4$).

The total expression for the radiant efficiency looks like:

$$n = (1 - r)\frac{h\nu_e}{E}S\eta$$ \hspace{1cm} \text{Equation 14}$$

where $r$ is the amount of radiation which is not absorbed, $\nu_e$ is the (averaged) frequency of the emitted radiation, $E$ is the energy required for the electron-hole pair creation, $S$ is the efficiency of transfer of electron-hole pair energy to the luminescent centre, and $\eta$ is the quantum efficiency of the luminescent centre. Assuming that all excitation energy is absorbed (i.e. $r=0$), that all electron-hole pair energy arrives at the luminescent centre (i.e. $S=1$), and that $\eta=100\%$, we arrive at the maximum radiant efficiency.

$$n_{\text{max}} = \frac{h\nu_e}{E} = \frac{h\nu_e}{\beta E g}$$ \hspace{1cm} \text{Equation 15}$$

The factor $h\nu_e$ accounts for the fact that the emitted energy will be less than the band gap energy. For example in the case of ZnS:Ag $h\nu_e=2.75\text{eV}$ and $E_g=3.8\text{eV}$ in the case of NaI:Tl $h\nu_e=3.02\text{eV}$ and $E_g=5.9\text{eV}$ and of La$_2$O$_2$S:Eu $h\nu_e=2.0\text{eV}$ and $E_g=4.4\text{eV}$, respectively (Blasse, 1994).

**Fig. 18:** Electron-hole pair creation.

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**The role of activator**

One of the most important properties of scintillation radiation detectors used in medical imaging systems is the compatibility between the scintillators light spectrum and the spectral sensitivity of the optical detector. Poor spectral compatibility causes a decrease in the detector’s sensitivity and image quality. To improve spectral compatibility, activators may be employed during scintillator preparation, so that light spectra are modified to fit the sensitivity of existing optical detectors. However, the type of
activator may seriously affect the emission performance of the scintillator. Specifically the type of activator affects:

- The decay time
- The intrinsic x-ray to light conversion efficiency
- The attenuation (absorption and scattering) of light created within the scintillator material, by means of wavelength light photons of shorter wavelengths are easier attenuated.

Both last factors affect the intensity of the emitted light and the quality of the image produced. The main employed activators are Ce$^{+3}$, Tb$^{+3}$, and Eu$^{+3}$ (Blasse, 1994; Kandarakis and Cavouras, 2000).

**Fluorescence-Phosphorescence**

![Fluorescence-Phosphorescence diagram](image)

**Fig. 19:** Fluorescence-Phosphorescence.

Scintillator materials emit light when are excited by x-rays. This phenomenon is called luminescence. If the excited electron reaches an impurity level the crystal could stay in this state for an interval of time and then emit radiation, or it could emit the radiation very quickly. If the emission is prompt ($10^{-8}$ s) it is called fluorescence, if it is delayed it is called phosphorescence. The energy of the phosphorescence radiation may be different from the energy of the fluorescence radiation. In general, phosphorescence that gives a long-lived emission of visible light is a nuisance in diagnostic radiology, since the screen will continue to glow long after being exposed to the x-rays. The colour of light emitted in fluorescence depends upon the chemical composition of the material and is also very dependent on the presence of small amounts of chemical impurities that may be present. For example, CsI:Na is a screen made from CsI with a small amount of Na impurity. This fluoresces over a broad range of wavelength with peak fluorescence at 420 nm. If the Na impurity is replaced by Tl (Thallium) then the screen gives its peak fluorescence at 550 nm (Johns and Cunningham, 1983.; Van Eijk, 1994).
The performance evaluation of phosphor screens

The performance evaluation of radiographic phosphor screens as x-ray to light converters comprise three main stages, the process of radiation absorption, the process of x-ray to light conversion and the process of light transmission to phosphor screen output. The phosphor’s efficiency as an x-ray to light converter is determined by the intensity of light emission (luminescence) with respect to the incident x-ray beam intensity. The phosphor’s coupling efficiency to optical photon detectors used in radiography (films, photodiode arrays) is evaluated by the emitted optical spectrum and how this spectrum is captured by optical detectors. Finally the image information transfer efficiency of the phosphor, giving the information content of the produced diagnostic image, is determined by the MTF, NTF and DQE.

The absolute efficiency

The absolute efficiency of a phosphor is given by the relation:

\[ n(E_0, w) = \frac{\Psi_\lambda(E_0, \lambda, w)}{X(E_0)} \]  

Equation 16

where \( \Psi_\lambda \) is the emitted light energy flux (energy of light per unit of area and time), \( X \) is the incident exposure rate that excites the phosphor to luminescence, \( E_0 \) denotes the maximum x-ray energy determined by the tube voltage, \( w \) is the coating weight of the phosphor and \( \lambda \) is the wavelength of the emitted light. AE depends on the following physical properties:

1. The x-ray absorption efficiency, which is determined by the x-ray absorption coefficient (\( \mu \)) and the coating weight (\( w \)) of the phosphor.
2. The intrinsic x-ray to light conversion efficiency (\( \eta_\lambda \)), which expresses the fraction of absorbed x-ray energy that is converted into light. \( \eta_\lambda \) depends on the fundamental energy band gap of the material, the site symmetry and other intrinsic properties of the phosphor.
3. The light transmission efficiency (\( g_\lambda \)), which expresses the fraction of light that is transmitted through the phosphor and is emitted by the phosphor surface. \( g_\lambda \) depends on the light attenuation properties of the material which are functions of light wavelength \( \lambda \).
The effective efficiency

The effective efficiency ($\eta_{\text{eff}}$) has been defined (Cavouras et al., 1998) by the formula:

$$n_{\text{eff}} = n_A \cdot a_s \quad \text{Equation 17}$$

Where

$$a_s = \frac{\int_{\lambda_2}^{\lambda_1} S_p(\lambda) S_D(\lambda) d\lambda}{\int_{\lambda_1}^{\lambda_2} S_p(\lambda) d\lambda} \quad \text{Equation 18}$$

where $S_p(\lambda)$ is the phosphor's emission spectrum and $S_D(\lambda)$ is the spectral sensitivity distribution of the optical detector used with the phosphor. The factor $a_s$ expresses the spectral compatibility of the phosphor's spectrum with respect to the optical detector (Cavouras et al., 1998).

Radiation detection

Quantum detection efficiency (QDE)

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 (Beutel et al., 2000). For polyenergetic X-rays the QDE of a scintillator layer of coating thickness $w$ is written as:

$$QDE(E) = \frac{\int_{E_0}^{E} \Phi_0(E)(1-e^{-(\mu_{\text{tot},E}(E)/\rho)W})dE}{\int_{E_0}^{E} \Phi_0(E)dE} \quad \text{Equation 19}$$

$\Phi_0(E)$ is the X-ray photon fluence (photons per unit of area) incident on the scintillator. The spectrum of $\Phi_0(E)$ may be given in terms of the X-ray energy fluence spectrum ($\psi_0 = \Phi_0(E)$ (Swank, 1973; Beutel et al., 2000)). $\mu_{\text{tot},E}(E)/\rho$ is the X-ray total mass attenuation coefficient of the scintillator (Hubbel, 1995; Storm and Israel, 1967). For monoenergetic X-rays the QDE of a scintillator layer of coating thickness $w$ is written as

$$QDE(E) = 1 - e^{-(\mu_{\text{tot},E}(E)/\rho)W} \quad \text{Equation 20}$$
Energy absorption efficiency (EAE)

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 (Beutel et al., 2000). EAE may be calculated by the relation

\[
EAE(E) = \frac{E_{0} \int_{0}^{E} \Phi_{0}(E) E \left( \frac{\mu_{\text{tot, en}}(E)}{\mu_{\text{tot, t}}(E)} \right) \left( 1 - e^{-\left( \frac{\mu_{\text{tot, t}}(E)}{\rho} \right) W} \right) dE}{E_{0} \int_{0}^{E} \Phi_{0}(E) E dE} \quad \text{Equation 21}
\]

\( E_{0} \) is the incident X-ray energy fluence and \( \mu_{\text{tot, en}} \) is the total mass energy absorption coefficient of the scintillator. \( \mu_{\text{tot, en}} \) includes all mechanisms of energy deposition locally at the point of X-ray interaction within the scintillators mass. All secondary photons, e.g., K-characteristic fluorescence X-rays, created just after the primary interaction effect, are assumed to be lost (Hubbel, 1994; Hubbel, 1995; Storm and Israel, 1967; Hubbell, 1999).

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).

For monoenergetic X-rays the EAE of a scintillator layer of coating thickness \( w \) is written as (Kandarakis et al., 2005).

\[
EAE(E) = \left( \frac{\mu_{\text{tot, en}}(E)}{\mu_{\text{tot, t}}(E)} \right) \left( 1 - e^{-\left( \frac{\mu_{\text{tot, t}}(E)}{\rho} \right) W} \right) \quad \text{Equation 22}
\]
Image transfer characteristics
Modulation transfer function (MTF)

An x-ray beam transmitted through any object can be represented as a series of sine waves, with each wave possessing a characteristic frequency and amplitude. The transmitted x-ray beam is said to be modulated by the object because the distribution of amplitudes and frequencies is influenced by the object, and hence contains information about the features of the object. The function of an imaging system is to translate the modulation of the transmitted x-ray beam into a visible image and to reproduce the modulation faithfully so that features of the object are apparent. The ability of an imaging system to fulfill this responsibility is described by the modulation transfer function (MTF) (Hendee, 2002).

The spatial-frequency transfer characteristics of linear and spatially invariant imaging systems can be characterized by the system’s modulation transfer function (MTF) (Cunningham, 1992).

Modulation transfer function (MTF) of an imaging system is defined as the absolute value of its optical transfer function, normalized to unity at spatial frequency zero (Neitzel et al., 2004).

\[
MTF(f) = \frac{M(u)}{M(0)}
\]

That is, the MTF of an imaging system is a measure of how well sine waves that describe the transmission of x-rays through an object are represented faithfully in the image (Hendee, 2002). A curve that represents the MTF as a function of spatial frequency is shown in the next figure (Kandarakis et al., 2005):

![MTF curves of Y3Al5O12:Ce for various scintillator screens at 100 kVp.](image)

In the illustration, the value of the MTF is unity (i.e., 100%) at low spatial frequencies, signifying that the imaging system reproduces low frequencies without distortion or loss of resolution. As the frequency increases, the MTF decreases until it reaches zero, signifying that the spatial frequencies are so high that the imaging system provides no reproduction at all.
The MTF is a useful descriptor of the spatial resolution of an imaging system. Other characteristics of the system are also important, including its ability to reveal subtle differences in subject contrast and to provide images with low levels of noise.

The MTF provides information about the clarity of information in images furnished by an imaging system, but it is not a complete descriptor of image clarity. The MTF of any imaging system is a product of the MTFs of the individual components of the system. That is, the MTF of the complete system can be computed if the MTFs of the individual components are known. This principle is illustrated in Example:

At a spatial frequency of 5 cycles per millimeter, the MTFs of components of an x-ray film–screen imaging system are focal spot, 0.9, motion, 0.8, and intensifying screen, 0.7. The MTF of the composite imaging system is $0.9 \times 0.8 \times 0.7 = 0.5$ (Hendee, 2002).

Three methods used to measure the LSF of medical x-ray imaging systems. Slit, square wave response Function (SWRF) and the edge techniques.

**Square-wave response function (SWRF) method**

The square wave method uses a bar pattern with progressively narrower patterns of dark and light to determine the approximate frequency response of a system. The bar pattern is placed in the image, either parallel to the x or y axes, or along a 45° diagonal, and an image is obtained. The pixel values behind the bar pattern are then analyzed to determine the amplitude of response at each of the discrete frequencies included in the bar pattern. This amplitude reflects the square wave response of the system, not the response to a sinusoid. The following equation is the square wave function.

$$E(x) = E_0 + \Delta E_0 \left( \frac{4}{\pi} \sum_{n=1,3,5,7\ldots} \frac{1}{n} \sin(2\pi nx) \right)$$

Equation 24

n=1,3,5,7…, $E_0$ is the mean exposure value and $\Delta E_0$ the modulation amplitude:

$$\Delta E_0 = (E_{\text{max}} - E_{\text{min}}) / 2$$

Equation 25

$\nu$ is the spatial frequency and $n$ an uneven integer. $E(x)$ can be simplified if the beam falls to a sinusoidal object:

$$E(x) = E_0 + \Delta E_0 \cos(2\pi nx)$$

Equation 26

$E_R(x)$ is the distribution or (modulation) of the exposure captured by the imaging system (eg film-screen):
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\[ E_n(x) = \lambda E_o + \lambda E_o \cdot MTF(\nu) \cos[2\pi nx - \phi(\nu)] \]  
Equation 27

It may be converted to a sinusoid response by using the following approximate formula:

\[ M(u) = \frac{\pi}{4} [M'(u) + \frac{1}{3} M'(3u) - \frac{1}{5} M'(5u) + ...] \]  
Equation 28

where \( M(u) \) is the sinusoid response at frequency \( u \) (i.e., the MTF) and \( M'(u) \) is the square wave "transfer function" derived from the bar pattern (Dainty and Shaw, 1974).

![Fig. 21 MTF measurement by the SWRF method.](image)

Contrast transfer function (CTF)

For analog imaging systems, such as a photographic film camera, it is possible to measure the sine wave MTF up to the frequency at which the MTF goes to zero, which is called the 'cut-off' spatial frequency in the optical field, equivalent to the 'band-limit' frequency in electrical engineering parlance. For this case, a simple series expansion formula derived nearly half a century ago (Coltman, 1954) will quite accurately convert the square wave CTF to its equivalent sine wave MTF, or vice versa. CTF is given as:

\[ CTF(f) = \frac{I_{max} - I_{min}}{I_{max} + I_{min}} \]  
Equation 29

Where \( I_{max} \) is the local maxima and \( I_{min} \) is the local contrast minima for a given frequency \( f \).

Given the CTF, the Coltman formula to determine the MTF, is

\[ MTF(f) = \frac{\pi}{4} \left[ C(f) + \frac{C(3f)}{3} - \frac{C(5f)}{5} + \frac{C(7f)}{7} + ... \right] \]

where, \( M(f) = \) sine wave MTF, \( C(f) = \) bar target CTF, \( f = \) spatial frequency.
The analyses and theory presented over the years in technical literature confirms that, for the same imaging system, the bar target derived spatial frequency response and sine target derived MTF are not equal to each other, but, as will be seen, there is a definite mathematical relationship between them. The difference between the two quantities is often highlighted by denoting the spatial frequency response obtained from a bar target as the CTF or square wave response, to distinguish it from the term 'MTF', which is reserved for the sine wave transfer function.

**Slit camera method**

The slit method measures the response of the system to an impulse function, but rather than a delta function, it uses a slit. Thus, the response of the system is given by the convolution of the PSF with the slit. The result is the line spread function (LSF). The slit method determines the MTF perpendicular to the axis of the slit by taking the Fourier amplitude of the LSF. (Technically, the MTF is a two-dimensional function, and the value of the MTF along one of the frequency axes is just the same as the Fourier amplitude of the convolution of the PSF with an infinite slit.) The slit method uses a slit placed at a shallow angle (typically 1.5-3°) with respect to the pixel matrix to measure the LSF at a sampling interval much finer that that provided by the pixel-to-pixel distance. Pixel values in the vicinity of the angled slit represent samplings of the line spread function (LSF) at distances equal to the length from the slit center to the pixel center. The LSF is synthesized as a plot of pixel value versus distance from the slit.

The image data are first converted to be proportional to detected exposure, based on the measured characteristic curve. (Note that if the digital detector is truly linear over the range of exposures used for MTF measurement, then it is unnecessary to convert pixel values to exposure.) Next, the integral of digital values across the slit (perpendicular to the direction of the slit) is computed for each point along the length of the slit. These perpendicular profile integrals are used to normalize all pixels in the vicinity of the slit. This normalization procedure corrects for variations in x-ray intensity attributable to imperfections in the slit. Such as from slight variations in slit width. All subsequent analyses of slit data use the normalized image values.

A plot is then made of the maximum pixel value in the perpendicular profile for each point along the length of the slit. The graph of maximum profile values is used to identify the range of the slit over which acceptable data are available for analysis. The local minima of this curve indicate the points at which the center of the slit is halfway between pixels. The distance in pixels between two of these local minima is used to determine the angle of the slit. Several of these angle determinations are averaged in order to give better precision.
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The LSF is next computed by plotting the image intensity versus distance from the center of the slit for each pixel in a region of interest surrounding the slit. It is necessary to fill in any missing values and resample the LSF so that an identical spacing is used between all points. The MTF is then computed by taking the Fourier transform of the LSF.

\[
MTF(f) = \left| F[LSF(x)] \right| = \left| \frac{1}{\sqrt{2\pi}} \int_{-\infty}^{\infty} LSF(x)e^{i2\pi fx} dx \right|
\]

Equation 30

The zero-frequency value of the FT is used to normalize the MTF to "1.0 at zero frequency. The MTF must be corrected for the finite width of the slit, by dividing by sinc(\(\alpha\)), where \(\alpha\) is the estimated slit width including focal spot blurring (Fujita, 1992; Peppler and Dobbins, 1999; Cunningham, 1992).

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Fig. 22: Cross section of the bevelled slit camera (Radiation Measurements, Inc Carla D. Bradford).

Edge spread function method

One established method to determine the MTF is based on the use of a sharp edge that is imaged to produce an edge spread function (ESF). The ESF is then differentiated to obtain the line spread function (LSF), from which the MTF is calculated by a Fourier transform (Neitzel et al., 2004).

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Fig. 23: MTF measurement by the ESF method.
An edge test device with a well-defined edge is usually realized by carefully machining a thin piece of metal, e.g., lead, tungsten, or platinum. Material thicknesses of 0.1 to 0.25 mm are often used to allow easy manufacturing and handling as well as accurate alignment of the edge in the x-ray beam. Depending on the actual thickness of the material and on the beam quality used for imaging, the metal sheet may be either (almost) fully absorbing or semitransparent. When X-rays hit the edge test device, scattered radiation is inevitably generated. The scattered radiation may exit from the back side of the edge device toward the detector if the material is not thick enough to absorb all radiation. In this case the scattered radiation will superpose on the direct (transmitted or not attenuated) radiation forming the image of the edge transition; thus, the x-ray energy profile at the detector surface differs from the desired step-like profile (figure 24). The measured ESF will therefore be different from the true detector ESF and, consequently, the determined MTF will not be the true detector MTF (Neitzel et al., 2004; Samei et al., 2005).

**Fig. 24:** Distribution of primary and scattered radiation for a semitransparent edge test device (schematically).
Noise power spectrum (NPS)

The signal-to-noise ratio imposes the fundamental limitation to object perceptibility in a digital radiograph because image contrast can be manipulated during the display of digitally acquired radiographic images. Thus, noise characterization plays an increasingly central role in the evaluation of system performance in medical imaging. The noise power spectrum (NPS) is a spectral decomposition of the variance. As such, the NPS of a digital radiographic image provides an estimate of the spatial frequency dependence of the pixel-to-pixel fluctuations present in the image. Such fluctuations are due to the shot (quantum) noise in the x-ray quanta incident on the detector, and any noise introduced by the series of conversions and transmissions of quanta in the cascaded stages between detector input and output (Williams et al., 1999). The NPS is defined as the Fourier transform (FT) of the autocorrelation function, defined in one dimension as:

$$C(x) = \lim_{L \to \infty} \frac{1}{L} \int_{-L/2}^{L/2} p(x + \tau)p^*(\tau) d\tau$$  \hspace{1cm} \text{Equation 31}$$

where $p(x)$ is the value (pixel value for a digital image of the one-dimensional image (1D) at position $x$, and $p^*(x)$ is its complex conjugate. Since $p(x)$ is real, $p(x) = p^*(x)$. The Fourier transform of $C(x)$ is then

$$S(u) = \int_{-\infty}^{\infty} C(x)e^{-2\pi i u x} dx = \lim_{L \to \infty} \frac{1}{L} \int_{-L/2}^{L/2} p^*(\tau)e^{2\pi i u \tau} \int_{-\infty}^{\infty} p(x + \tau)e^{-2\pi i u x} dx e^{-2\pi i u \tau} d\tau$$

$$= \lim_{L \to \infty} \frac{1}{L} \int_{-L/2}^{L/2} p^*(\tau)e^{2\pi i u \tau} \int_{-\infty}^{\infty} p(x)e^{-2\pi i u x} dx d\tau = \lim_{L \to \infty} \frac{1}{L} \int_{-L/2}^{L/2} p^*(\tau)e^{2\pi i u \tau} d\tau P(u)$$  \hspace{1cm} \text{Equation 32}$$

Inspection of Equations 31 and 32 shows that:

$$S(u) \equiv FT\{C(x)\} = \lim_{L \to \infty} \frac{1}{L} |P(u)|^2$$  \hspace{1cm} \text{Equation 33}$$

Thus, the NPS may be calculated either from the Fourier transform of the autocorrelation function (the indirect method), or from the square of the modulus of the Fourier transform of the data itself (the direct method). Note that since the data $p(x)$ are real, that $P(-u) = P^*(u)$. That is, the NPS at a given negative frequency is equal in magnitude to the NPS at the corresponding positive frequency. Extending the expression of equation 33 to two dimensions:

$$S(u, v) \equiv FT\{C(x, y)\} = \lim_{L \to \infty} \frac{1}{L} |P(u, v)|^2$$  \hspace{1cm} \text{Equation 34}$$

With the advent of the fast Fourier transform (FFT) and fast computers, indirect calculation of the NPS via the autocorrelation function has largely been replaced by the direct method. If the stationary random process being characterized by the NPS is also ergodic, as is typically the case for radiographic image
noise, then the spectral estimate is found by ensemble averaging. That is, the final NPS is the average of spectra obtained from a series of uniform irradiation images. Then, equation 34 becomes:

\[ S(u, v) = \lim_{X,Y \to \infty} \left\{ \frac{1}{X \cdot Y} \int_{-X/2}^{X/2} \int_{-Y/2}^{Y/2} p(x, y) \times e^{-2\pi i (ux+vy)} \, dx \, dy \right\} \]

where the pointed brackets denote ensemble averaging. The digital representation of the NPS is:

\[ S_{\text{dig}}(u) = \frac{A_{\text{nic}}}{N_x N_y M} \sum_{i=0}^{M-1} \left( \sum_{n_y=0}^{N_y-1} \sigma_{n_x, n_y} \right)^2 \]

The NPS\text{dig} is thus the average discrete Fourier transform of the average signal variation in the x direction scaled by the pixel size and the number of pixels under consideration. Examples of the latter are gain variances in the conversion of x-ray quanta to light quanta in a phosphor (or to electron–hole pairs in a solid-state detector), statistical fluctuations in the transmission of optical quanta between a scintillator and a photodetector, and additive noise sources such as preamplifier noise. The NPS is a much more complete description of image noise than is quantification of integrated (total) noise via simple measurement of the rms pixel fluctuations, because it gives information on the distribution in frequency space of the noise power. An understanding of the frequency content of image noise can provide insight regarding its clinical impact. For example, in mammography, excess high-frequency noise may render the detection of microcalcifications impossible. If desired, the total variance can be obtained by integrating the NPS over spatial frequency (Williams et al., 1999).

**Noise transfer function (NTF)**

NTF expresses the x-ray quantum noise transfer from input to output as a function of spatial frequency and it is associated with quantum noise content in the resulting radiographic image.

The noise transfer function (NTF) is defined as

\[ \text{NTF}(E_0, u, T) = \left( \frac{\text{NPS}(E_0, u, T)}{\text{NPS}(E_0, 0, T)} \right)^{1/2} \]

Where \( \text{NPS}(E_0, u, T) \) is the spatial frequency depended noise power spectrum and \( \text{NPS}(E_0, 0, T) \) is the zero frequency noise power spectrum.
Detective quantum efficiency (DQE)

While the radiologist is the ultimate authority to judge the diagnostic content of medical x-ray images, objective means are needed to characterize the image quality of x-ray detection systems. Striving towards this goal has gone on for decades. In recent years, the consensus within the community, including researchers, government authorities, and manufacturers, has begun to converge. There was and is a strong need to establish a quantity which is well defined and portable, and which facilitates a comparison of different imaging systems and supports the development of new detecting devices. These will lead to broad acceptance of detective quantum efficiency (DQE) as the fundamentally significant performance parameter that characterizes a detection system. The usefulness of DQE lies in the fact that it can be interpreted as the efficiency of a system to transmit the information that it receives (Stierstorfer and Spahn, 1999).

It is expressed as a function of spatial frequency. It is valid for shift-invariant imaging systems with wide-sense stationary (WSS) noise processes. It can predict the performance of the ideal observer for simple noise-limited visual detection tasks and can be determined objectively from image data. Determination of the DQE requires measurements of the noise-power spectrum (NPS) (Cunningham et al., 2001).

Definition of DQE

The detective quantum efficiency (DQE) is defined as

\[
DQE(u, v) = \frac{(SNR_{out})^2}{(SNR_{in})^2}
\]

Equation 38

where \(SNR_{in}\) is the signal-to-noise ratio (SNR) for an ideal detector that preserves all information in the radiation stream without adding any noise.

The zero spatial frequency detective quantum efficiency, DQE(0), is defined as (Dainty and Shaw 1974):

\[
DQE(0) = \frac{(SNR_{out}(0))^2}{(SNR_{in}(0))^2}
\]

Equation 39

where \(SNR^2_{out}(0)\) is the zero spatial frequency signal-to-noise ratio at the output of the system and \(SNR^2_{in}(0)\) equals the number of x-rays incident on a pixel of the detector (Mah et al., 1999; Cahn et al., 1999). One accepted expression for the two-dimensional DQE is given by:

______________________________ 43 ______________________________
where $\overline{d}$ is the average pixel value in a “flat-field” dark-subtracted image [unitless], MTF(u, v) is the two dimensional system modulation-transfer function (MTF) [unitless], $q$ is the average density of x-ray quanta incident on the system [mm$^{-2}$] while the image is acquired, and NPS(u, v) is the two-dimensional NPS measured from image data [mm$^2$] (Cunningham et al., 2001;Neitzel et al., 2004;Ranger et al., 2005). It should be noted that the theoretical maximum DQE at zero spatial frequency is limited by the X-ray absorption efficiency of the detector material. The DQE(0) can never exceed the absorption efficiency (QDE): $DQE(0) \leq QDE$

The theoretical limit of the MTF is determined by the pixel size of the imaging sensor. Since the sharpest response of the imaging system to a delta signal is as wide as one pixel the maximum MTF is the Fourier transform of a step function with a width of one pixel:

$$MTF_{\text{max}}(f) = \frac{1}{\alpha} \left| \int_{-\alpha/2}^{\alpha/2} e^{i2\pi fx} dx \right| = \frac{\sin \pi f \alpha}{\pi f \alpha}$$

Equation 41

where $\alpha$ is the width of one pixel. Then the theoretical maximum frequency dependent DQE can be determined as:

$$DQE_{\text{max}}(f) = QDE \times \frac{\sin^2 \pi f \alpha}{(\pi f \alpha)^2}$$

Equation 42

Practical example of how DQE impacts object discrimination. The same test object is exposed to x-rays on both film and a direct digital detector. The first image has high resolution and low noise. The second image has high detection (DQE). The detector with the higher DQE will show more of the visible objects as darker spots (Dickson 2003).
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Materials and Methods

In the present study several phosphor screens were prepared by sedimentation techniques. MTF, NTF, and DQE were determined considering that the phosphor screen is divided into a number of consecutive thin fluorescent layers (Nishikawa and Yaffe, 1990). Calculations of image transfer characteristics were performed under either general radiographic conditions, using a W / 2 mm Al anode-filter combination x-ray spectrum, or under mammographic conditions, using a Mo/ Mo anode-filter combination x-ray spectrum. MTF is a function describing the efficiency of signal transfer as a function of spatial frequency and it is indicative of image sharpness and spatial resolution deterioration from the input to the output of an imaging system. NTF expresses the x-ray quantum noise transfer from input to output as a function of spatial frequency and it is associated with quantum noise content in the resulting radiographic image. DQE expresses the degradation of signal to noise ratio (SNR) from input to output and it is indicative of image information content (Dick and Motz, 1981).

Experiment

Preparation of the fluorescent screens by the method of sedimentation

Preparation of Lu$_2$SiO$_5$:Ce screens

The screens, necessary for the experiments, were prepared by sedimentation of a mixture consisting of: a) 2.0 L deionised water, b) 25 ml Na$_2$SiO$_3$ aqueous solution (index of refraction 1.353) and c) the appropriate amount of phosphor (1.0 g to 4.5 g), on suitably selected substrates. Na$_2$SiO$_3$ acts as a binder between the phosphor grains and between the phosphor layer and the substrate. The substrates were in the form of discs made of fused silica (spectrosil B), 30 mm in diameter and 2.5 mm thick. The thickness (coating density) of the screens varied from 25 to 172 mg/cm$^2$. Screens of lower thickness were not studied because of their very low efficiency, while screens of higher thickness were not studied because of their very low spatial resolution (Giakoumakis et al., 1990; Kandarakis et al., 1996).

![Fig. 26: LSO:Ce screens in their protecting case.](image)

The following table shows prepared screens data.
Table 1: LSO:Ce screens data.

<table>
<thead>
<tr>
<th>Screen</th>
<th>Phosphor Weight (gr)</th>
<th>Substrate Weight (gr)</th>
<th>Screen Weight (gr)</th>
<th>Screen diameter (cm)</th>
<th>Screen coating density (mg/cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>LSO</td>
<td>1.004</td>
<td>3.135</td>
<td>3.312</td>
<td>3</td>
<td>25.0</td>
</tr>
<tr>
<td>LSO</td>
<td>1.509</td>
<td>3.763</td>
<td>4.168</td>
<td>3</td>
<td>57.3</td>
</tr>
<tr>
<td>LSO</td>
<td>1.599</td>
<td>3.200</td>
<td>3.648</td>
<td>3</td>
<td>63.4</td>
</tr>
<tr>
<td>LSO</td>
<td>2.506</td>
<td>3.136</td>
<td>3.832</td>
<td>3</td>
<td>98.5</td>
</tr>
<tr>
<td>LSO</td>
<td>2.501</td>
<td>3.206</td>
<td>3.972</td>
<td>3</td>
<td>108.4</td>
</tr>
<tr>
<td>LSO</td>
<td>4.501</td>
<td>3.192</td>
<td>4.411</td>
<td>3</td>
<td>172.5</td>
</tr>
</tbody>
</table>

The screen coating density (mg/cm²) can be found according to next equation:

$$E = \frac{m_1 - m_2}{S}$$  

Equation 43

![Sedimentation tube.](image)

Fig. 27: Sedimentation tube.

Where E is the screen coating density (mg/cm²), \(m_1\) , \(m_2\) are the masses of the substrate after the sedimentation with the phosphor material on it and before sedimentation respectively (in mg). \(S\) is the substrate surface (in cm²). The next figure shows the variation of screen coating density (mg/cm²) with the amount of phosphor used to prepare the screen. The curve fits the measured data. From this curve the amount of phosphor for a specific screen coating density can be easily found.

![Screen coating thickness versus phosphor weight.](image)

Fig. 28: Screen coating thickness versus phosphor weight.
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The following plot shows the percentage (%) of the phosphor powder that was finally deposited on the substrate by the sedimentation process, for various values of screen coating density.

![Graph showing percentage deposition vs. coating thickness.]

**Fig. 29:** Percentage (%) of the powder that was deposited on the screen.

**Spectral matching factors**

To determine both the mean light photon energy $\bar{E}_\lambda$ and the spectral matching factor $\alpha_s$ the emitted light of the LSO:Ce powder phosphor was measured by an Oriel grating optical spectrometer (Ocean Optics Inc., HR2000) while the spectral sensitivities of the optical detectors were obtained from manufacturers' data. Spectrometer light measurements were performed under X-ray excitation. The light emitted by the irradiated LSO:Ce powder phosphor was transferred to the spectrometer through a 2.0 m long, 400 $\mu$m fiber optic, (Avantes Inc. FCB-UV400-2, Colorado, USA). Corrections for light signal degradation due to fiber optic light losses were taken into account.
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Experimental measurement of the modulation transfer function (MTF)

Square-wave response function (SWRF) method

The phosphor was used in the form of a thin layer (test screen) with coating weight of 25 mg/cm² in order to simulate the intensifying screens employed in X-ray mammography. The Modulation Transfer Function of the screens was determined according to the SWRF method (see chapter 1.10.1). The screen was brought in close contact with a radiographic film (Kodak T-Mat) enclosed in a light tight cassette.

![Fig. 30: KODAK T-MAT film curve.](image)

The film-screen combination was irradiated by X-rays on a General Electric Senographe DMR Plus mammographic unit (molybdenum anode-molybdenum filter). The exposure conditions employed in the experiments are: 27 kV and 63mAs (linear region of the film curve). Tube voltage was checked using an RMI model 240 multifunction meter.

In accordance with the square-wave response function (SWRF) method (Barnes, 1979; ICRU, 1986), for MTF determination, an MTF test pattern (typ-53, Nuclear Associates), with lead lines of various widths, corresponding to spatial frequencies from 0.25 to 10 lp/mm, was used. In reflection mode measurements, the film was placed behind the test pattern and in front of the screen. In transmission mode the film was placed behind the screen. The pattern-film–screen system, corresponding to each screen, was exposed to X-rays under conditions identical to those employed for light flux measurements. The films were then developed to obtain the pattern images, as optical density variations across pattern strips (see next figure), which by definition are the SWRFs (Barnes, 1979; ICRU, 1986). The film processor was an Agfa Scopix LR 5200 operated at a temperature of 36 °C and at 90 s processing time. Then the film images were digitized on an Agfa Duoscan scanner with scanning parameters 1000dpi, 8bit. The MTF’s were finally calculated from the digitized image density variations (digital SWRF) across directions...
vertical with respect to the images of the test pattern lines, employing Coltman’s formula (see chapter 1.10.1) (Barnes, 1979; ICRU, 1986).

\[ MTF(f) = \frac{\pi}{4} \left[ C(f) + \frac{C(3f)}{3} - \frac{C(5f)}{5} + \frac{C(7f)}{7} + \ldots \right] \]

The latter gives the MTF as a function of SWRF. The MTF data, obtained in this way, were corrected by dividing the system MTF with the MTF of the scanner and the MTF of the film (Kandarakis et al., 2001; Kandarakis et al., 2005).

Fig. 31: Square-wave response function (SWRF) method for the experimentally determination of the modulation transfer function.
Noise power spectrum (NPS)

NPS was determined following the typical auto-correlation function technique (see chapter 1.10.2) (Williams et al., 1999; Kandarakis et al., 2001). The latter was determined after uniform irradiation of the phosphor screen brought in close contact with the film. Exposure geometry, x-ray tube voltages and radiographic film selected were as in the case of MTF determination. After irradiation and film development, the film images were digitized employing the same technique previously described for MTF measurements. Six regions of interest of 128x128 pixels were selected and image density profiles along pixel rows were obtained.

![Fig. 32: Region of interest of 128x128 pixels for the measurement of NPS.](image)

A self developed Low-Pass filter was applied to each profile data, which were then subjected to Fast Fourier Transform (FFT) and squared. This process was repeated for each row and data obtained were averaged. The noise power spectrum obtained in this way; besides the phosphor screen-film noise, comprises the noise due to the digitising scanner. Film NPS was also measured and subtracted from NPS data to determine the screen NPS. In order to measure the film NPS a film was passed through a film Sensitometer (PTW SensodensiX Combined Device) and then processed by the Agfa Scopix LR 5200 film processor as mentioned above. The optical density of the film’s step which had almost the same value with the screens NPS image was selected in order to measure the film NPS. Film NPS then measured as above and subtracted from the screens NPS.

Noise transfer function (NTF)

The noise transfer function was obtained through the noise power spectrum. NTF was then calculated as the square root of the NPS normalized to zero spatial frequency (Kalivas et al., 2006).
Detective quantum efficiency (DQE)

The Detective Quantum Efficiency at exposure $E$ and frequency $f$ is not measured directly but calculated from the measured data according to the next equation:

$$DQE(f) = \frac{MTF(f)^2}{NTF(f) \cdot q \cdot E} \quad \text{Equation 44}$$

$E$ is the detector incident exposure in mR, and $q$ is the estimate of the number of incident x-ray quanta per unit area per mR or the SNR$^2_{in}$ per unit exposure (Ranger et al. 2005). (The theoretical model predictions, shows that the values of incident x-ray quanta per unit area and the SNR$^2_{in}$ are equal). $q$ can be also defined as the conversion factor between dose and photon fluence per unit area. The values of $q$ and $E$ obtained from the measured values of (air KERMA and the calculated values of the Mammographic x-ray spectra with variable Air KERMA), $q$ is given in $\text{mm}^{-2}\mu\text{Gy}^{-1}$ and can be calculated by subtraction of the photon fluence (photons/mm$^2$) with the air KERMA at a specific x-ray tube voltage value ($\mu\text{Gy}$) (Ranger et al. 2005). The special unit of exposure is roentgen (R), while the SI unit is $2.58 \times 10^{-4}$ C/kg with 1R=$2.58 \times 10^{-4}$ C/kg. Thus:

$$K_{\text{col}}(\text{air}) = (2.58 \times 10^{-4}) \cdot \left[ \frac{C}{\mu\text{Gy}} \right] X = [0.876 \frac{\text{Gy}}{\text{R}}] X \quad \text{Equation 45}$$

With the exposure $X$, in roentgens (Podgorsak 2003). Detective Quantum Efficiency calculated both for reflection and transmission mode. The measured value of air KERMA (at 27 kVp 63 mAs, 30mm, Perspex filter) was 329 $\mu\text{Gy}$. Then by using the previous equation the exposure $E$ in mR (Ranger et al. 2005) is $3.76 \times 10^{-5}$ mR. The method for DQE calculation from measured quantities contains systematic errors due to different methods for each quantity measurement and the relative error that every measurement introduces.
Theoretical interpretation

Image information transfer efficiency

To assess the quality of the radiographic image produced by the Lu₂SiO₅:Ce screen, the MTF, NPS, NTF, DQE and the zero-frequency DQE were evaluated.

Polyenergetic x-ray spectra

For the production of polyenergetic Radiographic x-ray spectra, the Tungsten anode spectral model interpolating polynomials (TASMIP) was used. The 11 spectra with kVp ranging from 40–140 keV excluding 60 keV have a low energy cut-off of about 10 keV, and thus the lowest energy resolved in the TASMIP technique was 10 keV. The 11 spectral entries at the 20 keV bin were fit to a general polynomial equation of \( n \)th order:

\[
\text{If } E \leq \text{kVp}, \quad F(E) = a_0[E] + a_1[E]kV + a_2[E]kV^2 + a_3[E]kV^3 + \ldots + a_n[E]kV^n, \\
\text{else } F(E) = 0.
\]

The least squares polynomial fitting routine ‘‘POLFIT’’ from Bevington used for determining the coefficients \( a_0 - a_n \). At 20 keV, there were 11 pairs of values \( \{\Phi_i, kV_i\} \) where \( i = 1,2,3,\ldots,11 \) that could be used in the fit to solve for coefficients \( a_n \). As the energy of the bin to be fit increased, fewer data were available for the fit procedure. For example, at 55 keV, only eight of the modified Fewell spectra produced x-ray photons in the 55 keV energy bin, since, of course, the 30, 40, and 50 kV spectra did not produce any 55 keV photons. Consequently, as the energy of the bin increased, the number of fluence values contributing to the polynomial fit was gradually reduced. Consequently, the order of the polynomial also was gradually reduced as energy increased, with orders starting at 4 for low energies going to order 2 at higher energies. Negative fluence values could be produced by the polynomial at energies higher than the kV of the spectrum; in this case the fluence was set to zero since negative fluence values have no physical meaning. The polynomial fitting technique was performed at all energies from 10 to 140 keV, in 1 keV steps.

For the production of polyenergetic Mammographic x-ray spectra the TASMIP, MASMIP and RASMIP techniques was used (Boone and Fewell, 1997). In this technique, measured spectra are parameterized using polynomial interpolation of spectral data. Using x-ray spectra measured for molybdenum, rhodium, and tungsten anode x-ray tubes at 13 different kVp (18, 20, 22,\ldots, 42 kV), a spectral model using
interpolating polynomials was used. For each x-ray energy, the x-ray photon fluence was fitted by using 2, 3, or 4 terms, depending on the energy polynomials, as a function of the applied tube voltage. Using the polynomial fit coefficients determined at each 0.5 keV interval in the x-ray spectrum. The molybdenum anode spectral model using interpolating polynomials is given the acronym MASMIP, and the rhodium and tungsten spectral models are called RASMIP and TASMIP, respectively.

Quantum detection and energy absorption efficiency (QDE & EAE)

For the calculation of the quantum detection and energy absorption efficiency the following formulas used, as mentioned earlier in the theory section:

\[
QDE(E) = \frac{\int_{0}^{E_0} \Phi_0(E)(1 - e^{-\left(\frac{\mu_{tot,t}(E)}{\rho} W\right)}dE}{\int_{0}^{E_0} \Phi_0(E)dE}
\]

Equation 19

\[
EAE(E) = \frac{\int_{0}^{E_0} \Phi_0(E)E \left(\frac{\mu_{tot,en}(E)}{\mu_{tot,t}(E)}\right)(1 - e^{-\left(\frac{\mu_{tot,t}(E)}{\rho} W\right)}dE}{\int_{0}^{E_0} \Phi_0(E)dE}
\]

Equation 21

Attenuation and absorption coefficients were calculated using tabulated data.

The intrinsic conversion efficiency

The intrinsic conversion efficiency was calculated by using the equation:

\[
\eta_e = \left(\frac{hc}{\lambda}\right) / \beta E_g
\]

Equation 46

where \(hc/\lambda\) is the average energy of emitted light photons. \(\beta E_g\) represents the average energy that must be transferred by a fast electron (e.g. a photoelectron) to create an electron-hole pair in the scintillator material (Blasse, 1994). \(E_g\) is the forbidden energy band-gap between the valence and the conduction energy bands of the scintillator material (Van Eijk, 2002; Blasse, 1994). \(\beta\) is a parameter related to energy losses to lattice vibrations. For Lu_2SiO_5:Ce scintillator, \(\beta=5.6\) and \(E_g=6.6 \text{ eV}\) (Van Eijk, 2002; Blasse, 1994).
The light transmission efficiency

The light transmission efficiency, $g_A(E, \sigma, \tau, \rho)$ of a scintillating screen may be expressed as follows:

$$g_A(E, \sigma, \tau, \rho) = \int_0^{W_0} \bar{\phi}_X(E, w) g_J(\sigma, \tau, \rho, w) dw dE \quad \text{Equation 47}$$

$W_0$ is the total screen thickness. For the purposes of analysis it has been considered that the screen was divided into a large number of superimposed elementary thin layers of thickness $dw$. Here $w$ denotes the depth of each thin layer from the screen surface. The function $\bar{\phi}_X(E, w)$ describes the relative probability of x-ray absorption at a depth $w$ from the screen surface as follows:

$$\bar{\phi}_X(E, w) = \frac{\mu(E) \exp[-\mu(E)w]dw}{\int_0^{\infty} \mu(E) \exp[-\mu(E)w]dw} \quad \text{Equation 48}$$

where $\mu(E)$ is the x-ray absorption coefficient calculated using tabulated data (Hubbel and Seltzer, 1995; Storm and Israel, 1967). The numerator in (equation 48) gives the probability of x-ray photon absorption at depth $w$. The denominator is equal to the total probability of absorption in a scintillator of thickness $w_0$. The function $g_J(\sigma, \tau, \rho)$ is given by the following relation:

$$g_J(\sigma, \tau, \rho) = \frac{\sigma \rho[(q \beta + \sigma)e^{qw} + (q \beta - \sigma)e^{-qw}]}{(q \beta + \sigma)(q \beta + \sigma \rho)e^{qW_0} - (q \beta - \sigma)(q \beta - \sigma \rho)e^{-qW_0}} \quad \text{Equation 49}$$

where $\sigma$ is the light attenuation coefficient of the scintillator, which is equal to the reciprocal of the light photon diffusion length and it is given as a function of the optical scattering coefficient ($s$), and the optical absorption coefficient ($a$), i.e.

$$\sigma = [\alpha(\alpha + 2s)]^{1/2} \quad \text{Equation 50}$$

In the spatial frequency domain, $\sigma$ is written as, $\sigma = \sigma_o^2 + 4\pi \nu^2 \quad \text{Equation 51}$

where $\sigma_o$ corresponds to zero-frequency (Ludwig, 1971; Swank, 1973). $\tau$ is the inverse relaxation length given as $\tau = \alpha + 2s \quad \text{Equation 52}$. 

$\rho_o, \rho_1$ are optical parameters expressing the reflection of light at the front and back scintillator surfaces:

$$\rho_n = \frac{(1 - r_n)}{(1 + r_n)} \quad \text{Equation 53},$$

where $r_n$ denotes the optical reflection coefficients at the front ($n=0$) and back ($n=1$) screen surfaces. In this study we have used $\rho_0=0.91, \rho_1=0.87$ (obtained from previous studies and the values of $\sigma$ and $\tau$ found by fitting. $q$ in (equation 49) is an optical parameter defined as:

$$q = (4\pi^2 \frac{\mu}{d})^2 + \sigma^2 \quad \text{Equation 54}$$
where $u$ is the spatial frequency and $d$ is the density of the Lu$_2$SiO$_5$:Ce phosphor coating in powder form ($d=3.7$ g cm$^{-3}$). The value of $d$ corresponds to a packing density of 50.6%, which is similar to that of commercial screens. Equation 55 corresponds to transmission mode. To describe reflection mode emission the coating thickness, in equation 56, was replaced by $(W_0-w)$ (Ludwig, 1971; Kandarakis et al., 2001; Cavouras et al., 2000; Kandarakis et al., 2005). The function $g_{2}(x,y,z)$ is determined as a solution to the photon diffusion differential equation describing light propagation through light scattering media.

$$
\begin{align*}
\frac{-\bar{g}_{2}(u,t)}{(q\beta + \sigma)(q\beta + \sigma\rho) e^{qW_0}} &= \frac{\sigma \rho [(q\beta + \sigma) e^{qW_0} + (q\beta - \sigma) e^{-qW_0}]}{(q\beta + \sigma)(q\beta + \sigma\rho) e^{qW_0} - (q\beta - \sigma)(q\beta - \sigma\rho) e^{-qW_0}} \\
\text{Equation 55}
\end{align*}
$$

for light quanta emitted at the screen surface opposing that of the x-ray beam incidence (transmission mode), and by (Nishikawa and Yaffe 1988)

$$
\begin{align*}
\frac{-\bar{g}_{2}(u,t)}{(q\beta + \sigma)(q\beta + \sigma\rho) e^{qW_0}} &= \frac{\sigma \rho [(q\beta + \sigma) e^{q(W_0-w)} + (q\beta - \sigma) e^{-q(W_0-w)}]}{(q\beta + \sigma)(q\beta + \sigma\rho) e^{qW_0} - (q\beta - \sigma)(q\beta - \sigma\rho) e^{-qW_0}} \\
\text{Equation 56}
\end{align*}
$$

for light quanta exiting the screen surface receiving the incident x-ray beam (reflection mode).
Output signal and signal transfer efficiency

The output signal of a scintillating screen is often expressed by either the emitted light energy fluence-$\Psi_A$ (light energy per unit of area) or the emitted light photon fluence-$\Phi_A$ (light photons per unit of area) (Ludwig, 1971; Swank, 1973).

Based on the previously defined intrinsic properties of the phosphor material, the total output signal of a phosphor screen may be expressed by the following relations:

$$\Psi_A(E_0) = \int_0^{E_0} \Psi_0(E)\eta_\epsilon(E)\eta_\Lambda g_A(E,\sigma,\tau) dE$$  \hspace{1cm} \text{Equation 57}

$$\Phi_A(E_0) = \int_0^{E_0} \Phi_0(E)\eta_\epsilon(E)m_\Lambda g_A(E,\sigma,\tau) dE$$  \hspace{1cm} \text{Equation 58}

where $\Psi_0(E)$ is the incident x-ray energy fluence spectral distribution (energy fluence per energy interval) and $\phi_0(E)$ is the incident x-ray photon fluence spectral distribution (photon fluence per energy interval). $E$ is the x-ray photon energy and $E_0$ is the maximum energy of the x-ray spectrum. $\eta_\epsilon(E)$ is the energy absorption efficiency (EAE). $\eta_\epsilon(E)$ is the quantum detection efficiency (QDE). $\eta_\Lambda$ is the intrinsic x-ray to light conversion efficiency. $m_\Lambda(E)$ is the intrinsic quantum conversion gain, $g_A$ is the light transmission efficiency, as described above (Ludwig, 1971; Swank, 1973).
Modulation transfer function

To describe the imaging properties, contrast and spatial resolution, of a scintillating screen, the signal transfer efficiency is often expressed by the modulation transfer function (MTF). This function has been defined as the spatial frequency-dependent output signal (photon fluence) normalized to zero frequency, as follows:

\[
M(E_0, \nu) = \frac{\Phi_\Lambda(E_0, \nu)}{\Phi_\Lambda(E_0, \nu = 0)}
\]

Equation 59

Output noise and noise transfer

The output noise of a scintillating screen is expressed by the variance in the emitted light photon fluence (number of photons per unit of area) over the screen emitting area. In the spatial frequency domain noise is expressed by the noise power spectrum (NPS). The latter may be obtained by the Fourier transform of data obtained after uniform irradiation of the screens (Dainty and Shaw, 1974; Kalivas et al., 2002). NPS has been also expressed in terms of \( \eta_\lambda, m_\lambda \) and \( g_\Lambda \) (Kalivas et al., 2002). NPS was calculated (Shaw and Van Metter, 1984) in terms of emitted optical quantum fluence \( \lambda \Phi \) by equation 57:

\[
W_\lambda(u, \nu) = \Phi_\lambda(E, w)[m(E, \lambda)\{1 + \frac{\varepsilon}{m_0}\}MTF^2(u, \nu) + 1]
\]

Equation 60

where \( m(E, \lambda) \) is the mean number of optical quanta emitted per X-ray quantum absorbed, \( m_0 \) is the mean number of optical quanta created within the phosphor layer per X-ray absorbed, \( \varepsilon \) is a parameter accounting for the excess of the variance in \( m_0 \) with respect to the Poisson distribution, given by (Shaw and Van Metter, 1984).

\[
\varepsilon = (\sigma^2(m_0)/m_0) - 1
\]

Equation 61

The number \( m(E, \lambda) \) of emitted optical quanta per X-ray absorbed was determined by dividing \( \Phi_\lambda \) by the absorbed fraction of \( \Phi_\lambda \). Following previous studies (Nishikawa and Yaffe, 1990; Shaw and Van Metter, 1984) it was assumed that \( m_0 \) follows Poisson distribution, then \( \sigma^2(m_0) = m_0 \) and hence \( \varepsilon = 0 \) (Cavouras et al., 2000).

Noise transfer function (NTF)

The noise transfer function (NTF) as previously defined in equation 37 is:

\[
NTF(E_0, u, T) = \left[\frac{NPS(E_0, u, T)}{NPS(E_0, 0, T)}\right]^{1/2}
\]
Signal to noise ratio and detective quantum efficiency

As it has been previously mentioned the detective quantum efficiency $\eta_D(v)$ of a scintillating screen has been defined by the relation $SNR_{out}^2 / SNR_{in}^2$ (Dainty, 1974). The square of the output noise is given by the noise power spectrum $W(E_0, v)$ (Dainty, 1974). Hence DQE may be written as follows:

$$\eta_D(v) = \frac{(\Phi(0)(E_0)M(E_0, v))^2}{W(E_0, v)SNR_{in}^2} \quad \text{Equation 62}$$

The emitted light photon fluence $\Phi(x)(E_0)$ has been expressed by equation 57. In addition it may be also expressed in terms of experimentally measurable quantities (absolute efficiency, exposure, mean light wavelength) using the relation:

$$\Phi(x) = \frac{(\eta_x X)/hc}{\lambda} \quad \text{Equation 63}$$

Where the numerator is equal to the light energy fluence $\Psi(x) = (\eta_x X)$ and the denominator is equal to the mean energy $(E_{\lambda})$ of the emitted light photons $(E_{\lambda} = hc/\lambda)$, $\lambda$ being the mean light wavelength determined from emission spectra measurements as follows:

$$hc\lambda^{-1} = hc\int\Phi(x)(\lambda)d\lambda / \int\Phi(\lambda)d\lambda \quad \text{Equation 64}$$

where $\phi(\lambda)$ is the scintillator's emission spectrum. The input signal to noise ratio ($SNR_{in}$), for an x-ray imaging detector, can be expressed by the relation:

$$SNR_{in}^2 = \left(\int_0^{E_0} \Phi_0(E)E dE\right)^2 / \int_0^{E_0} \Phi_0(E)E^2 dE \quad \text{Equation 65}$$

where the numerator is equal to the square of the first statistical moment of the distribution of x-ray photons. This distribution is expressed by the x-ray spectrum $\Phi_0(E)$ (Greening, 1985). This is also equal to the square of the total incident x-ray energy fluence (input signal). The denominator is equal to the second moment of the aforementioned distribution. This second moment has been considered to express the input quantum noise. Equation 65 is used for energy integrating detectors, i.e. the detector’s output signal is proportional to the x-ray energy fluence deposited within phosphor mass (Greening, 1985; Kundel et al., 2000). $\Phi_0(E)$ in equation 65 was determined by using: a tungsten anode x-ray spectrum model (for conventional radiography), a Mo anode x-ray spectrum model (for mammography), the incident x-ray exposure measurements and the exposure to photon fluence conversion formula (Beutel et al., 2000).
Zero frequency detective quantum efficiency

The zero frequency detective quantum efficiency ($\eta_D$), which is defined by the ratio $(\text{SNR}_{\text{output}}/\text{SNR}_{\text{input}})^2$, has been also expressed by the relation (Dick and Motz, 1981).

$$\eta_D(E_0, w_0) = \eta_Q(E_0, w_0) \eta_I(\sigma, w_0)$$  \hspace{1cm} \text{Equation 66}

where $I$ is a parameter expressing the statistical fluctuations in the number of emitted light photons per absorbed X-ray photon, often called the statistical factor or the information factor and recently Swank factor. $I$ is defined by the relation (Swank, 1973).

$$I(\sigma, w) = \frac{M_1^2(\sigma, w_0)}{M_2(\sigma, w_0) M_0(\sigma, w_0)}$$  \hspace{1cm} \text{Equation 67}

where

$$M_i(\sigma, w_0) = \int_0^{w_0} \psi_Q(E, w) [g(\sigma, w)]^i dw,$$  \hspace{1cm} \text{Equation 68}

$M_i$ are the zeroth, first and second moments of the statistical distribution of the number of light photons emitted per X-ray absorbed (Swan, 1973; Dick and Motz, 1981). $M_2$ increases with the width of the statistical distribution while for a perfect detector $I = 1$ (Kandarakis et al., 2001).
Additional measurements

Transmission measurement of the LSO:Ce crystal

![Fig. 33: PERKIN-ELMER Lambda 15 UV/VIS Spectrophotometer and the LSO:Ce (1X1X2 cm) crystal imported in it.](image)

The transmission measurement carried out with a PERKIN-ELMER Lambda 15 UV/VIS Spectrophotometer. The scan range, in nanometers, was determined from 200 to 700 nm. The interval was 0.10 nm and the slit 1 nm. The scan speed was 120 nm/min. The ordinate mode set-up to %T. Initially the Spectrophotometer runs a correction baseline procedure from 700 to 200 nm and vice-versa. Transmission measurements carried out for the LSO:Ce crystal and for the glass substrate which is used as substrate for the phosphor materials in the screens. The glass substrate was imported in the Spectrophotometer for the background measurement.

![Fig. 34: Glass substrate imported in the Spectrophotometer for the background measurement.](image)

The same procedure made also for the crystal. The glass substrate removed from the Spectrophotometer. For the measurements the height correction procedure was on. The energy gap of the crystal can be calculated as follows: \[ E_g = h c / \lambda_{cf} \] where \( h \) is the Plank’s constant, \( c \) is the velocity of light and \( \lambda_{cf} \) is the threshold (cutoff) wavelength of the transmission spectrum of the investigated crystal. The absorption edges of crystals doped with Ce are blue shifted compared with the bulk crystals. The energy gap, \( E_g \), is inversely proportional to the square of the particle radius, \[ E_g = E + \left( h^2 \pi^2 / 2 \mu R^2 \right) \] where \( E \) is the bulk band gap \( \hbar = h / 2\pi \), \( h \) the Plank’s constant and \( \mu \) is the effective mass of electron and hole (Battisha, 2002).
Scanning electron microscope SEM measurements
Morphology and size measurement

For observation of the particle size and morphology of the LSO:Ce phosphor, SEM micrographs were taken by using the Jeol JSM 5310 Scanning Electron Microscope (SEM). For SEM, LSO:Ce phosphor specimen was dehydrated in graded ethanol (50-100° GL) critical point dried in CO₂, coated with gold and examined under the scanning electron microscope. Vacuum conditions established for the specimen. The High voltage was set to 20 kV. The following zooming parameters used for the sequence of the sides of interest (x1000, x5000, x3500, x35000, x10000, x5000).

![LSO:Ce sample](image1)

**Fig. 35:** LSO:Ce sample.

Electron micrographs were obtained with digitalization of images in PC software.

![Jeol JSM 5310 Scanning Electron Microscope (SEM)](image2)

**Fig. 36:** Jeol JSM 5310 Scanning Electron Microscope (SEM).
Results and Discussion

In the following sections experimental and theoretical model results are presented for the following parameters:

Experimental results
Detector optical gain (DOG)

The detector optical gain (DOG) is defined as the number of emitted optical photons (NEP) per incident X-ray quantum. DOG is directly related to the image brightness that can be achieved for a given level of radiation fluence incident on the phosphor. Additionally, DOG may be expressed as a function of spatial frequency. In this case DOG is equivalent to the contrast transfer function (CTF) usually defined for films and screen-film systems used in medical radiography.

![Fig. 37: Variation of the detector optical gain with x-ray tube voltage for the 25 mg/cm² thick screen](image)

The value of \( \sigma \) for LSO:Ce was determined by considering the variation of this coefficient with the mean value of the emitted light wavelength. This variation was determined using data on \( \sigma \) and on emitted mean light wavelength, obtained from previous studies on various phosphors (Kandarakis et al., 1997; 2001a–c). It was found that \( \sigma \) varied according to the relation:

\[
\sigma = \alpha + \beta(\bar{\lambda})^{-1} + c(\bar{\lambda})^{-2}
\]

Equation 69

where \( a, b, c \) are fitted parameters. This variation of \( \sigma \) is in accordance with what was expected from the well-known light absorption and light-scattering laws, which state that light attenuation increases with decreasing wavelength (van de Hulst, 1957). Using previous equation, the value of \( \sigma \) for LSO:Ce was calculated to be \( \sigma = 75 \; \text{cm}^2/\text{g} \) given that \( \bar{\lambda} = 420 \; \text{nm} \). In a similar way \( \beta_0 \) was found equal 0.03.

The value of \( \sigma \) can be also determined by fitting to detector optical gain (DOG).
Chapter 3   Results and Discussion

Fig. 38: Variation of the light attenuation coefficient ($\sigma$) values as a function of light wavelength corresponding to various phosphors (indicative data points displayed).

**Air KERMA**

**Radiographic unit**

The next table shows the values of X-ray tube voltage (kVp) and Air KERMA (µGy) as they were measured by a multimeter Victoreen 4000 Cardinal Health at a Philips Optimus radiographic unit. Measurements obtained using 22 mm of Al filter, 63 mA tube current, 1 s time and 72.5 cm Source to Detector Distance (SDD).

<table>
<thead>
<tr>
<th>X-ray tube voltage (kVp)</th>
<th>Air KERMA (µGy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>40</td>
<td>9</td>
</tr>
<tr>
<td>50</td>
<td>19</td>
</tr>
<tr>
<td>60</td>
<td>64</td>
</tr>
<tr>
<td>70</td>
<td>150</td>
</tr>
<tr>
<td>80</td>
<td>297</td>
</tr>
<tr>
<td>90</td>
<td>491</td>
</tr>
<tr>
<td>100</td>
<td>745</td>
</tr>
<tr>
<td>110</td>
<td>1072</td>
</tr>
<tr>
<td>120</td>
<td>1440</td>
</tr>
<tr>
<td>130</td>
<td>1872</td>
</tr>
<tr>
<td>140</td>
<td>2361</td>
</tr>
</tbody>
</table>
Mammographic unit

The next table shows the values of X-ray tube voltage (kVp) and Air KERMA (μGy) as they were measured by a multimeter Victoreen 4000 Cardinal Health at a General Electric Senographe DMR Plus mammographic unit (molybdenum anode-molybdenum filter). Measurements obtained using 30 mm of Perspex filter, 63 mAs.

<table>
<thead>
<tr>
<th>X-ray tube voltage (kVp)</th>
<th>Air KERMA (μGy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>22</td>
<td>84</td>
</tr>
<tr>
<td>25</td>
<td>189</td>
</tr>
<tr>
<td>28</td>
<td>329</td>
</tr>
<tr>
<td>30</td>
<td>442</td>
</tr>
<tr>
<td>35</td>
<td>780</td>
</tr>
</tbody>
</table>

Fig. 39: Victoreen multimeter at 72.5 cm SDD.

Fig. 40: Air KERMA versus x-ray tube voltage.

Fig. 41: Air KERMA versus x-ray tube voltage.
X-ray response characteristic curve

The x-ray characteristic curve (signal versus exposure) is plotted in the next figures and shows a linear dependence between 1-372 mR/s for Radiography and 4.8-114 mR/s for Mammography. The linear no-threshold fit gave a reduced $R^2$ of 0.9961 and 0.9575 for radiography and mammography respectively, which is very close to the most likely reduced $R^2$ value (1) and indicates a satisfactory fit.

**Fig. 42, 43:** The characteristic curve of the phosphor in the fluoroscopic range of exposures. The circles are measurements, and the solid line is a linear no threshold fit to the data. The result shows that the system is linear over the range of exposures used in the experiment. The characteristic curve of the phosphor in the Mammographic range of exposures. The circles are measurements, and the solid line is a linear no threshold fit to the data. The result shows that the system is linear over the range of exposures used in the experiment.

Spectral matching factors

**Fig. 44:** Optical emission spectrum of LSO: Ce phosphor measured at various x-ray tube voltages (kVp) for 25mg/cm$^2$ coating thickness.

Figure 44 shows the measured light emission spectrum of the LSO:Ce phosphor under excitation by X-rays at various x-ray tube voltages. The peak value of the light spectrum was found at 420 nm. The long tail on the right part of the spectrum should be ascribed to the $5d \rightarrow 4f$ electronic transitions of the Ce$^{3+}$.
ion, which locates at Ce2 center. The Ce2 emission, much weaker and not well resolved with respect to Ce1 emission at (393-423 nm), occurs at 500 nm. LSO has two Lu ions with 6 and 7 oxygen ligands (and consequently two Ce\(^{3+}\)) activator sites, called Ce1 and Ce2. The figure is uncorrected and the measurement was conducted at room temperature (Van Eijk 2002; Pidol 2004). The emitted light energy fluence shows a tendency to increase with increasing tube voltage.

![Normalized spectral response of the LSO:Ce compared to various optical detectors versus wavelength.](image)

Table 9 shows the spectral matching factors of the LSO:Ce. It exhibits excellent compatibility with the AgfaGS and KodakGR radiographic films. In addition it was found adequately compatible with the Amorphous Silicon (AmorSi) photodiode (Michail et al., 2006).

**Table 4**: Spectral matching factors of the LSO:Ce screen.

<table>
<thead>
<tr>
<th>Optical Detectors</th>
<th>Lu(_2)SiO(_4).Ce</th>
</tr>
</thead>
<tbody>
<tr>
<td>GaAs</td>
<td>0.916</td>
</tr>
<tr>
<td>Si</td>
<td>0.320</td>
</tr>
<tr>
<td>AmorSi</td>
<td>0.580</td>
</tr>
<tr>
<td>MAMORAY</td>
<td>0.873</td>
</tr>
<tr>
<td>E/S 20</td>
<td>0.965</td>
</tr>
<tr>
<td>AgfaGS</td>
<td>0.960</td>
</tr>
<tr>
<td>KodakGR</td>
<td>0.965</td>
</tr>
<tr>
<td>FujiUM</td>
<td>0.896</td>
</tr>
</tbody>
</table>
Modulation transfer function (MTF)

Figure 46 shows experimental MTF curves of the LSO:Ce screen measured at 27 kVp (Mo-Mo anode filter combination), in both reflection and transmission mode (mammographic imaging conditions). In reflection mode MTF was found 27% higher at 10 lp/mm (e.g. 0.24 and 0.18 at 10 lp/mm for reflection and transmission respectively) (Michail et al., 2006). The superiority of reflection over transmission mode MTF values indicates the better resolution properties of conventional (reflection) over digital (transmission) mammography detectors for identical phosphor material and equal phosphor thickness.

Fig. 46: Experimentally determined MTF of the LSO:Ce powder phosphor screen (27 kVp 25 mg/cm²).

Figure 47 shows a comparison between the MTFs of the 25 mg/cm² LSO:Ce screen and the commercially employed ‘Kodak Min-R film-screen system’. This system is based on a 31.7 mg/cm² thick Gd₂O₂S:Tb phosphor, approximately corresponding to a quantum detection efficiency equal to that of the 25 mg/cm² LSO:Ce screen, as shown in the next Figure.
Fig. 47: Variation of calculated QDE of LSO:Ce and Gd$_2$O$_2$S:Tb with X-ray tube voltage for 25 mg/cm$^2$ and 31.7 mg/cm$^2$ powder screens.

Gd$_2$O$_2$S:Tb MTF data, (Bunch et al., 1987) were measured at 30 kVp X-ray tube voltage. The LSO:Ce screen was found with higher MTF than the Kodak Min-R. It must be noted however, that the Gd$_2$O$_2$S:Tb screen is thicker. This difference may be explained by: (i) the lower thickness of the LSO screen (ii) the lower light emission wavelength of the LSO at 420 nm. Low light wavelength photons are strongly absorbed within the phosphor mass, especially in lateral directions. This could lead into sharp output light distribution which improves screen resolution properties.

Fig. 48: Comparison of MTFs of LSO:Ce and Kodak Min-R screens as measured experimentally in reflection mode.


table

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Noise power spectrum (NPS)

Figure 49 shows the noise power spectrum (NPS) of the LSO:Ce phosphor screen. The noise power spectrum is more precisely termed the power spectral density, that is, the noise power per unit frequency. The quantity \( S(u,v)\,dudv \) (see equation 35) is the contribution to the variance from noise components with spatial frequencies between \( u \) and \( u+du \), and \( v \) and \( v+dv \). \( S(u,v) \) thus has dimensions of inverse spatial frequency squared. The correct scaling of the NPS can be verified by observing that the volume (area) under the 2D (1D) NPS equals the total variance. Thus, integration over frequency of spectra estimated using mean-zero sections from uniform illumination images should numerically equal the mean-square deviations from their average of the pixel values in those sections. Noise power spectra are sometimes presented normalized by the square of the large-area (average) output signal. This normalization has the effect of compensating for difference in gain between two systems which are being compared, and of simplifying the expression for the DQE. LSO:Ce compared to GDOS:Tb (Kodak Min-R) exhibit lower NPS (GDOS:Tb at 26 kVp 13.6 mR) but higher than CsI:Tl. (FOS-HR 26 kVp 13.2 mR) Gd2O2S-based mammographic phosphor screen Min-R and structured CsI:Tl scintillator (optimized for high spatial resolution FOS-HR) for mammographic exposures ranging from 2 to 40 mR at 26 kVp (Jee 2002). The lower NPS exhibited by the LSO:Ce compared to GDOS:Tb is due to the lower screen quantum gain.

Fig. 49: NPS of the 25 mg/cm² LSO:Ce screen at 27 kVp 63 mAs.

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Figure 50: Comparison of Noise power spectrum (NPS) for GDOS:Tb, CsI:Tl and LSO:Ce detectors.

Figure 51 shows optical density variation of the films step pattern. The optical density that corresponds to that of the screen film image, were used for the measurement, was 0.74.

**Noise transfer function (NTF)**

Figure 52 shows the noise transfer function (NTF) of the LSO:Ce phosphor screen. NTF decreases with spatial frequency, although at a slower rate than MTF. This is because noise is transferred more efficiently than signal in the higher spatial frequencies as it has been shown in previous publications (Lubberts 1968; Badano et al. 2004; Nishikawa and Yaffe, 1989).
Detective quantum efficiency (DQE)

Figure 53 shows the detective quantum efficiency (DQE) of the 25 mg/cm$^2$ LSO:Ce phosphor screen in reflection and transmission mode. Reflection mode DQE is superior than that of transmission mode. Zero frequency DQE for both mode’s are 0.54 which however, are lower than the corresponding detection efficiency which is 0.75. As it has previously mentioned DQE at zero spatial frequency is limited by the X-ray absorption efficiency of the detector material. The DQE(0) can never exceed the absorption efficiency (QDE): $DQE(0) \leq QDE$. The DQE of our LSO:Ce screen appears to be superior (figure 54), compared to published data for the GDOS:Tb in the Kodak Min-R screen and CsI:Tl, in the whole spatial frequency range (Bunch; Nishikawa Yaffe 1990; Zhao and Rowlans 2004). DQE of the LSO:Ce compared to GDOS:Tb (Min-R at 30 kVp, 31.7 mg/cm$^2$ ) and CsI:Tl. (FOS-HR 26 kVp 13.2 mR), is higher for spatial frequencies up to 4.2 lp/mm. Gd2O2S-based mammographic phosphor screen Min-R and structured CsI:Tl scintillator (optimized for high spatial resolution FOS-HR) for mammographic exposures (Nishikawa and Yaffe, 1989; Jee 2002).
Fig. 53: DQE of the 25 mg/cm$^2$ LSO:Ce screen at 27 kVp 63 mAs.

Fig. 54: Comparison of the 25 mg/cm$^2$ LSO:Ce screen at 27 kVp 63 mAs with GDOS:Tb Min-R at 30 kVp and CsI:Tl FOS-HR.
Theoretical model results

Polyenergetic x-ray spectra

Radiographic x-ray spectra with variable Air KERMA

The following x-ray spectra shown in the next figure, were calculated as described previously (see chapter 2.2.1). All spectra are filtered by 2 mm aluminum inherent filter (Al) and an additional 20 mm Al filtration, to simulate beam quality (x-ray spectrum shape) alteration by an average human body. Tungsten anode was used. Relative voltage ripple was 0.05 and Air KERMA from the measured values.

![TASMIP Spectra](image)

**Fig. 55:** Plot of the Photon Fluence (Photons/mm²) versus Energy (keV) in Radiographic conditions.

Mammographic x-ray spectra with variable Air KERMA

The following molybdenum anode x-ray spectra have been calculated as described previously. Spectra from 22 to 35 kVp were filtered by 30 µm Molybdenum inherent filter (Mo), 38 and 40 kVp were filtered by 25 µm Rhodium (Rh) and finally 42 kVp was filtered by 1mm of Aluminum. To take into account x-ray attenuation and spectrum shape alteration by human breast, the beam was transmitted through a 3.5 cm thick slab of Lucite.
Output signal and signal transfer efficiency

The output signal of a scintillating screen can be expressed by the emitted light photon fluence- $\Phi_A$ (light photons per unit of area) as it was described earlier. The next figure is a comparison of the incident x-ray photon fluence spectral distribution (photon fluence per energy interval) with the emitted light photon fluence- $\Phi_A$ (light photons per unit of area), as they were calculated using the model. This figure demonstrates the gain effect of the screen corresponding to various x-ray energies contained in the spectrum.

Fig. 56: Plot of the Photon Fluence (Photons /mm²) versus Energy (keV) in Mammographic conditions.

Fig. 57: Comparison of the input photon fluence with the emitted light photon fluence.
Quantum detection and energy absorption efficiency (QDE & EAE)

Shown in the next figures is the variation of calculated QDE and EAE with X-ray tube voltage for various scintillator layers (mg/cm$^2$). Calculations were performed for a range of X-ray tube voltages larger than that used in ordinary examinations. This was achieved in order to estimate scintillator detection efficiency under medical examinations requiring specialised equipment (e.g., breast and some type of chest examinations). The first point to note is that EAE differs significantly from QDE.

Energy Absorption Efficiency expresses strictly the fraction of incident radiation energy that is locally deposited at the point of primary photon interaction. All other types of secondary radiation, e.g., scattered, $K$ or $L$- fluorescence and bremsstrahlung, are excluded from the calculations. Hence only the useful energy for image formation is taken into account. The curves were calculated by assuming exponential X-ray absorption, determined by the screen thickness and the X-ray mass energy absorption coefficients, as described previously. The values of the energy absorption coefficients of Lu$_2$SiO$_5$:Ce, Gd$_2$O$_2$:Tb, CsI:Tl and YTaO$_4$:Nb scintillators were calculated from tabulated data on absorption coefficients.

X-ray energy absorption and quantum detection decreased approximately exponentially with increasing energy but increased with increasing coating thickness in both radiographic and mammographic energy range (figures. 58-67). At low X-ray tube voltages, thick screens (172 mg/cm$^2$) absorb relatively large fractions of incident X-ray energy (e.g., 0.8 for Radiographic conditions at 40 kVp in figure 61, which is converted into light energy. At higher voltages X-ray photons are more penetrating and X-ray energy absorption is lower (e.g., 0.46 at 70 kVp and 0.33 at 100 kVp for the 172 mg/cm$^2$ screen (Figure 61)). As it may be seen, at 40 kVp, EAE (0.8) is approximately 12% lower than QDE (0.91) (figures 58, 61). This deviation increases with increasing X-ray tube voltage, being approximately 41% at 100 kVp. In the low X-ray tube voltage region from 25 to 30 kVp, QDE takes values higher than 0.96 (at 172.5 mg/cm$^2$, figure 63. In the same region EAE ranges from 0.83 to 0.87 at the same coating thickness (figure 66). For higher voltages employed in general radiography and fluoroscopy QDE retains values higher than 0.52 for voltages up to 140 kVp (at 172.5 mg/cm$^2$) (figure 58). To compare the X-ray absorption performance of Lu$_2$SiO$_5$:Ce with currently employed materials, similar calculations were performed for Gd$_2$O$_2$:Tb, CsI:Tl and YTaO$_4$ scintillators, which are widely used in medical radiography, fluoroscopy, mammography and X-ray computed tomography (GDOS:Pr). Figures 68-70 illustrate the variation of calculated QDE and EAE with X-ray tube voltage for the 35 mg/cm$^2$ Lu$_2$SiO$_5$:Ce screen in the mammographic energy range. Calculations have shown that in the 30–40 kVp voltage range, the QDE of Lu$_2$SiO$_5$:Ce is very close to that of YTaO$_4$ and higher than those of CsI:Tl and Gd$_2$O$_2$:Tb (e.g., 0.86, 0.89, 0.70 and 0.80 at 25 kVp, respectively (figure 68). EAE of Lu$_2$SiO$_5$:Ce
is clearly higher than those of CsI:Tl, YTaO₄ and Gd₂O₂S:Tb (e.g., 0.75, 0.64, 0.64 and 0.73 at 25 kVp, respectively (figure 69). At higher voltages both Gd₂O₂S:Tb and CsI:Tl had higher energy absorption efficiency than Lu₂SiO₅:Ce. QDE had higher values than Gd₂O₂S:Tb and CsI:Tl for energies lower than 50 and higher than 80 kVp. Compared to YTaO₄, has higher values in the whole radiographic energy range. These findings were expected, considering the higher value of the radiation detection index $\rho Z^4_{eff}$ (=143x10⁶) of Lu₂SiO₅:Ce with respect to corresponding index values of Gd₂O₂S:Tb and CsI:Tl (103x10⁶, 38x10⁶, respectively). As it may be seen, at 28 kVp, EAE (0.71) is approximately 15% lower than QDE (0.83 figures 63, 66). Calculations also showed that the Lu₂SiO₅:Ce powder scintillator has approximately 7% higher values of QDE and 2% higher values of EAE than Gd₂O₂S: Tb (figures 68, 69). CsI:Tl scintillator has the lowest X-ray detection in this region of energy.

Quantum Detection and Energy Absorption Efficiency (QDE & EAE) results for the Radiographic Energy range.

![Graph](image)

**Fig. 58:** X-ray quantum detection efficiency (QDE) for Lu₂SiO₅:Ce screens.
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Fig. 59: X-ray monoenergetic quantum detection efficiency (QDE) at 140 kVp for Lu₂SiO₅:Ce screens.

Fig. 60: X-ray quantum detection efficiency (QDE) versus coating thickness for Lu₂SiO₅:Ce screens.
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Fig. 61: X-ray energy absorption efficiency (EAE) for Lu$_2$SiO$_5$:Ce screens.

Fig. 62: X-ray energy absorption efficiency (EAE) versus coating thickness for Lu$_2$SiO$_5$:Ce screens.

Quantum Detection and Energy Absorption Efficiency (QDE & EAE) results for the Mammographic Energy range.
Fig. 63: X-ray quantum detection efficiency (QDE) for Lu$_2$SiO$_5$:Ce screens.

Fig. 64: X-ray monoenergetic quantum detection efficiency (QDE) at 40 kVp for Lu$_2$SiO$_5$:Ce screens.
Fig. 65: X-ray quantum detection efficiency (QDE) versus coating thickness for Lu$_2$SiO$_5$:Ce screens.

Fig. 66: X-ray energy absorption efficiency (EAE) for Lu$_2$SiO$_5$:Ce screens.
Fig. 67: X-ray energy absorption efficiency (EAE) versus coating thickness for Lu$_2$SiO$_5$:Ce screens.

Fig. 68: Comparison of X-ray quantum detection efficiency (QDE) for Lu$_2$SiO$_5$:Ce versus Gd$_2$O$_2$:S:Tb, CsI:Tl and YTaO$_4$:Nb in mammographic conditions for 35 mg/cm$^2$ coating thickness screens.
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Fig. 69: Comparison of X-ray energy absorption efficiency (EAE) for Lu$_2$SiO$_5$:Ce versus Gd$_2$O$_2$:Tb, CsI:Tl and YTaO$_4$:Nb in mammographic conditions for 35 mg/cm$^2$ coating thickness screens.

Fig. 70: Comparison of X-ray monoenergetic quantum detection efficiency (QDE) for Lu$_2$SiO$_5$:Ce versus Gd$_2$O$_2$:Tb, CsI:Tl and YTaO$_4$:Nb in mammographic conditions for 35 mg/cm$^2$ coating thickness screens.
Fig. 71: Comparison of X-ray quantum detection efficiency (QDE) for Lu$_2$SiO$_5$:Ce versus Gd$_2$O$_2$:S:Tb, CsI:Tl and YTaO$_4$:Nb in radiographic conditions for 80 mg/cm$^2$ coating thickness screens.

Fig. 72: Comparison of X-ray monoenergetic quantum detection efficiency (QDE) for Lu$_2$SiO$_5$:Ce versus Gd$_2$O$_2$:S:Tb, CsI:Tl and YTaO$_4$:Nb in radiographic conditions for 80 mg/cm$^2$ coating thickness screens at 140 kVp.
Fig. 73: Comparison of X-ray energy absorption efficiency (EAE) for Lu$_2$SiO$_5$:Ce versus Gd$_2$O$_2$:Tb, CsI:Tl and YTaO$_4$:Nb in radiographic conditions for 80 mg/cm$^2$ coating thickness screens.
Modulation transfer function (MTF)

Modulation Transfer Function predictions given by the theoretical model are shown below. Figure 74 shows the MTF curves of several Lu$_2$SiO$_5$:Ce screens calculated in transmission mode by using optical parameters ($q, \sigma, \rho$ and $\beta$) determined with a combination of experimentally measured and calculated quantities. The calculated curves are at 90 kVp, using a W x-ray spectrum to simulate general radiographic conditions. It is observed that the MTF curves do not change significantly for screens thicker than 98 mg/cm$^2$. Figure 75 shows a comparison between the MTFs of Lu$_2$SiO$_5$:Ce, Gd$_2$O$_2$:Tb, CsI:Tl and YTaO$_4$:Nb screens, using a 90 kVp W spectrum with 80 mg/cm$^2$ screen thickness for general radiographic conditions. Spatial resolution and sharpness of Lu$_2$SiO$_5$:Ce are better than those of Gd$_2$O$_2$:Tb and CsI:Tl but lower than those of YTaO$_4$:Nb. These findings may be attributed to deviations in light output at various frequencies, which are due to differences in parameters $\mu(E)$, $\sigma$ and $\beta$ between the four materials. Almost the same findings are shown in figure 76 for these materials in mammographic conditions. For the comparison a 28 kVp Mo spectrum to simulate mammographic conditions, with 35 mg/cm$^2$ screen thickness were used. Again spatial resolution and sharpness of Lu$_2$SiO$_5$:Ce are better than those of Gd$_2$O$_2$:Tb and CsI:Tl but lower than those of YTaO$_4$:Nb.

![Figure 74: Lu$_2$SiO$_5$:Ce screens Modulation Transfer Function for various coating thicknesses at 90 kVp.](image)

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Fig. 75: Modulation Transfer Function comparison between Lu$_2$SiO$_5$:Ce, Gd$_2$O$_2$S:Tb, CsI:Tl and YTaO$_4$:Nb in radiographic conditions for 80 mg/cm$^2$ coating thickness screens at 90 kVp.

Fig. 76: Modulation Transfer Function comparison between Lu$_2$SiO$_5$:Ce, Gd$_2$O$_2$S:Tb, CsI:Tl and YTaO$_4$:Nb in mammographic conditions for 35 mg/cm$^2$ coating thickness screens at 28 kVp.

The next figures shows the MTF of the theoretical model compared to others from published data. The first is the MTF of a 88 mg/cm$^2$ (300 μm nominal thickness) CsI:Tl at 70 kVp. The dots are from a previous work (Zhao and Rowlands 2004) and the continuous line from this work. The figure shows the very good agreement of the two MTFs and is given for validation purposes. The second is the MTF of a GDOS:Tb screen used in the Kodak Min-R 2000/4316 (Kodak RP X-OMAT Chemistry, Eastman Kodak company) at 28 kVp. Again the predictions of the model are in very good agreement with the
YTaO₄: Nb validated with data published by Beutel (Beutel 1993) for a 80mg/cm² screen at 70 kVp. Again there is very good agreement between the theoretical model results and published data.

Fig. 77: Validation of the theoretical model’s MTF through comparison with previous work.

Fig. 78: Validation of the theoretical model’s MTF through comparison with previous work.

Fig. 79: Validation of the theoretical model’s MTF through comparison with previous work.
Comparison of the experimental MTF findings and model data

For comparison purposes this section demonstrates both the experimental and the model predicted MTF values. For the comparison, an LSO:Ce phosphor was used in the form of a thin layer (test screen) with coating weight of 25 mg/cm² in order to simulate the intensifying screens employed in X-ray mammography. The screen was brought in close contact with a radiographic film (Kodak T-Mat) enclosed in a light tight cassette. The film-screen combination was irradiated by X-rays on a General Electric Senographe DMR Plus mammographic unit (molybdenum anode-molybdenum filter). The exposure conditions employed in the experiments are: 27 kV and 63mAs. For the theoretical model MTF calculation 28 kVp Mo spectrum and a 25 mg/cm² screen was used. The light attenuation coefficient of the scintillator $\sigma$ was allowed to vary over a wide range in order to fit the calculated MTF values to the experimental measured values. Figure 80 shows experimental MTF curves of the LSO:Ce screen measured at 27 kVp in transmission mode compared with the theoretical model data. The agreement between model predictions and the experimental MTF is better at low frequencies while in medium and high frequencies ($f=15$ to $f=100$ mm⁻¹) the model overestimates the experimental values. This is due to the disadvantage of the model to produce accurate results for very thin screens and low x-ray tube voltages (Nishikawa and Yaffe, 1990; Swank, 1973). The theoretical model is valid under the following assumptions: (i) there are no discontinuities (in the sense of gross nonuniformities) in the properties of the screen, (ii) the probability of absorption is small compared with the probability of scattering and (iii) solutions are sought for points far from the source. Assumptions (i) and (ii) are valid for granular phosphors used in medical imaging. Assumption (iii) is valid conditionally. The overestimation of the model occurs because thin screens do not fully comply with assumptions (i) and (ii). That is: (i) The thin screen may not be perfectly homogeneous (i.e. presents discontinuities), due to lower uniformity in the phosphor grain deposition. (iii) The distance between the point of light creation within the phosphor mass and the screen output may not be adequate for the majority of the phosphor layers in thin screens. Figure 81 shows the experimental and the theoretical model MTF curves of the same screen in reflection mode. Results obtained in reflection mode showed higher MTF. In reflection mode the theoretical model simulation shows better agreement with the experimental MTF values than that of the transmission mode.
Fig. 80: Comparison of MTFs of a LSO:Ce powder phosphor as measured experimentally and as produced by the theoretical model in transmission mode.

Fig. 81: Comparison of MTFs of a LSO:Ce powder phosphor as measured experimentally and as produced by the theoretical model in reflection mode.
Noise power spectrum (NPS)

Figures 82 and 83 show a comparison between the NPS of Lu$_2$SiO$_5$:Ce, Gd$_2$O$_2$:S:Tb, CsI:Tl and YTaO$_4$:Nb screens, using similar conditions as in the case of MTF. Figure 82 shows NPS for general radiographic conditions (80 mg/cm$^2$ coating thickness screens at 90 kVp) and figure 83 shows NPS for mammographic conditions (35 mg/cm$^2$ coating thickness screens at 28 kVp). In the general radiographic conditions Lu$_2$SiO$_5$:Ce appears to have lower NPS than that of YTaO$_4$:Nb but higher than those of Gd$_2$O$_2$:S:Tb and CsI:Tl. The same findings shown in figure 83 for mammographic conditions. The generally lower NPS exhibited by the Gd$_2$O$_2$:S:Tb and CsI:Tl is due to the lower detector optical gain of these phosphors.

Fig. 82: Noise Power Spectrum (NPS) comparison between Lu$_2$SiO$_5$:Ce, Gd$_2$O$_2$:S:Tb, CsI:Tl and YTaO$_4$:Nb in radiographic conditions for 80 mg/cm$^2$ coating thickness screens at 90 kVp.

Fig. 83: Noise Power Spectrum (NPS) comparison between Lu$_2$SiO$_5$:Ce, Gd$_2$O$_2$:S:Tb, CsI:Tl and YTaO$_4$:Nb in mammographic conditions for 35 mg/cm$^2$ coating thickness screens at 28 kVp.
Noise transfer function (NTF)

Figures 85 and 86 show a comparison between the NTF of Lu$_2$SiO$_5$:Ce, Gd$_2$O$_2$:Tb, CsI:Tl and YTaO$_4$:Nb screens, using similar conditions as in the case of MTF. Figure 85 shows NTF for general radiographic conditions (80 mg/cm$^2$ coating thickness screens at 90 kVp) and figure 86 shows NTF for mammographic conditions (35 mg/cm$^2$ coating thickness screens at 28 kVp). The variation of NTF with spatial frequency for the four phosphors follows a pattern similar to that of the MTF. This is expected because MTF and NTF depend on the same optical and x-ray attenuation properties. The NTF values are higher than the corresponding MTF values, indicating that quantum noise is more efficiently transferred through the screen in the higher spatial frequencies. This may also be explained by the fact that NTF can be expressed as the weighted sum of the squares of the thin layers MTFs (Lubberts 1968; Nishikawa and Yaffe, 1990; Kalivas et al., 1968). These MTFs are affected by the shape of the output light bursts, as it has already been noted. The degradation of NTF with frequency however is slower than the corresponding MTF degradation. This effect has been previously explained by considering that the overall MTF is expressed as a weighted sum of the MTFs of each thin layer (Kandarakis et al.,1997; Kalivas et al. 2006). Furthermore the demonstrated phosphor screens NTF are just above (for the 90 kVp) and just below the K-edge (for the 28 kVp). The NTF just above the K-edge is poorer than the NTF just below the K-edge. This occurs because the presence of K X-rays reduces the transfer characteristic of the phosphor screen, since a fraction of the energy carried away by the K X-rays is reabsorbed in sites away from the site of the primary interaction. This in turn reduces total screen NTF. Furthermore, above the K-edge the absorption efficiency of the phosphor is highly increased, related to the absorption efficiency just below the K-edge. This has the consequence that the light photons are generated in layers near the input of the phosphor, thus their corresponding shape light bursts as the output is broadened. On the other hand, just below the K-edge there are no K X-rays and the absorption efficiency of the phosphor is reduced. Therefore, the interactions mainly occur in layers near the screen output leading to a narrower shape of the light bursts and, therefore, better transfer characteristics and higher NTF values. LSO:Ce and YTaO$_4$:Nb phosphors has K-edges at 63 and 67 keV respectively. GDOS:Tb and CsI:Tl has K-edges at 34 and 51 keV respectively. In radiographic imaging conditions the MTF values of these four phosphors are affected by the K X-rays in addition with the mammographic imaging conditions where all four phosphors are below the K X-rays.
Fig. 84: Schematically representation of the absorption efficiency above and below K-edge.

![Image of absorption efficiency above and below K-edge]

Fig. 85: Noise Transfer Function (NTF) comparison between Lu$_2$SiO$_5$::Ce, Gd$_2$O$_2$S::Tb, CsI:Tl and YTaO$_4$:Nb in radiographic conditions for 80 mg/cm$^2$ coating thickness screens at 90 kVp.

![Graph showing NTF comparison]

Fig. 86: Noise Transfer Function (NTF) comparison between Lu$_2$SiO$_5$::Ce, Gd$_2$O$_2$S::Tb, CsI:Tl and YTaO$_4$:Nb in mammographic conditions for 35 mg/cm$^2$ coating thickness screens at 28 kVp.

![Graph showing NTF comparison]
Zero frequency detective quantum efficiency (DQE(0))

The variation of DQE(0) with screen thickness for the Lu$_2$SiO$_5$:Ce phosphor is presented in figures 87 and 89 for the general radiographic and mammographic conditions respectively. DQE(0) was found to increase at low x-ray energies and at low to medium coating thicknesses. At low energies the quantum detection efficiency (QDE) is significant, giving high DQE(0) values. At low thicknesses, distances travelled and attenuation suffered by the generated optical pulses are small, thus, emitted optical pulses are of approximately equal magnitudes. This results in narrow optical pulse statistical distributions, which correspond to low values of statistical moment ($M_2$), inducing high DQE(0) values. At higher screen thicknesses, although the x-ray detection efficiency is increased, $M_2$ is relatively high, because of greater fluctuations in the amplitudes of the emerging optical pulses, thus causing a decrease in the statistical factor $I$ and an overall reduction in DQE(0). Lu$_2$SiO$_5$:Ce appears with higher DQE(0) than YTaO$_4$:Nb (higher QDE of the LSO:Ce than that of YTaO:Nb) in both radiographic and mammographic conditions and with higher DQE(0) than Gd$_2$O$_2$:Tb for screen coating thicknesses up to 63 mg/cm$^2$ for radiography and up to 40 mg/cm$^2$ for mammography. These is due to the lower value of light attenuation coefficient $\sigma$ of Gd$_2$O$_2$:Tb, giving a better optical pulse statistical response and the higher x-ray to light conversion efficiency ($\eta_c = 0.207$, 0.08 and 0.1 for Gd$_2$O$_2$:Tb, Lu$_2$SiO$_5$:Ce and YTaO$_4$:Nb respectively), producing larger number of optical photons ($m_0$) per x-ray detected. This increases the mean value of the optical pulse statistical distribution and, thus, the first moment $M_1$ resulting in higher DQE(0) (Kandarakis et al., 1997; Kandarakis et al., 2001).

Fig. 87: Zero-frequency detective quantum efficiency (DQE(0)) versus screen coating thickness of Lu$_2$SiO$_5$:Ce phosphors for x-ray tube voltages from 40 to 140 kVp.
Fig. 88: Zero-frequency detective quantum efficiency (DQE(0)) versus screen coating thickness of Lu$_2$SiO$_5$:Ce, Gd$_2$O$_5$:Tb, and YTaO$_4$:Nb phosphors for 90 kVp x-ray tube voltage.

Fig. 89: Zero-frequency detective quantum efficiency (DQE(0)) versus screen coating thickness of Lu$_2$SiO$_5$:Ce phosphors for x-ray tube voltages from 22 to 40 kVp.
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Fig. 90: Zero-frequency detective quantum efficiency (DQE(0)) versus screen coating thickness of Lu$_2$SiO$_5$:Ce, Gd$_2$O$_2$S:Tb, and YTaO$_4$:Nb phosphors for 28 kVp x-ray tube voltage.
Statistical factor I (Swank factor)

Shown in figures 91 through 96 are the variation of the statistical factor with screen coating thickness and x-ray tube voltage. The statistical factor decreases as the screen coating thickness and the x-ray tube voltage increases (figures 91-93 and 95). As mentioned before, at low thicknesses, distances travelled and attenuation suffered by the generated optical pulses are small, thus, emitted optical pulses are of approximately equal magnitudes. This results in narrow optical pulse statistical distributions, which correspond to low values of statistical variance ($M_2$), inducing high DQE(0) values. At higher screen thicknesses, although the x-ray detection efficiency is increased, $M_2$ is relatively high, because of greater fluctuations in the amplitudes of the emerging optical pulses, thus causing a decrease in the statistical factor I. Figures 94 and 96 show that Gd$_2$O$_2$S:Tb retains higher statistical factor values than those of Lu$_2$SiO$_5$:Ce and YTaO$_4$:Nb. This can be explained due to the fact that Gd$_2$O$_2$S:Tb emits green light in contrast to Lu$_2$SiO$_5$:Ce and YTaO$_4$:Nb which emits blue light.

Fig. 91: Variation of the Swank Factor versus screen coating thickness for various x-ray tube voltages.
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Fig. 92: Variation of the Swank Factor versus x-ray tube voltage for various screen coating thicknesses.

Fig. 93: Variation of the Swank Factor versus screen coating thickness for various x-ray mammography tube voltages.
Fig. 94: Variation of the Swank Factor versus screen coating thickness of Lu₂SiO₅:Ce, Gd₂O₂S:Tb, CsI:Tl and YTaO₄:Nb phosphors for 90 kVp x-ray tube voltage.

Fig. 95: Variation of the Swank Factor versus x-ray tube voltage for various screen coating thicknesses.

Fig. 96: Variation of the Swank Factor versus screen coating thickness of Lu₂SiO₅:Ce, Gd₂O₂S:Tb and YTaO₄:Nb phosphors for 28 kVp x-ray tube voltage.
Detective quantum efficiency (DQE)

DQE expresses the degradation of signal to noise ratio (SNR) from input to output and it is indicative of image information content (Dick and Motz, 1981). Figure 98 shows the variation of DQE with spatial frequency for various phosphor layers employed in our measurements. At zero spatial frequency, the 172 mg/cm$^2$ layer showed the highest DQE value in radiographic imaging conditions and at the 25 mg/cm$^2$ in mammographic imaging conditions, while at the medium and high frequency range, the DQE of the 25 mg/cm$^2$ layer was found higher for radiographic imaging conditions. Very low DQE values were obtained for the thickest phosphor layer of 172 mg/cm$^2$ at medium and high frequencies (at 28 kVp). DQE variation with coating weight and frequency is determined by the corresponding variations of $\Phi_\lambda$ and $m$, affected by coating weight, and MTF, affected by both frequency and coating weight. Thus, DQE is clearly decreasing with spatial frequency affected by a similar MTF behavior. However, at low spatial frequencies, MTF values are close to unity and hence the influence of the emitted optical quantum fluence $\Phi_\lambda$ on DQE is more evident. On the other hand, the thin layer of 25 mg/cm$^2$ displayed better DQE at medium and high frequencies because of its high MTF (Cavouras et al., 2000).

Fig. 97: Variation of DQE with spatial frequency for the 25, 57, 63, 98, 108, and 172 mg/cm$^2$ Lu$_2$SiO$_5$:Ce screen coating thicknesses (90 kVp).
Fig. 98: Variation of DQE with spatial frequency for the 25, 57, 63, 98, 108, and 172 mg/cm$^2$ Lu$_2$SiO$_5$:Ce screen coating thicknesses (28 kVp).

Fig. 99: Detective Quantum efficiency (DQE) comparison between Lu$_2$SiO$_5$:Ce, Gd$_2$O$_2$S:Tb, CsI:Tl and YTaO$_4$:Nb in radiographic conditions for 80 mg/cm$^2$ coating thickness screens at 90 kVp.
Fig. 100: comparison between Lu$_2$SiO$_5$:Ce, Gd$_2$O$_2$:Tb, CsI:Tl and YTaO$_4$:Nb in mammographic conditions for 35 mg/cm$^2$ coating thickness screens at 28 kVp.
Comparison of experimental DQE findings and model data

For comparison purposes this section demonstrates both the experimental and the model predicted DQE values. For the comparison, an LSO:Ce phosphor screen was used, with coating weight of 25 mg/cm² in order to simulate the intensifying screens employed in X-ray mammography. For the theoretical model MTF calculation 28 kVp Mo spectrum and a 25 mg/cm² screen was used just like for the MTF calculation. Figure 101 shows experimental DQE curves of the LSO:Ce screen measured at 27 kVp in reflection mode compared with the theoretical model data. The agreement between model predictions and the experimental DQE is better at low frequencies while in medium and high frequencies ($f=15$ to $f=100$ mm$^{-1}$) the model overestimates the experimental values. This is due to the disadvantage of the model to produce accurate results for very thin screens and low x-ray tube voltages (Nishikawa and Yaffe, 1990; Swank, 1973).

![Figure 101: Comparison of Experimental (27 kVp 63 mAs) DQE findings and model data (28 kVp) for the 25 mg/cm² LSO:Ce screen.](image-url)
Additional measurements

Transmission measurement in the LSO:Ce crystal

Optical transmission spectrum of the LSO:Ce crystal was measured using PERKIN-ELMER Lambda 15 UV/VIS Spectrophotometer, in the wavelength range 0.2 to 0.7 \( \mu \text{m} \). The next figure shows the optical transmission spectrum.

![Optical transmission spectrum of LSO:Ce crystal](image)

**Fig. 102:** Optical transmission spectrum of LSO:Ce crystal (10X10X20 mm).

Absorption and Transparency Measurements on the fused silica substrate (Spectrosil B)

The purpose of this measurement was to confirm that the emission wavelength of the phosphors doesn’t influence drastically the absorption and scattering properties of the substrate. E.g. the values of \( \rho_0 \cdot \rho_1 \) and consequently the light attenuation coefficient of the scintillator, \( \sigma \), which is included in the Swanks formula for the MTF calculation (described in the section 2.2.4) doesn’t change significantly from green light to blue.

![Diagram showing the maximum emission wavelength between LSO:Ce and GDOS:Tb phosphor materials](image)

**Fig. 103** Diagram showing the maximum emission wavelength between LSO:Ce and GDOS:Tb phosphor materials.
LSO:Ce phosphor exhibits maximum emission wavelength at 420 nm and GDOS:Tb at 545 nm respectively. Low light wavelength photons are strongly absorbed within matter, especially in lateral directions. This could lead into sharp output light distribution which improves screen resolution properties.

**Fig. 104:** sharp output light distribution of LSO:Ce compared to that of GDOS:Tb, which improves screen resolution properties.

If the absorption spectrum of the substrate shows a considerable change when shifting from the green region to blue then the values of $\rho_0$, $\rho_1$ and consequently the value of $\sigma$ should be changed. $\rho_0$ and $\rho_1$ values has been measured previously for phosphors emitting in the green region, like GDOS:Tb. (Ludwig, 1971; Kandarakis et al., 2001).

These values were $\rho_0 = 0.91$, $\rho_1 = 0.87$. Figure 47 shows that at 420 nm (LSO:Ce’s emission wavelength) the transparency of the substrate is 92.92% and at 545 nm (GDOS:Tb’s emission wavelength) the transparency is 93.30%. These findings reveals that emission wavelength doesn’t influence drastically the reflection and scatter properties of the substrate, consequently the value of $\sigma$ will be unaffected because the values of $\rho_0$ and $\rho_1$ are practically the same even for LSO:Ce (blue light).

**Fig. 105:** Optical transmission and absorption of the fused silica substrate (Spectrosil B).
Scanning electron microscope SEM measurements
Morphology and size measurement

For verification of the particle size and morphology, SEM micrographs of phosphor powders were taken by using the Jeol JSM 5310 Scanning Electron Microscope (SEM). Some fragments of SEM images of LSO:Ce phosphor powder are presented in figure 106. Qualitatively the particle sizes of LSO:Ce phosphor has a mean grain size of 8 μm.

Fig. 106: 1. SEM images from four sites of interest of LSO:Ce phosphor. 2. (a) Electron images of the LSO:Ce phosphor (b) Silicon only (c) Cerium only (d) Lutetium only. 3. Real particle distribution in solid state reaction for LSO:Ce (a) phosphor and every substance separately (b) Cerium, (c) Lutetium (d) Silicon.

Fig. 107: Line map of the LSO:Ce phosphor with the relative appearance of it’s substance.
Conclusions

In the present study, six LSO:Ce powder scintillating screens with coating thicknesses 25, 57, 63, 98, 108 and 172 mg/cm$^2$ were prepared and examined under X-ray mammographic and radiographic imaging conditions.

Experimental MTF curves of the LSO:Ce screen measured at 27 kVp showed that reflection mode MTF was approximately 40% higher than transmission mode. LSO:Ce screen was found with higher MTF than the commercially used Kodak Min-R screen. The superiority of reflection over transmission mode MTF values indicates the better resolution properties of conventional (reflection) over digital (transmission) mammography detectors for identical phosphor material and equal phosphor thickness.

The NPS of LSO:Ce was found lower than the corresponding NPS of GdOS:Tb (Kodak Min-R at 26 kVp 13.6 mR) but higher than CsI:Tl. (FOS-HR 26 kVp 13.2 mR). The lower NPS exhibited by the LSO:Ce is due to the lower screen quantum gain.

Reflection mode DQE is superior than transmission mode. For both modes zero frequency DQE is 0.54 which however, is lower than the corresponding detection efficiency which is 0.75. The DQE of our LSO:Ce screen appears to be superior, as compared to published data for the GDOS:Tb in the Kodak Min-R screen and CsI:Tl, in the whole spatial frequency range (Bunch et al.1993; Nishikawa Yaffe 1990; Zhao and Rowlans 2004). DQE of the LSO:Ce compared to GDOS:Tb (Min-R at 30 kVp, 31.7 mg/cm$^2$) and CsI:Tl. (FOS-HR 26 kVp 13.2 mR), is higher for spatial frequencies up to 4.2 lp/mm.

The X-ray QDE was found higher than currently employed materials (e.g. Gd$_2$O$_2$S:Tb, CsI:Tl but lower than YTaO$_4$:Nb) for detection of X-rays used in mammographic applications. X-ray energy absorption efficiency (EAE) was found higher than currently employed materials (e.g. Gd$_2$O$_2$S:Tb, CsI:Tl and YTaO$_4$:Nb) for detection of X-rays used in mammographic applications.

The LSO:Ce powder scintillator (25 mg/cm$^2$) has approximately 10% higher values of QDE and 4.5% higher values of EAE than Gd$_2$O$_2$S:Tb The X-ray quantum detection efficiency (QDE) was found higher than currently employed materials (e.g. Gd$_2$O$_2$S:Tb, CsI:Tl and YTaO$_4$:Nb) for detection of X-rays used in radiographic applications for tube voltages lower than 50 kVp and higher than 80 kVp.

Theoretical model predictions gave the following findings: Spatial resolution and sharpness of Lu$_2$SiO$_5$:Ce are better than those of Gd$_2$O$_2$S:Tb and CsI:Tl but worse than that of YTaO$_4$:Nb. These findings may be attributed to deviations in light output at various frequencies, which are due to differences in parameters $\mu(E)$, $\sigma$ and $\beta$ between the four materials. Almost the same findings hold
for these materials in mammographic conditions. Again spatial resolution and sharpness of Lu₂SiO₅:Ce are better than those of Gd₂O₂S:Tb and CsI:Tl but lower than those of YTaO₄:Nb.

For the comparison of the experimental and the model predicted MTF values, light attenuation coefficient \( \sigma \) of the scintillator was allowed to vary over a wide range of values, in order to fit the calculated values to the experimental ones.

The validation of the theoretical model showed very good agreement, compared to published data, for x-ray tube voltages used in general radiographic conditions. The agreement between model predictions and the experimental MTF values is better at low frequencies while in medium and high frequencies \((f=15\) to \(f=100\) mm\(^{-1}\)) the model overestimates the experimental values (Transmission mode) for x-ray tube voltages used in mammographic conditions. This is due to the inability of the model to produce accurate results for very thin screens and low x-ray tube voltages. (Nishikawa and Yaffe, 1990).

In reflection mode the model simulation shows better agreement with the experimental MTF values than that of the transmission mode.

LSO:Ce and YTaO₄:Nb phosphors have K-edges at 63 and 67 keV respectively. GDOS:Tb and CsI:Tl has K-edges at 36 and 50.2 keV respectively. In radiographic imaging conditions the MTF values of these four phosphors are affected by the K X-rays. However at 28 kVp used for the calculation, the MTFs of all of the examined phosphors are unaffected by the K X-rays.

In the general radiographic conditions Lu₂SiO₅:Ce appears to have lower NPS than that of YTaO₄:Nb but higher than those of Gd₂O₂S:Tb and CsI:Tl. The same findings hold for mammographic conditions. The generally lower NPS exhibited by the Gd₂O₂S:Tb and CsI:Tl is due to the lower detector optical gain of these phosphors.

The variation of NTF with spatial frequency for the four phosphors follows a pattern similar to that of the MTF. This is expected because MTF and NTF depend on the same optical and x-ray attenuation properties. The NTF values are higher than the corresponding MTF values, indicating that quantum noise is more efficiently transferred through the screen in the higher spatial frequencies. The degradation of NTF with frequency however is slower that the corresponding MTF degradation.

DQE(0) was found increased at low x-ray energies and at low to medium coating thicknesses. At low energies the QDE is significant, giving high DQE(0) values. At low thicknesses, distances travelled and attenuation suffered by the generated optical pulses are small, thus, emitted optical pulses are of approximately equal magnitudes. This results in narrow optical pulse statistical distributions, which correspond to low values of statistical variance \((M_2)\), inducing high DQE(0) values. At higher screen thicknesses, although the x-ray detection efficiency is increased, \(M_2\) is relatively high, because of greater fluctuations in the amplitudes of the emerging optical pulses, thus causing a decrease in the statistical factor I and an overall reduction in DQE(0). LSO:Ce appears with higher DQE(0) than YTaO₄:Nb in
both radiographic and mammographic conditions and with higher DQE(0) than Gd$_2$O$_2$S:Tb for screen coating thicknesses up to 63 mg/cm$^2$ for radiography and up to 40 mg/cm$^2$ for mammography. These is due to the lower value of light attenuation coefficient of Gd$_2$O$_2$S:Tb, giving a better optical pulse statistical response and the higher x-ray to light conversion efficiency ($\approx$0.207, 0.08 and 0.1 for Gd$_2$O$_2$S:Tb, LSO:Ce and YTaO$_4$:Nb respectively), producing larger number of optical photons ($m_0$) per x-ray detected. This increases the mean value of the optical pulse statistical distribution and, thus, the first moment $M_1$ resulting in higher DQE(0).

The statistical factor decreases as the screen coating thickness and the x-ray tube voltage increases. As mentioned before, at low thicknesses, distances travelled and attenuation suffered by the generated optical pulses are small, thus, emitted optical pulses are of approximately equal magnitudes. This results in narrow optical pulse statistical distributions, which correspond to low values of statistical variance ($M_2$), inducing high DQE(0) values. At higher screen thicknesses, although the x-ray detection efficiency is increased, $M_2$ is relatively high, because of greater fluctuations in the amplitudes of the emerging optical pulses, thus causing a decrease in the statistical factor 1. Gd$_2$O$_2$S:Tb retains higher statistical factor values than those of LSO:Ce and YTaO$_4$:Nb. This can be explained due to the fact that Gd$_2$O$_2$S:Tb emits green light in the contrary with LSO:Ce and YTaO$_4$:Nb which emits blue light.

At zero spatial frequency, the 172 mg/cm$^2$ layer showed the highest DQE value in radiographic imaging conditions and at the 25 mg/cm$^2$ in mammographic imaging conditions, while at the medium and high frequency range, the DQE of the 25 mg/cm$^2$ layer was found higher for radiographic imaging conditions. Very low DQE values were obtained for the thickest phosphor layer of 172 mg/cm$^2$ at medium and high frequencies (at 28 kVp). DQE variation with coating weight and frequency is determined by the corresponding variations of $\Phi_4$ and $m$, affected by coating weight, and MTF, affected by both frequency and coating weight. Thus, DQE is clearly decreasing with spatial frequency affected by a similar MTF behavior. However, at low spatial frequencies, MTF values are close to unity and hence the influence of the emitted optical quantum fluence $\Phi_4$ on DQE is more evident. On the other hand, the thin layer of 25 mg/cm$^2$ displayed better DQE at medium and high frequencies because of its very high MTF (Cavouras et al., 2000).

The agreement between theoretical model predictions and the experimental DQE is better at low frequencies while in medium and high frequencies ($f=15$ to $f=100$ mm$^{-1}$) the model overestimates the experimental values. This is due to the disadvantage of the model to produce accurate results for very thin screens and low x-ray tube voltages (Nishikawa and Yaffe, 1990; Swank, 1973).
The emission spectrum of LSO:Ce screen showed excellent spectral compatibility with currently used detectors and taking also into account its very fast response it could be considered for applications in X-ray mammographic imaging systems.

The value of the intrinsic conversion efficiency ($n_c=0.08$), is lower than the corresponding value of Gd$_2$O$_2$S:Tb ($n_c=0.2$) but approximately equal to the values of CsI:Tl, CsI:Tl and NaI:Tl phosphors ($n_c=0.10$), used in a large variety of radiation detectors.

Emission wavelength doesn’t influence drastically the reflection and scatter properties of the substrate, so the same values for $\sigma$, $\rho_0$, and $\rho_1$ can be used even for LSO:Ce (blue light) as those used for Gd$_2$O$_2$S:Tb. Qualitatively the particle sizes of LSO:Ce phosphor has a mean grain size of 8 $\mu m$, found by Scanning Electron Microscope (SEM) measurements.

LSO:Ce exhibits excellent compatibility with the AgfaGS and KodakGR radiographic films. It was also found adequately compatible with the Amorphous Silicon (AmorSi) photodiode.

The peak value of the light spectrum was found at 420 nm. The long tail on the right part of the spectrum should be ascribed to the $5d \rightarrow 4f$ electronic transitions of the Ce$^{3+}$ ion, which locates at Ce2 center. The Ce2 emission, much weaker and not well resolved than Ce2 emission at (393-423 nm), occurs at 500 nm. LSO:Ce maybe a competitive alternative in terms of detection and image transfer characteristics to the currently used detectors e.g. GdOS:Tb and CsI:Na. A major drawback of the LSO:Ce is it’s intrinsic background activity ($318$ counts $s^{-1} cm^{-3}$) due to 176Lu in the compound and it’s high cost compared to the other two materials. The theoretical model’s predictions reveals that YTaO4:Nb should be a very promising phosphor and it’s performance should be investigated in a future work.
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