Effective detective quantum efficiency in digital radiography: A phantom study on chest image

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**Purpose**

The detective quantum efficiency (DQE) is widely used as a standard to characterize all kinds of X-ray detectors.

The eDQE is a measurement of a detector's efficiency within a total X-ray system and as such considers the effects of focal spot size, scattered radiation, magnification and scatter reduction techniques.

The efficiency of low tube potentials for chest imaging is also reported based on the receiver operating characteristic method. The aim of this study was to measure the eDQEs for various tube voltages, with and without the use of an anti-scatter moving grid, by using a phantom simulating the human chest.
Methods and Materials

1. Equipment

Images were acquired using a Digital diagnost TH (Philips, Netherlands) silicon-based indirect flat-panel detector. The matrix size of the images was 3000 × 3000, and the pixel pitch was 0.143 mm. A phantom designed by the Food and Drug Administration for use in the Nationwide Evaluation of X-ray Trends (NEXT) program was used. The phantom simulates the attenuation and the scatter properties of an average human chest and consists of 25.4 × 25.4 cm$^2$ pieces of clear acrylic and aluminum with a 19 cm air gap that emulates the thoracic cavity. The configuration of the phantom is detailed in Fig. 1. The anti-scatter grid that we tested had a density of 36 line/Cm, a ratio of 12:1 and a focus distance of 180 cm.

2. Dosimetry

Exposure factors of 117 kV and 3.2 mAs with a focus-to-detector distance of 180 cm are the general radiographic conditions used with digital radiography (DR) for an average-sized adult chest examination in Korea. The effective dose for an average sized adult was calculated as 4.9 µSv by using PCXMC version 1.5.1 (STUK, Finland) and general radiographic conditions. For different tube voltages, 81, 90, 102, 109, and 117 kVp, the entrance dose was measured and converted into an effective dose by using each conversion factor which was obtained as the effective dose per mGy. The milliampere-second values for each tube voltage were adjusted until the intended effective dose was reached.

The incident air kerma was measured using a calibrated ionization chamber (Radcal 10 × 9 - 60 ion chamber/Radcal 9095 radiation monitor, Radcal Corporation, Monrovia, USA) for six exposure conditions. The incident air kerma of the detector position was estimated from the phantom' position based on an inverse square law. The chamber was located in front of the phantom surface without the phantom in the beam. This set up allowed us to measure the air kerma without backscatter from the phantom, the so-called incident air kerma.

3. Linearity

The linearity of the detector response is a graph of the detector signal versus the entrance exposure. The flat field images were acquired for different exposure conditions (five different tube potentials and with/without an anti-scatter grid). The raw images were acquired using a standard set-up (Fig. 2). The points on the characteristic curve are the average detector signal level in the region of interest (ROI), which measured 125 × 125
mm² and was positioned over the center of the phantom, versus the averages of three exposure measurements.

4. Scatter Fraction (SF)

The SF is defined as the ratio of the scattered radiation to the total radiation:

\[ SF = \frac{S}{S + P} \]

where S stands for pixel value in areas without contribution from primary radiation and S + P stands for pixel values in pixels with both primary (P) and scattered radiation (S).

The scatter fractions (SF) of the systems were measured using a beam stop array device composed of a 14 × 16 array of 224, 6mm-thick and 3mm-diameter lead cylinders. The SF measurement requires three steps. First, three raw images of the beam' stop array device placed in front of the phantom were taken for each measurement condition (Fig. 3). Then, all mean pixel values behind the 16 beam stops at the center of the image were measured as scattered radiation. Each ROI (15 × 15 pixels) eliminated pixels bordering the beam stop edge to prevent penumbral shadow effects. Finally, the average of all mean pixel values with the same size ROI on either side of the 16 beam stops was measured as total radiation, including both primary and scattered radiation.

5. Transmission Fraction (TF)

The transmission fraction was calculated using the ratio of the air kerma with and without the phantom. The X-ray beam was collimated to be slightly larger than the ionization chamber size. The air kerma was measured with and without the phantom between the X-ray tube and the ionization chamber for each exposure condition.

6. Effective Normalized Noise Power Spectrum (eNNPS)

The noise properties of the imaging system were acquired using only the phantom with the same set-up as illustrated in Fig. 2. Images of the phantom were obtained at an exposure level of \( K_0 \), where \( K_0 \) is the normal level of exposure at each measurement condition. The eNPS was determined using the four raw images to obtain at least 4 million independent pixels in the 125 × 125 mm ROI.

The image processing followed established steps to deduce the eNPS. The four phantom images were averaged and subtracted from the extracted image which was one of the four images used for averaging to remove the background trends due to the anode-heel effect and system noise. The center area of the processed image was divided into 16 overlapping sub-ROIs 360 × 360 in size. The two-dimensional (2D) eNNPS was
calculated using the Fourier transform and was normalized by dividing by the square of the mean large area signal. The one dimensional (1D) eNNPS was obtained from the horizontal axis by using the four rows above and below, excluding the axis row.

7. Effective Modulation Transfer Function (eMTF)

The resolution properties of the system were measured using the edge method. A schematic diagram is illustrated in Fig. 2. The phantom was positioned just adjacent to the detector’ cover plate. A edge test device was placed in contact with, and centered on, the phantom, at a 2~4° angle relative to the vertical axis of the phantom. This measurement geometry involved the effects of the focal spot blur penumbra, magnification, scattered radiation, and anti-scatter grid. For each of the different exposure conditions, three images of the slit were acquired at 3.2 $K_0$, where $K_0$ is the incident air kerma corrected to the detector plane and averaged. The eMTF was calculated using the Fourier transform, from which the data were obtained by measuring the line spread function of the system. The details of this method have been previously described.

8. eDQE

The eDQE of the DR system was measured incorporating the influences of scatter, magnification, phantom transmission fraction, anti-scatter grid, eMTF, eNNPS, incident exposure, and number of photons per mm$^2$-$\mu$Gy. The q values were estimated using an X-ray spectral simulation program (SRS-78). The eDQE was determined using the following equations: (Fig4)

\[
\text{eDQE} = \frac{q}{u' \cdot m \cdot (FOD \cdot FDD)}
\]

where $K_0$ = incident air kerma corrected to the detector plane ($\mu$Gy); $q$ = number of photons per mm$^2$-$\mu$Gy; $u$ is the spatial frequency in the detector plane (mm$^{-1}$); $u'$ is the spatial frequency corrected to the object plane (mm$^{-1}$); $m$ = magnification; FOD = focus to object distance (cm); FDD = focus-to-detector distance (cm).
Images for this section:

**Fig. 1:** (a) Average-sized chest phantom. (b) Geometry and configuration in a diagonal view. Al = aluminum.

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Fig. 2: (a) Experimental standard set-up of the phantom with a DR detector and anti-scatter moving grid and (b) resolution measurement set-up using a edge device.

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Fig. 3: (a) Side-view schematic diagram for measurement of SF, (b) an image of beam stop array.

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\[ eDQE(f') = \frac{SNR_{out}^2 (f')}{SNR_{in}^2 (f')} \]

\[ SNR_{out}^2 (f') = \frac{eMTF(f')^2 (1 - SF)^2}{eNNPS(f')} \]

\[ SNR_{in}^2 (f') = TF \times K \times q \]

\[ eDQE(f') = \frac{eMTF(f')^2 (1 - SF)^2}{eNNPS(f') \times TF \times K \times q} \]

\[ f' = mf = \frac{FOD}{FDD} f \]

**Fig. 4:** Equations of the eDQE

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Results

The system showed excellent linearity between the mean pixel value and the exposure at each tube potential ($R^2 = 1$). The relationships were an important requirement of linear system analysis.

Figure 5 shows the scatter fraction with and without an anti-scatter moving grid at each tube potential. The scatter fraction with a grid was lower than without a grid by approximately 50% for all tube potentials.

Figure 6 shows that the transmission fractions for the phantom increased with increasing tube potential.

Figure 7 shows the eMTF results with and without the use of an anti-scatter moving grid at various tube voltages. The eMTF only slightly depended on tube potential regardless whether or not a grid was used. The variation in the resolution as a function of tube potentials was a little bigger with a grid than without a grid.

Figure 8 illustrates the eNNPS at exposure levels of $K_0$ for 81 and 90 kVp. The results using high exposure conditions are not reported as they were very similar to those using a low exposure condition. For all tube voltages, a higher magnitude of the eNNPS was acquired with an anti-scatter moving grid than that without a grid.

Figure 9 shows the eDQE with and without an anti-scatter moving grid at different tube potentials. The tendency of conventional DQE will be different from the eDQE if we evaluate the conventional DQE by considering only the detector effect. The eDQE was higher without the use of an anti-scatter grid for most tube potentials. Furthermore, the eDQE was larger at low tube potentials than at high tube potentials both with and without the use of a grid.
Fig. 5: Scatter fraction for the phantom with and without an anti-scatter grid.

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Fig. 6: Transmission fraction for the phantom at each tube potential.

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Fig. 7: Measured eMTF (a) with and (b) without an antiscatter grid at various tube potentials.
**Fig. 8:** Measured eNNPS (a) with and (b) without an antiscatter grid at various tube potentials.

**Fig. 9:** eDQE measured (a) with and (b) without an antiscatter grid at different tube potentials.
Conclusion

This study used the eDQE to evaluate exposure conditions, including tube potentials and scatter reduction techniques, for chest examination. The environment in which we acquired images included scattered radiation, magnification, and focal spot blur. Our results suggest that the use of low tube potentials, such as 81 or 90 kVp, without an anti-scatter grid represents the most appropriate exposure conditions for DR in chest imaging. In conclusion, the eDQE reflects the real exam environment better than the general DQE because it represents by the effects of a realistic X-ray system.
References


