Is the image quality of conventional chest radiography obtained from a two-layer flat panel detector affected by the internal structure of the detector?

Shinya Takarabe a, b, Taku Kuramoto c, *, Yusuke Shibayama a, Yuzo Yamasaki d, Yoshiyuki Kitamura d, Hideki Yoshikawa a, Toyoyuki Kato a

a Division of Radiology, Department of Medical Technology, Kyushu University Hospital, 3-1-1 Maidashi, Higashi-ku, Fukuoka 812-8582, Japan
b Department of Oral and Maxillofacial Radiology, Faculty of Dental Science, Kyushu University, 3-1-1 Maidashi, Higashi-ku, Fukuoka 812-8582, Japan
c Department of Radiological Technology, Faculty of Health Sciences, Kobe Tokiwa University, 2-6-2 Otarucho, Nagaokaku, Kobe 653-0838, Japan
d Department of Clinical Radiology, Graduate School of Medical Sciences, Kyushu University, 3-1-1, Maidashi, Higashi-ku, Fukuoka 812-8582, Japan

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ABSTRACT

Purpose: Recently developed and commercialized dual-layer flat panel detectors (DL-FPDs) with two indirect scintillators are capable of acquiring dual-energy X-ray images. However, in clinical practice, they are utilized to perform conventional radiography using diagnostic X-rays with a wide energy spectrum. The two layers of the DL-FPD may affect the obtained image quality, even when only using one layer for conventional image acquisition, and these effects are yet to be substantiated. Therefore, in this study, we quantitatively evaluated the image quality of a conventional chest radiography using DL-FPD and visually verified the characteristics of the chest anthropomorphic phantom images.

Methods: The physical characteristics of the system were evaluated using the pre-sampled modulation transfer function (MTF), normalized noise power spectrum (NNPS), and detective quantum efficiency (DQE), for beam quality RQA 7 and RQA 9. In addition, the subjective visibility of the anthropomorphic chest phantom and simple objects images were compared with those of a conventional single-layer flat-panel detector (SL-FPD).

Results: No significant differences were found in the MTF between the SL-FPD and DL-FPD images. In addition, a higher DQE was observed at some exposure doses and in the high spatial frequency regions wherein NNPSs were lower for DL-FPD than for SL-FPD. Furthermore, no significant differences were found in the subjective visibility of the chest phantoms in each system.

Conclusions: We concluded that the image quality of the conventional radiography acquired with DL-FPD is comparable to or better than that of the SL-FPD.

Introduction

Flat panel detectors (FPDs) are a predominant technology in digital radiography [1]. Indirect FPD (I-FPD) systems are now widely used for medical radiography in clinical practice [2]. An I-FPD comprises of an X-ray scintillator (CsI:Tl or Gd2O2S:Tb, hereon referred to as CsI and GOS, respectively) [3] coupled with a two-dimensional array of pixelated photodiodes and thin-film transistors (TFTs). The scintillator converts the energy of the absorbed X-rays into visible light. Thereafter, the photodiode array converts it to charge and is read out to form a digital image. Both FPD systems that employ irradiation side sampling (ISS) have been developed and are now commercially available [4].

Further technological advances have recently led to the development of bilayer FPDs that consist of two indirect scintillators stacked on top of each other. In a previous study, Lu et al. [5] implemented a prototype dual-layer FPD (DL-FPD) with a copper metal filter that was stacked between two CsI scintillators. In contrast, DL-FPD with two X-ray scintillators, CsI and GOS, was developed and made commercially available by FUJIFILM Co. in 2018 [6]. This DL-FPD enabled single-exposure dual-energy (DE) imaging acquisitions wherein the top-layer scintillator (CsI) absorbs low-energy X-rays, and the bottom-layer scintillator (GOS) stops the transmitted high-energy X-rays. Previous studies have

* Corresponding author.
E-mail address: t.kura13@gmail.com (T. Kuramoto).

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reported that DE chest radiography improved the detection of lung cancer screenings as compared to conventional radiography using diagnostic X-rays with a wide energy spectrum [6]. Nevertheless, to guarantee signal-to-noise ratio and reduce the graininess of DE imaging, the exposure dose needs to be approximately 20% higher than that of the conventional radiography [6]. Therefore, in clinical practice, chest images are acquired using conventional radiography as well as DE X-ray imaging. For conventional radiography using DL-FPD, only the upper scintillator of the two layers is utilized for image acquisition. However, since the two scintillator layers are very close to each other, the scattered radiation generated by the lower scintillator may affect the acquired image. A previous study [5] reported that the MTF of the images obtained with the bottom scintillator layer of the prototype DL-FPD decreased at low frequencies due to internal scattering. However, the structural details of the commercially available DL-FPDs have not been published by the manufacturers. In addition, the effect of the top scintillator layer on the image quality has not been investigated. Therefore, the objective of this study is to evaluate the image characteristics of the conventional chest radiography using DL-FPD. In addition, anthropomorphic phantoms of the chest have been utilized to visually confirm the image quality of the DL-FPD system.

Materials and methods

Equipment

A general radiographic system that combined an X-ray generator (UD150B-40, Shimadzu Co., Kyoto, Japan) and an X-ray tube (0.6/1.2P324DK-125, Shimadzu Co., Kyoto, Japan) was used for all experiments. Table 1 lists the specifications of the digital imaging systems used in this study. Two types of FPD systems (DR CALNEO Smart C47, FUJIFILM Co., Tokyo, Japan; and DR CALNEO Dual, FUJIFILM Co., Ltd., Tokyo, Japan) were utilized. These two systems differed with respect to their characteristics of their scintillator layers: SL-FPD and DL-FPD, respectively. These FPDs were used the ISS method for sampling wherein the edge and the FPD, and the horizontal and vertical directions correspond to the rows of pixels in the FPD, respectively. MTFs were derived by averaging three independent measurements in each RQA beam quality. All measurements were obtained by averaging three independent measurements.

Table 1

<table>
<thead>
<tr>
<th>Specifications for digital imaging systems used in this study.</th>
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<tr>
<td>System</td>
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<tr>
<td>Product name</td>
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<tr>
<td>Manufacturer</td>
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<tr>
<td>Scintillator</td>
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<td></td>
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<tr>
<td>Sampling method</td>
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<tr>
<td>Pixel size (mm)</td>
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<tr>
<td>Field size (inch)</td>
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<tr>
<td>Matrix size</td>
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<td>Bit depth (bits)</td>
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X-ray beam quality

To evaluate the physical image quality of the detector, we used X-ray beam qualities as per the International Electrotechnical Commission (IEC) (62220-1-1) standard [7] of RQA 7 (tube voltage: 90 kV, additional filtration: 35.0 mm Al, half-value layer: 9.2 mm Al) and RQA 9 (tube voltage: 120 kV, additional filtration: 42.0 mm Al, half-value layer: 11.3 mm Al). The source image receptor distance (SID) was fixed at 150 cm for determining the X-ray beam quality. The distance between the detector and the walls of the room was set to 50 cm to minimize the effect of backscattering. Dosimetry was performed using a calibrated X-ray measurement system (Accu-Gold + sensor, 10 × 6-6 chamber, Radcal Co., Monrovia, CA, USA).

Image quality

The digital characteristic curves, pre-sampled modulation transfer function (MTF), normalized noise power spectrum (NNPS), and detection quantum efficiency (DQE) of each FPD system were measured using both RQA beam qualities. All measurements were obtained by averaging three independent measurements.

The digital characteristic curves were measured using a combination of distance and timescale methods. SID was respectively set to 100 cm and 200 cm to acquire uniform exposure images of the high-dose and low-dose ranges. A linear relationship between the displayed exposure time of the X-ray generator, and the relative exposure dose was confirmed before the measurements. The digital characteristic curves for the entire dynamic range of the FPDs, used for linearization in the MTF and NNPS measurements, were obtained by combining the ranges of the high-dose and low-dose curves. The edge method with a tungsten edge device (100 × 100 × 1.0 mm; Toshiba Materials Co., Ltd., Kanagawa, Japan) was used to measure the MTF of the raw image, and SID was set to 200 cm to acquire these images. The exposure conditions were based on the radiation dose required to achieve approximately 80% of the maximum pixel value in each FPD system for each RQA beam quality. The two MTFs were calculated according to different arrangements of the edge and the FPD, and the horizontal and vertical directions corresponded to the rows of pixels in the FPD, respectively. MTFs were derived by averaging three independent measurements in each direction.

NNPS of the obtained images was calculated using 2D fast Fourier transform by uniform exposure under various exposure conditions. To minimize the effect of statistical fluctuations, over four million pixels were used for NNPS calculations. The exposure dose on the detector was varied from 8.8 μGy (1.0 mR) to up to 43.8 μGy (5.0 mR). DQEs were calculated by the following equation:

\[
DQE(\mu, \nu) = \frac{MTF^2(\mu, \nu)}{E \times q \times NNPS(\mu, \nu)}
\]

where \( \mu \) and \( \nu \) represent the spatial frequencies in the horizontal and vertical directions, respectively; \( E \) represents the entrance air kerma to the detector (\( \mu G \)); and \( q \) is an estimate of the square of the signal-to-noise ratio per unit area per unit exposure, assumed to be 32,490 (RQA 7) and 31,087 (RQA 9) (1/mm×Gy) based on IEC 62220-1-1 [7]. The Nyquist frequencies of the FPD systems were 3.33 cycles/mm.

Phantom images

Two types of phantoms were used to compare the visibility of the images. One was a simple object of acrylic beads with a diameter of 2.5 mm on a 9 cm lucite phantom that simulated X-ray attenuation in a peripheral lung [8], while the other was a multi-purpose anthropomorphic male chest phantom (N1 “LUNGMAN,” Kyoto Kagaku Co., Ltd., Kyoto, Japan) with an 8 mm acrylic ball and a 100 HU (Kyoto Kagaku Co., Ltd., Kyoto, Japan) [9]. This acrylic ball simulated lung nodules and was superimposed on the pulmonary area of the chest phantom [10]. Two types of phantom images were acquired using each FPD system with an X-ray stand and anti-scatter grids (grid ratio of 12:1, density of 40 lines/cm, focusing distance of 180 cm, interspace material was aluminum; Mitaya Manufacturing Co., Ltd., Saitama, Japan). These images were acquired with SID of 180 cm at various exposure conditions. The surface dose on the phantoms was varied from 8.8 μGy (1.0 mR) to 43.8 μGy (5.0 mR) at both 90 kV and 120 kV. All radiographs were acquired using a collimated X-ray beam over a fixed area of 30 × 40 cm², and image processing was adopted for chest radiography since it...
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Results

Figure 1 displays the digital characteristic curves that indicate the relationship between the exposure dose and pixel value of the raw images. These curves exhibit different characteristics for different systems and beam qualities; however, they both displayed linear tendencies over a wide range of exposures.

Figure 2a and b display the results of comparing the MTFs for each system at beam quality RQA 7 and RQA 9, respectively. Table 2 displays the MTF values at a spatial frequency of 1.0, 2.0, and 3.0 cycles/mm for the abovementioned RQA beam qualities in the SL-FPD and DL-FPD systems. Since there were no noteworthy differences between the horizontal and vertical MTFs for all conditions, only the MTFs in the horizontal direction were specified. For beam quality RQA 7, there were no observable differences in the MTFs between the SL-FPD and DL-FPD (Fig. 2a). For beam quality RQA 9, MTF of SL-FPD was slightly lower than that of DL-FPD after 1.4 cycles/mm (Fig. 2b). However, these differences were negligible at less than approximately 3.1% of the MTF value for each RQA beam quality.

Figure 3 displays the NNPS values of each FPD system under various exposure doses of RQA 7 and RQA 9 beam qualities. Since there were no noteworthy differences in the horizontal and vertical NNPS values for each FPD system, only those in the horizontal direction are displayed in this figure. The NNPS values in all exposure conditions for both systems decreased as the exposure increased. In addition, the NNPS of each system exhibited different trends as the dose increased. The NNPS values of DL-FPD at 3.0 cycles/mm were up to about 20% lower than those of SL-FPD under 43.8 μGy (5.0 mR) for each RQA beam quality.

Figure 4a, b, and c display the DQE values in the horizontal direction for each FPD system for each RQA beam quality at 8.8 μGy (1.0 mR), 26.3 μGy (3.0 mR) and 43.8 μGy (5.0 mR), respectively. In each system, the DQE values of RQA 9 were lower than those of RQA 7. Similar results were obtained under other exposure doses as well. For 3.0 cycle/mm at 43.8 μGy (5.0 mR), the DQE values of each RQA beam quality of DL-FPD were about 30% higher than those of SL-FPD; a similar tendency was observed in the vertical direction as well.

is commonly used in clinical practice. The image quality of each FPD system was compared by two radiologists with 15 years of experience in interpreting digital images. All viewings were performed under subdued and uniform lighting conditions (approximately 30 lx) on a calibrated three-megapixel medical color liquid crystal display (LCD) (Radiforce RX340, Eizo Co., Ishikawa, Japan). The subjective evaluation by the radiologist was performed on the same type of monitors. The pair of images acquired from DL- and SL-FPD with the same dose and beam quality was displayed on each monitor. The radiologist observed these images under same exposure condition on each monitor simultaneously. The only difference in the observed images was the FPD system used. These radiologists were not informed about image acquisition condition . Each statement was to be answered with a three-grade evaluation (DL-FPD is better, equivalent and SL-FPD is better).

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**Table 2**

<table>
<thead>
<tr>
<th>Beam quality</th>
<th>Spatial frequency (cycles/mm)</th>
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<tr>
<td></td>
<td>1.0</td>
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<tr>
<td>SL-FPD</td>
<td></td>
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<tr>
<td>RQA 7</td>
<td>0.765</td>
</tr>
<tr>
<td>RQA 9</td>
<td>0.741</td>
</tr>
<tr>
<td>DL-FPD</td>
<td></td>
</tr>
<tr>
<td>RQA 7</td>
<td>0.751</td>
</tr>
<tr>
<td>RQA 9</td>
<td>0.741</td>
</tr>
</tbody>
</table>

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Fig. 1. Relationship between the relative exposure dose and pixel value of each FPD system with two RQA beam qualities: RQA 7 and RQA 9. The solid and dotted lines indicate the characteristic curves of the SL-FPD and DL-FPD systems, respectively.

Fig. 2. Comparison of the pre-sampled MTF values in the horizontal direction for the two FPD systems with different beam qualities: (a) RQA 7 and (b) RQA 9. The solid and dotted lines indicate the MTF of the SL-FPD and DL-FPD systems, respectively.
Figs. 5 and 6 present two types of phantom images for each FPD system under various exposure conditions. As a result of comparing images obtained with different FPDs under same exposure conditions, the two radiologists answered “equivalent” in subjective visibility for all exposure conditions.

**Discussion**

We evaluated the image characteristics of the SL-FPD and DL-FPD systems under chest radiography conditions using conventional radiography and confirmed our findings using phantom images.

There were no significant differences in the MTF values between SL-FPD and DL-FPD under two types of beam qualities, namely RQA 7 and RQA 9. (Fig. 2a and b, and Table 2). The MTF values of RQA 9 were slightly lower than those of RQA 7 for both the SL-FPD and DL-FPD systems (Fig. 2a and b) due to the increased effective energy caused by the beam hardening effect of the 42.0 mm Al filter and a high tube voltage [11,12]. The absorption efficiency of the CsI scintillator decreases continuously with increasing X-ray energy [13]. High-energy X-rays accumulate energy in the deep part of the CsI scintillator, and the diffused light from the deep part enters the photodiode that is arranged in front of the incident X-ray direction. As a result, in this study, the MTF values of both FPD systems were degraded for RQA 9.

The NNPS values of SL-FPD and DL-FPD exhibited almost the same values at 8.8 μGy (1.0 mR); however, in the spatial frequency region above 1.8 cycles/mm at 26.3 μGy (3.0 mR) and 43.8 μGy (5.0 mR), DL-FPD exhibited lower values than SL-FPD. Due to this, at 8.8 μGy (1.0 mR), the DQE values of both systems were nearly identical. At 26.3 μGy (3.0 mR) and 43.8 μGy (5.0 mR), DL-FPD exhibited higher DQEs than SL-FPD in the spatial frequency region above 1.0 cycles/mm. Based on these results, we discussed the details regarding the structure of DL-FPD that have not been published by the manufacturer. The high DQE values of DL-FPD indicates an increased thickness of the top scintillator layer (CsI). Conventionally, the MTF value worsens as the scintillator thickness increases [14]; however, this was not observed in the obtained results. Therefore, we inferred that there was no substantial difference in the thickness of the scintillator layers between SL-FPD and DL-FPD. In other words, the quality of the scintillator itself may have been improved. In addition, it may be due to individual differences in the CsI scintillators [3]. Nevertheless, it is
worth noting that the DQE values of DL-FPD were higher than that of the SL-FPD. As a result of comparing images obtained with the SL-FPD and DL-FPD systems under same exposure conditions and beam qualities, the two radiologists answered “equivalent” in subjective visibility for all exposure conditions. Although there were some differences in the DQE factors under physical evaluation, they were not directly reflected in the results of the visual assessment. Therefore, the difference in the DQE values did not necessarily affect the results of the visual evaluation [15]. This leads us to believe that image processing, the only parameter not considered, is a possible cause of the discrepancy. From the results, we concluded that the image quality of conventional radiography obtained from the DL-FPD system is comparable to or better than that of the SL-FPD system.

This study has several limitations. First, it does not consider any imaging site other than the chest. This is important because when the imaging site is changed, the tissue to be observed (bone tissue, soft tissue, etc.) also changes and therefore, the quality of X-ray required also varies. Therefore, strictly speaking, our results cannot be generalized to all forms of radiography. However, since chest radiography is the most commonly performed diagnostic imaging examination worldwide, the results of this study is expected to be useful for many patients. Second, the two phantoms used in this study have fixed thickness. Phantoms of different thicknesses may result in variations in other parameters such as scattered doses and image visibilities. However, optimizing these parameters was not an aim of this study and therefore, it was not explored. This theme will, however, be a topic of our future research. Thirdly, a three-grade evaluation was used for the subjective visual evaluation in this study. In the future, to perform more detailed evaluation, Likert evaluation with an increased number of grades must be performed. Finally, we did not perform quantitative visual evaluation using clinical images, such as, the Scheffe paired comparison method and receiver operating characteristic curve analysis in this study. Although these methods are advantageous in proving a visibility difference between the images to be compared, they are not suitable for determining the equivalence of visibilities. The observations of the radiologists using the

Fig. 4. Comparison of the DQE values of each FPD system in the horizontal direction with two types RQA beam quality (RQA 7 (solid line) and RQA 9 (dotted line)) at (a) 8.8 μGy (1.0 mR), (b) 26.3 μGy (3.0 mR) and (c) 43.8 μGy (5.0 mR). The black and red lines indicate the DQE of the SL-FPD and DL-FPD systems, respectively. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)
Fig. 5. Close-up radiographs of five acrylic beads with a diameter of 2.5 mm put on a 9 cm lucite phantom for DL-FPD and SL-FPD with 90 kV and 120 kV and different doses (8.8 μGy (1.0 mR), 26.3 μGy (3.0 mR), and 43.8 μGy (5.0 mR)).

Fig. 6. Close-up radiographs of simulated nodules (arrows) over pulmonary area in the chest phantom for DL-FPD and SL-FPD with 90 kV and 120 kV with different doses (8.8 μGy (1.0 mR), 26.3 μGy (3.0 mR), and 43.8 μGy (5.0 mR)).

Conclusion

We quantitatively evaluated the image characteristics of the high-resolution medical monitor adequately reflect the influence of the physical measurements obtained in this study on the clinical images.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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