

Optimization of Tube Voltage in Digital Chest Radiography: Integrating Exposure Index and Image Noise Using an Anthropomorphic Phantom

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Abstract Tube voltage (kVp) influences detector exposure and objective image-quality characteristics in digital radiography; however, the integration of standardized exposure indicators with region-specific image noise and signal-to-noise ratio (SNR) across anatomically distinct thoracic regions remains limited. This study evaluated the effects of tube-voltage variation on Exposure Index (EI), Deviation Index (DI), image noise, and SNR in digital chest radiography. An anthropomorphic PBU-50 thorax phantom was imaged using a Samsung XGEO GF50 digital radiography system at eleven tube-voltage settings from 50 to 100 kVp in 5-kVp increments while maintaining a constant 8 mAs. Three independent acquisitions were performed at each setting, yielding 33 radiographic images. EI and DI were extracted from DICOM metadata, while mean pixel value, image noise, and SNR were evaluated in five anatomical regions of interest: clavicle, scapula, costae, sternum, and lung field. Descriptive statistics, linear and quadratic regression, and Pearson and Spearman correlation analyses were performed. EI increased from 73.95 ± 0.28 at 50 kVp to 1980.91 ± 16.78 at 100 kVp, while DI increased from -4.01 ± 0.02 to $+10.28 \pm 0.04$. Image noise generally decreased with increasing kVp, with reductions from 50 to 100 kVp ranging from 6.9% in the scapula to 58.8% in the sternum. SNR exhibited region-dependent nonlinear responses, and quadratic models better characterized several kVp-dependent relationships. At 60 kVp, EI was 216.02 ± 1.68 , and DI was $+0.65 \pm 0.03$, representing the only investigated condition within the predefined $-1 \leq DI \leq +1$ interval while providing higher SNR than at 50 kVp across all ROIs. These findings demonstrate an anatomy-dependent trade-off between detector exposure and objective image quality. The 60-kVp condition should therefore be considered a candidate detector-exposure condition for further optimization using patient-dose measurements and diagnostic image-quality assessment.

Keyword Digital Radiography, Exposure Index, Deviation Index, Image Noise, Signal-to-Noise Ratio, Tube Voltage

1. Introduction

Chest radiography is an essential diagnostic imaging modality for detecting lung cancer, trauma, pneumonia, airway infections, and other pulmonary diseases [1], [2], [3]. Its clinical use increased substantially during the COVID-19 pandemic for screening, diagnosis, and patient monitoring [4][5], and international evidence demonstrates its important role in diagnosis and follow-up imaging [6]. Adequate radiographic image quality is essential for reliable diagnosis and for minimizing repeat examinations that may unnecessarily increase patient radiation exposure [7]. Consequently, optimizing exposure protocols in digital radiography is vital to deliver superior diagnostic utility while adhering strictly to the As Low As Reasonably Achievable (ALARA) principle [1]. In digital radiography (DR), tube voltage (kVp) determines X-ray beam energy and penetrability and consequently influences detector exposure, subject contrast, and image-quality characteristics [8][9]. To standardize detector-exposure monitoring, the International

Electrotechnical Commission introduced the Exposure Index (EI), Target Exposure Index (EIT), and Deviation Index (DI) through IEC 62494-1 [9], [10], [11]. EI provides a quantitative indicator of detector response within the relevant image region, whereas DI expresses the logarithmic deviation of the measured EI from the examination-specific EIT [12][13]. These indicators provide useful feedback for identifying exposure conditions that differ from the intended detector exposure, but they should not be interpreted as direct measures of patient radiation dose. Their interpretation also depends on the examination protocol, beam quality, detector characteristics, and manufacturer-specific implementation [10] [13].

Despite the availability of standardized exposure indicators, their clinical interpretation and application remain challenging. Lewis et al. [11] reported that only approximately 50% of digital radiographs were within the manufacturer-recommended EI range and that 27% were overexposed, illustrating the potential for exposure creep

when exposure parameters are increased to suppress image noise. Johansen et al. [10] further reported that 71% of radiographers in Singapore and 78% in Norway demonstrated only an intermediate understanding of EI, while 46%–47% experienced difficulty distinguishing measured EI from EIT. Moreover, exposure-indicator behavior is affected by beam quality, field size, AEC response, generator performance, image-processing algorithms, and other system-dependent factors [13]. Nevertheless, appropriate DI-based feedback has been shown to improve exposure consistency, with previous studies reporting an increase in examinations within a targeted DI range from 42% to 81% [12][14].

Previous investigations have demonstrated that image noise, EI, and detector response are influenced by exposure conditions, but these parameters are frequently evaluated separately or using global image-quality measures [15][16]. Patient-based studies are additionally affected by variations in anatomical thickness, positioning, and pathology, which can complicate isolation of the specific effect of tube-voltage variation [17][18][19]. Although previous thoracic imaging studies have investigated the influence of tube voltage on CNR, sharpness, and overall image quality [8][20], relatively few studies have integrated standardized EI/DI metrics with region-specific noise and SNR measurements across anatomically distinct thoracic ROIs under controlled exposure conditions. Global noise assessment may therefore obscure local variations caused by differences in attenuation between structures such as the lung field, ribs, sternum, clavicle, and scapula. This creates a need for a controlled multi-ROI approach that evaluates detector-exposure indicators together with localized, objective image-quality responses.

Accordingly, this study aimed to evaluate the effects of tube-voltage variation from 50 to 100 kVp at a constant 8 mAs on EI, DI, image noise, and SNR in digital chest radiography using an anthropomorphic thorax phantom. Image noise and SNR were assessed independently in five anatomical ROIs: the clavicle, scapula, costae, sternum, and lung field. The novelty of this study lies in integrating kVp-dependent EI and DI behavior with anatomy-specific noise and SNR responses within a controlled multi-ROI framework. The study therefore provides: 1) quantitative characterization of the relationship between tube voltage and standardized detector-exposure indicators; 2) region-specific evaluation of noise and SNR across thoracic structures with different attenuation characteristics; and 3) an integrated framework for identifying a candidate detector-exposure condition that warrants further optimization using patient-dose measurements and diagnostic image-quality assessment. The dataset, data collection, data processing, and data analysis are covered in part II of this study. The results of the kVp variation toward noise, EI, and SNR are shown in Section III. The interpretation, comparison, and limitations of the results with those of other studies are covered in Section IV. The aims, key findings, and future work are rewritten in Section V, Conclusions.

II. Materials and Method

A. Experimental Setup & Equipment

The experimental study was conducted at the Radiology Laboratory of Universitas 'Aisyiyah Yogyakarta using a Samsung XGEO GF50 digital radiography (DR) system with a wireless flat-panel detector (S4335-W) and a whole-body anthropomorphic PBU-50 phantom (Kyoto Kagaku Co., Ltd., Japan). The phantom was selected because its attenuation characteristics are comparable to those of human soft tissue and bone, thereby providing stable and reproducible imaging conditions. Digital radiographic images were acquired at eleven kVp, ranging from 50 to 100 kVp in 5-kVp increments. Tube current and exposure time were maintained at 200 mA, 0.04 s (8 mAs). The Source-to-Image Distance (SID) was fixed at 180 cm, a 10:1 anti-scatter grid was used, and the collimation field was maintained at 35 × 43 cm. To evaluate experimental variability and acquisition repeatability, three independent radiographic acquisitions were performed at each tube-voltage setting. A total of 33 radiographic images were therefore included in the final analysis (11 kVp settings × 3 independent acquisitions).

The Samsung XGEO GF50 workstation's Digital Imaging and Communications in Medicine (DICOM) metadata provided the Exposure Index (EI) directly in accordance with the IEC 62494-1 standard. Using the RadiAnt DICOM Viewer ROI analysis tool, image assessed as the standard deviation of pixel values measured directly on the generated raw images inside five ROIs, including the costae, lung field, sternum, clavicle, and scapula. The ROI placement, as shown in Fig. 1, followed the methodology described by [21]. These ROIs were selected to represent anatomical structures with different attenuation characteristics, enabling a comprehensive evaluation of the effects of kVp variations on detector response and image quality.

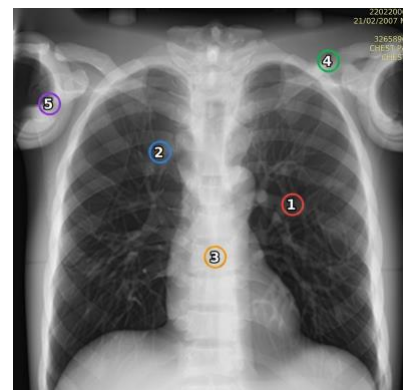


Fig. 1. Placement of five Regions of Interest (ROIs) for image noise measurement on the thorax phantom radiograph. ROI locations: (1) rib, (2) lung field, (3) sternum, (4) clavicle, and (5) scapula.

B. Image Acquisition & Experimental Procedure

Data collection was performed using a digital radiography acquisition protocol with a Samsung XGEO GF50 digital radiography (DR) system. The anthropomorphic PBU-50 thorax phantom was positioned upright against the bucky stand in a routine posterior-anterior (PA) projection, with the mid-coronal plane parallel to the image receptor, as shown in Fig. 2. A source-to-image distance (SID) of 180

cm and a 10:1 anti-scatter grid were used. Automatic exposure control (AEC) was not activated. The collimation field was adjusted to 35×43 cm to adequately cover the thoracic region of the phantom. Digital radiographic images were acquired at eleven tube voltage settings: 50, 55, 60, 65, 70, 75, 80, 85, 90, 95, and 100 kVp, with 5-kVp increments. The tube current and exposure time were kept constant at 200 mA and 0.04 s, respectively, corresponding to an exposure of 8 mAs.

All acquired images were stored in DICOM format to preserve the image data and associated examination metadata. Inspection of the original DICOM metadata confirmed that the images were classified as "FOR PRESENTATION" under the DICOM Presentation Intent Type attribute (0008,0068). The Exposure Index (EI), Target Exposure Index (EIT), and Deviation Index (DI) were obtained directly from the Digital Imaging and Communications in Medicine (DICOM) metadata generated by the Samsung XGEO GF50 digital radiography system. The corresponding standardized DICOM attributes were Exposure Index (0018,1411), Target Exposure Index (0018,1412), and Deviation Index (0018,1413). The EIT used in this study was 186, as recorded directly in the DICOM metadata for the selected chest PA examination protocol.



Fig. 2. Experimental setup and positioning of the anthropomorphic PBU-50 thorax phantom for digital radiography image acquisition.

The acquired digital radiographic images were processed to obtain the Exposure Index (EI) and image noise values for each kVp setting. The EI was extracted directly from the DICOM metadata generated by the Samsung XGEO GF50 workstation. According to IEC 62494-1[13], the Exposure Index measures the detector response to radiation within the relevant image region. The EI is calculated using the equipment-specific inverse calibration function, as shown in Eq. (1) [13], [22].

$$EI = c_0 \cdot g(V) \quad (1)$$

where EI is the Exposure Index, c_0 is a constant with a value of $100\mu\text{Gy}^{-1}$, $g(V)$ is the equipment-specific inverse calibration function, and V is the Value of Interest representing the central tendency of the original data within the relevant image region. In IEC 62494-1, the calibration of the Exposure Index is established by relating the EI to the image receptor air kerma under specified calibration conditions, as expressed in Eq. (2) [22].

$$EI = c_0 \cdot K_{CAL} \quad (2)$$

where K_{CAL} is the image receptor air kerma under the specified calibration conditions, expressed in μGy , and c_0 is a constant with a value of $100\mu\text{Gy}^{-1}$. This relationship represents the calibration basis of the Exposure Index, in which the EI is standardized according to the image receptor air kerma. In the present study, direct measurement of K_{CAL} was not performed; instead, the EI values were obtained directly from the DICOM metadata generated by the digital radiography system. Thus, the EI values used for analysis represent the system-generated values based on the manufacturer-specific calibration function. The Target Exposure Index (EIT) represents the reference Exposure Index corresponding to the intended detector exposure for a specific radiographic examination. The EIT was used as the reference value for evaluating whether the measured EI was below, close to, or above the intended detector exposure. The EIT value was obtained from the target exposure reference used by the digital radiography system for the examination. The Deviation Index (DI) was used to quantify the deviation of the measured EI from the EIT [13] and was calculated using Eq. (3) [22].

$$DI = 10 \log_{10} \left(\frac{EI}{EIT} \right) \quad (3)$$

where EIT stands for Target Exposure Index, EI for Measured Exposure Index, and DI for Deviation Index. The measured EI matches the intended exposure when the DI value is 0. Detector exposure below the target is indicated by a negative DI value, while exposure above the target is shown by a positive DI value. In this study, DI values between -1 and $+1$ were considered to represent exposure close to the target, while values below -1 and above $+1$ indicated detector exposure below target and detector exposure above target, respectively.

C. Data Processing

A predefined detector-exposure criterion was used to identify a candidate exposure condition rather than a definitive clinical optimum. A condition was considered target-conforming when the mean DI was within -1 to $+1$. Image noise and SNR were then evaluated jointly to characterize the corresponding objective image-quality response. Because patient radiation dose and diagnostic observer performance were not measured, the selected condition was interpreted as a candidate detector-exposure condition for further optimization rather than an optimal clinical chest-radiography technique. The mean pixel value was calculated for each anatomical ROI to represent the average pixel intensity within the selected region as expressed in Eq. (4) [23].

$$\bar{x} = \frac{1}{N} \sum_{i=1}^N x_i \quad (4)$$

where $\bar{x}$ is the mean pixel value within the ROI, x_i is the pixel value of the i th pixel, and N is the total number of pixels contained within the ROI. Image noise was quantified using the standard deviation (SD) of pixel intensity values measured within each Region of Interest (ROI). The SD was calculated using Eq. (5) [23].

$$SD = \sqrt{\frac{\sum_{i=1}^n (x_i - \bar{x})^2}{n-1}} \quad (5)$$

where SD represents the image noise expressed as the standard deviation, x_i is the intensity value of the i th pixel, $\bar{x}$ is the mean pixel intensity within the ROI, and n is the

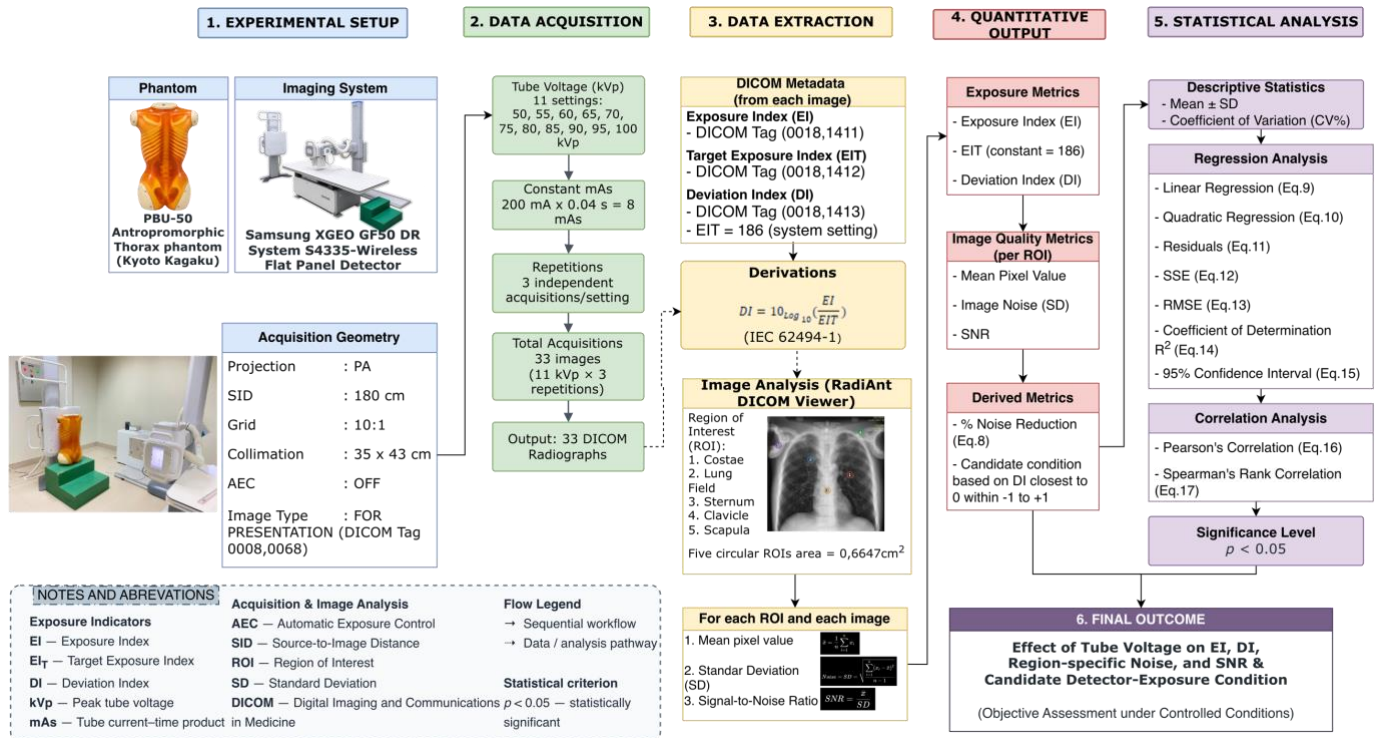


Fig. 3. Experimental and analytical workflow for evaluating the effects of tube-voltage variation on exposure index (EI), deviation index (DI), region-specific image noise, and signal-to-noise ratio (SNR), and for identifying a candidate detector-exposure condition under controlled experimental conditions.

total number of pixels in the ROI. A higher SD value indicates greater variation in pixel intensity and, consequently, a higher level of image noise. Image noise was evaluated using the Region of Interest (ROI) analysis tool by placing five ROIs using RadiAnt DICOM Viewer. Each ROI was positioned within a homogeneous anatomical region to minimize the influence of structural edges and anatomical boundaries on the standard deviation (SD) measurement. All ROIs were circular in shape and had an identical area of 0.6647 cm². The same ROI size, shape, and area were maintained across the three repeated acquisitions and all kVp settings. Signal-to-noise ratio was calculated separately for each anatomical ROI and each independent acquisition using the mean pixel value and corresponding SD obtained from the same ROI using Eq. (6).

$$SNR = \frac{\bar{x}}{SD} \quad (6)$$

where $\bar{x}$ represents the mean pixel value and SD represents the standard deviation of pixel values within the same ROI.

D. Statistical Analysis

Statistical analyses were performed using the complete repeated-acquisition dataset consisting of 33 observations. Descriptive statistics were calculated for EI, DI, image noise, and SNR. For each kVp setting, measurements from the three independent acquisitions were summarized as mean ± standard deviation. Repeatability was evaluated using the coefficient of variation (CV) Eq. (7) [24].

$$CV = \frac{SD}{Mean} \times 100\% \quad (7)$$

where SD and Mean represent the standard deviation and mean of the three independent acquisitions at the same

tube-voltage setting. CV was calculated separately for EI, image noise, and SNR. The results were presented in tables and graphical representations to illustrate the trends in detector exposure and image quality associated with variations in tube voltage. In digital radiography, image noise, represented by the standard deviation of pixel values within a homogeneous region, is a key indicator of image quality, with lower noise values indicating superior image quality [16]. The percentage reduction in image noise between 50 and 100 kVp was calculated separately for each anatomical ROI to quantify the relative change in noise across the investigated tube-voltage range. The calculation was performed using Eq. (8), where $Noise_{50}$ and $Noise_{100}$ represent the mean ROI-specific image-noise values obtained at 50 and 100 kVp, respectively.

$$Noise\ Reduction(\%) = \frac{Noise_{50} - Noise_{100}}{Noise_{50}} \times 100\% \quad (8)$$

Tube voltage was considered the primary experimental predictor because kVp was the parameter deliberately manipulated during image acquisition. Initial linear regression was performed to characterize the direction of association between kVp and each outcome. Scatterplots and regression diagnostics indicated nonlinear patterns in several outcomes; therefore, second-order polynomial regression was subsequently performed in Eq. (9) [25].

$$Y_i = \beta_0 + \beta_1(X_i) + \varepsilon_i \quad (9)$$

where Y_i is the observed outcome for the i^{th} observation, X_i represents tube voltage, β_0 is the intercept; β_1 represents the linear component, and ε_i represents the residual error for the i^{th} observation. Because visual inspection of the scatterplots and regression diagnostics

indicated nonlinear patterns in several outcomes, a second-order polynomial (quadratic) regression model was subsequently fitted to characterize potential curvature in the kVp-dependent responses. The quadratic regression model was expressed as Eq. (10) [25].

$$Y = \beta_0 + \beta_1(kVp) + \beta_2(kVp^2) + \varepsilon_i \quad (10)$$

where Y represents EI, DI, region-specific image noise, or SNR; β_0 is the intercept; β_1 represents the linear component; β_2 represents the quadratic component. The goodness-of-fit of the regression models was evaluated using regression residuals, the sum of squared errors (SSE), root mean square error (RMSE), and the coefficient of determination R^2 . The residual e_i was defined as the difference between the observed value Y_i and the corresponding value predicted by the regression model $\hat{Y}_i$, as expressed in Eq. (11) [25].

The overall magnitude of unexplained variation was quantified using SSE, calculated as the sum of the squared residuals according to Eq. (12) [25]. RMSE was subsequently calculated to express the average magnitude of the prediction error in the same unit as the corresponding outcome variable, as shown in Eq. (13) [25]. The coefficient of determination R^2 was used to quantify the proportion of variability in the observed outcome explained by the regression model, as defined in Eq. (14) [25]. Lower SSE and RMSE values indicate smaller prediction errors, whereas an R^2 value closer to 1 indicates a greater proportion of the observed variability explained by the model. These measures, together with residual inspection and the statistical significance of the quadratic term, were used to compare the linear and quadratic regression models.

$$e_i = Y_i - \hat{Y}_i \quad (11)$$

$$SSE = \sum_{i=1}^n (Y_i - \hat{Y}_i)^2 = \sum_{i=1}^n e_i^2 \quad (12)$$

$$RMSE = \sqrt{\frac{1}{n} \sum_{i=1}^n (Y_i - \hat{Y}_i)^2} \quad (13)$$

$$R^2 = 1 - \frac{\sum_{i=1}^n (Y_i - \hat{Y}_i)^2}{\sum_{i=1}^n (Y_i - \bar{Y})^2} \quad (14)$$

A 95% confidence interval (95% CI) was calculated for each regression coefficient (β_j) using Eq. (15) [25].

$$CI_{95\%}(\beta_j) = \hat{\beta}_j \pm t_{0.975, n-p} \times SE(\hat{\beta}_j) \quad (15)$$

where $(\hat{\beta}_j)$ is the estimated coefficient, $t_{\{0.975, n-p\}}$ is the critical value from the t-distribution with $n - p$ degrees of freedom, $SE(\hat{\beta}_j)$ is the standard error of the estimate, and n is the number of observations, and p is the number of estimated regression parameters, including the intercept. Linear associations between exposure index (EI), region-specific image noise, and signal-to-noise ratio (SNR) were evaluated using Pearson's correlation coefficient (r) Eq. (16) [26].

$$r = \frac{\sum_{i=1}^n (x_i - \bar{x})(y_i - \bar{y})}{\sqrt{\sum_{i=1}^n (x_i - \bar{x})^2 \sum_{i=1}^n (y_i - \bar{y})^2}} \quad (16)$$

Spearman's rank correlation was additionally used as a non-parametric sensitivity analysis to evaluate monotonic associations and reduce dependence on distributional assumptions. For data without tied ranks, Spearman's coefficient ρ_s can be expressed as Eq. (17) [26].

$$\rho_s = \frac{\sum_{i=1}^n (R_{x,i} - \bar{R}_x)(R_{y,i} - \bar{R}_y)}{\sqrt{\sum_{i=1}^n (R_{x,i} - \bar{R}_x)^2 \sum_{i=1}^n (R_{y,i} - \bar{R}_y)^2}} \quad (17)$$

where $d_i = R_{x,i} - R_{y,i}$ is the difference between the ranks of paired observations. All statistical hypothesis tests were two-tailed, with a significance level of $\alpha = 0.05$. Regression coefficients were reported with 95% confidence intervals and corresponding p-values. Linear and quadratic models were compared using the coefficient of determination R^2 , RMSE, residual inspection, and the statistical significance of the quadratic term. Pearson correlation was used to characterize linear associations, whereas Spearman rank correlation was used as a complementary sensitivity analysis for monotonic relationships. All statistical analyses were performed using IBM SPSS Statistics version 26.

The complete experimental and analytical workflow of this study is summarized in Fig. 3. The workflow comprised six sequential stages: experimental setup using the anthropomorphic thorax phantom and DR system; image acquisition at eleven tube-voltage settings with three independent repetitions; extraction of EI, EIT, and DI from DICOM metadata; ROI-based calculation of mean pixel value, image noise, and SNR; derivation of percentage noise reduction and the candidate detector-exposure condition; and statistical evaluation using descriptive, regression, and correlation analyses. This structured workflow was designed to ensure consistent data acquisition and analysis across all investigated kVp settings and anatomical ROIs.

III. Results

A. Repeatability of Repeated Acquisitions

Three independent radiographic acquisitions were performed at each of the eleven tube-voltage settings, resulting in 33 acquisitions. Exposure Index (EI), Deviation Index (DI), image noise, and signal-to-noise ratio (SNR) were evaluated for each acquisition. The repeated measurements demonstrated low within-condition variability. The mean coefficients of variation (CVs) for image noise were 0.80%, 0.81%, 0.47%, 1.02%, and 0.85% for the clavicle, scapula, costae, sternum, and lung field, respectively, while the mean CV for EI was 0.55%. The corresponding mean CVs for SNR were 1.17%, 1.45%, 2.24%, 1.04%, and 2.20%, respectively. These results indicate good measurement repeatability across the repeated acquisitions.

B. Effect of Tube Voltage on Exposure Index and Deviation Index

The relationship between kilovoltage peak (kVp), EI, and DI is presented in Fig. 4. Overall, the EI increased progressively with increasing tube voltage, indicating greater detector exposure at higher kVp settings. The mean EI and DI values obtained from three independent acquisitions at each tube-voltage setting are presented in Table 1. Mean EI increased from 73.95 ± 0.28 at 50 kVp to 1980.91 ± 16.78 at 100 kVp. The corresponding DI increased from -4.01 ± 0.02 to $+10.28 \pm 0.04$.

The DI was used to evaluate the deviation of the measured EI. A DI value close to 0 indicates that the detector exposure is closest to the target exposure,

Table 1. Mean Exposure Index (EI) and Deviation Index (DI) obtained from three independent acquisitions at each tube-voltage setting

kVp	EI, mean $\pm$ SD	DI, mean $\pm$ SD
50	73.95 $\pm$ 0.28	-4.01 $\pm$ 0.02
55	131.55 $\pm$ 0.00	-1.50 $\pm$ 0.00
60	216.02 $\pm$ 1.68	+0.65 $\pm$ 0.03
65	320.87 $\pm$ 1.95	+2.37 $\pm$ 0.03
70	461.17 $\pm$ 4.23	+3.94 $\pm$ 0.04
75	629.13 $\pm$ 5.60	+5.29 $\pm$ 0.04
80	826.54 $\pm$ 1.22	+6.48 $\pm$ 0.01
85	1064.56 $\pm$ 2.22	+7.58 $\pm$ 0.01
90	1338.02 $\pm$ 4.08	+8.57 $\pm$ 0.01
95	1653.23 $\pm$ 17.00	+9.49 $\pm$ 0.05
100	1980.91 $\pm$ 16.78	+10.28 $\pm$ 0.04

whereas negative and positive values indicate detector exposure below and above the target, respectively. At lower tube-voltage settings, mean DI values were -4.01 at 50 kVp and -1.50 at 55 kVp. At 60 kVp, the mean DI reached +0.65 $\pm$ 0.03, representing the value closest to zero among the investigated conditions. At 65 kVp, DI increased to +2.37 and continued to increase progressively to +10.28 at 100 kVp. According to the predefined criterion of $-1 \leq DI \leq +1$, 60 kVp was the only investigated setting within the target interval.

The relationship between kVp and EI was strongly nonlinear. A quadratic regression model provided an excellent fit ($R^2 = 0.9998$). The quadratic coefficient was statistically significant ($\beta_2 = 0.6426$; 95% CI: 0.6269–0.6583; $p < 0.001$). Similarly, DI demonstrated a strong nonlinear relationship with kVp ($R^2 = 0.9987$), and the quadratic coefficient was also significant ($\beta_2 = -0.003435$; 95% CI: -0.003704 to -0.003165; $p < 0.001$).

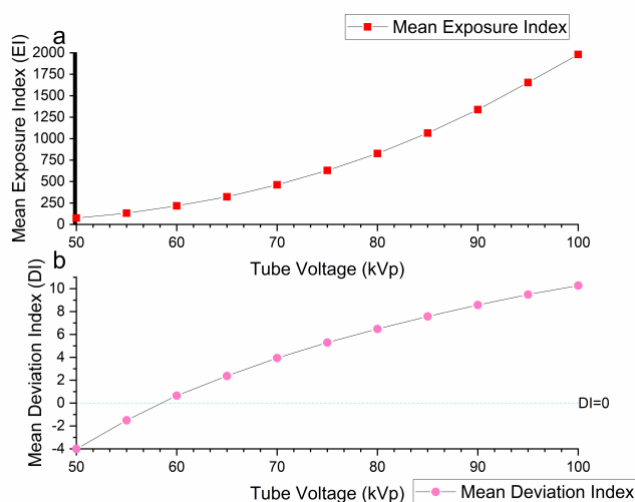


Fig. 4. Changes in Exposure Index and Deviation Index as a function of tube voltage in digital chest radiography. (a) Exposure Index and (b) Deviation Index

C. Effect of Tube Voltage on Region-Specific Image Noise

Image noise, expressed as the standard deviation of pixel values within each ROI, generally decreased with increasing tube voltage; however, the magnitude and pattern of reduction differed among anatomical regions. The standard deviation (SD) measured within five anatomical ROIs, clavicle, lung field, sternum, costae, and scapula. The image noise values generally showed a decreasing tendency with increasing tube voltage, although fluctuations were observed across kVp settings and among the anatomical regions, as shown in Fig. 5. Between 50 and 100 kVp, mean noise decreased from 866.24 to 601.19 in the clavicle, corresponding to a 30.6% reduction. In the costal region, noise decreased from 798.88 to 651.55, representing an 18.4% reduction. The sternum demonstrated the largest reduction, from 418.93 at 50 kVp to 172.59 at 100 kVp, corresponding to a 58.8% decrease. Lung-field noise decreased from 498.80 to 398.06, representing a 20.2% reduction. The scapula showed the smallest overall change, decreasing from 870.34 at 50 kVp to 810.05 at 100 kVp, corresponding to a 6.9% reduction.

The noise response was not uniformly linear. Quadratic regression provided a better fit than simple linear regression for all anatomical regions. The coefficients of determination were $R^2 = 0.968$ for the clavicle, $R^2 = 0.918$ for the scapula, $R^2 = 0.976$ for the costae, $R^2 = 0.966$ for the sternum, and $R^2 = 0.851$ for the lung field. All quadratic models were statistically significant ($p < 0.001$ for the overall models), as shown in Table 2. The quadratic terms were statistically significant for the clavicle ($\beta_2 = +0.1289$; 95% CI: 0.0973–0.1604; $p < 0.001$), scapula ($\beta_2 = +0.1150$; 95% CI: 0.0983–0.1317; $p < 0.001$), costae ($\beta_2 = +0.0787$; 95% CI: 0.0670–0.0904; $p < 0.001$), sternum ($\beta_2 = -0.0366$; 95% CI: -0.0639 to -0.0094; $p = 0.010$), and lung field ($\beta_2 = +0.0983$; 95% CI: 0.0772–0.1194; $p < 0.001$). These results demonstrate that the effect of increasing kVp on measured pixel-value variability differed according to anatomical ROI.

D. Effect of Tube Voltage on Signal-to-Noise Ratio

SNR also demonstrated a region-dependent response to increasing tube voltage. Mean SNR increased from 10.27 at 50 kVp to a maximum of 16.19 at 90 kVp in the clavicle, followed by a slight reduction to 15.33 at 100 kVp. In the scapula, SNR increased from 7.29 at 50 kVp to a maximum of 9.49 at 90 kVp and subsequently decreased to 8.49 at 100 kVp. The costal ROI demonstrated an increase from 4.07 at 50 kVp to a maximum of 5.25 at 85 kVp, followed by a decrease to 4.31 at 100 kVp. Similarly, lung-field SNR increased from 5.88 at 50 kVp to a maximum of 9.29 at 85 kVp before decreasing to 7.80 at 100 kVp. In contrast, the sternum demonstrated a continuous increase in SNR, from 34.76 at 50 kVp to 80.54 at 100 kVp.

Quadratic regression provided a substantially better representation of the SNR response than a simple linear model. The quadratic models explained 95.4%, 91.9%, 84.0%, 98.3%, and 94.2% of SNR variability in the clavicle, scapula, costae, sternum, and lung field,

Table 2. Quadratic regression analysis of tube voltage and region-specific image noise

ROI	β_0	β_1	β_2 (95% CI)	R ²	p for β_2
Clavicle	1859.35	-25.590	+0.1289 (0.0973–0.1604)	0.968	<0.001
Scapula	1517.50	-18.586	+0.1150 (0.0983–0.1317)	0.918	<0.001
Costae	1308.60	-14.375	+0.0787 (0.0670–0.0904)	0.976	<0.001
Sternum	512.01	+0.067	-0.0366 (-0.0639 to -0.0094)	0.966	0.010
Lung field	1031.81	-16.050	+0.0983 (0.0772–0.1194)	0.851	<0.001

respectively. As shown in Table 3 the quadratic coefficients were significant for all ROIs: clavicle $\beta_2 = -0.00246$ (95% CI: -0.00324 to -0.00168; $p < 0.001$), scapula $\beta_2 = -0.00201$ (95% CI: -0.00235 to -0.00166; $p < 0.001$), costae $\beta_2 = -0.00144$ (95% CI: -0.00170 to -0.00117; $p < 0.001$), sternum $\beta_2 = +0.01956$ (95% CI: 0.01598–0.02314; $p < 0.001$), and lung field $\beta_2 = -0.00324$ (95% CI: -0.00366 to -0.00281; $p < 0.001$). Thus, except for the sternum, increasing tube voltage did not produce an unlimited increase in SNR. Instead, the clavicle, scapula, costae, and lung field demonstrated an increase followed by a plateau or reduction at higher tube voltages.

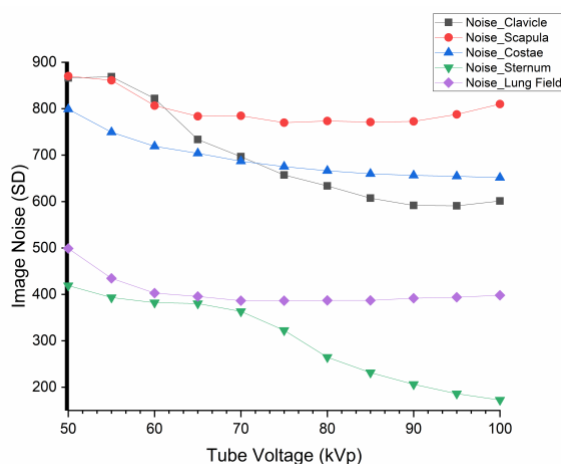


Fig. 5. Effect of tube voltage on image noise across different anatomical regions in digital chest radiography

E. Relationship between Image Noise, Exposure Index (EI), and SNR

Exploratory correlation analysis demonstrated negative associations between EI and image noise in all anatomical ROIs. Pearson correlation coefficients were -0.856 for the clavicle, -0.426 for the scapula, -0.795 for the costae, -0.972 for the sternum, and -0.467 for the lung field. All correlations were statistically significant ($p < 0.05$). Spearman analysis similarly demonstrated significant negative monotonic relationships between EI and image noise, with p values of -0.949, -0.485, -0.990, -0.993, and -0.481 for the clavicle, scapula, costae, sternum, and lung field, respectively (all $p < 0.01$). EI was positively associated with SNR in the clavicle ($r = 0.858$, $p < 0.001$), scapula ($r = 0.575$, $p < 0.001$), sternum ($r = 0.990$, $p < 0.001$), and lung field ($r = 0.511$, $p = 0.002$). The Pearson relationship between EI and costal SNR was

not statistically significant ($r = 0.237$, $p = 0.183$), consistent with the non-monotonic SNR response observed in this region. Because EI increased systematically as a function of tube voltage, these correlations were interpreted as secondary associations rather than evidence of an independent causal relationship between EI and image noise or SNR.

The integrated results demonstrated a trade-off between detector exposure and objective image-quality metrics. At 60 kVp, EI was 216.02 ± 1.68 , and DI was $+0.65 \pm 0.03$, representing the only investigated condition within the predefined DI interval of -1 to +1. At this setting, SNR was higher than at 50 kVp in all five anatomical regions. Increasing tube voltage beyond 60 kVp generally resulted in further reductions in image noise and improvements in SNR. However, these improvements were accompanied by progressively increasing EI and strongly positive DI values. For example, the highest SNR values in the clavicle and scapula occurred at 90 kVp, where DI was +8.57, while maximum costal and lung-field SNR occurred at 85 kVp, where DI was +7.58. Therefore, the tube-voltage conditions producing the highest SNR did not correspond to detector exposure closest to the target. Among the investigated conditions, 60 kVp represented the setting that most closely matched the predefined detector-exposure target while providing higher SNR than the lower-kVp conditions. This finding should not be interpreted as evidence that 60 kVp represents the optimal clinical chest-radiography technique because patient radiation dose and diagnostic observer performance were not evaluated.

IV. Discussion

A. Effect of Tube Voltage on Detector Exposure

The repeated-acquisition experiment demonstrated a substantial and systematic increase in detector exposure with increasing tube voltage under constant 8-mAs conditions. Mean EI increased from 73.95 at 50 kVp to 1980.91 at 100 kVp, while DI shifted from -4.01 to +10.28. The relationship was better represented by a quadratic model ($R^2 = 0.9998$ for EI and $R^2 = 0.9987$ for DI) than by a simple linear model, indicating that the detector response increased progressively across the investigated tube-voltage range. Among the investigated conditions, 60 kVp produced an EI of 216.02 ± 1.68 and a DI of $+0.65 \pm 0.03$, and it was the only condition within the predefined $-1 \leq DI \leq +1$ interval. Importantly, this finding represents changes in detector exposure and should not be interpreted as an equivalent change in patient radiation dose.

Table 3. Quadratic regression analysis of tube voltage and region-specific signal-to-noise ratio

ROI	β_0	β_1	β_2 (95% CI)	R^2	p for β_2
Clavicle	-9.189	+0.4977	-0.00246 (-0.00324 to -0.00168)	0.954	<0.001
Scapula	-4.650	+0.3347	-0.00201 (-0.00235 to -0.00166)	0.919	<0.001
Costae	-3.741	+0.2264	-0.00144 (-0.00170 to -0.00117)	0.840	<0.001
Sternum	+84.336	-1.9642	+0.01956 (0.01598-0.02314)	0.983	<0.001
Lung field	-12.607	+0.5311	-0.00324 (-0.00366 to -0.00281)	0.942	<0.001

The increase in EI is consistent with the role of EI as an indicator of detector exposure, which is influenced by exposure parameters and the amount of radiation incident on the detector [27] [28]. The kVp dependence of EI observed in this study is consistent with Tsalaftoutas et al. [13], who demonstrated that EI increased with increasing tube voltage even when incident air kerma (IAK) at the detector was maintained constant. Their experiments used 70, 100, and 120 kV and further demonstrated that EI was influenced by imaged anatomy, examination protocol, field size, grid configuration, and manufacturer-specific characteristics. Consequently, the EI of 216.02 observed at 60 kVp in the present study should be regarded as specific to the Samsung XGEO GF50 system and the investigated chest protocol rather than as a universal target EI for chest radiography. This interpretation is also supported by Tanaka et al. [29], who found that, under simulated chest-radiography conditions, EI at 120 kV was significantly higher than at 70 kV when incident detector doses were identical, showing that EI can vary with beam quality independently of incident detector dose.

The clinical interpretation of DI requires consideration of the target exposure index and examination-specific conditions. Dave et al. [22], in the AAPM Task Group 232 report, demonstrated substantial variability in clinical DI distributions, with standard deviations ranging from 1.3 to 3.6 and fewer than 50% of radiographs falling within the $-1 \leq DI \leq +1$ interval. Similarly, Creeden and Curtis [30] used a manufacturer-recommended optimal DI range of -3 to $+2$ in a large clinical dataset and demonstrated that systematic DI monitoring could support optimization of exposure parameters and reduction of default mAs. Park et al. [17] further demonstrated that DI interpretation depends strongly on appropriate EIT determination, reporting differences of up to 2.03 in DI when clinical EIT was not adjusted according to changes in diagnostic reference levels. Collectively, these studies indicate that DI should be interpreted as an examination- and system-specific exposure feedback indicator rather than according to a universal numerical acceptance range. Therefore, the $-1 \leq DI \leq +1$ interval applied in the present experimental study represents a predefined criterion for identifying detector exposure closest to the EIT, rather than a universal clinical acceptance or repeat threshold.

B. Effect of Tube Voltage on Region-Specific Image Noise

Increasing tube voltage was generally associated with reduced region-specific image noise, but the magnitude of the reduction differed substantially among anatomical

regions. From 50 to 100 kVp, noise decreased by 30.6% in the clavicle, 18.4% in the costae, 58.8% in the sternum, and 20.2% in the lung field, whereas the scapula showed only a 6.9% net reduction. The strongest fitted relationship was observed in the costae (quadratic $R^2 = 0.976$), followed by the clavicle ($R^2 = 0.968$) and sternum ($R^2 = 0.966$), indicating that the response of image noise to increasing tube voltage was region-dependent rather than uniform across the thoracic image. The particularly large reduction in sternal noise, compared with the relatively small change in the scapula, demonstrates that a single global noise measurement could conceal substantial spatial differences in objective image quality. The reduction in noise can be explained primarily by changes in photon statistics. Increasing tube voltage increases the energy and penetrating capability of the X-ray beam and can increase the number of photons reaching the detector under the experimental conditions [31], [32].

These findings are consistent with previous studies demonstrating that radiographic noise and image quality are influenced by both beam quality and anatomical region. Tanaka et al. [29] demonstrated that radiographic noise could not be predicted solely from EI; under simulated chest-radiography conditions, similar EI values produced different Wiener-spectrum characteristics when tube voltage was varied, indicating that beam quality independently influences radiographic noise. Chukorn et al. [33] further demonstrated substantial regional variation in a digital chest phantom: at the same exposure condition of 100 kVp and 0.4 mAs, SNR values were 45.03 at the fourth rib, 36.54 at the eighth thoracic vertebra, and 18.47 at the left diaphragm. These studies support the present finding that increasing tube voltage affects image noise non-uniformly across the thoracic image. The substantially greater noise reduction observed in the sternum (58.8%) than in the scapula (6.9%) therefore highlights the importance of multiple anatomically defined ROIs rather than relying on a single global noise measurement. This variation suggests that image noise is influenced not only by tube voltage but also by anatomical attenuation, local signal characteristics, scatter, and ROI location.

C. Integrated Interpretation of EI, DI, Noise and SNR

The integrated analysis of EI, DI, noise, and SNR indicates that the condition producing the highest SNR did not coincide with the detector exposure closest to the predefined target. At 60 kVp, EI was 216.02 ± 1.68 and DI was $+0.65 \pm 0.03$, representing the only condition within $-1 \leq DI \leq +1$, with higher SNR than at 50 kVp across all five ROIs. However, further increases in kVp generally

Table 4. Comparison of previous studies on exposure indicators and image quality in digital radiography

Author(s), Year	Method	Research Scope	Main Findings
Tsalafoutas et al., 2026 [13]	Experimental evaluation using two DR systems with variations in kVp, mAs, AEC, grid, filtration, field size, and anatomy	Factors influencing EI and EIT under different acquisition and system conditions	EI was influenced by tube voltage, anatomy, acquisition parameters, and manufacturer, demonstrating that EI is not determined solely by detector exposure
Tanaka et al., 2026 [29]	Simulated chest and abdominal radiography; EI and radiographic noise evaluated under different tube-voltage and grid conditions	Relationship between EI and radiographic noise under different beam-quality conditions	Similar EI values did not necessarily correspond to equivalent radiographic-noise characteristics
Dave et al., 2018 [22]	Multicenter clinical evaluation of standardized EI and DI across different radiographic examinations	Clinical EI/DI distributions and establishment of reference and action levels	DI distributions varied among examinations; fewer than 50% of images were within the previously proposed $-1 \leq DI \leq +1$ interval
Creeden and Curtis, 2020 [30]	Clinical exposure-log analysis with iterative modification of default exposure factors based on DI	DI-based optimization of radiographic exposure protocols	DI monitoring reduced examinations outside the manufacturer-recommended range and enabled reduction of default exposure factors
Park et al., 2022 [17]	Clinical evaluation of EI and entrance surface dose for extremity radiography with EIT determined using DRLs	Relationship among EI, EIT, DI, and diagnostic reference levels	EI was related to detector exposure and beam quality; inappropriate EIT adjustment resulted in DI differences of up to 2.03
Chukorn et al., 2022 [33]	Chest-phantom study evaluating radiation dose, qualitative image quality, SNR, and CNR at selected anatomical regions	Exposure optimization and region-specific image-quality assessment in digital chest radiography	Image quality varied among anatomical regions; reported SNR values were 45.03 at the fourth rib, 36.54 at T8, and 18.47 at the left diaphragm
Present study, 2026	PBU-50 anthropomorphic thorax phantom; 50–100 kVp in 5-kVp increments at constant 8 mAs; EI, DI, noise, and SNR evaluated across five thoracic ROIs	Integrated evaluation of detector-exposure indicators and region-specific objective image quality under controlled kVp variation	EI increased nonlinearly with kVp, while noise and SNR responses were ROI-dependent. Noise reduction ranged from 6.9% to 58.8%; 60 kVp yielded EI = 216.02 ± 1.68 and DI = $+0.65 \pm 0.03$ and was the only investigated condition within $-1 \leq DI \leq +1$

reduced noise and increased SNR; maximum clavicular and scapular SNR occurred at 90 kVp (DI +8.57), whereas maximum costal and lung-field SNR occurred at 85 kVp (DI +7.58). Thus, maximizing SNR alone would favor detector exposures substantially above the system-defined EIT, indicating that SNR maximization and detector-exposure conformity represent different optimization objectives.

To provide a structured comparison with previous investigations, the principal methodological characteristics and findings of relevant studies are summarized in Table 4. This interpretation is supported by quantitative findings from previous studies. Tsalafoutas et al. [13] demonstrated EI dependence on tube voltage at 70, 100, and 120 kV even when incident air kerma at the detector was maintained constant, while Tanaka et al. [29] reported significantly higher EI at 120 kV than at 70 kV under equivalent incident-detector-dose conditions. Dave et al. [22] reported clinical DI standard deviations of 1.3–3.6, with fewer than 50% of radiographs within $-1 \leq DI \leq +1$, whereas Creeden and Curtis [30] applied a manufacturer-recommended DI range of -3 to

+2 for clinical exposure optimization. Park et al. [17] further reported DI differences of up to 2.03 when EIT was not appropriately adjusted to changes in diagnostic reference levels. Collectively, these findings support interpreting the present DI of +0.65 at 60 kVp as the condition closest to the system- and examination-specific EIT rather than as a universal clinical acceptance threshold.

The region-specific findings further demonstrate why EI and DI should be interpreted together with localized image-quality metrics. Chukorn et al. [33] reported SNR values of 45.03 at the fourth rib, 36.54 at the eighth thoracic vertebra, and 18.47 at the left diaphragm under the same exposure condition of 100 kVp and 0.4 mAs, demonstrating substantial anatomical variation in SNR. Similarly, the present study showed different SNR maximal among ROIs, while noise reductions from 50 to 100 kVp ranged from only 6.9% in the scapula to 58.8% in the sternum. These findings indicate that changes in detector exposure do not produce spatially uniform image-quality responses and that a single global metric may obscure important regional differences. The multi-

ROI approach therefore provides complementary information for evaluating the effects of tube-voltage variation across thoracic structures with different attenuation characteristics.

The observed EI–noise and EI–SNR correlations should nevertheless be interpreted cautiously because EI and image-quality metrics changed simultaneously as functions of kVp. Previous studies [29] [34] [35] similarly indicate that EI alone does not fully characterize radiographic noise because beam quality and acquisition conditions independently influence detector response. Accordingly, kVp was considered the primary experimental predictor, whereas EI–noise and EI–SNR relationships were treated as secondary associations rather than independent causal effects. Among the investigated conditions, 60 kVp therefore represents the setting closest to the predefined detector-exposure target while providing improved SNR relative to the lower-kVp condition. However, because patient dose, CNR, contrast-detail performance, and diagnostic observer assessment were not evaluated, this condition should be regarded as a candidate for further investigation rather than a definitive clinical optimum.

D. Methodological Implications and Limitations

Three independent acquisitions at each kVp produced low coefficients of variation for EI, noise, and SNR, indicating good repeatability and reducing the likelihood that the observed trends were caused solely by stochastic acquisition variability. The repeated-acquisition design therefore addresses an important limitation of the preliminary dataset. However, the experiment was performed using a single anthropomorphic phantom, one DR system, and constant 8 mAs, so the results characterize the combined effects of changing beam quality and detector exposure rather than an exposure-compensated high-kVp technique. From a practical perspective, the multi-ROI approach may complement EI/DI-based quality assurance by identifying local image-quality responses that are not represented by detector-exposure indicators alone. The marked difference in noise reduction between the sternum (58.8%) and scapula (6.9%) demonstrates that changes in exposure do not produce uniform image-quality responses across thoracic regions. Consequently, radiographers should interpret EI and DI together with region-specific image-quality characteristics when evaluating exposure protocols, rather than increasing exposure solely to maximize SNR. Incorporating anatomically defined ROIs into protocol evaluation may therefore provide an objective framework for comparing exposure settings, identifying region-specific image-quality limitations, and supporting quality-assurance decisions while reducing the risk of unnecessary detector exposure.

The EI and DI are useful exposure-monitoring tools, but they cannot replace patient-dose measurements. The present study did not measure entrance-surface dose (ESD), incident air kerma (IAK), effective dose, CNR, contrast-detail performance, MTF/NPS, or diagnostic observer performance; therefore, the findings should not be interpreted as establishing a definitive clinical optimization protocol. Future studies should evaluate multiple phantom sizes or patient-equivalent thicknesses

and additional DR systems while integrating EI/DI with patient-dose measurements, CNR, spatial-resolution and noise-power-spectrum analyses, and blinded diagnostic-quality assessment. Such investigations are required to determine whether the detector-exposure and region-specific image-quality relationships observed under controlled phantom conditions can be translated into clinically applicable chest-radiography optimization strategies.

V. Conclusion

This study aimed to evaluate the effects of tube-voltage variation on Exposure Index (EI), Deviation Index (DI), image noise, and Signal-to-Noise Ratio (SNR) across different anatomical regions in digital chest radiography and to identify an exposure condition that warrants further investigation for optimization. The findings demonstrated that increasing kVp produced a nonlinear increase in EI and generally reduced image noise, although the magnitude of noise and SNR responses differed among anatomical ROIs. Among the investigated conditions, 60 kVp produced an EI of 216.02 ± 1.68 and DI of $+0.65 \pm 0.03$, representing the condition closest to the predefined detector-exposure target while providing higher SNR than at 50 kVp across all five ROIs. However, the highest SNR values occurred at higher tube voltages with substantially positive DI values, demonstrating a trade-off between conformity to the target detector exposure and maximization of region-specific SNR. These findings support the use of multi-ROI assessment together with EI and DI to characterize exposure-dependent image-quality responses, rather than relying on a single global image-quality or detector-exposure indicator.

The 60-kVp condition should be considered a candidate for further investigation rather than a definitive clinical optimum because EI and DI indicate detector exposure, not patient radiation dose. The present findings are also limited to one anthropomorphic phantom, one DR system, and constant 8-mAs acquisition conditions. Future studies should incorporate direct patient-dose measurements, including ESD or IAK, multiple phantom sizes or patient-equivalent thicknesses, additional DR systems, CNR, MTF/NPS and contrast-detail assessments, and blinded diagnostic-quality evaluation. These investigations are necessary to establish the generalizability and clinical applicability of the multi-ROI framework for chest-radiography optimization.

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Author Contributions

The study was developed and designed by Jihan Ana Rahima, who also carried out the experiments, gathered and processed data, carried out statistical analysis, analyzed the findings, and wrote the first draft of the publication. Anisa Nur Istiqomah supervised the research, contributed to the study design and methodology, interpreted the findings, and critically reviewed and revised the manuscript. Muhamad Fa'ik rigorously examined the work for significant intellectual substance, contributed to the interpretation of the results, and offered crucial scientific advice. Each author agreed to be responsible for all elements of the work, evaluated and approved the final paper, and took part in the discussion of the findings.

Ethical Approval

Human subjects, animals, therapeutic procedures, human biological materials, and the gathering of personally identifiable information were all excluded from this study. An anthropomorphic thorax phantom (PBU-50) was used exclusively for all experimental methods in the Radiology Laboratory of Universitas 'Aisiyiah Yogyakarta under carefully monitored laboratory conditions. Therefore, this study did not require formal ethical approval or informed consent.

Consent for Publication

Not applicable

Competing Interests

The authors affirm that they have no competing interests, either financial or non-financial, that could have affected the way this study was conducted, analyzed, interpreted, or reported.

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