Published in final edited form as:

Proc SPIE Int Soc Opt Eng. 2016 ; 9783: . doi:10.1117/12.2217085.

Theoretical and Monte Carlo optimization of a stacked three-layer flat-panel x-ray imager for applications in multi-spectral medical imaging

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Abstract

We propose a new design of a stacked three-layer flat-panel x-ray detector for dual-energy (DE) imaging. Each layer consists of its own scintillator of individual thickness and an underlying thin-film-transistor-based flat-panel. Three images are obtained simultaneously in the detector during the same x-ray exposure, thereby eliminating any motion artifacts. The detector operation is two-fold: a conventional radiography image can be obtained by combining all three layers' images, while a DE subtraction image can be obtained from the front and back layers' images, where the middle layer acts as a mid-filter that helps achieve spectral separation. We proceed to optimize the detector parameters for two sample imaging tasks that could particularly benefit from this new detector by obtaining the best possible signal to noise ratio per root entrance exposure using well-established theoretical models adapted to fit our new design. These results are compared to a conventional DE temporal subtraction detector and a single-shot DE subtraction detector with a copper mid-filter, both of which underwent the same theoretical optimization. The findings are then validated using advanced Monte Carlo simulations for all optimized detector setups. Given the performance expected from initial results and the recent decrease in price for digital x-ray detectors, the simplicity of the three-layer stacked imager approach appears promising to usher in a new generation of multi-spectral digital x-ray diagnostics.

Keywords

Dual Energy; X-ray; Diagnosis; Multilayer; Flat-Panel Detector

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1 Introduction

Several radiography diagnostic and screening techniques depend on the visualization of small objects with high attenuation coefficient embedded in an inhomogeneous soft-tissue environment. These include coronary angiography¹—where a contrast agent is added to blood vessels in the heart to assess cardiovascular diseases—and calcification detection in chest radiography^{2,3}—where the benignity of pulmonary nodules is characterized by its calcium content. Enhancing the detectability of the objects of interest of these techniques is, therefore, very desirable.

A technique that has long been proposed to achieve this is that of dual-energy (DE) subtraction imaging,^{4,5} which exploits the difference in the energy-dependence of the x-ray attenuation coefficient of different tissue types to remove the soft-tissue component from a radiographic image and hence enhance visualization of the objects of interest. DE imaging has been successfully applied to both coronary angiography^{6,7} and calcified nodule detection in chest radiography^{8–11} to improve detection.

DE systems work by obtaining one low-energy and one high-energy image, and performing a weighted subtraction to combine them into a third image in which the inhomogeneity of the soft-tissue structure has been removed.^{12,13} This spectral separation is typically achieved by kVp switching,¹⁴ where a first image is taken using a low x-ray tube voltage (kVp) and, immediately after, a second image is obtained with a high kVp. While this minimizes the overlap in the images' spectra, the temporal separation between the two images makes this technique very sensitive to motion artifacts, an issue of particular importance when dealing with both cardiac¹⁵ and pulmonary imaging.¹⁰ kVp switching also requires an x-ray source capable of producing two very different spectra in a time as short as 150 ms to 200 ms,¹⁴ and thus is not compatible with typical chest radiography setups.

An alternative method of obtaining spectral separation in DE imaging is the single-shot approach, in which both images are obtained simultaneously. This is accomplished by stacking two sensor layers vertically to form a double-layer detector in what is known as a sandwich configuration.^{15–18} The spectral separation results from a combination of the intrinsic higher probability of high-energy photons to go through the top layer unabsorbed, and the presence of a metal beam-hardening mid-filter in between the two layers. Single-shot techniques are thus much more tolerant to both patient and cardiac motion than kVp switching, and are compatible with most current x-ray sources. However, the presence of the hardening mid-filter means that a portion of the x-rays are wasted in it, resulting in lower patient dose efficiency.

To address this issue, we propose a three-layer detector to be used for single-shot DE imaging. Each layer of the detector consists of an amorphous silicon (a-Si) flat-panel sensor coupled to a scintillator layer of specific thickness (fig. 1). Much like in the standard sandwich configuration, the top and bottom layers will form the low and high energy images respectively, but unlike said configuration, the new middle layer will not only act as a filter and harden the beam before it reaches the bottom layer but also act as a third sensitive layer.

Our proposed detector can then operate in two distinct modes. The first is as a regular radiographic detector, in which the signals from all its sensitive layers are simply added to form a single image. In this mode, no dose delivered to the patient is wasted in a mid-filter and the image obtained will display information from the entire x-ray spectrum used. The second mode is that as a DE imager, in which only the signals from the top and bottom layers are used to acquire an image without the inhomogeneity of the soft-tissue. These two modes are not mutually exclusive, and both image types could be obtained simultaneously with only one exposure. The purpose of this report is to evaluate the theoretical performance of this three-layer detector when in its DE imager mode. An analytical model is developed to calculate the performance of such a detector, with which its optimal parameters are found. The results are later validated using photon transfer Monte Carlo simulations.

2 Theory

2.1 Signal and Noise for Single-Layer Imagers

Each one of the layers in our proposed three-layer imager can be considered as a single x-ray flat-panel detector (FPD). Therefore, it is necessary to analyze the SNR of a single detector to understand how to calculate it for the three-layered imager, and thus evaluate its performance.

We consider a single exposure of an x-ray spectrum $\Phi(E)$ (photon=(cm² keV)) through an object and to a scintillator-based FPD (fig. 2), where two x-ray paths stand out: one that will only go through soft tissue of thickness t_s (cm) before reaching the detector, and one that will go through an iodinated vessel or calcification (depending on the application) of thickness t_h (cm) as well as its surrounding soft tissue before reaching the detector. The former will result in the detector signal S_s while the latter will result in S_h .

We can calculate the expected mean signals $\overline{S}_s$ and $\overline{S}_h$ for a pixel of size A (cm²) located entirely in one of the main paths using¹⁹

$$\overline{S}_s = A \int_0^\infty \Phi(E) e^{-\mu_s(E)t_s} \eta(E) \overline{Q}_{CsI}(E) dE \quad (1)$$

$$\overline{S}_h = A \int_0^\infty \Phi(E) e^{-\mu_s(E)(t_s-t_h)-\mu_h(E)t_h} \eta(E) \overline{Q}_{CsI}(E) dE \quad (2)$$

where $\mu_s(E)$ and $\mu_h(E)$ are the x-ray attenuation coefficients (cm⁻¹) of soft tissue and the iodinated vessel or calcification respectively at an x-ray photon energy E (keV), $\eta(E)$ is the x-ray absorption ratio of the scintillator (i.e. its quantum efficiency), and $\overline{Q}_{CsI}(E)$ is the mean of the scintillator gain function. Note that we assume, for simplicity, that the attenuation coefficients are constant throughout each of the tissue types.

The quantum efficiency of the scintillator is given by

$$\eta(E) = 1 - e^{-\mu_{CsI}(E)t_{CsI}} \quad (3)$$

where $\mu_{CsI}(E)$ is the attenuation coefficient of the scintillator (also considered constant) and t_{CsI} is the thickness of the scintillator. It is, essentially, the portion of incoming x-ray photons that will be absorbed by the scintillator.

The scintillator gain function, $\tilde{Q}_{CsI}(E)$, is a random variable that corresponds to the number of optical photons generated in the scintillator due to an incoming x-ray of energy E . This means that the generated signals $\tilde{S}_s$ and $\tilde{S}_h$ are compound random variables,^{4,19} since the number of photons that will reach the pixel, N , will follow a Poisson distribution. However, since the number of scintillator photons generated for a typical x-ray is small (e.g. a cesium iodide scintillator can generate ~1500 optical photons for a 25 keV photon^{20–22}) when compared to N , its contribution to the signal variance can be neglected and only its mean value needs to be considered.²³ This value, in turn, can be simply expressed as

$$\overline{Q}_{CsI}(E) = \gamma(E) E_{abs}(E) \quad (4)$$

where $\gamma(E)$ (photons=keV) is the mean number of photons generated and collected in the scintillator due to an absorbed photon of energy E , and $E_{abs}(E)$ is the energy absorbed in the scintillator due to an x-ray photon of energy E interacting in it. For common inorganic scintillators, this gain is nearly proportional to the deposited radiation,²⁴ which removes the energy dependence in the number of photons generated ($\gamma(E) = \gamma$) and reduces eq. (4) to $\overline{Q}_{CsI}(E) = \gamma E_{abs}(E)$.

With this in mind, it is now possible to write an expression for the expected variance in each one of these signals that result from the contributions of these two random variables. This can be achieved by summing in quadrature the individual variances from all the components of the energy spectrum, since these can be considered independent from one another.^{23,25,26}

$$\sigma_s'^2 = A \int_0^\infty \Phi(E) e^{-\mu_s(E)t_s} \eta(E) \overline{Q}_{CsI}^2(E) dE \quad (5)$$

$$\sigma_h'^2 = A \int_0^\infty \Phi(E) e^{-\mu_s(E)(t_s-t_h) - \mu_h(E)t_h} \eta(E) \overline{Q}_{CsI}^2(E) dE \quad (6)$$

2.1.1 K-Edge Considerations—When the energy of the x-ray photons incident on the scintillator exceeds the K-edge energy of the absorber, some of their energy will be lost in escaping characteristic x-ray emission.²³ This is accounted for in $E_{abs}(E)$ as a part of the scintillator gain $Q_{CsI}(E)$. E_{abs} has previously been calculated²⁵ to be linearly dependent on E both below and above the K-edge energies, while showing considerable dips at those energies.

2.1.2 Scintillator Blur—The needle-like structure of CsI:Tl scintillators has been shown to guide light towards the top of and bottom surfaces of the phosphor screen and reduce lateral spread, thereby increasing spatial resolution.²⁷ However, theoretical and simulation analyses of these structured scintillators²⁸ have found that about 17 % of the light quanta generated in the scintillator still travels laterally and contributes to optical blur. This spatial spreading of the signal will affect the resolution of the imager and reduce the spatial noise variance.

We use an analytical model to approximate this noise reduction that begins by considering how the signal of an infinitely thin monoenergetic x-ray pencil beam of energy E will spread in a CsI:Tl structured scintillator of thickness t_{CsI} before reaching the detector plane. This is known as the Point Response Function (PRF) of the system and represents its deterministic scintillator blur.²⁹ Its value at any point on the detector (x, y) will be a function of these parameters, and so $PRF = PRF(x, y, E, t_{CsI})$.

Next, we calculate how this scintillator blur will affect the detector's signal variance. Obviously, this effect will be very dependent on the pixel pitch, p , of the detector. If we consider the signal caused by an x-ray beam incident over an entire square pixel centred at $x = y = 0$, its spread to an arbitrary position in the scintillator (x, y) can be calculated by integrating PRF contributions from each area element in the pixel and will be expressed by the Interpixel Spread Function, or

$$ISF(x, y, E, t_{CsI}) = \int_{-\frac{p}{2}}^{\frac{p}{2}} \int_{-\frac{p}{2}}^{\frac{p}{2}} PRF(x - x', y - y', E, t_{CsI}) dx' dy' \quad (7)$$

The ISF can be used to calculate how the signal incident over a pixel located at $i = j = 0$ will spread to its neighbouring pixels (see fig. 3). The spread will induce an increase signal of $S_{i,j}$ for the i^{th}, j^{th} pixel of the detector array, where this can be calculated using

$$\Delta S_{i,j}(E, t_{CsI}) = \int_{p(j-\frac{1}{2})}^{p(j+\frac{1}{2})} \int_{p(i-\frac{1}{2})}^{p(i+\frac{1}{2})} ISF(x, y, E, t_{CsI}) dx dy \quad (8)$$

Lastly, a normalized version of $S(S')$ can be thought of as the smoothing kernel that is applied to an unblurred image as a result of the signal spread, assuming that we are operating in a large-pixel regime where it will be fairly uniform. We fit S' to a two-dimensional discrete Gaussian kernel of standard deviation σ_G since we know that its convolution with a zero-frequency image with normal noise will reduce the image's standard deviation by a factor of $\sim (2\sigma_G \sqrt{\pi})$.³⁰ The standard deviation of signals generated by a monoenergetic x-ray source can now be updated to include scintillator blur as

$$\sigma_s \approx \frac{\sigma'_s}{2\sigma_G^s \sqrt{\pi}} \quad \sigma_h \approx \frac{\sigma'_h}{2\sigma_G^h \sqrt{\pi}} \quad (9)$$

where σ_G^s and σ_G^h are the standard deviations for the fitted Gaussian kernels of S for the expected soft tissue and iodinated vessel or calcification signals, respectively.

As previously mentioned, this treatment is only applicable to the case of a monoenergetic source. However, it can be easily expanded to any source spectrum by first defining an energy-dependent $\sigma_G(E)$, which, at every energy E , would represent the σ_G obtained through this process using a pencil beam of that energy. Full polyenergetic signal variances for $\tilde{S}_s$ and $\tilde{S}_h$ can be obtained by including the continuous $\sigma_G^2(E) / 4\pi$ as part of the integrand in eqs. (5) and (6) (see eqs. (14) to (17) for further details).

2.2 Energy-Subtraction Imaging

One of the operational modes of the proposed detector is as a dual-energy (DE) imager. DE imaging combines two images: one obtained with low-energy (LE) and one with high-energy (HE) photons.⁴ A third, enhanced image is thereby obtained that aims to isolate the object's high-attenuation-coefficient tissue component (an iodinated vessel or a calcification in our implementation) from its soft-tissue background component. Thus, the contrast within the latter is removed and the detectability of the former can be greatly increased.

This background suppression is attained by exploiting the different rates of change of the tissue's mass attenuation coefficients (μ/ρ) across the diagnostic x-ray energy range.⁴ Said phenomenon can be seen in fig. 4b, where the difference in μ/ρ between iodine and soft tissue or bone and soft tissue is clearly higher in the low end of the spectrum (~ 40 keV) than in the high end (~ 100 keV). By intelligently combining the LE and HE image through logarithmic subtraction (see section 2.2.1), the enhanced image with a suppressed background is obtained.

In its DE mode, the proposed detector functions as a one-shot sandwich detector. The top layer will have a thinner scintillator and will thus probabilistically absorb more of the lower end of the incident x-ray spectrum. The thicker scintillator of the middle layer will assume the role typically taken by a metal mid-filter (as shown in fig. 1) and will harden the remainder of the beam before it reaches the bottom layer. fig. 4a illustrates this effect, where more lower energy photons will be incident on the top layer, while the bottom layer will see more of those of higher energy. This achieves the energy separation needed for obtaining the LE and HE images without the need to waste photons in a metal mid-filter.

The mean value for those pixels at the top layer (S_t) for each of the paths of interest (s and h) will follow eqs. (1) and (2), as this layer is equivalent to a single-layered detector. The signals for the bottom (S_b) layer, however, need to expand on these equations to include the filtration due to the top and middle layers. These will therefore become

$$\bar{S}_{t,s} = A \int_0^\infty \Phi(E) e^{-\mu_s(E)t_s} \eta_t(E) \bar{Q}_{Csl}(E) dE \quad (10)$$

$$\bar{S}_{t,h} = A \int_0^\infty \Phi(E) e^{-\mu_s(E)(t_s-t_h) - \mu_h(E)t_h} \eta_t(E) \bar{Q}_{CsI}(E) dE \quad (11)$$

$$\bar{S}_{b,s} = A \int_0^\infty \Phi(E) e^{-\mu_s(E)t_s} (1 - \eta_t(E))(1 - \eta_m(E)) \eta_b(E) \bar{Q}_{CsI}(E) dE \quad (12)$$

$$\bar{S}_{b,h} = A \int_0^\infty \Phi(E) e^{-\mu_s(E)(t_s-t_h) - \mu_h(E)t_h} (1 - \eta_t(E))(1 - \eta_m(E)) \eta_b(E) \bar{Q}_{CsI}(E) dE \quad (13)$$

where $\eta_i = 1 - e^{-\mu_{CsI}(E)t_i}$ ($i=t, m, b$) are the top, middle and bottom layers' quantum efficiencies. The respective variances of these signals will also need to account for the detector design, and can be calculated by expanding eqs. (5), (6) and (9) into

$$\sigma_{t,s}^2 \approx \frac{A}{4\pi} \int_0^\infty \Phi(E) e^{-\mu_s(E)t_s} \eta_t(E) \bar{Q}_{CsI}^2(E) \sigma_G^{-2}(E, t_t, p) dE \quad (14)$$

$$\sigma_{t,h}^2 \approx \frac{A}{4\pi} \int_0^\infty \Phi(E) e^{-\mu_s(E)(t_s-t_h) - \mu_h(E)t_h} \eta_t(E) \bar{Q}_{CsI}^2(E) \sigma_G^{-2}(E, t_t, p) dE \quad (15)$$

$$\sigma_{b,s}^2 \approx \frac{A}{4\pi} \int_0^\infty \Phi(E) e^{-\mu_s(E)t_s} (1 - \eta_t(E))(1 - \eta_m(E)) \eta_b(E) \bar{Q}_{CsI}^2(E) \sigma_G^{-2}(E, t_b, p) dE \quad (16)$$

$$\sigma_{b,h}^2 \approx \frac{A}{4\pi} \int_0^\infty \Phi(E) e^{-\mu_s(E)(t_s-t_h) - \mu_h(E)t_h} (1 - \eta_t(E))(1 - \eta_m(E)) \eta_b(E) \bar{Q}_{CsI}^2(E) \sigma_G^{-2}(E, t_b, p) dE \quad (17)$$

where $\sigma_G(E, t_{CsI}, p)$ represents the standard deviations for a Gaussian kernels fitted to a S' of a monoenergetic exposure of energy E , a detector of thickness t_{CsI} and a pixel pitch p . The addition of this term and the $1/4\pi$ factor effectively accounts for scintillator blurring.

2.2.1 Logarithmic Subtraction—The combining of these signals into the enhanced DE image was done by using logarithmic subtraction.¹² This method, while fairly simple, has proven to be quite effective at obtaining the desired result. In it, a weighting factor w is defined such that soft-tissue contrast is canceled in the enhanced image and is ideally given by the ratio of its attenuation coefficients at high and low energy,

$$w = \frac{\mu_s(E_H)}{\mu_s(E_L)} \quad (18)$$

where E_H and E_L were chosen as the average energies reaching the bottom and top detector layers when no object is being imaged. These can be calculated using

$$E_L = \frac{\int_0^\infty \Phi(E) E dE}{\int_0^\infty \Phi(E) dE} \quad E_H = \frac{\int_0^\infty \Phi(E) (1 - \eta_t(E)) (1 - \eta_m(E)) E dE}{\int_0^\infty \Phi(E) (1 - \eta_t(E)) (1 - \eta_m(E)) dE} \quad (19)$$

Note that due to the broad spectrum of the x-ray source and to the absorption in the object, in practice, the value of w should be slightly tweaked to obtain the best possible cancellation, as the ideal 100 % cancellation cannot be achieved.

Using w , the DE enhanced image (S_{DE}) obtained through logarithmic subtraction can be easily calculated using³²

$$S_{DE}(x, y) = \ln(S_b(x, y)) - w \ln(S_t(x, y)) \quad (20)$$

2.2.2 Signal Difference and Noise—The merit of an enhanced image will depend on the contrast between the two tissue types once the inhomogeneity within soft-tissue has been canceled. Said contrast is defined as the Signal Difference (SD) between two pixels located fully in either one of the paths of interest described in section 2.1, and can be calculated with

$$SD = S_{DE,h} - S_{DE,s} \quad (21)$$

where $S_{DE,h}$ and $S_{DE,s}$ are the values in the enhanced image for these pixels. Using eq. (20), SD can be expanded into

$$SD = \ln(S_{b,h}) - w \ln(S_{t,h}) - [\ln(S_{b,s}) - w \ln(S_{t,s})] = \ln\left(\frac{S_{b,h}}{S_{b,s}}\right) - w \ln\left(\frac{S_{t,h}}{S_{t,s}}\right) \quad (22)$$

The mean value of SD ($\overline{SD}$) can be easily calculated by combining eqs. (10) to (13) and (22). In turn, we can now calculate its variance, σ_{SD}^2 , as a measure of its noise. Assuming no cross-talk between the detector layers, the noises in the top and bottom layers are uncorrelated,³³ and thus σ_{SD}^2 can be calculated using^{7,32}

$$\sigma_{SD}^2 = \left(\frac{1}{\overline{S}_{b,h}^2} \sigma_{b,h}^2 + \frac{1}{\overline{S}_{b,s}^2} \sigma_{b,s}^2 \right) + w^2 \left(\frac{1}{\overline{S}_{t,h}^2} \sigma_{t,h}^2 + \frac{1}{\overline{S}_{t,s}^2} \sigma_{t,s}^2 \right) \quad (23)$$

It is now possible to define a parameter that can evaluate the detectability of the object of interest in its noisy background. Said metric is the Signal Difference to Noise Ratio (SDNR) of the logarithmic subtraction image, and it can be computed as

$$\text{SDNR} = \frac{\overline{\text{SD}}}{\sqrt{\sigma_{SD}^2}} \quad (24)$$

2.2.3 Figure of Merit—While SDNR is a good single metric to evaluate the performance of a particular set of detector parameters, it is easy to see its one limitation. Given that SD will increase as the total exposure of the image increases, solely following SDNR will lead to favouring certain configurations that result in higher patient dosages—particularly those with higher peak x-ray tube voltage (kVp). To aid in finding a better performance metric, we begin by calculating the exposure X (R) as the total energy deposited in air during the imaging process^{34,35}

$$X = \frac{C_R}{\overline{W}_{air}} \int_0^\infty \Phi(E) E \left(\frac{\mu_{en}(E)}{\rho} \right)_{air} dE \quad (25)$$

where $C_R \cong 6.201 \times 10^{-10}$ R g J/(keV C) is a unit conversion factor, $\overline{W}_{air} \cong 33.97$ J/C is the energy required to be deposited in air for the production of one ion pair averaged across the diagnostic energy range, and $(\mu_{en}(E)/\rho)_{air}$ is the mass energy-absorption coefficient of air at energy E .

We can finally define a single-valued figure of merit (FoM) that will measure the performance of a detector as a DE imager. Given that SDNR will increase with the square root of exposure, our exposure-independent FoM becomes

$$\text{FoM} = \frac{\text{SDNR}}{\sqrt{X}} \quad (26)$$

3 Optimization Using Analytical Model

To find the ideal configuration of detector and x-ray tube parameters, the FoM was computed as described for the reasonable parameter ranges shown below in table 1. These ranges were selected by considering current FPD, scintillator and x-ray tube technologies. Two applications were independently investigated, and the optimal values for the parameters were found for each case. The first was that of DE angiography and cardiac imaging, where iodinated vessels of increasing iodine mass loading (I_m) located in 20 cm of soft tissue were considered. The second was calcification detection in chest radiography, where small calcium nodules of varying thickness (t_c) in the same soft tissue were evaluated.

In these calculations, the x-ray spectrum of a tungsten anode tube was calculated for different kVp values using Spektr,³⁶ a Matlab™ implementation of the TASMIP³⁷ (tungsten

anode spectral model using interpolating polynomials) empirical model, since this model has proven to be one of the most accurate at predicting x-ray tube spectra.³⁸ The source filtering with different materials and thicknesses t_f was computed using the built-in filtering capabilities of Spektr. All mass attenuation and mass energy-absorption coefficients, as well as all material and tissue densities, were obtained from the National Institute of Standards and Technology (NIST) measurement compilation.³¹

We chose Cesium Iodide as our scintillator material due to its maturity, availability and ability to be grown structurally. This material is also attractive due to its common and established integration with FPD technology. The packing efficiency of the CsI:Tl columnar scintillator was taken to be 75 %, as reported in previous works.²²

To be able to quickly calculate the effects of scintillator blur for specific parameters, the following computational model was used. First, Monte Carlo simulations were performed using hybridMANTIS (see section 4) to obtain $\text{PRF}(x, y, E)$ values for several model screens of thicknesses in the range of 50 μm to 800 μm at 10 μm increments. These simulations used a monoenergetic pencil beam source ranging in energy from 10 keV to 150 keV at 5 keV increments, with very small detector pixels to approximate a continuous function as best as possible.

Next, following the process described in section 2.1.2, values for $\sigma_G(E, t_{\text{CsI}}, p)$ were calculated and fitted for the energy and scintillator thicknesses mentioned, and for a pixel pitch range of 50 μm to 150 μm at 10 μm increments. These were then interpolated using natural neighbour interpolation to obtain a continuous function for $\sigma_G(E, t_{\text{CsI}}, p)$. fig. 5 shows a subset of these results to demonstrate that σ_G is a fairly well-behaved and thus these interpolations were sensible.

Certain constraints were imposed on the detector design when finding the optimal set of parameters. First, the pixel pitch p was set to 100 μm , as it is a reasonable state-of-the-art pixel size for an a-Si FPD. Second, the bottom detector layer is meant to absorb as much as possible of the high-energy x-rays that reach it without introducing too much signal spread, and thus its scintillator thickness t_b was not varied but rather set to a reasonable high-end limit of 500 μm imposed by fabrication and blurring constraints. With that, the parameters that resulted in the maximum possible FoM for both applications were found, and are presented in section 5.

4 Monte Carlo Validations

The final results from the theoretical method developed above were validated using x-ray photon transfer Monte Carlo simulations. These were carried out in hybridMANTIS,³⁹ a software package for modeling indirect x-ray detectors with structural scintillators. hybridMANTIS uses the PENELOPE⁴⁰ package for its geometry, x-ray and electron transport, and source modeling, but adds optical photon transfer in the scintillator to better model a columnar structure. Using this package, we were able to model all layers of our proposed detector with optimized parameters and verify the FoM analytical calculations for each case.

For the iodinated vessel application, six blood vessels of increasing I_{ml} were modeled as cuboids with width and depth of 5 mm. Their materials were set to a mixture of ICRP-defined blood⁴¹ and increasing concentrations of an iodine contrast medium molecule. In the case of the calcification application, five nodules were simulated as cubes of size t_c at 1 mm increments. For both cases, these objects were placed in the centre of a 20 cm-thick slab of ICRP-defined soft-tissue. Each layer of the detector was modeled as a 0.5 mm glass substrate, a 1 μm aluminum layer to simulate a thin-film transistor layer, and a structured CsI scintillator layer immediately above. fig. 6 shows an example of the images obtained in these simulations and how they are combined to form a DE image.

5 Results and Discussion

5.1 Source and Filtering

The first variable that was optimized pertaining to the x-ray source is the peak tube voltage. Increasing the kVp resulted in a steady increase of our FoM, as shown for both application cases in fig. 7. Hence, the source should be operated at the highest kVp possible. As a reasonable upper-limit for typical tungsten-anode sources, this value was set to 120 kV for all further simulations.

The second set of parameters related to the x-ray source that were optimized are the filtration materials and their thicknesses. The two elements out of the six modeled that were found to independently increase the FoM the most were aluminum (Al) and copper (Cu). fig. 8a shows an example of the different positive effect for both filter elements. Adding Al filtering will result in a continuous FoM increase but with diminishing returns, while the Cu effect will peak at ~ 0.2 mm and decrease with higher filter thickness t_f . It is important to note that these two filtration benefits cannot be combined, but rather the optimization of one will remove almost all positive effects of the other. Thus, sole Al filtration becomes the obvious practical solution, since there is no need to precisely optimize t_f to achieve maximum benefit. Similarly to choosing kVp, a sensible value of $t_f = 3$ mm of Al was chosen for all remaining calculations.

The remaining filter elements studied were rhodium (Rh), molybdenum (Mo), silver (Ag) and tin (Sn). As can be seen in the example in fig. 8b, most of these presented a similar trend to Cu, but with less of a positive effect and a narrower peak with a quicker decline. As such, these elements do not present a good alternative to Al filtering. Several combinations of different thicknesses of all filters were also studied, but only minute FoM improvement were found in the best of cases. The practicality of combining and optimizing several filters is certainly not worth the small improvements, which could also be achieved by slightly increasing the Al thickness.

5.2 Scintillator Thicknesses

Perhaps the most important parameters that were optimized are those that are required at the fabrication stage and not simply at the time of operation. Specifically, the thickness of the top and middle scintillators must be chosen such that they are thin enough to allow for x-rays to penetrate them and reach the next layers but thick enough to both harden the x-ray

beam and absorb sufficient photons for the creation of a good image. To find the optimal values, both t_t and t_m were allowed to independently vary along their respective ranges in the analytical model developed. These figures show the heavy dependence that the optimal values for both thickness parameters will have on the attenuation of the object of interest—and hence on I_{ml} —, particularly when dealing with t_m . This same effect is also observed for t_c in calcification detection.

The scintillator thicknesses that maximize our FoM for either applications are detailed in table 2. It is clear that any application that spans these large ranges for the I_{ml} or t_c parameters will require a detector that compromises on the values of these thicknesses to best suit the entire range. The thickness values we deemed most appropriate for the parameter ranges we investigated were 210 μm and 560 μm for t_t and t_m respectively when considering the iodinated vessel detection application, and 255 μm and 440 μm for calcification detection.

5.3 Comparison with other Dual-Energy Detectors

To evaluate the performance of the proposed detector as a DE imager, we compared it to the two established DE technologies. The first is dual-layer detector with a metal mid-filter, and the second is a kVp switching single-layer detector. Both of these were optimized using an analytical model similar to the one developed for the three-layer detector. The former only differed in that the thickness of the metal (copper) mid-filter had to be optimized as opposed to the thickness of the middle scintillator. The latter, on the other hand, differed in that it required both low and high kVp values and the ratio of patient expose of the low to the high image to be optimized, but only had a sole scintillator thickness parameter.

Using the optimized parameters, the FoM was calculated with the respective analytical models and Monte Carlo simulations for both applications in the I_{ml} and t_c ranges of interest for all three detector types. As shown in fig. 10, the Monte Carlo results follow the same trends as the presented analytical model, corroborating its effectiveness as a first-order optimization method. However, a mean difference of 32 % is observed between the two approaches, since the Monte Carlo simulations take into account several effects that the analytical model doesn't—most notably cross scatter between the detectors and re-emissions of the scintillator materials.

It is clear that the three-layer proposed detector performs comparably to an established single-shot detector. Thus, our detector has the advantage over other single-shot imagers of being able to obtain an image that includes the signal from its middle layer without sacrificing performance as a DE imager. Given these results and the recent decrease in cost of building each flat-panel sensor, the three-layer stacked detector presents a promising alternative to current single-shot imagers with a clear advantage and no performance downsides.

As expected,^{18,42} the poorer spectral separation of single-shot imaging resulted in lower FoM results for the three-layer detector when compared to an optimized kVp switching one. However, it must be noted that our FoM does not take into account the motion artifacts to

which kVp switching imagers will be susceptible, not to mention the potential further applications of our proposed detector.

6 Future Works

It is possible to build on the work presented in this report and expand further on the potential of the three-layer stacked detector. First, this report has analyzed the performance of our proposed three-layer stacked detector as DE imager according to the chosen FoM. Perhaps the largest issue with this figure is that it will benefit from individual image blurring, as it is based on the variance of the signal at certain regions. The noise associated with each signal will therefore decrease as a consequence of blurring. However, blurring can have adverse effects on the detectability of certain objects, especially for very thick scintillators and large signal spreads. To account for this, the optimization and comparison procedure detailed here could be repeated while considering other figures of merit that better describe the possible adverse effects of blurring. To do so, these figures must take into account the loss in spatial frequency resolution due to blurring and thus must describe the frequency response of the system. Likely candidates of this type of metrics include the modulation transfer function, the noise equivalent quanta, and the detective quantum efficiency.^{43,44}

Different possible optimization pathways could also be considered, especially if full control over the type of x-ray source used is assumed. Other anode materials aside from tungsten might provide more optimal spectra for these applications. The possibility of using multiple-anode sources (where multiple spectra can be generated at different kVp simultaneously) is particularly appealing, as it could provide the higher spectral separation needed for single-shot imaging.

Lastly, further applications of the three-layer detector should also be investigated. Of particular interest are those that are made available thanks to the new image obtained in the middle sensor. This newly available information has never been present in single-shot detectors before, and hence its potential uses are plentiful. Such possible applications include creating false color radiographic images in which each layer of the detector is somehow mapped to one of the RGB channels. Other methods of combining all three images to improve detectability should also be investigated.

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Fig 1. Diagram depicting the three-layer design of the proposed detector, where each layer is composed of a scintillator layer of thicknesses t_t , t_m and t_b respectively coupled to a flat-panel sensor.

Fig 2.
Diagram depicting the two possible x-ray paths of main interest.

Fig 3.

Diagram illustrating signal spread in a scintillator. The x-rays incident over the entire region above the middle pixel ($i = j = 0$) result in a signal that is spread once it reaches the detector plane, represented by the ISF. The signal will eventually be discretized into pixels, each with a resulting value $\Delta S_{i,j}$.

Fig 4.
 a Normalized spectra incident on a typical configuration of the proposed detector with CsI:Tl as its scintillating material. b Mass attenuation coefficients over the diagnostic energy range for iodine, cortical bone and soft tissue.³¹

Fig 5.

Calculated values of the standard deviation $\sigma_G(E, t_{CsI}, p)$ of the fitted Gaussian kernel used to model $S(E, t_{CsI})$ as a function of scintillator thickness (left), x-ray energy (middle), and pixel pitch (right).

Fig 6.

Example images of iodine vessels of increasing I_{ml} from left to right obtained with Monte Carlo simulations. S_s and s are obtained from the right-most part of the image. S_{DE} is calculated using eq. (20).

Fig 7. Effects of x-ray source tube voltage on the FoM for a typical case in both applications considered.

Fig 8.

Effects of source filtering on the FoM for filter elements a Al and Cu, and b Rh, Mo, Ag and Sn. All models include a 0.1 mm intrinsic Al filtering, as most commercial tungsten sources do. While both figures show the application example of the detection of an iodinated vessel of 50 mg/cm^2 mass loading in 20 cm of soft tissue, the same trends were observed for both of the applications studied and all of their parameters.

Fig 9.

FoM for varying a top scintillator thickness and b middle scintillator thickness for the iodinated vessel detection application at various I_{ml} values. Each curve is independently normalized to its maximum value as higher I_{ml} values will intrinsically result in a higher FoM. When varying t_t in a, t_m was fixed at its compromise value, and vice versa for b.

Fig 10.

Comparison of optimized FoM results for a conventional kVp switching detector, a copper mid-filter single-shot detector (two-layer) and our newly proposed three-layer detector for the a iodinated vessel detection and b calcification detection application. Lines show the analytical model's result, while triangles, crosses and circles show the respective Monte Carlo simulation results.

Table 1

Detector and x-ray source (top 5 rows), and application (bottom 2 rows) parameters and ranges used to optimize the theoretical performance of the proposed detector.

Symbol	Definition	Range
kVp	Peak x-ray tube voltage	50 kV to 140 kV
t_t	Thickness of top scintillator	50 μm to 300 μm
t_m	Thickness of middle scintillator	100 μm to 800 μm
t_f	Thickness of source filtering	0 mm to 5 mm
	Source filtering material	Al, Cu, Rh, Mo, Ag, Sn
I_{ml}	Vessel's iodine mass loading	10 $\text{mg}=\text{cm}^2$ to 75 $\text{mg}=\text{cm}^2$
t_c	Thickness of calcification	1 mm to 5 mm

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Table 2

Optimized scintillator thicknesses when they are both allowed to vary independently for a range of application parameters. The last row shows in bold the chosen compromised values for the thicknesses.

$I_m(\text{mg}\cdot\text{cm}^{-2})$	Iodinated Vessels			Calcifications		
	$t_l(\mu\text{m})$	$t_m(\mu\text{m})$	$t_c(\text{mm})$	$t_l(\mu\text{m})$	$t_m(\mu\text{m})$	
10	209	501	1	254	430	
25	210	527	2	254	431	
50	210	572	3	255	436	
60	210	589	4	255	443	
75	211	614	5	255	450	
	210	560		255	440	