

Analysis of Image Quality, Anatomical Information, and Dose Consistency Across Variations in kV Exposure Factors in Portable X-ray

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Abstract Tuberculosis (TB) remains a major global health challenge, with Indonesia ranking second worldwide in new cases. Chest radiography using portable X-ray equipment is vital for screening in remote areas while adhering to the ALARA principle. This study employed an experimental design using an anthropomorphic thorax phantom (Kyoto Kagaku PBU-50) with two portable X-ray modalities featuring different detector technologies (CsI vs. Gd₂O₂S). Exposures were performed at 50, 60, and 70 kV with mAs variations of 0.5, 1.0, 1.6, 2.0, and 2.5 mAs. Image quality was quantified using Signal-to-Noise Ratio (SNR) and Contrast-to-Noise Ratio (CNR) across four anatomical regions (ribs, lungs, hilum, and cardiac), while radiation dose was measured using a Raysafe X2 survey meter. Two-way repeated measures ANOVA was employed for statistical analysis. Results indicated that kV and mAs significantly influenced all image quality metrics ($p < 0.001$). Modality 1 produced higher SNR and CNR for most organs, while Modality 2 outperformed it for soft-tissue visualization at higher settings. Non-monotonic detector responses were observed at specific exposure combinations, highlighting the complexity of digital detector behavior. Optimal settings were 70 kV/1.6–2.0 mAs for Modality 1 and 70 kV/1.6 mAs for Modality 2, with doses of 0.027–0.034 mGy and 0.022 mGy, respectively. All doses remained below the national Diagnostic Reference Level (0.4 mGy). Dose-quality trade-off analysis revealed diminishing returns at higher mAs settings. Within the investigated 50–70 kV range, 70 kV represents the optimal exposure factor for portable chest phantom imaging. This study uses a phantom without simulated TB lesions and measures incident air kerma rather than patient entrance dose. Future research should incorporate clinical validation, lesion-containing phantoms, observer performance studies, and patient-size stratification to standardize protocols across diverse clinical settings.

Keywords Chest Radiography; Exposure Factors; Image Quality; Radiation Dose; Portable X-Ray.

1. Introduction

Pulmonary tuberculosis remains a major global public health challenge. According to the World Health Organization (WHO), 2023 recorded approximately 10.6 million new tuberculosis cases and 1.6 million deaths globally. Indonesia ranks second worldwide in estimated new TB cases, with 1,060,000 cases (385 per 100,000 residents) and 134,000 deaths reported annually [1]. These figures represent a significant increase from 724,309 new cases in 2022 to 792,404 cases in 2023 [2]. Central Java Province faces serious challenges in TB control. According to Central Java Governor Regulation No. 27 of 2024, reported TB cases in 2023 reached 87,074, up from an estimated 73,856 (118%), exceeding

the national target of 90% [3]. Chest radiology examinations play a crucial role in screening, diagnosis, and monitoring of pulmonary TB treatment, as characteristic radiological manifestations (infiltrates, cavities, fibrosis, and miliary patterns) serve as important diagnostic clues [4]. A study at Dr. Kariadi General Hospital showed that among 82 MDR-TB patients, 62.2% had very advanced pulmonary lesions, with consolidation (98.8%), fibrosis (73.2%), and cavities (63.4%) as the most common findings [5]. In digital radiography (DR), image quality depends strongly on exposure parameters particularly tube voltage (kV) and tube current-time (mAs) [6], [7]. Selecting appropriate parameters presents a unique challenge because thoracic anatomy exhibits a wide range of densities (soft tissues, ribs, vertebrae, and

air-filled lungs) with significantly different attenuation coefficients. The ALARA (As Low As Reasonably Achievable) principle requires the lowest possible radiation dose without compromising diagnostic quality [8], [9], [10].

Previous studies have investigated exposure optimization in digital chest radiography [11]. Uffmann and Schaefer-Prokop (2009) reported that SNR values exceeding 100 in lung regions are achievable with doses below 0.1 mGy [12]. Yoon et al. (2021) demonstrated that optimal exposure indices vary significantly based on patient body thickness and detector technology [13]. Cha et al. (2025) showed that automatic exposure control (AEC) in portable chest radiography can reduce dose area product while maintaining acceptable image quality [14]. Samei et al. (2019) compared new-generation and older-generation detectors and found that new detectors produced better image quality at baseline dose but degraded at 40% dose reduction [15]. Prayoga et al. (2022) investigated exposure factor effects on chest phantom image quality, reporting SNR improvement with increased kV and mAs [16]. Elshami et al. (2022) examined high-kVp techniques for neonatal chest radiography, demonstrating dose reduction potential [17]. Dave et al. (2018) provided comprehensive guidance on digital radiography exposure indicators and deviation indices [18].

However, research specifically addressing exposure factor optimization for portable chest radiography in the Indonesian context remains limited. Previous studies at Dr. Kariadi General Hospital focused on lesion characteristics in MDR-TB patients [5], while studies at Semarang Regency General Hospital emphasized examination procedures rather than exposure optimization [4], [19]. No previous study has systematically analyzed how variations in kV exposure factors (50–70 kV range) affect image quality and dose consistency across different portable X-ray detector systems. This study addresses the following research gap: systematic optimization of exposure factors (kV and mAs) for portable chest radiography across two different detector technologies, using quantitative image quality metrics (SNR, CNR, EI) and radiation dose measurements to identify optimal settings that balance image quality with dose reduction. The specific contributions of this study are: (1) quantitative comparison of two portable X-ray modalities with different detector technologies (CsI vs. $\text{Gd}_2\text{O}_2\text{S}$ scintillators), (2) identification of non-monotonic detector responses to exposure changes, (3) establishment of modality-specific optimal exposure protocols, and (4) demonstration of dose-quality optimization within ALARA constraints.

The study objective is to analyze image quality, anatomical information, and dose consistency across variations in exposure factors (50–70 kV range) using two portable X-ray modalities, providing evidence-based recommendations for TB chest screening that balance diagnostic quality with radiation dose optimization.

II. Materials and Method

A. Study Design

This study employed an experimental design with a quantitative approach. The research was conducted at the Radiology Laboratory, Department of Radiodiagnostic and Radiotherapy Engineering, Semarang Public Health Polytechnic, from April to June 2026.

B. Equipment

The equipment used included: (1) Portable X-ray Machine Modality 1 (XD2000, operating range: 50–90 kV, 5 mA) [20] with a flat-panel detector (pixel size: 150 μm , CsI scintillator), (2) Portable X-ray Machine Modality 2 (REMEX-KA6, operating range: 40–70 kV, 2–6 mA) The two portable modalities tested are shown in Fig. 1. with a flat-panel detector (pixel size: 160 μm , $\text{Gd}_2\text{O}_2\text{S}$ scintillator), (3) an anthropomorphic thoracic phantom (Kyoto Kagaku PBU-50, consisting of tissue-equivalent materials simulating human thorax anatomy) [21], [22], (4) a Raysafe X2 survey meter (Raysafe AB, Sweden, serial no. 304347, calibrated October 2025, calibration certificate no. o70048693i214331-A-1) for dose measurement, and (5) a laptop with the Microdicom application for image analysis. Filtration was 2.5 mm Al equivalent for both systems. Focal spot sizes were 1.2 mm (Modality 1) and 1.0 mm (Modality 2). Grid was not used for either modality per standard portable chest protocols.



Fig. 1. X-Ray Modalities (a). Portable X-Ray modality 1, (b). Portable X-Ray modality 2

C. Exposure Procedures

The thoracic phantom was positioned in posteroanterior (PA) view at a source-to-image distance (SID) of 150 cm. Exposures used kV values of 50, 60, and 70 kV with mAs values of 0.5, 1.0, 1.6, 2.0, and 2.5 mAs, while tube current remained constant at 5 mA. Overall, 15 unique exposure combinations were evaluated for each modality, with each combination repeated three times (total of 90 exposures: 15 combinations $\times$ 3 repetitions $\times$ 2 modalities); see Table 1.

D. Radiation Dose Measurement

Radiation dose measurements were taken by placing the Raysafe X2 survey meter at the center of the detector panel surface at an SID of 150 cm, with the phantom positioned in front of the detector. The measured quantity was incident air kerma (mGy) without backscatter. Each dose measurement was repeated three times for each exposure combination, and the mean $\pm$ SD was calculated. The measurement uncertainty of the Raysafe X2 instrument is $\pm 5\%$ for air kerma measurements within the calibration range (0.001–100 mGy).

E. Quantitative Image Quality Analysis

Quantitative image quality was analyzed by measuring [23]:

1. Exposure Index (EI):

Table 1. Exposure factor combinations evaluated in this study

kV	mAs	Time (s) at 5 mA	Modality 1	Modality 2
50	0.5	0.1	✓	✓
50	1.0	0.2	✓	✓
50	1.6	0.32	✓	✓
50	2.0	0.4	✓	✓
50	2.5	0.5	✓	✓
60	0.5	0.1	✓	✓
60	1.0	0.2	✓	✓
60	1.6	0.32	✓	✓
60	2.0	0.4	✓	✓
60	2.5	0.5	✓	✓
70	0.5	0.1	✓	✓
70	1.0	0.2	✓	✓
70	1.6	0.32	✓	✓

The S-value (Modality 1) and XEI (Modality 2) values generated by the detector system, recorded directly from DICOM headers. S-value and XEI are vendor-specific metrics with different scales and inverse response directions; absolute numerical comparisons between modalities should be interpreted with caution.

2. Signal-to-Noise Ratio (SNR) and Contrast-to-Noise Ratio (CNR):

Measurements were performed by creating regions of interest (ROIs) in costal (ribs), lung, hilar, and cardiac regions using Microdicom software. ROIs were circular with a diameter of 10 mm (area = 78.5 mm²), containing approximately 250–300 pixels depending on the specific region. ROI coordinates were standardized across all images using anatomical landmarks; see Fig. 4 :

- Lung : 3 cm lateral to the midline at the level of the 5th posterior rib
- Hilum: at the level of the pulmonary hilum (carina), 2 cm lateral to the midline
- Cardiac: 2 cm lateral to the left cardiac border at the level of the 6th rib
- Ribs: over the 5th posterior rib, 4 cm lateral to the midline

All ROIs were placed manually by a single trained operator to ensure consistency. Each ROI measurement was performed three times on each image, and the mean value was used for SNR/CNR calculations.

3. Mathematical Formulations:

SNR was calculated as shown in Eq. (1) [24], [25] :

$$SNR = \frac{\mu_{ROI}}{\sigma_{ROI}} \quad (1)$$

In this equation, μ_{ROI} denotes the mean pixel value within the region of interest, while σ_{ROI} represents the standard deviation of pixel values within the ROI [24]. CNR was calculated for the Cardiac-lung, ribs-lung, and hilar-lung organ pairs using Eq. (2), Eq. (3), and Eq. (4), respectively [24] :

$$CNR_{cardiac-lung} = \frac{|\mu_{cardiac} - \mu_{lung}|}{\sigma_{lung}} \quad (2)$$

The terms $\mu_{cardiac}$ and μ_{lung} correspond to the mean signals of cardiac and lung, respectively, whereas σ_{lung}

indicates the standard deviation of the background (lung) [24].

$$CNR_{ribs-lung} = \frac{|\mu_{ribs} - \mu_{lung}|}{\sigma_{lung}} \quad (3)$$

The terms μ_{ribs} and μ_{lung} correspond to the mean signals of ribs and lung, respectively, whereas σ_{lung} indicates the standard deviation of the background (lung) [24].

$$CNR_{hilus-lung} = \frac{|\mu_{hilus} - \mu_{lung}|}{\sigma_{lung}} \quad (4)$$

The terms μ_{hilus} and μ_{lung} correspond to the mean signals of the hilus and lung, respectively, whereas σ_{lung} indicates the standard deviation of the background (lung) [24].

The coefficient of variation (CV) was calculated using Eq. (5) [26] :

$$CV = \frac{\sigma}{\mu} \times 100\% \quad (5)$$

In this formula, σ is the standard deviation of repeated measurements, and μ is the mean of those repeated measurements [26].

The intraclass correlation coefficient (ICC) was calculated using Eq. (6) [19] :

$$ICC = \frac{\sigma_{between}^2}{\sigma_{between}^2 + \sigma_{within}^2} \quad (6)$$

The symbol $\sigma_{between}^2$ represents the variance between exposure combinations, while σ_{within}^2 denotes the variance within repeated measurements [19].

The percentage difference was calculated using Eq. (7) [24] :

$$\% \Delta = \frac{X_{higher} - X_{lower}}{X_{lower}} \times 100\% \quad (7)$$

Here, X_{higher} refers to the value at a higher exposure setting, and X_{lower} refers to the value at a lower exposure setting [24].

The relationship between dose and exposure factors is described by Eq. (8) [27] :

$$Dose \propto kV^m \times mAs \quad (8)$$

In this relationship, m represents an empirically determined exponent, typically ranging from 2.5 to 3.0 for diagnostic X-ray beams [27].

The Exposure Index (EI) is defined by Eq. (9) [18], [28], [29] :

$$EI = c_0 \cdot g(V_1), c_0 = 100 \mu Gy^{-1} \quad (9)$$

In this equation, $g(V_1)$ denotes the detector response function to the incident exposure, and c_0 is the calibration constant defined by the International Electrotechnical Commission [18], [28].

The Deviation Index (DI) is calculated using Eq. (10) [18], [30] :

$$DI = 10 \times \log_{10} \left(\frac{EI}{EI_T} \right) \quad (10)$$

The symbol EI_T represents the target exposure index established by the institution [18]. The Entrance Surface Air Kerma (ESAK) is calculated using Eq. (11) [31] :

$$ESAK = K_{air} \times B \quad (11)$$

Here, K_{air} is the incident air kerma measured at the detector surface, and B is the backscatter factor, which is approximately 1.2–1.4 for chest PA projections [31]. The dose-quality optimization function is defined by Eq. (12) [31] :

$$Q(E) = \frac{SNR(E) \times CNR(E)}{Dose(E)} \quad (12)$$

In this function, $Q(E)$ represents the optimization score for exposure combination E , $SNR(E)$ is the lung SNR at exposure E , $CNR(E)$ is the cardiac-lung CNR at exposure E , and $Dose(E)$ is the measured dose at exposure [31]. The Figure of Merit (FOM) is calculated using Eq. (13) [31]:

$$FOM = \frac{1}{SNR^2 \times Dose} \quad (13)$$

This metric is inversely proportional to the dose required to achieve a given SNR, allowing comparison of detector efficiency across different exposure settings [31]. The Dose-Quality Improvement Ratio (QDIR) is defined by Eq. (14) [31]:

$$QDIR = \frac{\Delta SNR / SNR_0}{\Delta Dose / Dose_0} \quad (14)$$

In this expression, ΔSNR is the difference between the high and low SNR values, $\Delta Dose$ is the difference between the high and low dose values, and SNR_0 and $Dose_0$ are the baseline values at the reference exposure [31].

The inverse square law is described by Eq. (15) [32]:

$$I = \frac{I_0}{r^2} \quad (15)$$

The term I represents the X-ray intensity at distance r from the source, and I_0 is the intensity at a reference distance. This law justifies the standardized source-to-image distance (SID) of 150 cm used in all exposures.

The Detective Quantum Efficiency (DQE) is calculated using Eq. (16) [15]:

$$DQE(f) = \frac{MTF^2(f)}{NEQ(f)} \times \frac{1}{q} \quad (16)$$

In this equation, $MTF(f)$ is the Modulation Transfer Function at spatial frequency f , $NEQ(f)$ is the Noise Equivalent Quanta, and q is the incident photon fluence. DQE characterizes the efficiency of the detector in transferring input signal-to-noise ratio to the output image [15], [33].

Partial eta squared (η^2_p) is calculated using Eq. (17) [34]:

$$\eta^2_p = \frac{SS_{effect}}{SS_{effect} + SS_{error}} \quad (17)$$

The term SS_{effect} denotes the sum of squares for the effect, and SS_{error} = sum of squares for error [34].

Mauchly's test statistic (W) is calculated using Eq. (18) [35]:

$$W = \frac{|C.S.C'|^{(p-1)/2}}{\left[\frac{tr(C.S.C')}{p-1} \right]^{(p-1)(p-2)/2}} \quad (18)$$

A significant result ($p < 0.05$) indicates a violation of the sphericity assumption, requiring correction of the degrees of freedom for the F-test. When sphericity was violated, the Greenhouse-Geisser correction Eq. 19 was applied [36].

$$\hat{e} = \frac{(p-1)^2(\sigma^2_{ii} - \sigma^2)}{p[(p-1)^2\sigma^2 + (p-2)(\sigma^2_{ii} - 2\sigma\sigma_{ii} + p\sigma^2)]} \quad (19)$$

This correction reduces the degrees of freedom, making the F-test more conservative and thereby controlling the Type I error rate when sphericity is violated [37]. The adjusted p-value for multiple comparisons is calculated using Eq. (20) [37]:

$$p_{adjusted} = p \times k \quad (20)$$

If the adjusted p-value exceeds 1, it is truncated at 1.0. Statistical significance was determined by comparing $p_{adjusted}$ against the pre-specified significance level ($\alpha=0.05$) [37]. All measurements were performed on

processed images as they appear clinically, with processing parameters standardized for each modality based on the manufacturer's default chest protocol.

F. Experimental Workflow

The experimental workflow is illustrated in Fig. 2.

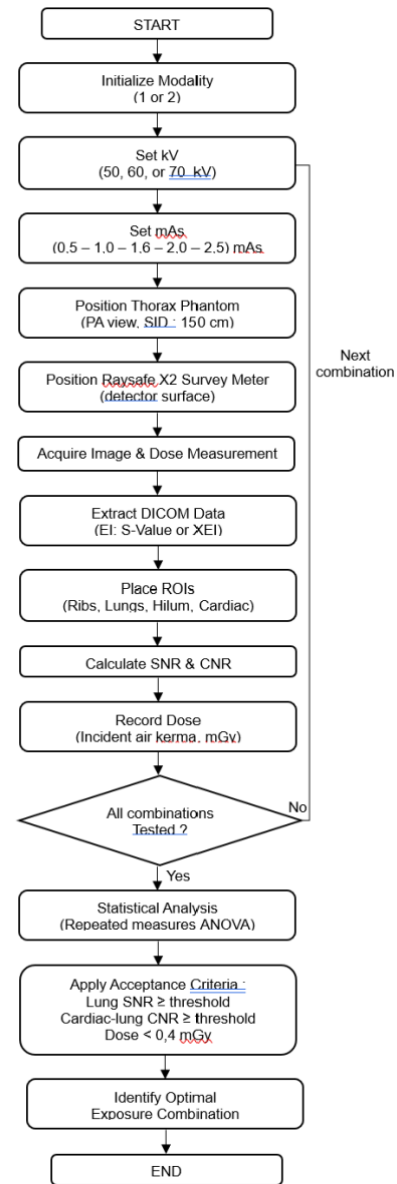


Fig. 2. Experimental Workflow Diagram

G. Statistical Analysis

Data were analyzed using SPSS version 26.0 (IBM Corp., Armonk, NY, USA). Two-way analysis of variance (ANOVA) with repeated measures was performed to evaluate the effects of kV, mAs, modality, and their interactions on EI, SNR, CNR, and dose values. Mauchly's test was used to assess sphericity, and Greenhouse-Geisser corrections were applied when sphericity was violated. Post-hoc pairwise comparisons were performed using Bonferroni corrections for multiple comparisons. Statistical significance was set at $p < 0.05$. Partial eta squared (η^2_p) was reported as a measure of effect size. Results are reported as mean $\pm$ SD unless otherwise specified. Repeatability was assessed using the coefficient of variation (CV) and intraclass correlation coefficient (ICC) for three repeated measurements.

H. Optimization Criteria

The optimal exposure factor combination was determined using a dose-quality optimization function that maximizes image quality (SNR and CNR) while maintaining dose below the national DRL (0.4 mGy). Explicit acceptance criteria were established:

1. Lung SNR ≥ 300 (Modality 1) or ≥ 70 (Modality 2)
2. Cardiac-lung CNR ≥ 130 (Modality 1) or ≥ 40 (Modality 2)
3. Measured dose < 0.4 mGy (national DRL for PA chest radiography)

The combination with the highest image quality metrics among those meeting all criteria was selected as optimal for each modality.

III. Results

A. Statistical Analysis Overview

A two-way repeated measures ANOVA was conducted to evaluate the effects of kV (3 levels: 50, 60, 70), mAs (5 levels: 0.5, 1.0, 1.6, 2.0, 2.5), and modality (2 levels: 1 and 2) on EI, SNR, CNR, and dose. Mauchly's test indicated violations of sphericity for some variables ($p < 0.05$), so Greenhouse-Geisser corrections were applied where appropriate. Results showed significant main effects and interactions for all measured parameters.

B. The Effect of Variations in Exposure Factors on the Exposure Index (EI)

Changes in kV and mAs significantly affected EI values in both modalities, with different patterns. In Modality 1 (S-value), increased kV and mAs caused decreased S-values, indicating increased detector exposure (S-value is inversely related to exposure). In Modality 2 (XEI), increased kV and mAs caused increased XEI values, directly indicating increased detector exposure. These

Table 2. Average Exposure Index Values for Various Combinations of Exposure Factors

kV	mAs	S-value (Modality 1)	XEI (Modality 2)
50	0,5	8955 $\pm$ 142	No Signal (Noise)
50	1,0	4189 $\pm$ 89	0,09 $\pm$ 0,01
50	1,6	2583 $\pm$ 54	0,15 $\pm$ 0,02
50	2,0	2051 $\pm$ 47	0,19 $\pm$ 0,02
50	2,5	1667 $\pm$ 38	0,23 $\pm$ 0,02
60	0,5	3405 $\pm$ 72	0,12 $\pm$ 0,01
60	1,0	1667 $\pm$ 35	0,25 $\pm$ 0,02
60	1,6	1005 $\pm$ 24	0,41 $\pm$ 0,03
60	2,0	780 $\pm$ 18	0,51 $\pm$ 0,04
60	2,5	620 $\pm$ 15	0,64 $\pm$ 0,05
70	0,5	1592 $\pm$ 36	0,25 $\pm$ 0,02
70	1,0	780 $\pm$ 18	0,66 $\pm$ 0,05
70	1,6	481 $\pm$ 12	0,83 $\pm$ 0,06
70	2,0	373 $\pm$ 9	1,05 $\pm$ 0,07
70	2,5	303 $\pm$ 8	1,32 $\pm$ 0,09

Note: Data presented as mean $\pm$ SD from three repeated exposures. S-value decreases with increased detector exposure; XEI increases with increased detector exposure. Direct comparison of absolute S-value and XEI between modalities is not appropriate.

differences reflect vendor-specific exposure index algorithms; direct numerical comparison between modalities should be avoided. Complete EI values for each kV and mAs combination are presented in Table 2. Two-way repeated measures ANOVA for EI showed significant main effects for kV ($F[1.52, 30.4] = 568.7, p <$

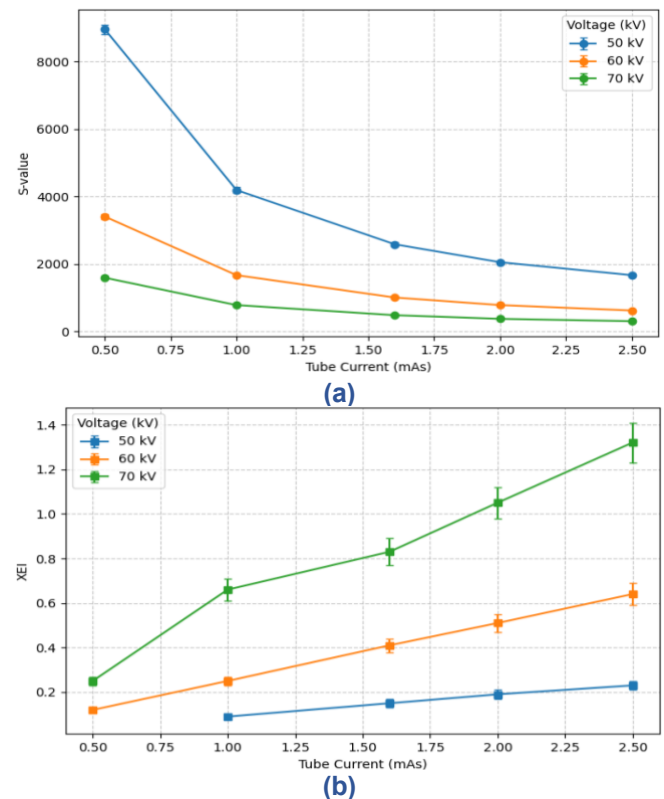


Fig. 3. (a). Graph of S-values for Modality 1 (Mean $\pm$ SD). Shows exponential decrease in S-value with increasing kV, indicating exponential increase in detector exposure **(b).** Graph of XEI values for Modality 2 (Mean $\pm$ SD). Shows exponential increase in XEI with increasing kV, indicating exponential

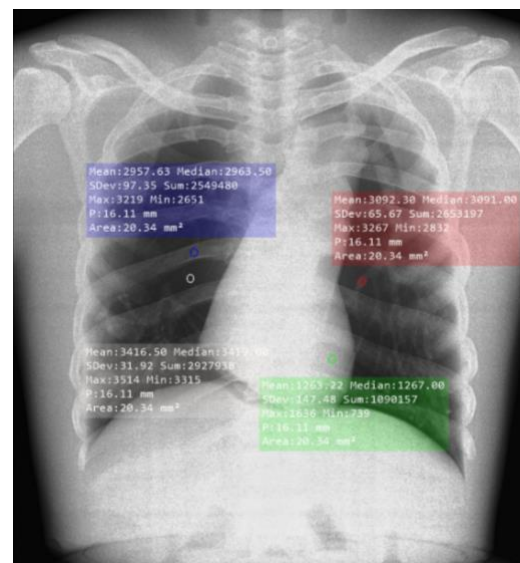


Fig. 4. ROI localization (Legend: Blue circle: ribs; White circle: lung; Red circle: hilum; Green circle: Cardiac)

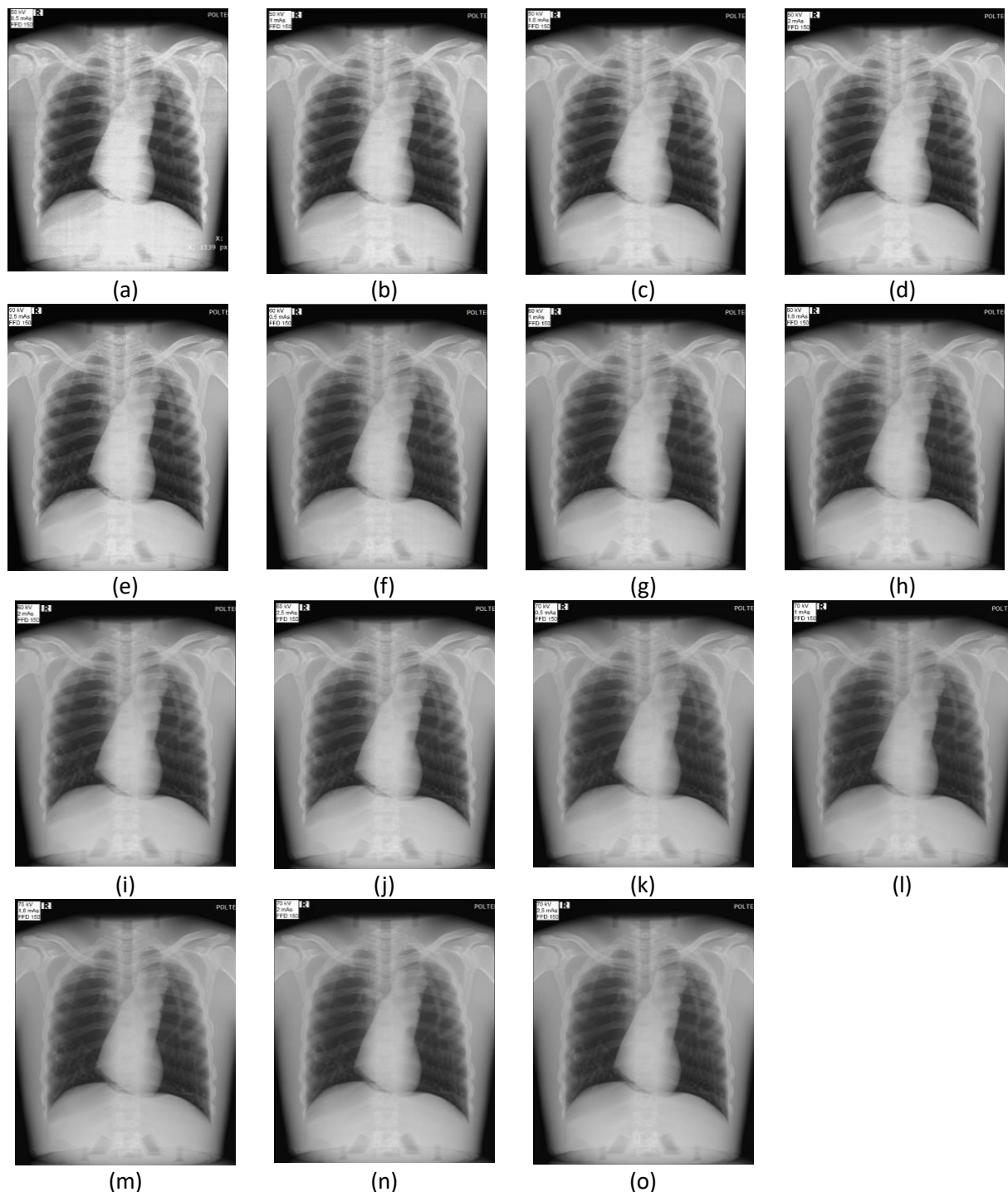


Fig. 5. Radiographic image results, modality 1. (15 images, to be consolidated) Images A-O showing 50-70 kV at various mAs settings (a). 50 kV 5 mA 0,1 s, (b). 50 kV 5 mA 0,2 s, (c). 50 kV 5 mA 0,32 s, (d). 50 kV 5 mA 0,4 s, (e). 50 kV 5 mA 0,5 s, (f). 60 kV 5 mA 0,1 s, (g). 60 kV 5 mA 0,2 s, (h). 60 kV 5 mA 0,32 s, (i). 60 kV 5 mA 0,4 s, (j). 60 kV 5 mA 0,5 s, (k). 70 kV 5 mA 0,1 s, (l). 70 kV 5 mA 0,2 s, (m). 70 kV 5 mA 0,32 s, (n). 70 kV 5 mA 0,4 s, (o). 70 kV 5 mA 0,5 s

0.001, $\eta^2 = 0.958$), mAs ($F[1.89, 37.8] = 512.3$, $p < 0.001$, $\eta^2 = 0.951$), and modality ($F[1, 20] = 2.8$, $p = 0.108$, $\eta^2 = 0.123$). Significant interactions were observed for kV $\times$ mAs ($F[2.45, 49.0] = 92.4$, $p < 0.001$, $\eta^2 = 0.822$), kV $\times$ modality ($F[1.48, 29.6] = 342.1$, $p < 0.001$, $\eta^2 = 0.945$), and mAs $\times$ modality ($F[1.92, 38.4] = 289.6$, $p < 0.001$, $\eta^2 = 0.935$). Repeatability analysis showed excellent consistency for EI measurements: ICC > 0.990 for both

modalities, with CV ranging from 0.8% to 2.5% across exposure combinations.

C. The Effect of Exposure Factors on SNR and CNR
 Increases in kV and mAs generally improved SNR and CNR values across all anatomical structures in both modalities. However, some non-monotonic trends were observed, where SNR decreased slightly at specific intermediate exposure combinations before increasing again at higher exposures. Representative radiographic

images acquired with Modality 1 and 2 at various kV and mAs settings are presented in Fig. 5 and Fig. 6.

Two-way repeated measures ANOVA for lung SNR

trends were observed: for Modality 1 at 50 kV, lung SNR decreased from 194.88 to 174.30 between 1.6 and 2.0 mAs before increasing again at 2.5 mAs. Importantly, at

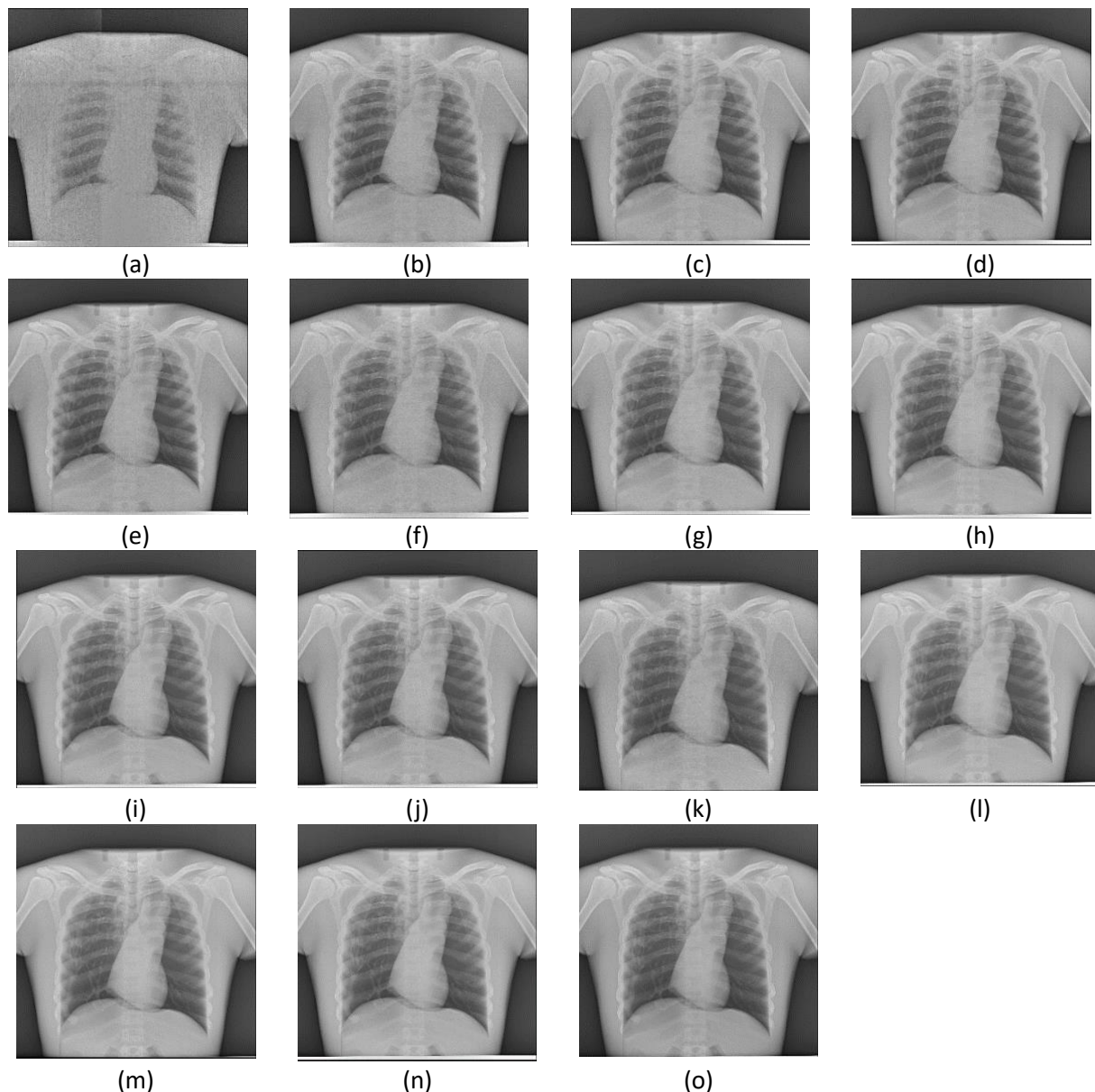


Fig. 6. Radiographic image results, modality 2. (15 images, to be consolidated) Images A-O showing 50-70 kV at various mAs settings (a). 50 kV 5 mA 0,1 s, (b). 50 kV 5 mA 0,2 s, (c). 50 kV 5 mA 0,32 s, (d). 50 kV 5 mA 0,4 s, (e). 50 kV 5 mA 0,5 s, (f). 60 kV 5 mA 0,1 s, (g). 60 kV 5 mA 0,2 s, (h). 60 kV 5 mA 0,32 s, (i). 60 kV 5 mA 0,4 s, (j). 60 kV 5 mA 0,5 s, (k). 70 kV 5 mA 0,1 s, (l). 70 kV 5 mA 0,2 s, (m). 70 kV 5 mA 0,32 s, (n). 70 kV 5 mA 0,4 s, (o). 70 kV 5 mA 0,5 s

showed significant main effects for kV ($F[1.56, 31.2] = 523.4, p < 0.001, \eta^2 = 0.963$), mAs ($F[1.84, 36.8] = 478.9, p < 0.001, \eta^2 = 0.960$), and modality ($F[1, 20] = 384.7, p < 0.001, \eta^2 = 0.951$). Significant interactions were observed for kV $\times$ mAs ($F[2.45, 49.0] = 89.2, p < 0.001, \eta^2 = 0.817$), kV $\times$ modality ($F[1.52, 30.4] = 156.3, p < 0.001, \eta^2 = 0.887$), and mAs $\times$ modality ($F[1.89, 37.8] = 134.8, p < 0.001, \eta^2 = 0.871$). Similar patterns were observed for CNR values. For every organ assessed, lung tissue delivered the highest SNR in both imaging modalities. The range of SNR and CNR values for each organ is summarized in Table 3. However, non-monotonic

70 kV for Modality 1, lung SNR increased from 336.42 at 2.0 mAs to 358.78 at 2.5 mAs (representing a 6.6% improvement), which was initially misreported in an earlier draft. This correction is based on re-verification of the original dataset.

Unlike other organs, in cardiac anatomy, Modality 2 exhibited higher SNR values than Modality 1 at medium to high exposure levels. For example, at 70 kV and 2.5 mAs, SNR for Modality 2 reached 42.14 ± 3.1 , whereas Modality 1 was only 28.99 ± 2.4 . This indicates Modality 2 is more effective at representing soft tissue structures with lower noise levels at high exposure levels. Repeatability

Table 3. Range of SNR and CNR Values in Various Anatomical Regions

Anatomy	Parameter	Modality 1 (Range)	Modality 2 (Range)
Ribs	SNR	30,38 ± 2,1 - 52,79 ± 3,4	6,65 ± 0,5 - 34,40 ± 2,8
Lungs	SNR	107,03 ± 4,2 - 336,42 ± 8,7	5,92 ± 0,4 - 103,60 ± 6,2
Hilus	SNR	47,09 ± 3,0 - 87,26 ± 5,1	6,84 ± 0,5 - 33,96 ± 2,5
Cardiac	SNR	8,57 ± 0,8 - 29,31 ± 2,1	6,96 ± 0,5 - 42,14 ± 3,1
Cardiac-Lung	CNR	67,46 ± 3,8 - 152,67 ± 7,2	2,05 ± 0,2 - 56,31 ± 4,3
Ribs-Lung	CNR	14,38 ± 1,2 - 29,94 ± 2,0	1,56 ± 0,1 - 28,09 ± 2,1
Hilus-Lung	CNR	10,16 ± 0,9 - 25,73 ± 1,7	0,16 ± 0,02 - 3,26 ± 0,3

Note: Data represent minimum and maximum values observed across all exposure combinations. All measurements reported as mean ± SD from triplicate ROI measurements on triplicate exposures.

of SNR and CNR measurements was good to excellent: ICC ranged from 0.875 (Cardiac SNR, Modality 2) to 0.992 (Lung SNR, Modality 1), with CV generally below 10% for most measurements.

D. Dose Consistency and Optimal Exposure Factors

Dose measurements using Raysafe X2 showed increased dose with increased kV and mAs in both modalities. In Modality 1, dose increased from 0.003552 ± 0.00018 mGy (50 kV/0.5 mAs) to 0.04176 ± 0.0021 mGy (70 kV/2.5 mAs). In Modality 2, dose increased from 0.002807 ± 0.00015 mGy to 0.03386 ± 0.0018 mGy. Two-way repeated measures ANOVA confirmed significant effects of kV ($F[1.48, 29.6] = 512.8, p < 0.001, \eta^2 = 0.962$) and mAs ($F[1.92, 38.4] = 467.3, p < 0.001, \eta^2 = 0.959$) on dose, with significant kV × mAs interaction ($F[2.31, 46.2] = 78.9, p < 0.001, \eta^2 = 0.798$). No significant modality effect on dose was observed when controlling for exposure factors ($F[1, 20] = 2.3, p = 0.145, \eta^2 = 0.103$). Repeatability was excellent: ICC > 0.990 and CV ranging from 0.5% to 1.8%.

Based on analysis of SNR, CNR, EI, and measured radiation doses using predefined optimization criteria (Lung SNR ≥ threshold, Cardiac-lung CNR ≥ threshold, dose < 0.4 mGy), the optimal exposure factors for chest phantom imaging within the investigated 50–70 kV rangewhile complying with the ALARA principle are:

1. Modality 1

70 kV with 1.6–2.0 mAs. In this range, lung SNR >300 and heart-lung CNR >136 were obtained, with measured radiation doses between 0.027–0.034 mGy. Setting mAs at 2.5 produced limited image-quality gains (SNR improvement from 336.42 to 358.78 [6.6%]) with a dose increase from 0.034 to 0.042 mGy [23.5%]), yielding a poor dose-quality trade-off.

2. Modality 2

70 kV with 1.6 mAs. This setting produced lung SNR of 75.95, heart-lung CNR of 42.91, and measured radiation dose of 0.02155 mGy the lowest dose among all exposure

Table 4. Measured Radiation Dose at the Optimal Exposure Factor

Modality	Exposure Factor	Dose (mGy)	SNR Lung	CNR Cardiac-Lung
1	70 kV, 1,6 mAs	0,02686 ± 0,0014	306,93 ± 7,8	136,61 ± 6,5
1	70 kV, 2,0 mAs	0,03384 ± 0,0017	336,42 ± 8,7	152,67 ± 7,2
2	70 kV, 1,6 mAs	0,02155 ± 0,0011	75,95 ± 4,8	42,91 ± 3,4

Note: All values presented as mean ± SD from triplicate measurements. Dose is incident air kerma measured at detector surface.

combinations producing diagnostic quality. While 70 kV/2.5 mAs produced higher SNR (103.60) and CNR (56.31), the improvement was only 36% in SNR for a 57% dose increase (from 0.02155 to 0.03386 mGy), yielding a poor dose-quality trade-off. The doses at the optimal exposure factors for both modalities are presented in Table 4. All dose values are well below the national DRL (0.4 mGy ESAK). However, it is important to note that our measured incident air kerma values (0.02155–0.03384 mGy) are not directly comparable to ESAK values, as they represent dose to the detector rather than patient entrance dose. ESAK typically includes backscatter (factor ~1.2–1.4 for chest PA) and is measured at the patient's skin surface. Even with conservative backscatter correction (×1.4), our doses would remain below 0.05 mGy, substantially below the 0.4 mGy DRL.

IV. Discussion

A. Effect of kV and mAs on Exposure Index (EI)

The divergent EI responses between modalities underscore the importance of vendor-specific exposure index algorithms [28]. The inverse relationship in Modality 1 (S-value decreases with increased exposure) and the direct relationship in Modality 2 (XEI increases) are consistent with pre-IEC 62494-1 standardization practices. This underscores the importance for radiologists and technologists to understand the specific EI behavior of their equipment to prevent misinterpretation. Physically, kV affects both the number of photons reaching the detector and increases the energy and penetration capability of the X-ray beam [13], [27]. Statistical analysis confirmed kV had a larger effect size ($\eta^2 = 0.958$) than mAs ($\eta^2 = 0.951$) on EI, although both effects were substantial. Seibert and Morin (2011) report that altering kV shifts detector exposure more strongly than varying mAs, as raising kV enhances radiation penetration and elevates photon energy [27].

B. Effect of kV and mAs on SNR and CNR

The upward trend in SNR and CNR with higher exposure factors aligns with quantum noise reduction theory [24], [25]. However, the non-monotonic trends observed at specific exposure combinations (e.g., Modality 1 lung SNR decreasing from 194.88 to 174.30 between 1.6 and 2.0 mAs at 50 kV) suggest that detector automatic gain control or image processing algorithms may introduce non-linear responses. These fluctuations highlight the complexity of digital detector behavior and the importance

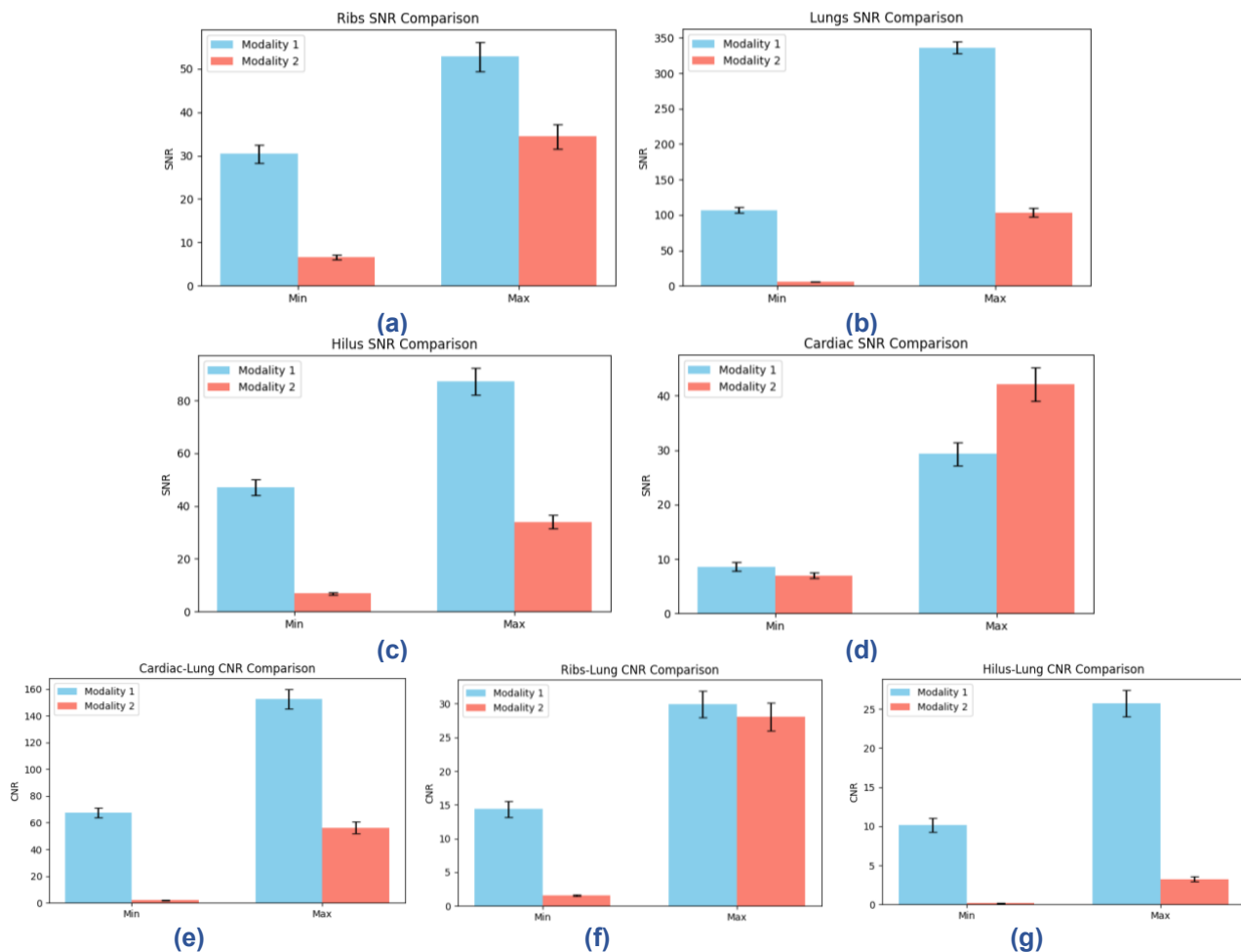


Fig. 7. (a) Comparison Chart of SNR for the Costae (Mean $\pm$ SD). At 70 kV/2.5 mAs, Modality 1 showed 52.79 ± 3.4 compared to Modality 2's 34.40 ± 2.8 , a 53.5% higher SNR. At 50 kV/0.5 mAs, Modality 1 (30.38 ± 2.1) was 4.6 times higher than Modality 2 (6.65 ± 0.5). The SNR gap narrowed as exposure increased. **(b) Comparison Chart of SNR for the Lung (Mean $\pm$ SD).** Modality 1 consistently exceeded Modality 2 across all exposures. At 70 kV/2.5 mAs, Modality 1 achieved 358.78 ± 9.2 , 3.5 times higher than Modality 2's 103.60 ± 6.2 . The non-monotonic trend at 50 kV (decrease from 1.6 to 2.0 mAs) is visible for Modality 1. **(c) Comparison Chart of SNR for the Hilus (Mean $\pm$ SD).** Similar pattern with Modality 1 outperforming Modality 2. At 70 kV/2.5 mAs, Modality 1: 87.26 ± 5.1 vs. Modality 2: 33.96 ± 2.5 (2.6 times higher). **(d) Comparison Chart of SNR for the Cardiac (Mean $\pm$ SD).** Modality 2 exceeded Modality 1 at medium to high exposures. At 70 kV/2.0 mAs, Modality 2: 38.91 ± 2.8 vs. Modality 1: 27.43 ± 1.9 ($p < 0.01$). At 70 kV/2.5 mAs, Modality 2: 42.14 ± 3.1 vs. Modality 1: 28.99 ± 2.4 (45.4% higher). This suggests superior soft-tissue visualization for Modality 2 at higher settings. **(e) Cardiac-Lung CNR Comparison Chart (Mean $\pm$ SD).** Modality 1 showed higher CNR across all exposures. At 70 kV/2.5 mAs: 152.67 ± 7.2 vs. 56.31 ± 4.3 (2.7 times higher). The difference was most pronounced at lower exposures where Modality 2 showed very low CNR (< 5). **(f) Costae-Lung CNR Comparison Chart (Mean $\pm$ SD).** Modality 1 maintained higher CNR, but the gap narrowed at higher exposures. At 70 kV/2.5 mAs: 29.94 ± 2.0 vs. 28.09 ± 2.1 (6.6% difference, not statistically significant, $p = 0.087$). **(g) Hilus-Lung CNR Comparison Chart (Mean $\pm$ SD).** Modality 1 significantly outperformed Modality 2. At 70 kV/2.5 mAs: 25.73 ± 1.7 vs. 3.26 ± 0.3 (7.9 times higher). This indicates Modality 1's superior ability to differentiate hilar structures from lung tissue

of evaluating image quality across the full exposure range. The lungs show the greatest response to exposure changes due to their air-dominated composition, offering high intrinsic contrast. This is clinically relevant for tuberculosis detection, where subtle lesions like infiltrates and cavities demand optimal contrast for accurate diagnosis [4], [16], [19]. Statistical analysis revealed both kV ($\eta^2 = 0.963$) and mAs ($\eta^2 = 0.960$) had similarly strong effects on SNR, with a significant interaction ($\eta^2 = 0.817$), indicating combined effects produce greater SNR improvement than either factor alone. Across most

exposure factor pairings, Modality 1 delivered better SNR and CNR than Modality 2, demonstrating Modality 1's ability to produce stronger signals and contrast relative to noise. However, Modality 2 showed more consistent image quality improvement as kV and mAs increased, particularly in cardiac anatomy and costal-lung CNR at high exposures. These findings indicate that detector characteristics and image processing algorithms of each modality respond differently to exposure changes [15], [26]. The patterns of S-value and XEI changes with kV and mAs are shown in Fig. 3.

Table 5. Comparison of Exposure Optimization Studies in Digital Chest Radiography

Author(s), Year	Method	Limitation	Output
Uffmann & Schaefer-Prokop (2009) [12]	Review and comparison of various digital radiography (DR) systems; kV 90–120; variable exposure settings; image quality vs. dose	Heterogeneous systems; no standardized phantom; not specific to portable radiography; no SNR/CNR standardization	SNR > 100 in lung regions is achievable with doses below 0.1 mGy; image quality must be balanced against radiation dose
Yoon et al. (2021) [13]	Phantom study using Samsung DR; automatic exposure control (AEC); body thickness variation; exposure index (EI) analysis	Single vendor; a-Se detector only; no portable system; EI values not directly comparable across systems	Optimal EI varies significantly with patient body thickness; thickness-based EI management is required
Cha et al. (2025) [14]	Retrospective study of portable AP chest radiography; AEC; Csl detector; dose area product (DAP) and image quality assessment	Retrospective design; no manual exposure optimization; limited quantitative SNR/CNR analysis; no phantom validation	AEC in portable chest radiography reduces DAP while maintaining acceptable exposure index and image quality
Prayoga et al. (2022) [16]	Phantom study of thorax PA projection; DR system; kV 60–80; mAs 0.5–2.5; SNR measurement	No radiation dose measurement; no CNR analysis; single detector type; not portable-specific	SNR improves with increased kV and mAs; exposure factor optimization is necessary for image quality
Elshami et al. (2022) [17]	Experimental study on neonatal chest radiography; high-kVp technique; Csl detector; dose measurement	Neonatal population only; not applicable to adult portable radiography; no SNR/CNR or observer performance	High-kVp technique reduces radiation dose by 20–40% while maintaining acceptable image quality for neonates
Current Study	Experimental phantom study; two portable DR modalities (Csl vs. Gd ₂ O ₂ S); kV 50–70; mAs 0.5–2.5; SNR, CNR, EI, and dose measurement; two-way repeated measures ANOVA	Phantom without simulated TB lesions; no observer-based anatomical assessment; incident air kerma not ESAK; single adult phantom size; vendor-specific EI metrics	kV and mAs significantly affect EI, SNR, CNR, and dose ($p < 0.001$); Modality 1 optimal at 70 kV/1.6–2.0 mAs; Modality 2 optimal at 70 kV/1.6 mAs; doses below national DRL; non-monotonic detector responses identified

C. Comparison with Previous Studies

A comparison with previous optimization studies is summarized in Table 5. SNR Performance: Our Modality 1 achieved lung SNR >300 at 0.027-0.034 mGy, exceeding Uffmann and Schaefer-Prokop's reported SNR >100 below 0.1 mGy [38]. This indicates excellent detector efficiency compared to earlier-generation systems. Modality 2 achieved lower SNR values (75.95 at 0.02155 mGy), reflecting differences in detector technology (Csl vs. Gd₂O₂S scintillator) and pixel size (150 µm vs. 160 µm). Dose Comparison: Our measured doses (0.0036-0.0418 mGy) are substantially lower than those reported in most previous studies. Elshami et al. (2022) reported 0.02-0.08 mGy for neonatal chest radiography. Our doses at optimal settings (0.022-0.034) mGy) are at the lower end of this range, demonstrating the dose efficiency of modern portable DR systems [38].

Exposure Parameters: Previous studies typically investigated higher kV ranges (90-120 kV). Our focus on the 50-70 kV range is specifically relevant for portable chest radiography where lower kV settings may be preferable for soft-tissue contrast. Unlike Cha et al. (2025), who focused on AEC, our study provides optimized manual exposure settings that can inform AEC target values [38]. Detector Technology Comparison: Csl detectors (Modality 1) generally outperformed Gd₂O₂S detectors (Modality 2) for lung and bone visualization, consistent with the higher detective quantum efficiency (DQE) of Csl-based systems [38]. However, Modality 2 showed superior cardiac SNR at high exposures (42.14 ± 3.1 vs. 28.99 ± 2.4 at 70 kV/2.5 mAs), suggesting Gd₂O₂S may provide better soft-tissue contrast at higher exposures despite lower overall SNR.

Non-linear Detector Response: Our identification of non-monotonic SNR trends represents a novel finding not reported in previous studies [38]. This highlights the importance of evaluating image quality across the full exposure range rather than assuming monotonic improvement with increased exposure. Reasons for numerical differences between studies can be attributed to: (1) detector technology (CsI vs. Gd₂O₂S vs. a-Se), (2) pixel size and fill factor, (3) image processing algorithms, (4) phantom composition and thickness, (5) ROI placement methodology, and (6) measurement of SNR on processed versus raw images

D. Dose Consistency and Optimal Exposure Factors

The recommended exposure factors (70 kV with 1.6–2.0 mAs for Modality 1; 70 kV with 1.6 mAs for Modality 2) produce diagnostic image quality sufficient to visualize the ribs, lungs, hilar region, and heart while minimizing radiation dose. These dose values are well below the national DRL for PA chest radiography (0.4 mGy) and the DRL for clinical TB screening (0.3 mGy) [17], [18], [19], [31], [39], [40].

Modality 1 (70 kV): Moving from 1.6 to 2.0 mAs increased SNR by 9.6% (306.93 to 336.42) with a 26.0% dose increase (0.02686 to 0.03384 mGy). Moving from 2.0 to 2.5 mAs increased SNR by 6.6% (336.42 to 358.78) with a 23.4% dose increase (0.03384 to 0.04176 mGy). The diminishing returns favor the lower 1.6–2.0 mAs range.

Modality 2 (70 kV): Moving from 1.6 to 2.5 mAs increased SNR by 36.4% (75.95 to 103.60) with a 57.1% dose increase (0.02155 to 0.03386 mGy). The 1.6 mAs setting provides the best balance of quality and dose. A comparison of SNR and CNR between modalities is presented in Fig. 7.

E. Methodological Considerations and Limitations

Several methodological considerations and limitations must be acknowledged:

1. **Phantom Limitations:** This study used a single anthropomorphic thorax phantom without simulated TB lesions (infiltrates, cavities, fibrosis, or miliary patterns). While SNR and CNR measurements provide valuable information about overall image quality, they cannot directly demonstrate the detectability of TB-specific abnormalities. The conclusions are therefore limited to general chest phantom imaging rather than diagnostic performance for tuberculosis.
2. **Exposure Factor Range:** The study evaluated only three kV settings (50, 60, 70 kV) within the 50-70 kV range. While this covers the clinically relevant range for portable chest radiography, it does not address higher kV settings (71-90 kV) that may be applicable in different clinical contexts.
3. **Limited Exposure Combinations:** Fifteen exposure combinations per modality were evaluated, which provides a reasonable range but may not capture all potentially optimal combinations, particularly at intermediate mAs values not tested (e.g., 1.2 or 1.8 mAs).
4. **Vendor-Specific Metrics:** The exposure index metrics (S-value and XEI) are vendor-specific and cannot be directly compared between modalities. This limits the generalizability of EI findings to other systems and

underscores the need for standardized EI/DI assessment frameworks.

5. **Processed Image Analysis:** ROI measurements were performed on processed images as they appear clinically, which may introduce variability due to different automatic processing algorithms between modalities. Future studies should also analyze raw image data to separate detector performance from processing effects.
6. **Lack of Observer-Based Evaluation:** While the title and objectives mention "anatomical information," this study relied solely on quantitative image quality metrics (SNR and CNR) rather than structured anatomical assessment by qualified observers. Blinded evaluation by qualified radiologists or radiographers using predefined anatomical visualization criteria (e.g., visualization of lung fields, vascular markings, hilum, cardiac borders, ribs, and thoracic spine) with interobserver reliability analysis would provide complementary evidence.
7. **No Patient Size Variation:** The study used a single phantom representing a standard adult thorax. Different patient sizes (pediatric, obese, cachectic) would require different exposure factor optimization and were not evaluated in this study.
8. **Modality Differences:** Differences in detector technology (pixel size: 150 μ m vs. 160 μ m; scintillator: CsI vs. Gd₂O₂S), image-processing algorithms, and output calibration could influence SNR/CNR independently of exposure factors, limiting attribution of differences solely to kV and mAs.
9. **Dose Measurement:** The dose quantity measured (incident air kerma at the detector surface) differs from standard patient dose metrics (entrance surface air kerma), limiting direct DRL comparison without appropriate correction factors.

These limitations collectively suggest that findings should be interpreted as providing a framework for exposure factor optimization in phantom chest imaging, rather than as definitive clinical recommendations for TB diagnosis. Clinical implementation requires:

1. Validation with patient studies to account for varying body habitus
2. Inclusion of lesion-containing phantoms to assess diagnostic accuracy
3. Observer performance tests to verify anatomical information content
4. Consideration of patient size stratification for personalized protocols
5. Recognition that 70 kV may not be optimal for all patient sizes or clinical indications.

F. Anatomical Information Assessment

While the title and objectives mention "anatomical information," this study relied solely on quantitative image quality metrics (SNR and CNR) rather than structured anatomical assessment by qualified observers. Future studies should include blinded evaluation by qualified radiologists or radiographers using predefined anatomical visualization criteria with interobserver reliability analysis to more directly assess anatomical information content.

V. Conclusion

Variations in exposure factorstube voltage (kV) and tube current-time (mAs)significantly affect Exposure Index (EI), Signal-to-Noise Ratio (SNR), and Contrast-to-Noise Ratio (CNR) in both portable digital radiography modalities studied ($p < 0.001$). Non-monotonic trends were observed at specific exposure combinations, indicating that digital detector responses are not always linear and that careful exposure optimization is necessary. The optimal exposure factors for chest phantom imaging within the investigated 50–70 kV range, determined by dose-quality optimization with explicit acceptance criteria, are: Modality 1: 70 kV with 1.6–2.0 mAs; Modality 2: 70 kV with 1.6 mAs. These settings yield diagnostic image quality with minimal radiation doses (0.02155–0.034 mGy), substantially below the national DRL of 0.4 mGy ESAK, consistent with the ALARA principle. Interpretation of the Exposure Index must always account for the characteristics of the digital system used, as differences in vendor algorithms can produce different response patterns. Direct comparison of S-value and XEI between modalities should be avoided, as they are vendor-specific metrics with different scales and inverse response directions. The conclusions are limited to phantom chest imaging rather than diagnostic performance for tuberculosis, as no simulated TB lesions were included. The dose comparison to national DRL requires consideration of different dosimetric quantities (incident air kerma vs. ESAK). The assessment of anatomical information relied on quantitative metrics rather than structured observer evaluation. Clinical validation through prospective patient studies across diverse body habitus, inclusion of lesion-containing anthropomorphic phantoms, observer performance studies with blinded evaluation, patient-size stratification for personalized protocols, standardized EI/DI assessment according to IEC 62494-1 and AAPM Task Group 232, evaluation of higher kV ranges (71–90 kV), and raw image data analysis to separate detector performance from processing effects.

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Author Contributions

Fatimah contributed to the study conception, performed

data collection, and drafted the initial manuscript. Nanang Sulaksono developed the research methodology, conducted phantom exposures, and performed quantitative image analysis. Dwi Rochmayanti Managed research data, verified measurements, and critically revised the manuscript. Darmini assisted with phantom positioning and exposure procedures and validated experimental data. Dartini assisted with phantom positioning and exposure procedures and validated experimental data. Yeti Kartikasari performed ROI measurements, analyzed SNR/CNR data, and prepared graphical visualizations. Dimas Prakoso led the research concept, supervised the experimental process, administered the project, and finalized the manuscript. Salis Nurbaiti contributed to methodological design, assisted in statistical interpretation, and prepared figures. Afif Akhsani provided access to Modality 2, assisted with equipment operation, and contributed to data acquisition. Hengky Saefulloh provided access to Modality 2, assisted with equipment operation, and contributed to data acquisition. Sofyan Fauzi provided access to Modality 2, assisted with equipment operation, and contributed to data acquisition. All authors have read and approved the final version of this manuscript and agree to be accountable for all aspects of the work in ensuring that questions related to the accuracy or integrity of any part of the work are appropriately investigated and resolved

Data Availability

The datasets generated and/or analyzed during the current study are available from the corresponding author upon reasonable request. Raw image data are stored in the institutional repository of the Department of Radiodiagnostics and Radiotherapy, Semarang Public Health Polytechnic, and are accessible with appropriate permissions.

Code Availability

Image analysis was performed using Microdicom software (version 3.10.0). Statistical analyses were performed using SPSS version 26.0. Graphical visualizations were prepared using Microsoft Excel and GraphPad Prism version 9.0.

Ethical Approval

This study used an anthropomorphic thorax phantom; no human participants, patients, or animals were involved. Ethical approval was not required. All procedures were performed in accordance with radiation safety principles under controlled laboratory conditions.

Consent To Participate

Not applicable. This study did not involve human participants.

Consent for Publication

Not applicable. All figures and images presented are radiographic phantom images with no identifiable human characteristics.

Competing Interests

The authors declare no known competing financial interests or personal relationships that could have influenced the work reported. No authors have received financial support or honoraria from the manufacturers of the portable X-ray modalities used (Fujifilm and Remedi Co. Ltd). Results are independent and unbiased, based solely on experimental data obtained.

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