Feasibility of Active Sandwich Detectors for Single-Shot Dual-Energy Imaging

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ABSTRACT

We revisit the doubly-layered sandwich detector configuration for single-shot dual-energy x-ray imaging. In order to understand its proper operation, we investigated the contrast-to-noise performance in terms of the x-ray beam setup using the Monte Carlo methods. Using a pair of active photodiode arrays coupled to phosphor screens, we have built a sandwich detector. For better spectral separation between the projection images obtained from the front and rear detectors during a single x-ray exposure, we inserted a copper sheet between two detectors. We have successfully obtained soft tissue- and bone-enhanced images for a postmortem mouse with the developed sandwich detector using weighted logarithmic subtraction, and the image quality was comparable to those achieved by the conventional kVp-switching technique. Although some problems to be mitigated for the optimal and practical use, for example, the scatter effect and image registration, are still left, the performance of the sandwich detector for single-shot dual-energy x-ray imaging is promising. We expect that the active sandwich detector will provide motion-artifact-free dual-energy images with a reasonable image quality.

Keywords: Dual-energy imaging, sandwich detector, single shot, flat-panel detector, CNR, Monte Carlo

1. INTRODUCTION

Dual-energy (DE) imaging can remove overlapping anatomical structures that obscure the detection and characterization of lesions in radiographs by discriminating or enhancing the material content (e.g., bone or soft tissue). Although the theoretical framework describing DE imaging had been initialized in the 1970s, its renewed interest has been in the spotlight since large-area flat-panel detectors (FPDs) capable of real-time imaging with high detective quantum efficiency had been commercially available. DE x-ray imaging based on the fast kVp-switching technique (also known as "double-shot" or "double-exposure" technique), which acquires low- and high-energy projections in successive exposures, is therefore now available in clinic. However, this double-shot technique is susceptible to motion artifacts resulting from anatomical mismatch between two exposures. A great amount of effort based on image processing has been devoted to remove or reduce motion artifacts. Otherwise, the "single-shot" or "single-exposure" technique, which acquires low- and high-energy projections simultaneously from two detectors arranged in front and rear layers, could be an alternative.

We revisit the doubly-layered detector configuration or the "sandwich" detector for DE x-ray imaging. While sandwich detectors based on "passive" film-screen combinations or storage phosphors were reported, there has not been sufficient attention paid to the use of "active" solid-state detectors. In this study, we investigate a feasibility of the sandwich detector based on FPDs. To demonstrate its performance, we estimated the contrast-to-noise ratio (CNR) performances with respect to various x-ray beam setups using the simple Monte Carlo method and experimentally obtained DE (soft tissue- and bone-enhanced) images for a postmortem mouse. For comparisons, we also performed the Monte Carlo simulations and measured DE images using the conventional double-shot technique.

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2. MATERIALS AND METHODS

2.1 Sandwich detector

We built the sandwich detector by stacking two FPDs, as shown in Fig. 1. Each FPD consisted of a scintillator for converting x-ray into optical photons and a photodiode array for detecting the optical photons. As scintillation layers, we employed commercial phosphor screens based on terbium-doped gadolinium oxysulfide (Gd$_2$O$_2$S:Tb) phosphors. Considering the attenuation of x-ray photons through the front FPD, we installed a thicker scintillator at the rear FPD for higher sensitivity: Min-R$^{TM}$ 2000 (phosphor thickness = 0.084 mm) and Lanex$^{TM}$ Fast Front (0.11 mm) for the front and rear FPDs, respectively. The matrix-addressed active photodiode array (RadEye$^{TM}$, Teledyne DALSA Inc., Sunnyvale, CA) was fabricated by a complementary metal-oxide-semiconductor (CMOS) process and it featured a small 0.048 mm-sized pixel arranged in 1024×512 format. Therefore, the active imaging area was about 50×25 mm. We placed a 0.2 mm-thick copper (Cu) sheet between the two FPDs to enhance x-ray beam hardening through it, as shown in Fig. 1(c); thereby, increasing the energy separation between the projections obtained from the front and rear FPDs. The sandwich detector was installed in a light-tight box made of aluminum (Al) which opened a 1 mm-thick polycarbonate window for x-ray irradiation as shown in Figs. 1(a) and (b).

Figure 1. (a) Computer-aided design drawing and (b) picture of sandwich detector configuration. (c) Schematic illustration of x-ray spectra obtained from the front and rear detectors in the sandwich detector.

Figure 2. Pictures of (a) the postmortem mouse phantom and (b) the experimental setup for DE x-ray imaging.
2.2 Mouse imaging

We obtained images of a postmortem mouse (~40 g), whose blood was replaced by para-formaldehyde, as shown in Fig. 2(a), using the developed sandwich detector and the tungsten (W)-target x-ray source (E7239X, Toshiba, Japan) which incorporated a 1 mm-thick Al filter. Figure 2(b) shows the experimental setup for DE imaging. The source-to-detector distance was set to 1200 mm and 70 kVp spectrum was used for the sandwich detector. For a comparison, DE images using the double-shot technique were also obtained at 40/70 kVp combination. To obtain DE images, we applied weighted log-subtraction arithmetic to low- and high-energy projections.\(^2\) We determined the weighting factor when it resulted in the minimum contrast between the material to be subtracted and the background.

2.3 Monte Carlo simulations

For the proper operation of the sandwich detector, we performed the Monte Carlo simulations with respect to various operation parameters such as x-ray tube voltage (kVp) and intermediate Cu filter thickness. All x-ray spectra were generated using an in-house MATLAB\textsuperscript{®} routine for the W-target x-ray tube and the simulation framework was similar to a previous work.\(^6\) The number of incident x-ray photons in each energy interval (1 keV) was determined for the desired exposure (in mR) using a Poisson random number generator. As shown in Fig. 3(a), a virtual numerical phantom with four polyurethane (PU) disks and an Al bar embedded in polymethyl methacrylate (PMMA) was used for mimicking a mouse. The PU disks and Al bar mimicked soft tissue and bone, respectively, and the Al bar overlay two PU disks. For the virtual phantom, a 128×128 grid of 0.16×0.16-mm detector element was simulated, giving a 20.48×20.48-mm image. Monte Carlo simulation geometry is illustrated in Fig. 3(b) and the detailed physical parameters for the numerical phantom are summarized in Table 1.

Transmissions through each phantom components were calculated using tabulated values of the attenuation coefficients \(\mu\) and the thickness \(t\) for each material \(j\) such that \(e^{-\sum_{j}^{\mu_{j}t_{j}}}\). To simulate detector signal at each pixel of the front and rear detectors, respectively, we weighted each transmitted x-ray photon by its corresponding

Table 1. Physical parameters describing numerical mouse phantom.

<table>
<thead>
<tr>
<th>Material</th>
<th>Formula</th>
<th>Thickness (cm)</th>
<th>Density (g/cm(^3))</th>
</tr>
</thead>
<tbody>
<tr>
<td>PMMA</td>
<td>(C_5H_8O_2)</td>
<td>3.0</td>
<td>1.18</td>
</tr>
<tr>
<td>Aluminum</td>
<td>Al</td>
<td>0.1</td>
<td>2.70</td>
</tr>
<tr>
<td>Polyurethane</td>
<td>(C_{25}H_{42}N_2O_6)</td>
<td>0.3</td>
<td>0.59</td>
</tr>
</tbody>
</table>
quantum efficiency $\alpha(E)$ and energy absorption $A(E)$ of the given detector and then summed over the entire spectral distribution $N(E)$:

$$d_F = \int_0^{\infty} N(E) e^{-\sum_j \mu_j t_j} \alpha_F(E) A_F(E) dE$$

(1)

and

$$d_R = \int_0^{\infty} N(E) e^{-\sum_j \mu_j t_j} T(E) \alpha_R(E) A_R(E) dE.$$  

(2)

$T(E)$ denotes the transmission of x-ray photons through the front detector including the silicon wafer (0.725 mm), the Al$_2$O$_3$ substrate (1 mm), and the Cu filter. We assumed that $A(E)$ was the photon energy times the ratio of linear attenuation coefficients between energy absorption and total attenuation. DE images were then obtained using weighted logarithmic subtraction of two images from the front and rear detectors.

### 2.4 Contrast-to-noise ratio

For the quantitative evaluation of image quality, we calculated the CNR of the enhanced material $j$ against the background, which was normalized by the exposure $X$ used for imaging, such that

$$\frac{\text{CNR}_j}{\sqrt{X}} = \frac{|\bar{d}_j - \bar{d}_{bgn}|}{\sqrt{X \left( \sigma_j^2 + \sigma_{bgn}^2 \right)}},$$

(3)

where $\bar{d}$ and $\sigma$ denote the mean pixel value and standard deviation of the regions of interest (ROIs). For comparisons, the CNR of DE images obtained from the conventional dual-shot technique was also estimated.

### 3. RESULTS

#### 3.1 Simulated images and CNR performances

Figures 4, 5, and 6 summarize the projection, soft tissue (PU)-enhanced, and bone (Al)-enhanced images, respectively, with respect to various energy separation $\Delta E$ using the Monte Carlo simulation results. The energy
Figure 5. Calculated PU (tissue)-enhanced images from the Monte Carlo simulation results with respect to various energy separation $\Delta E$. (a) The single-shot technique. (b) The double-shot technique. Thick-line boxes denote the image showing the best image quality with naked eyes.

Figure 6. Calculated Al (bone)-enhanced images from the Monte Carlo simulation results with respect to various energy separation $\Delta E$. (a) The single-shot technique. (b) The double-shot technique. Thick-line boxes denote the image showing the best image quality with naked eyes.
Figure 7. Exposure-normalized CNRs in (a) tissue- and (b) bone-enhanced images. Open and solid symbols denote the simulations and measurements, respectively. Lines describe the trend curves for simulations: $y = ax$ and $y = bx^2 + cx$ for the double- and single-shot techniques, respectively.

The separation $\Delta E$ is the difference between the mean energies of spectra measured from the front and rear detectors in the single-shot technique and from the detector for low- and high-kVp settings in the double-shot technique. $\Delta E$ in the double-shot technique was determined by the kVp setting, and $\Delta E$ in the single-shot technique was determined by the kVp setting as well as the thickness of Cu filter.

As shown in Fig. 4, the projection image obtained from the rear detector becomes noisy as $\Delta E$ increases and kVp decreases because of the starvation in the number of x-ray photons detected as a result of the attenuation of low-energy photons and/or the low quantum absorption efficiency of the rear detector for high-energy photons. According to the simulation results, the sandwich detector well yields the DE images, as shown in Figs. 5, and 6. We note that the single-shot technique provides the DE image with the best image quality at certain $\Delta E$ (see the images highlighted by thick-line boxes in Figs. 5 and 6), whereas the DE image quality obtained from the double-shot technique improves with increasing $\Delta E$. This observation is more clearly shown in the CNR performance as plotted Fig. 7.

Figure 7 shows the exposure-normalized CNR performance calculated from the simulated DE images as a function of $\Delta E$. The inset images included in Fig. 7 show the ROIs for the calculation of CNR values. The double-shot technique shows the almost linear dependency of CNR on $\Delta E$, whereas the single-shot technique shows the maximum values of CNR with respect to $\Delta E$, which represents the presence of optimal Cu filter thickness. On the other hand, the increase in $\Delta E$ by separate exposures in the double-shot technique improves the image quality of DE images. Considering the CNR values and their uncertainties, we note that both DE techniques are more effective for the Al-enhanced (i.e., bone) images than the PU-enhanced (i.e., soft tissue) images.

### 3.2 Measured mouse images

Figure 8 shows projections, tissue- and bone-enhanced mouse images at 70 kVp obtained from the developed sandwich detector. For comparisons, images obtained from the double-shot technique for 40/70 kVp settings are included. The image qualities between two techniques are comparable to each other. CNR values were measured for both tissue- and bone-enhanced images obtained from the sandwich detector and plotted in Fig. 7 with solid symbols. The measured CNR values are well included within the range of simulated values. We note that the line patterns shown in the projection obtained from the rear detector [Fig. 8(b)] and thus DE images [Fig. 8(c) and (d)] obtained from the single-shot technique are glue stains for adhesion between the CMOS photodiode array and the ceramic substrate.
Figure 8. (a) and (b) are the projections of a postmortem mouse obtained from the front and rear detectors including a 0.2 mm-thick Cu filter, respectively, at 70 kVp ($\Delta E = 6.3$ keV). (c) and (d) are the tissue- and bone-enhanced images calculated from the projections (a) and (b). The dotted boxes in (c) and (d) denote the ROIs for the calculation of CNR performances. (e) and (f) are the tissue- and bone-enhanced images using the kVp-switching technique at 40/70 kVp ($\Delta E = 12.2$ keV).

4. DISCUSSION AND CONCLUSION

The double-shot technique is more flexible in the optimal design in terms of energy separation rather than the single-shot technique, because the double-shot technique has large margins in two kVp settings, beam current and exposure time to produce the optimum bone and tissue subtraction. However, there is still room for further enhancement in image quality of the single-shot technique. Proper registration of images for subtraction taking into account misalignment and magnification between the front and rear detectors will improve the image quality. Rejection or correction of scatter x-ray photons in images due to the sandwich detector configuration including the filter layer is another consideration for the improvement of image quality.

The decrease of CNR at higher $\Delta E$ in the single-shot technique is probably due to the significant reduction in the number of x-ray photons with lower energies detected at the rear detector because of their high attenuation through thick Cu filter as clearly shown in Fig. 4 and/or the low quantum absorption efficiency of the rear detector for hardened x-ray photons. Therefore, there exists the optimal thickness of intermediate filter or the operation parameters. Although the Monte Carlo simulations in this study neglected the contribution of scattered x-ray photons to projection images, we further suspect that noise in the projections from the rear detector due to scattered x-ray photons will increase with increasing $\Delta E$ as mentioned above. In addition, there would be correlation between two projections obtained from the front and rear detectors due to the scattered and fluorescence x-ray photons. It is doubt whether this correlation affects the image quality in dual-energy images positively or negatively at this moment. We are developing a theoretical model describing the signal and noise performance in the single-shot technique including the image correlation issue. We are investigating the effect of scatter and fluorescence photons using a commercial Monte Carlo code as well. Considering all these issues, we should design the sandwich detector, such as the converter materials (e.g., scintillator and...
photoconductor) for the front and rear detectors and their thicknesses, and the intermediate filter material and thickness, for the optimal and practical use. Although we used the scintillator-based flat-panel detectors for the sandwich detector, the photoconductor-based detectors can also be used. Combination of the scintillator- and photoconductor-based detectors might be another option for a better image quality as we arrange them accounting for which position (e.g., front or rear) requires higher sensitivity or higher spatial resolution. The active sandwich detector demonstrated in this study was built in laboratory scale. We believe that a careful integration of two detectors, each of that would be optimized in terms of sensitivity and spatial resolution, in the detector fabrication level in industry will provide the better image quality.

We demonstrated the feasibility of sandwich detector configuration for the single-shot dual-energy imaging that is immune to motion artifact and might require a smaller patient dose, compared to the conventional double-shot dual-energy imaging.

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