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Effects of the energy-separation filter on the performance of each detector layer in the sandwich detector for single-shot dual-energy imaging

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ABSTRACT: A novel sandwich-style single-shot detector has been built by stacking two indirect-conversion flat-panel detectors for preclinical dual-energy mouse imaging. Although this single-shot method is more immune to motion artifacts compared with the conventional dual-shot method (i.e., fast kVp switching), it may suffer from reduced image quality because of poor spectral separation between the two detectors. Spectral separation can be improved by using an intermediate filter between the two detector layers. Adversely, the filter reduces the number of x-ray photons reaching the rear detector, hence probably increasing image noise. For a better design and practical use of the sandwich detector for single-shot dual-energy imaging, imaging performances of each detector layer in the sandwich detector are investigated for various spectral-separation extents and applied tube voltages. The imaging performances include the modulation-transfer function, the Wiener noise-power spectrum, and the detective quantum efficiency. According to the experimental results, impacts of the intermediate filter on the imaging performances of each detector layer are marginal. The detailed experimental results are shown in this study.

KEYWORDS: X-ray detectors; X-ray radiography and digital radiography (DR)
1 Introduction

An x-ray projection image is typically formed by x-ray spectrum transmitted through a 3D object. Therefore, information presented by the x-ray projection image is incomplete because it is expressed as an average value over both depth and x-ray energy [1]. As a result, the projection imaging provides poorer lesion conspicuity compared with the space-discriminating computed tomography (CT) [2] and tomosynthesis [3], and the energy-discriminating photon-counting imaging [4]. Multi-energy CT/tomosynthesis may provide a far better conspicuity in the resultant images [5, 6]. However, these modalities require additional equipment for motion, specialized x-ray imaging detectors, and post-image processing.

A relatively simple method for an improved conspicuity in projection images is to subtract two images obtained at two different energies. By rapidly switching the kilo-voltage (kVp) applied to the x-ray tube, a dual-energy imaging (DEI) method is commercially available [7]. However, this dual-shot DEI is susceptible to motion artifacts due to the time delay between successive exposures and may increase the patient dose [8, 9].

As an alternative method, the single-shot DEI method with a multilayer (sandwich) detector has been introduced [10, 11]. For a single x-ray exposure, the front detector primarily uses low-energy x-ray photons while the rear detector primarily uses high-energy x-ray photons. Although the single-shot method is more tolerant of patient motion, it generally suffers from reduced image quality (e.g., contrast-to-noise ratio, CNR) compared to the dual-shot method because of poor spectral separation [12, 13]. Spectral separation can be improved by using intermediate filtration material between the two detector layers. Adversely, the filter can reduce the number of x-ray photons reaching the rear detector, hence increasing image noise. The use of intermediate filter in the sandwich detector trades off the image quality of DE images.

For a better design and practical use of the sandwich detector for single-shot DEI, therefore, it is important to investigate of its imaging performance for various spectral-separation extents. In
Figure 1. (a) CAD drawing and a picture describing the developed sandwich detector. (b) A picture showing the experimental bench to measure the detector performance.

In this study, the modulation-transfer function (MTF), the Wiener noise-power spectrum (NPS), and the detective quantum efficiency (DQE) of each detector layer in the sandwich configuration are measured with respect to various combinations of kVp’s and filter thicknesses.

2 Materials and methods

2.1 Detective quantum efficiency

The DQE of the \( j \)th detector layer (\( j \) designates \( F \) for the front or \( R \) for the rear detector, respectively) is given by [14]

\[
\text{DQE}_j(u) = \frac{\bar{q}_j G_j^2 \text{MTF}_j^2(u)}{\text{NPS}_j(u)} = \frac{\text{MTF}_j^2(u)}{\bar{q}_j \left[ \text{NPS}_j(u)/\bar{d}_j^2 \right]},
\]

(2.1)

where \( \bar{q} \) \text{ (mm}^{-2}\text{)}\) denotes the average incident photon fluence: \( \bar{q}_F = \bar{q} \) and \( \bar{q}_R = \bar{q}_F \tau \), where \( \tau \) represents the transmittance through the front detector and intermediate filter layers. \( G \) \text{ (DN mm}^2\text{)}\) is the detector gain relating \( \bar{q} \) to the average pixel signal \( \bar{d} \) in units of digital number (DN). Hence, the bracketed term in the denominator of equation (2.1) represents the normalized NPS (NNPS). \( u \) is the spatial frequency variable corresponding to \( x \) in the space domain.

2.2 Sandwich detectors

As shown in figure 1(a), the sandwich detector consisted of two flat-panel detectors (FPDs) and copper (Cu) sheet. The FPDs used the same photodiode array (RadEye1\textsuperscript{TM}, Teledyne Rad-icon Imaging Corp., Sunnyvale, CA) and gadolinium oxysulfide (Gd\textsubscript{2}O\textsubscript{2}S:Tb) phosphor screens (Carestream Health Inc., Rochester, NY) with different thicknesses: thinner (~34 mg cm\textsuperscript{-2}) for the front and thicker (~67 mg cm\textsuperscript{-2}) for the rear detector layers, respectively. The photodiode array had 0.048 mm-sized pixels arranged in 1024\times512 format. The thickness of Cu filter, \( t_{Cu} \), was varied from 0.1 to 0.5 mm.

2.3 Measurement and analysis

The sandwich detector was placed at a distance of 1 m from the focal spot of the x-ray tube (E7239X, Toshiba, Japan), as shown in figure 1(b). The tungsten-anode x-ray tube had 2.4 mm aluminum...
Figure 2. Summary of the sandwich detector responses. (a) $\bar{d}_F$ with $t_{Cu} = 0.3$ mm as a function of $X_F$ for various kVp’s. (b) $\bar{d}_R$ with $t_{Cu} = 0.3$ mm as a function of $X_F$ for various kVp’s. (c) $\bar{d}_R$ as a function of $X_R$ for various $t_{Cu}$ at 60 kVp. (d) The total transmittance factor as a function of $t_{Cu}$ for various kVp’s. (e) The transmittance due only to the Cu filter layers as a function of $t_{Cu}$ for various kVp’s. (f) The gains of each of detector layers as a function of kVp. Error bars indicate the average standard deviation of the measurements for an exposure or a kVp. Linear solid lines shown in (a), (b), and (c) denote least-squares regression analyses using first-order polynomials with zero-valued intercept, whereas dotted lines shown in (d), (e), and (f) denote trend curves.

equivalent filtration and the tube voltage was varied between 50 and 90 kVp. The exposure time was fixed to be 0.2 seconds while the each detector read out images in 3 seconds. Entrance exposure at the front detector, $X_F$ (mR), was measured with a calibrated ion chamber (Piranha R&F/M 605, RTI Electronics AB, Sweden) and $X_R$ at the rear detector after attenuation by the front detector and intermediate filter. The incident x-ray photon fluence was estimated using the calculated photon fluence per unit exposure $\bar{q}_0$ and the measured exposures [15].

The detector response was measured by calculating the average pixel signal $\bar{d}$ in central 128x128-pixel regions of 10 images obtained at each exposure level. The detector gain $G$ was estimated by dividing the slope of detector response curves, which were obtained for various incident x-ray exposures, by $\bar{q}_0$.

The MTF was evaluated from the images obtained for an edge-knife phantom. The edge profiles were extracted from the edge images, and then they were differentiated to give rise to line-spread
functions. By performing fast Fourier transformations (FFTs) to the line-spread functions, we obtained the MTF results.

The NNPS was evaluated by applying the 2D FFT to the zero-mean flat-field images normalized by the mean values. The 1D NNPS curves were obtained by extracting profiles from the 2D NNPS along a given direction.

The DQE was calculated using equation (2.1) with the measured MTF and NNPS, and with the estimated $\tilde{q}$. Because the MTF and NPS curves extracted along the directions parallel to the pixel addressing line ($x$ in the space and $u$ in the Fourier domains) and the data readout line ($y$ and $v$) of the photodiode array were the almost same to each other [16], only $u$-directional performances including DQE are shown here.

DE images were obtained of a postmortem mouse using the sandwich detector by combining two measured projections obtained from the front and rear detectors for various kVp and $t_{Cu}$ combinations. Image quality of the bone-enhanced DE images was assessed by calculating a figure of merit (FOM) defined as the squared CNR of enhanced bone region relative to background, normalized by the exposure used for imaging [11]:

$$FOM = \frac{\text{CNR}^2}{X_F} = \frac{|\tilde{d}_{\text{bone}} - \tilde{d}_{\text{bgn}}|^2}{X_F \left(\sigma_{\text{bone}}^2 + \sigma_{\text{bgn}}^2\right)}$$

(2.2)

where $\tilde{d}$ and $\sigma$ denote the pixel mean value and standard deviation of the regions of interest, respectively.

3 Results and discussion

As shown in figures 2(a) and 2(b), the responses of both detector layers were linear to exposure levels at the entrance surface of the sandwich detector (i.e., at the entrance of the front detector), and the sensitivity (DN mR$^{-1}$) was increased with increasing kVp. The front detector response was observed to be independent of $t_{Cu}$. The rear detector response was also nearly independent of $t_{Cu}$ when the transmittance was considered for the incident exposure at the rear detector, as shown in figure 2(c). Least-squares regression analysis showed that more than 99.82% of the data could be described by a first-order polynomial with a zero-valued intercept. Strictly speaking, however, the sensitivity of the rear detector was slightly increased with increasing $t_{Cu}$ (e.g., 6.3% larger sensitivity when $t_{Cu} = 0.5$ mm compared to the average sensitivity), probably due to the beam hardening effect. The authors note that the averaged relative errors of the measured responses of the front and rear detectors were about 0.84% and 1.04%, respectively. The measured transmittance through the front detector including intermediate filter layers is summarized in figure 2(d). The transmittance due only to the intermediate filter layers is shown in figure 2(e). While the attenuation of x-ray photon fluence due to the intermediate filter layers accounts for about half of the total attenuation when $t_{Cu} = 0.1$ mm, it accounts for more than 70% when $t_{Cu} > 0.1$ mm.

The average gain (DN mm$^{-2}$) of each detector as a function of kVp is shown in figure 2(e). The rear detector with a thicker phosphor showed a larger gain than the front detector.

As shown in figures 3(a) and 3(b), the MTF performances of each detector layer in the sandwich detector were nearly the same as those measured from their single-detector configurations. The MTFs were mainly characterized by the phosphor thickness; the thinner (front) phosphor showed a
Figure 3. Summary of the Fourier-based metrics measured for each detector layer. (a) The MTF results for various $t_{Cu}$ at 60 kVp. (b) The MTF results with $t_{Cu} = 0.3$ mm for various kVp’s. (c) The exposure-corrected NNPS results with $t_{Cu} = 0.3$ mm for various exposures at 60 kVp. (d) The exposure-corrected NNPS results with $t_{Cu} = 0.3$ mm for similar exposure levels at different kVp’s. (e) The DQE results with $t_{Cu} = 0.3$ mm for various exposures at 60 kVp. (f) The DQE results with $t_{Cu} = 0.3$ mm for similar exposure levels at different kVp’s. For comparison, ideal MTFs describing geometrical pixel response are included in (a) and (b). Respective performances measured from each of the front and rear detectors in a single-detector configuration are shown in (a), (c), and (e).
better MTF than the thicker (rear) one. Slight degradation in the rear-detector MTF was observed compared with the MTF measured from its single-detector configuration, and the reason was probably explained by the scattered/fluorescent x-ray photons due to the sandwich configuration. The effects of $t_{Cu}$ and kVp on the MTF performances were nearly negligible. For comparison, the ‘sinc’ function of the photodiode array, which describes the pixel aperture response in the Fourier domain, is also plotted. To calculate the sinc function, the authors assumed the fill factor of the photodiode array to be 0.8.

Figures 3(c) and 3(d) show the NNPS performances of each detector layer. Assuming that noise is due only to statistical fluctuations associated with the detection of a limited number of quanta [17], all the measured NNPS results were multiplied by the exposure levels used to remove the exposure-dependency of NNPS. Overall, the noise-power spectral densities of each detector were decreased with increasing spatial frequency. For the spatial frequencies less than $\sim 4 \text{ mm}^{-1}$, the roll-off speed of noise density of the rear detector was larger than that of the front detector. As shown in figure 3(c), the effects of sandwich configuration on the NNPS were marginal considering the measurement errors. On the other hand, the NNPS was degraded with increasing kVp, as shown in 3(d).

MTF and NPS performances of each detector layer were well reflected into the DQE performance of each detector layer, as shown in figures 3(e) and 3(f). As shown in figures 3(e), the DQE of each detector layer was almost independent of exposures used in this study and the effects of sandwich configuration on the DQE were marginal. On the other hand, the increase in kVp degraded both the DQE performances of the two detectors, as shown in figure 3(f), because the quantum efficiency lowered as the kVp increased [18]. The increase of variation in the measured pixel signal might further degrade the DQE performances with increasing kVp (i.e., decrease of the Swank noise factor at higher kVp) [18, 19].

Figure 4 shows bone-enhanced DE images obtained for a postmortem mouse using the sandwich detector for various combinations of kVp and $t_{Cu}$. DE images were obtained using weighted logarithmic subtraction of two images obtained from the front and rear detectors. The weighting factor was determined by minimizing contrast between the material to be subtracted and background [11]. All the images were displayed with the level of their mean value and a window of two times their standard deviation. For the maxillary bone and neighboring background regions as designated by the boxes in the mouse image (90 kVp and $t_{Cu} = 0.5 \text{ mm}$), the FOMs were calculated using equation (2.2), and the results are shown in figure 4. The maximum FOM was observed at the combination of 60 kVp and $t_{Cu} = 0.3 \text{ mm}$. The line patterns observed in the images are glue stains due to adhesion between the photodiode array and ceramic substrate [11].

4 Limitations

This study represents the performance evaluation of each detector layer (i.e., the front and rear detectors) consisting of the sandwich detector, but does not report the performance of the sandwich detector itself. The concept of DE MTF, NPS, and DQE describing the performance of sandwich detectors was introduced by Richard and Siewerdsen [20]. For example, the DE MTF was determined from the weighted average of the low and high-energy MTFs. However, this DE MTF concept could not describe the unsharp masking effect in single-shot DE images [11] that was resultant from subtraction of two images obtained from the front and rear detectors having re-
Figure 4. Bone-enhanced DE images obtained for a postmortem mouse using the sandwich detector for various combinations of kVp and $t_{Cu}$. The FOM includes the CNR between the maxillary bone and neighboring background regions as indicated by the boxes in the mouse image (90 kVp and $t_{Cu} = 0.5$ mm).

spective different phosphor thicknesses. Similarly, the DE NPS assumed that two images obtained from the two detector layers were independent to each other, but which is not true. Furthermore, the conventional DQE without consideration of anatomical background clutter may underestimate the sandwich detector performance [21]. Therefore, an alternative DE DQE model is necessary to correctly describe the true performance of the sandwich detector, and which will be a separate future study. On the other hand, this study is in pursuit of investigating the effects of spectral-separation filter on the imaging performances of each detector layer in the sandwich detector, and the authors believe that this work is also helpful to design a better sandwich detector.

In this study, only bone-enhanced images for a postmortem mouse have been demonstrated with the sandwich detector. The soft tissue-enhanced imaging based on the DEI technique may be more popularly required for clinical applications (e.g., detection and characterization of lung nodules in chest examinations) than the bone-enhanced imaging [22]. However, the bone-enhanced images can also improve characterization through analysis of calcification in chest examinations [23]. Although the soft tissue-enhanced images for a postmortem mouse was presented in previous studies [10, 11], more quantitative evaluation on the soft-tissue imaging performance using the sandwich detector will be shown in near future.

5 Conclusion

The use of a filter between front and rear detector layers in the sandwich detector for single-shot dual-energy x-ray imaging is essential to increase the x-ray spectral separation between the two detectors. The copper filter can significantly attenuate the x-ray photon fluence reaching the rear detector when its thickness is greater than 0.1 mm. From the experimental investigation, however, its impact on imaging performances, such as MTF, NPS, and DQE, is marginal. The authors concern the high-frequency noise performance of the rear detector can be vulnerable to the additive electronic noise because the number of secondary quanta lessens at the high frequency. For the
optimal design of a sandwich detector, it is required to define the DQE in dual-energy images obtained from the sandwich detector. Quantitative analysis of the imaging performance in dual-energy images with the help of the cascaded linear-systems model will be performed in a future study.

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References


