Modeling indirect detectors for performance optimization of a digital mammographic detector for dual energy applications

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Abstract. Dual Energy imaging is a promising method for visualizing masses and microcalcifications in digital mammography. The advent of two X-ray energies (low and high) requires a suitable detector. The scope of this work is to determine optimum detector parameters for dual energy applications. The detector was modeled through the linear cascaded (LCS) theory. It was assumed that a phosphor material was coupled to a CMOS photodetector (indirect detection). The pixel size was 22.5 μm. The phosphor thickness was allowed to vary between 20mg/cm² and 160mg/cm². The phosphor materials examined were Gd₂O₂S:Tb and Gd₂O₂S:Eu. Two Tungsten (W) anode X-ray spectra at 35 kV (filtered with 100 μm Palladium (Pd)) and 70 kV (filtered with 800 μm Ytterbium (Yb)), corresponding to low and high energy respectively, were considered to be incident on the detector. For each combination the contrast-to-noise ratio (CNR) and the detector optical gain (DOG), showing the sensitivity of the detector, were calculated. The 40 mg/cm² and 70 mg/cm² Gd₂O₂S:Tb exhibited the higher DOG values for the low and high energy correspondingly. Higher CNR between microcalcification and mammary gland exhibited the 70mg/cm² and the 100mg/cm² Gd₂O₂S:Tb for the low and the high energy correspondingly.

1. Introduction
Breast cancer, which is a common cause of death among female population, may manifest as microcalcifications. In X-ray mammography, medical diagnosis (including screening techniques) relies on the detection and visualization of such microcalcifications (μCs) and/or soft tissue masses. The early detection of breast cancer has been shown to decrease breast cancer mortality [1]. Dual-energy subtraction imaging techniques [2-8] offer an alternative approach to the detection and visualization of μCs. With this technique, high- and low-energy images are separately acquired and
“subtracted” from each other in a weighted fashion to cancel out the cluttered tissue structure so as to decrease the obscurness from overlapping tissue structures. Current Full-Field Digital Mammography (FFDM) systems, are based on either direct X-ray detectors using amorphous Selenium (a-Se) or indirect X-ray detectors using scintillators coupled to amorphous Silicon (a-Si) sensors [9,10]. Detector modelling has been carried out to determine the optimum detector design for the best detector performance in optimizing X-ray detection and signal generation. One method to determine the optimum detector parameters is linear cascaded systems theory (LCS). This theory calculates the output of a detector as a series of cascaded stages. These stages describe the statistics of signal carrier interactions and are divided into gain stages and blur stages [11-17]. In this study, the aforementioned theory was used in order to calculate the optimum phosphor thickness of an indirect detector for CNR and DOG maximization.

2. Materials and Methods

In this study the LCS theory was used. This theory calculates the output of a detector as a series of cascaded stages. Every stage has a frequency domain input $S_{in}(u)$, where $u$ is the spatial frequency, a mean input value $\bar{x}_{in}$, a frequency domain output $S_{out}(u)$ and a mean output $\bar{x}_{out}$. Every gain stage is characterized by a statistical mean value $\bar{q}$ and variance $\sigma^2_q$, while every blur stage is characterized by a Modulation Transfer Function $MTF(u)$. The blur stages are either stochastic or deterministic. The frequency dependent output and the mean output signal for each stage can be calculated from the following relationships: (i) $S_{out}(u) = \bar{q} S_{in}(u) + \bar{x}_{out} \sigma^2_q$ and $\bar{x}_{out} = \bar{x}_{in}$, for gain stages, (ii) $S_{out}(u) = (S_{in}(u) - \bar{x}_{in})MTF^2(u) + \bar{x}_{in}$ and $\bar{x}_{out} = \bar{x}_{in}$ for stochastic blur stages and (iii) $S_{out}(u) = S_{in}(u)MTF^2(u)$ and $\bar{x}_{out} = \bar{x}_{in}$ for deterministic blur stages [11-17]. In this work the following stages were considered: the X-ray absorption in the phosphor material, the optical photon production per absorbed X-ray, the optical photon escape and spread to the output, the impingement of the optical photons at the CMOS surface and the production of electrons at the CMOS output. If an X-ray fluence $\Phi_x(E)$, of energy $E$ is incident on the detector, then it is absorbed exponentially in different depths of the phosphor. The probability of X-ray interaction is assumed to follow binomial distribution with a mean probability $M_x(E,t)$ per incident X-ray photon, where $M_x(E,t) = e^{-\mu(E)\Delta t}$, with $\mu(E)$ being the total attenuation coefficient of the material. The absorbed X-ray energy is transformed into optical photons with a Poisson process with a mean value $\bar{m}_x(E)$, expressing the number of optical photons per absorbed X-ray photon. $\bar{m}_x(E)$ can be calculated as $\bar{m}_x(E) = n_C E / E_x$, where $n_C$ is the intrinsic conversion efficiency of X-ray power to optical photons power. $E_x$ is the optical photons energy. A fraction $G(t)$of these photons, escape to the output. $G(t)$ is assumed to follow binomial distribution with mean value $\overline{G(t)}$ and variance $\overline{G(t)(1-\overline{G(t)})}$. The optical photons are spread to the output. This spread is characterized by $MTF(u,t)$. By combining the above stages through LCS theory the Noise Power Spectrum ($NPS(u)$) and the mean number of optical quanta escaping the detector, $M_t$, can be calculated [13-16]. The optical photons impinge at the CMOS pixel with size $a_{pix}$. Only a fraction of these photons are detected. This is a gain stage characterized by a pixel fill factor $f_{fill}$ and an effective pixel area, $a_{pix}$ where $a_{pix} = \frac{f_{fill} a_{pix}^2}{f_{fill}}$. This process follows binomial distribution with a mean value $f_{fill}$ and a variance $f_{fill} (1-f_{fill})$. Due to spectral matching a fraction of these photons will be actually detected. This is a gain stage with a probability characterized by a mean value $\bar{a}_s$ and a variance $\bar{a}_s (1-\bar{a}_s)$. The optical photons are absorbed and electron-hole pairs are created. It was assumed that this process also follows a binomial distribution with a mean probability $\bar{Q}_p$ and a variance $\bar{Q}_p (1-\bar{Q}_p)$. A
fraction of these pairs will reach the detector output. It was considered that this process follows a Poisson distribution with a mean probability value \( \overline{Q}_e \) and a variance \( \overline{Q}_e (1 - \overline{Q}_e) \). By applying LCS theory in the above stages the total detector NPS \( (\text{NPS}_I(u)) \) is equal to:

\[
\text{NPS}_I(u) = \left( \frac{ff a_s Q_p Q_e}{\overline{Q}_e} \right)^2 \text{NPS}(u) + \overline{M} \cdot \frac{ff a_s Q_p Q_e}{\overline{Q}_e} (1 - \frac{ff a_s Q_p Q_e}{\overline{Q}_e})
\]

The total number of electrons reaching the output \( (\overline{X}_e) \) equals to \( \overline{X}_e = \overline{M} \cdot \frac{ff a_s Q_p Q_e}{\overline{Q}_e} \). The electrons are spread to the output. This is a blur stage characterized by CMOS MTF \( (MTF_{inh}) \). The final NPS equals to

\[
\text{NPS}_f(u) = \left[ \text{NPS}_I(u) - \overline{X}_e \right] MTF_{inh}^2(u) + \overline{X}_e \]  

where it was assumed that \( MTF_{inh}(u) = \sin c(\pi a_{inh} u) \).

Finally the signal is integrated due to the finite pixel size. This is a deterministic blur stage characterized by \( MTF_{pix}(u) \) where \( MTF_{pix}(u) = \sin c(\pi a_{pix} u) \) [13-16]. An additional useful parameter is the number of optical photons per incident X-ray (DOG). DOG expresses the sensitivity of the scintillator and is expressed as the total number of signal carriers produced per incident X-ray photon. With regards to the dual energy optimization, CNR was utilized as an index for image evaluation. CNR was calculated as

\[
\text{CNR} = \frac{X_{e, bg} - X_{e, micro}}{\sqrt{\sigma_{bg}^2 + \sigma_{micro}^2}}
\]

where \( X_{e, bg} \) is the signal passing through 4cm PMMA and \( X_{e, micro} \) is the signal passing through microcalcification. In addition, \( \sigma^2 \) is the corresponding signal variance calculated as

\[
\sigma^2 = \sum_{u=0}^{\infty} \text{NPS}_f(u)
\]

The model was tested with two different CMOS indirect detectors each one comprised of a RadEye CMOS photoreceptor and Gd\(_2\)O\(_3\):Tb or Gd\(_2\)O\(_3\):Eu scintillator. The latter has been evaluated as a probable candidate for imaging detectors [19-23]. Different surfaces densities of the scintillator were considered in the range between 20mg/cm\(^2\) and 160 mg/cm\(^2\). The CNR was tested for microcalcification thicknesses of 100 \( \mu m \) and 200 \( \mu m \). The low energy spectrum was obtained by filtering a standard 35kV spectrum with Pd of 100 \( \mu m \) thickness and the high energy spectrum was obtained by filtering a standard 70kV spectrum with Yb of 800 \( \mu m \) thickness. The data used for calculating the equations were obtained from literature [13-27].

**Results and Discussion**

The low and high energy spectra are presented in Figure 1. It may be observed from the calculated data that the addition of Pd and Yb filtration introduces a sharp edge at 25keV and 62keV respectively. This is due to the k-edge of these two materials at approximately 24.4keV and 61.3keV respectively. These sharp edges can minimize the effect of variance in assessing material differences and composition.

![Figure 1](image.png)

**Figure 1.** The low energy spectrum filtered with 100\( \mu m \) Pd and the high energy spectrum filtered with 800\( \mu m \) Yb.
Figure 2 present the CNR for the low and the high energy spectra for different Gd$_2$O$_2$S:Tb and Gd$_2$O$_2$S:Eu surface densities respectively. The thickness of the microcalcification was 200 $\mu$m. It can be observed from Figure 2 that Gd$_2$O$_2$S:Tb phosphor based X-ray detector exhibits better CNR than Gd$_2$O$_2$S:Eu phosphor based one. This occurs because CNR is the ratio of signal differences over the total noise variance. The signal in Gd$_2$O$_2$S:Tb detector is higher than that of Gd$_2$O$_2$S:Eu mainly due to the higher intrinsic conversion efficiency of the former which is 0.15 over 0.06. In addition the total number of electrons produced per keV absorbed (i.e. $m_e$) equals to 61 for Gd$_2$O$_2$S:Tb and 30 for Gd$_2$O$_2$S:Eu. Therefore the total optical photon output for Gd$_2$O$_2$S:Tb is higher than Gd$_2$O$_2$S:Eu. As it may be observed from Figure 2 the optimum phosphor thicknesses for Gd$_2$O$_2$S:Tb, are 70 mg/cm$^2$ and 100 mg/cm$^2$ for the low and the high X-ray energy respectively, while for Gd$_2$O$_2$S:Eu the corresponding thicknesses are 50 mg/cm$^2$ and 80 mg/cm$^2$. It is interesting to notice that the surfaces densities corresponding to higher DOG values were calculated similar to Figure 2, but approximately 30 mg/cm$^2$ lower. This occurs because the screen sensitivity with respect to thickness is affected by the X-ray absorption properties, as well as, the optical photon escape properties from the phosphor. When the thickness is increased the X-ray absorption probability also increases but the optical photon escape probability, as it is calculated by equation 3, decreases. Therefore, for each X-ray photon energy an optimum thickness can be associated. In our case, if detectability (i.e CNR) is considered, the best surface densities are between 70 mg/cm$^2$ and 100 mg/cm$^2$ for Gd$_2$O$_2$S:Tb and 50 mg/cm$^2$ and 80 mg/cm$^2$ for Gd$_2$O$_2$S:Eu.

The presented methodology was also tested for microcalcification thickness of 100 $\mu$m. The results were qualitatively the same. However the corresponding CNR values were lower due to the smaller difference of X-ray absorption probability between the100$\mu$m microcalcification and the mammary gland.

3. Conclusions
In this work, optimum indirect detector parameters, with respect to scintillation thickness for dual energy applications, were studied. The detector was modeled through the LCS theory, assuming Gd$_2$O$_2$S:Tb and Gd$_2$O$_2$S:Eu scintillators coupled to a CMOS photodetector. The 40mg/cm$^2$ and the 70mg/cm$^2$ Gd$_2$O$_2$S:Tb exhibited the higher DOG values for the low and the high energy correspondingly. Higher CNR between 200 $\mu$m microcalcification and mammary gland exhibited the 70mg/cm$^2$ and the 100mg/cm$^2$ Gd$_2$O$_2$S:Tb for the low and the high energy respectively. The results were qualitatively the same for 100 $\mu$m microcalcification but the corresponding CNR values were lower due to the lower absorption differences between the microcalcification and the mammary gland.

Figure 2. The CNR of Gd2O2S:Tb and Gd2O2S:Eu based detector for various surface densities.
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4. References

[1] Smigel K 1995 *J. Natl. Cancer Inst.* **87** 1940
[2] Johns P and Yaffe M 1985 *Med. Phys.* **12** 289
[3] Johns P, Drost D, Yaffe M and Fenster A 1985 *Med. Phys.* **12** 297
[4] Boone M, Shaber G and Tetczky M 1990 *Med. Phys.* **17** 665
[5] Boone M 1991 *Invest. Radiol.* **26** 521
[6] Bettle S and Cowen A 1994 *Phys. Med. Biol.* **39** 1989
[7] Asaga T, Masuzawa C, Yoshida A and Mattsuura H 1995 *J. Digit. Imag.* **8** 70
[8] Shaw C and Gur D 1992 *J. Digit. Imag.* **5** 262
[9] Lemacks M, Kappadath S, Shaw C, Liu X and Whitman G 2002 *Med. Phys.* **29** 1739
[10] Jong R, Yaffe M, Skarpotiakis M, Shumak R, Danjoux N, Gunsekera A and Plewes D 2003 *Rad.* **228** 842
[11] Meter R and Rabbani M 1990 *Med. Phys.* **17** 65
[12] El-Mohri Y, Antonuk L, Zhao Q, Wang Y, Hong Du Y and Sawant A 2007 *Med. Phys.* **34** 315
[13] Nishikawa R and Yaffe M 1990 *Med. Phys.* **17** 894
[14] Kim H, Jun S, Ko J, Cho G and Graeve T 2008 *IEEE Trans. Nucl. Sci.* **55** 1357
[15] Liaparinos P, Kalyvas N, Kandarakis I and Cavouras D 2013 *Nucl. Instrum. and Meth. A* **697** 87
[16] Michail C, Spyropoulou V, Fountos G, Kalyvas N, Valais I, Kandarakis I and G Panayiotakis 2011 *IEEE Trans. Nucl. Sci.* **58** 314
[17] Lin C, Mathur B and Chang M 2002 *IEEE Trans. Nucl. Sci.* **49** 754
[18] Ji W and Rowlands J 1990 *Med. Phys.* **25** 2148
[19] Michail C, Fountos G, Liaparinos P, Kalyvas N, Valais I, Kandarakis I and Panayiotakis G 2010 *Med. Phys.* **37** 3694
[20] Michail C, Fountos G, Valais I, Kalyvas N, Liaparinos P, Kandarakis I and Panayiotakis G 2011 *IEEE Trans. Nucl. Sci.* **58** 2503
[21] Michail C, Valais I, Toutountzias A, Kalyvas N, Fountos G, David S, Kandarakis I and Panayiotakis G 2008 *IEEE Trans. Nucl. Sci.* **55** 3703
[22] Seferis I, Michail C, Valais I, Fountos G, Kalyvas N, Stromatia F, Oikonomou G, Kandarakis I and Panayiotakis G 2013 *Nucl. Instr. and Meth. Phys. Res.* **729** 307
[23] Seferis I, Michail C, Valais I, Zeler J, Liaparinos P, Fountos G, Kalyvas N, David S, Stromatia F, Zych E, Kandarakis I and Panayiotakis G 2014 *J. of Lum.* **151** 229
[24] https://w9.siemens.com/cms/oemproducts/Home/X-rayToolbox/spektrum/Pages/Default.aspx
[25] Kalyvas N, Liaparinos P, Michail C, David S, Fountos G, Wójtowich M, Zych E and Kandarakis I 2012 *Appl. Phys.* **106** 131
[26] http://www.rad-icon.com/products-software.php
[27] Rad-icon Imaging Corp. Remote Radeye data sheet