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Luminescence in medical image science

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ABSTRACT

Radiation detection in Medical Imaging is mostly based on the use of luminescent materials (scintillators and phosphors) coupled to optical sensors. Materials are employed in the form of granular screens, structured (needle-like) crystals and single crystal transparent blocks. Storage phosphors are also incorporated in some x-ray imaging plates. Description of detector performance is currently based on quality metrics, such as the Luminescence efficiency, the Modulation Transfer Function (MTF), the Noise Power Spectrum (NPS) and the Detective Quantum Efficiency (DQE) can be defined and evaluated. The aforementioned metrics are experimental evaluated for various materials in the form of screens. A software was designed (MINORE v1) to present image quality measurements in a graphical user interface (GUI) environment. Luminescence efficiency, signal and noise analysis are valuable tools for the evaluation of luminescent materials as candidates for medical imaging detectors.

1. Introduction

Depending on the their physical principles, Medical Imaging disciplines are classified into those based on *Ionizing Radiations*, i.e. x-ray Diagnostic Radiology and Nuclear Medicine, and those based on *non – Ionizing* effects such as Magnetic Resonance Imaging (MRI), Ultrasound (US), Optical Imaging, etc. Techniques of Imaging follow two basic directions: (i) *projection imaging*, where a three dimensional object is projected on to a two dimensional area and (ii) *tomographic imaging*, where cross sectional images are mathematically reconstructed from data obtained through measurements on a three dimensional object. From the medical point of view imaging methods are separated into two branches: (i) *morphological Imaging* depicting shape, dimensions, coordinates and mechanical movement of anatomical structures and (ii) *functional imaging* aiming to the detection of biological mechanisms down to the molecular level [1–7].

Detection of Ionizing Radiation for Medical Imaging purposes is mostly based on the use of luminescent materials, i.e. scintillators and phosphors, coupled to optical sensors of various technologies, including hydrogenated amorphous silicon (a-Si:H) photodiode arrays, complementary metal oxide semiconductors (CMOS), charge coupled devices (CCD) (for x-ray imaging) and photomultipliers (PMT), avalanche photodiodes (APD), Silicon Multipliers (SiPM) (for det3ection of radionuclides). Luminescence based detectors are often referred to as *indirect detection* imaging systems while detectors employing photoconductors (e.g. amorphous selenium) or semiconductors (e.g. cadmium telluride) are characterized as *direct detection* systems [1,3–7]. Luminescence materials are also used for Radiation Dosimetry purposes during medical imaging examinations, i.e. thermoluminescent crystals are preferred in most cases.

Scintillators and phosphor materials are employed under various forms: (a) large area granular screens, employed in traditional projection imaging including Digital Radiography and Fluoroscopy, (b) structured (needle-like) crystals incorporated in the form of large area thin layers into Digital Imaging Detectors (i.e. Digital Radiography flat panels, etc.), (c) single crystal transparent blocks traditionally applied in nuclear medicine and storage phosphors are also incorporated in some x-ray imaging plates. Luminescence efficiency, decay time, spatial and spectral distribution of emitted light are principal properties for ranking the suitability of scintillators for Medical Imaging. Traditionally Tb activated materials have shown superior efficiency while Ce activated crystals or powders exhibit faster response [1,6,7].

On the basis of their principle of operation detectors are divided into two main categories: (a) *energy integrating devices* and (b) *photon counting devices*. Detectors operated in energy integrating mode produce an output signal directly proportional to the total radiation energy absorbed within the scintillator mass (or equivalently within the mass of some other type of x-ray converter) [6,7].

On the other hand detectors operated in photon counting mode produce a series of temporally separated output pulses, each one representing an x-ray or gamma ray photon absorbed in the detector. The amplitude V of each pulse is directly proportional to the incident photon energy.

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Energy integrating systems are mostly employed in Diagnostic Radiology and in Radiation Therapy (Portal Imaging) and use granular or structured scintillators coupled to flat panel amorphous silicon photodiode arrays. In Nuclear Medicine detectors are principally photons counting devices based on single crystal scintillators coupled to photomultipliers or photodiode arrays.

2. Materials and methods

2.1. Theories

The output signal of an energy integrating system can be expressed in the following form:

$$S_{out}(x,y) = \int_0^{kV} \frac{d\Psi(E)}{dE} \prod_i g_i(E,x,y)dE \quad (1)$$

where Ψ denotes the energy fluence of the incoming radiation (e.g. x-rays), E is the energy of x-ray photons and $d\Psi/dE$ is the spectral density of the radiation spectrum integrated over the energy scale, g_i are gain factors expressing the conversion of the input signal at the various stages of signal transmission through the detector system, e.g. g_1 represents the absorption of x-rays (quantum detection efficiency), g_2 is the intrinsic conversion efficiency of absorbed x-ray energy into light, g_3 is the light transmission efficiency, i.e. conversion of light created within a scintillator mass into light emitted by the scintillator surface, g_4 expresses the light diffusion and spreading during transmission to the scintillator surfaces, g_5 is the conversion of light into electrons in the optical sensor of the detector, etc. If only the scintillator (phosphor) is considered, (i.e. without taking into account the optical sensor and the electronic part of the detector), then the output signal S_{out} is equal to the emitted light energy flux Ψ_Λ. However in most studies related to theoretical and experimental analysis of imaging systems, the light photon fluence Φ_Λ (number of photons per unit of area) has been traditionally employed for measurements and calculations. In these cases the spectral density is also expressed in number of photons per unit of area ($d\Phi/dE$) and the gain g_2 of the second conversion stage expresses the number of light photons created within the scintillator mass per absorbed x-ray, $g_2 = m_\lambda = \eta_C E/h\nu$, where η_C is the intrinsic conversion efficiency, given in terms of the energy gap between valence and conduction bands and $h\nu$ denotes the mean energy of emitted light photons [3,4,8,9].

Description of detector performance is currently based on signal and noise analysis in both space and spatial frequency domains. Within this framework basic quality metrics, such as the emission efficiency, the Modulation Transfer Function (MTF), the Noise Power Spectrum (NPS) and the Detective Quantum Efficiency (DQE) can be defined and evaluated. This type of analysis is traditionally based on the theory of *Linear Cascaded Systems Theory* (LCS) in which a system (detector) is assumed to be *linear*, *shift invariant* and *ergodic*. A full imaging system can be *decomposed* into *cascaded stages*, each one representing a particular physical process contributing to image formation (Eq. (1)). The *output* (both signal and noise) of a previous stage is seen as the input to the next stage, where *input signal* is the incident x-ray fluence, *output signal* is the emitted light fluence, *input noise* is the variance in spatial distribution of incident photons and *output noise is the variance in the spatial distribution of emitted photons*. The stages are characterized by their particular transfer characteristics and can be divided into: (i) *quantum gain stages*, expressing the conversion and change in the number of carriers (photons, electrons) and (ii) *spreading or blurring stages*, corresponding to the spreading of carriers (e.g. isotropic light emission and light scattering). Depending on the statistics of signals, stages can be also characterized as either *Stochastic* or *Deterministic*. Gain stages show an *average gain* (gi), and an *average gain variance* (σg_i^2). Blurring stages are characterized by a *Modulation Transfer Function-MTF* ($Ti(u,v)$), which expresses the degree of spreading and contributes to the spatial resolution in the final image [9,10]. In addition to LCS, the *Signal Detection Theory* (SDT) is also employed for imaging detector analysis. SDT assumes that *a human observer detects signals in a noisy background*.

The particular physical properties of scintillators, as related to radiation detection and light transmission, have been described by theoretical models either based on the differential diffusion equation or on Monte Carlo techniques [14–19]. In the interpretation based on the diffusion equation (Hamaker–Ludwig and Swank theories), the light transmission efficiency (in granular scintillators), i.e. g_3 and g_4 in Eq. (1), is expressed as follows:

$$g_3(E)g_4(E) = \int_0^{w_0} \overline{\phi}_X(E,w)\overline{g}_\lambda(\sigma,\tau,\rho,w)dw\,dE \quad (2)$$

w_0 is the total scintillator (in the form of 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 Δw. Here w denotes the depth of each thin layer from the screen surface. The function $\overline{\phi}_X(E,w)$ describes the relative probability of x-ray absorption at a depth w from the screen surface. The function $\overline{g}_\lambda(\sigma,\tau,\rho)$ has been defined as a solution to the photon diffusion differential equation (as interpreted by the Hamaker–Ludwig and Swank theoretical models) [15–17] describing light propagation through light scattering media

$$\overline{g}_\lambda(\sigma,\tau,\rho) = \frac{\tau\rho_1[(\sigma+\tau\rho_0)e^{\sigma w}+(\sigma-\tau\rho_0)e^{-\sigma w}]}{(\sigma+\tau\rho_0)(\sigma+\tau\rho_1)e^{\sigma w_0}-(\sigma-\tau\rho_0)(\sigma-\tau\rho_1)e^{-\sigma w_0}} \quad (3)$$

where σ and τ are reciprocal of the light photon diffusion length and the inverse relaxation length, which are functions of the optical attenuation (absorption and scattering) coefficients. In the spatial frequency (ν) domain, σ is written as $\sigma^2 = \sigma_0^2 + 4\pi\nu^2$, where σ_0 corresponds to zero-frequency. ρ_0, ρ_1 are parameters expressing the reflection of light at the front and back scintillator surfaces. According to the LCS analysis g_3 correspond to a deterministic gain stage expressing the fraction of light transmitted through the scintillator (emitted over created light), while g_4 expresses a spreading and stochastic stage of signal conversion (i.e. due to isotropic light creation and light scattering). In the space domain $\nu = 0$, $\sigma = \sigma_0$ and $g_4 = 0$.

2.2. Definitions of quality metrics

In our experimental studies various physical quantities expressing luminescence emission efficiency and image quality were evaluated under clinical conditions for various materials in the form of screens.

The emission efficiency of a scintillating screen has been expressed by the x-ray luminescence efficiency, defined as the ratio of the emitted light energy fluence over the incident x-ray energy fluence ($\eta_\Psi = \Psi_\Lambda/\Psi_0$) and corresponds to three stages $g_1(E)g_2(E)g_3(E)$ (in Eq. (1)), integrated over the x-ray spectrum and divided by Ψ_Λ. To be consistent with clinical experimental conditions the x-ray luminescence efficiency is often expressed through the absolute efficiency-AE, (η_A), defined as the ratio of emitted light energy fluence, Ψ_Λ, over the incident exposure X [15,19–25],

$$\eta_A = \Psi_\Lambda/X \quad (4)$$

AE is expressed in efficiency units (EU, 1 EU = $\mu W\ m^{-1}$/mR).

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) [2,9–11],