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A theoretical investigation of spectra utilization for a CMOS based indirect 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. Currently commercially available detectors may be suitable for dual energy mammographic applications. The scope of this work was to theoretically examine the performance of the Radeye CMOS digital indirect detector under three low- and high-energy spectral pairs. The detector was modeled through the linear system theory. The pixel size was equal to 22.5μm and the phosphor material of the detector was a 33.9 mg/cm^2 Gd_2O_2S:Tb phosphor screen. The examined spectral pairs were (i) a 40kV W/Ag (0.01cm) and a 70kV W/Cu (0.1cm) target/filter combinations, (ii) a 40kV W/Cd (0.013cm) and a 70kV W/Cu (0.1cm) target/filter combinations and (iii) a 40kV W/Pd (0.008cm) and a 70kV W/Cu (0.1cm) target/filter combinations. For each combination the Detective Quantum Efficiency (DQE), showing the signal to noise ratio transfer, the detector optical gain (DOG), showing the sensitivity of the detector and the coefficient of variation (CV) of the detector output signal were calculated. The second combination exhibited slightly higher DOG (326 photons per X-ray) and lower CV (0.755%) values. In terms of electron output from the RadEye CMOS, the first two combinations demonstrated comparable DQE values; however the second combination provided an increase of 6.5% in the electron output.

1. Introduction

Breast cancer, which is a common cause of death among female population, may manifest as microcalcifications. Modern breast examination techniques, includes irradiation with dual energy spectra [1-5]. Dual-energy subtraction imaging techniques offer an alternative approach to the detection and visualization of microcalcifications. 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 obscurity from overlapping tissue structures [6]. Although this technique reduces the contrast to noise ratio of the final image, it makes microcalcifications better visualized [1-5]. Digital mammography utilizes direct or indirect detection methods. The latter uses scintillators coupled to amorphous Silicon (a-Si) sensors [7-8]. Detector modelling has been carried out to determine the detector design and the incident X-ray spectra for an optimum detector performance. 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 [7-10]. These stages describe the statistics of signal carrier interactions and are divided into gain stages and blur stages. In this study, the aforementioned theory was used in order to investigate the performance of three low energy (LE) and high energy (HE) X-ray spectra combinations incident at a commercially available CMOS detector. The performance was evaluated through image quality metrics like the detective quantum efficiency (DQE), showing the signal-to-noise transfer, the detector optical gain (DOG) showing the detector output per incident X-ray and the coefficient of variation (CV) of the output signal [7-10].

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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 σ_q^2, while every blur stage is characterized by a Modulation Transfer Function $MTF(u)$. The blur stages are either stochastic or deterministic. The frequency depended output and the mean output signal for each stage can be calculated from the following relationships: (i) $S_{out}(u) = \bar{q}^2 S_{in}(u) + \bar{x}_{in}\sigma_q^2$ and $\bar{x}_{out} = \bar{q}\bar{x}_{in}$, 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 [7-10]. 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. A more extensive analysis of the above stages can be found in current literature. Through these stages the total Noise Power Spectrum (NPS(u)) was calculated, were u is the spatial frequency. In addition the total signal output in electrons M was determined. DOG was calculated as [10]:

$$DOG = \frac{M}{\sum_E f(E)} \tag{1}$$

where *f(E)* is the incident X-ray photons of energy E,

$$CV = \frac{\sqrt{\sum_u NPS(u)}}{M} \tag{2}$$

and [11]:

$$DQE(u) = \frac{[M \cdot MTF(u)]^2}{\sum_E f(E) \cdot NPS(u)} \tag{3}$$

where *MTF(u)* is the modulation transfer function.

The X-ray spectra combinations tested were obtained by considering polyenergetic X-rays filtered with various filter materials and thicknesses [12]. An analytical model was developed for the calculation of the calcification signal to noise ratio (SNR_{tc}) and the mean glandular dose (MGD) for various LE and HE filter combinations. The filters selection was based on the maximization of the SNR_{tc}/MGD ratio. This work is presented in an accompanied paper [13].
The data used for calculating the equations were obtained from literature [7,8,10,11]. The model was applied to a commercially available indirect CMOS detector (RadEye CMOS), incorporating a 34mg/cm^2 Gd_2O_2S:Tb phosphor screen in close contact with a 22.5μm pixel size photodetector array [14-15].

3. Results and Discussion

The low and high energy spectra combinations are presented in Figure 1. It may be observed from the spectra that there is a small spectral overlap in the range between 35 keV and 40 keV.
In table 1 the DOG and CV values for the presented spectra is shown.

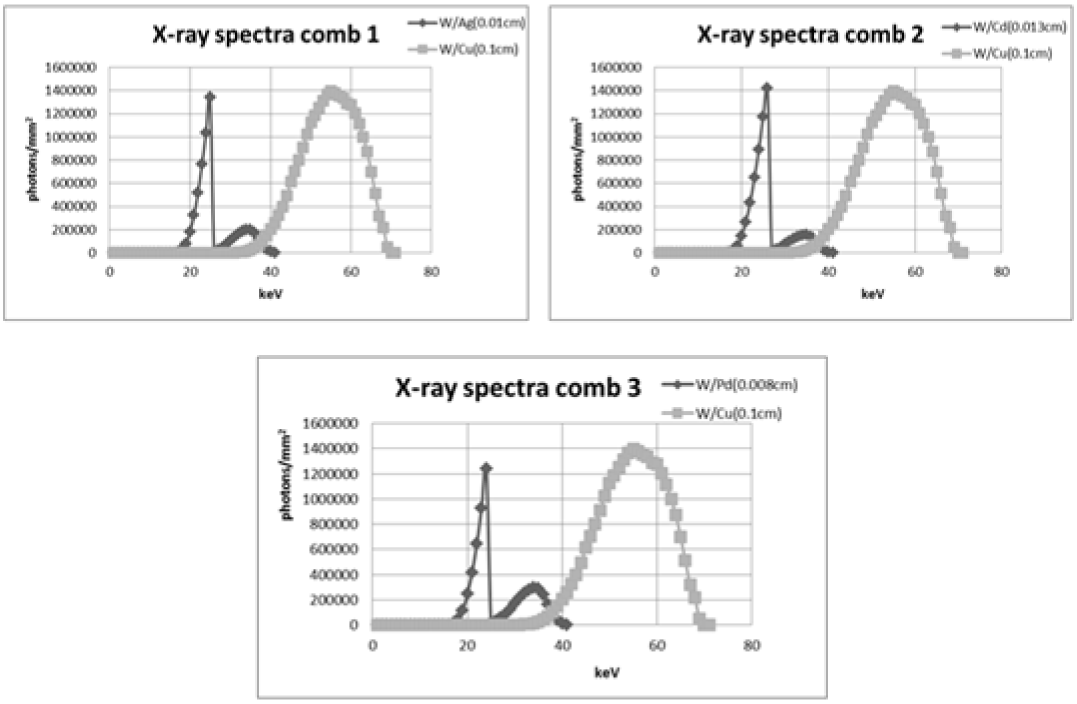

Figure 1. The used low and high energy spectra combinations.

Table 1. The DOG and CV values for the X-ray spectra under consideration.

X-ray spectrum (target/filter)	DOG (electrons/X-ray photon)	CV (%)
70 kVp W / Cu (0.1cm)	0.080	0.50
40 kVp W/ Ag (0.01cm)	0.060	0.78
40 kVp W/ Cd (0.013 cm)	0.060	0.76
40 kVp W/Pd (0.008cm)	0.058	0.79

It can be observed from table 1 that for the low energy spectra the larger CV and the lowest DOG values have been calculated for the W/Pd (0.008 cm) target/filter combination. From the other two LE combinations the 40 kVp W/Cd (0.013 cm) is slightly better than the 40 kVp W/Ag (0.01cm), due to its lower calculated CV value. In contrast the DOG value is higher for the HE spectrum. Although the thin Gd_2O_2S:Tb phosphor screen (34 mg/cm^2) of the RadEye CMOS sensor provides better absorption characteristics the phosphor intrinsic gain (optical photons produced per absorbed X-ray) is higher for higher X-ray energies deposited.

Figure 2 present the DQE for the low and the high energy spectra for the detector under consideration. It can be observed from Figure 2 that the W/Cd (0.013 cm) LE target/filter exhibit slightly better DQE values per spatial frequency than the other LE combinations. In addition the lowest DQE values are that of the HE combination due to the reduced X-ray absorption of the 34mg.cm^2 Gd_2O_2S:Tb phosphor screen.

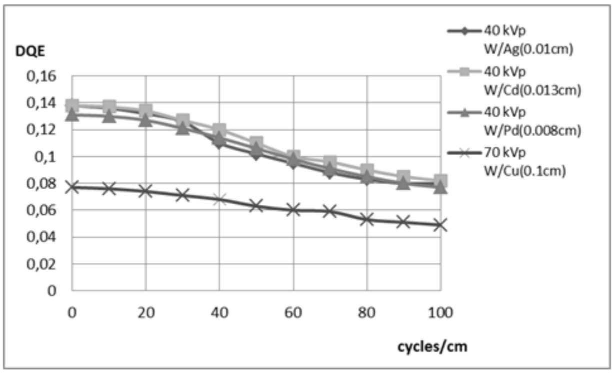

Figure 2. The DQE of the spectral components used for the dual energy spectra combinations.

4. Conclusions

In this work, the applicability of three target/filter combinations impinging on a commercially available RadEye CMOS detector was evaluated in terms of DOG, CV and DQE. It was found that the 40kVp W/ Cd (0.013 cm) provided the best low energy component. The low DQE and DOG values are mainly attributed (i) to the thin mammographic screen, which is designed for smaller X-ray energies.

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