

Chapter 2

COMPARISON OF MAMMOGRAPHY RADIATION DOSE VALUES OBTAINED FROM DIRECT INCIDENT AIR KERMA MEASUREMENTS WITH VALUES FROM MEASURED X-RAY SPECTRAL DATA

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Abstract

The application of X-rays and ionising radiations for diagnostic radiology requires that the procedure is justified and optimised and that the exposure to the patient is kept as low as possible, without compromising image information. X-ray mammography is considered to be the most sensitive technique currently available for early detection of breast cancer. The magnitude of the absorbed radiation dose to the breast from mammography X-ray beams forms an important part of the quality control of the mammographic examination since it gives an indication of the performance of the mammographic imaging system as well as an estimated risk to the patient. In this work mean glandular dose (MGD) values were obtained at various tube potentials and tube loadings (TL) using direct measurements of the incident air kerma (ESAK) at the surface of a standard breast phantom and also from spectral measurements acquired with a solid-state detector. Comparisons of the MGD values thus derived are presented and the relationship between MGD, phantom thickness, image quality and tube operating parameters is discussed.

2.1. Introduction

Breast cancer is reported to be the second highest cause of cancer deaths in women today (after lung cancer), with an estimation of 400,600 deaths from breast cancer for 2001 (GLOBOCAN, 2000). Breast cancer is also the most common cancer among women, excluding nonmelanoma skin cancers. In 2001, World Health Organization (WHO) predicted more than 1.2 million new breast cancer cases (globally) while in United States, both the American Cancer Society and the National Cancer Institute estimated approximately 192,200 new cases of invasive breast cancer cases for 2001. Male breast cancer has also been reported

but this accounts for less than 1% of the total (American Cancer Society, 2001; National Cancer Institute, 1999). Statistics on mammography indicate that the incidence of breast cancers per 100,000 women increased by approximately 4% during the 1980s and levelled off to about 100 cases per 100,000 women in the 1990s. The reported death rates from breast cancer also declined significantly between 1992 and 1996. Medical experts attribute the reduction in breast cancer death to earlier detection and more effective treatment.

The use of X-ray mammography in the screening of asymptomatic women has become very common in many parts of the world. The objective of the screening exercise is to detect breast cancer at an early stage to reduce breast cancer mortality. There is however, a small but non-negligible risk of radiation-induced carcinogenesis associated with an X-ray examination of the female breast (NCRP, 1986). An accurate knowledge of the output of an X-ray mammography tube is essential in medical diagnostics for ensuring accurate dose level estimation and for the provision of a useful check on the diagnostic adequacy of the imaging technique. Breast radiation dose assessments are therefore included in most national quality assurance programmes for X-ray mammography. Notable among the national protocols and institutions, which require this inclusion, are: NCRP (1986), AAPM (1990) and CEC (1996).

The glandular tissue is the most vulnerable of the tissues making up the breast. Thus, among the different dosimetric quantities used in risk assessment, the mean glandular dose (MGD) is the best indicator of the patient risk (NCRP, 1986) and it has been accepted by the national protocols and institutions mentioned earlier and many others as the preferred quantity for the measure of potential risk from mammography. Several other institutions have also suggested

MGD as the preferred quantity for the measure of potential risk from mammography (ICRP, 1987; NCRP, 1986; IPSM, 1989; IPSM, 1994). Measurements of absorbed radiation dose from mammography have been carried out by a number of investigators using a variety of mammographic techniques (Young et al., 1996; Klein et al., 1997 and Dance et al., 1999). The standard method of estimating the MGD dose on patients undergoing mammography X-ray examinations is based on ESAK measurements without backscatter and the conversion to glandular dose using appropriate conversion factors depending on the type of phantom used (IPSM, 1994). The air kerma value may be determined either for patients or for a standard breast phantom; polymethyl methacrylate (PMMA) is normally used as breast substitute phantom. The MGD is then obtained either by determining the tube loading or from the exposure parameters required to obtain the recommended optical density of the exposed standard phantom. Two possibilities exist; either (i) the tube output at this tube loading is measured or, (ii) a direct radiation dose assessment using thermoluminescent dosimeters (TLDs) placed on the breast can determine the dose (K_s) to the entrance surface of the breast (this will include all backscattered radiation).

Spectral information is essential for dosimetric purposes. An accurate knowledge of the X-ray spectral output forms the basis of almost all image quality simulations thereby enabling system designers to predict patient dose more accurately. Direct measurement of X-ray spectra is usually performed with a high-purity germanium detector with the signal output being processed by conventional electronics and a multichannel analyzer (MCA) to obtain the photon energy distribution. Spectrum correction techniques are then applied to the acquired photon distribution to generate the photon fluence spectrum from which the MGD is

calculated (Calicchia et al., 1996). MGD values have also been calculated using Monte Carlo techniques (Wu et al., 1991).

Unlike previously published works, this contribution provides an evaluation and comparison of the values for the important MGD parameter as obtained by the two methods outlined above; the standard incident air kerma direct method and the spectral method that utilises the energy fluence spectral data incident on standard phantoms of varying thicknesses. X-ray spectra were determined for a molybdenum anode with molybdenum filter at settings and exposure parameters normally used in routine mammography. The study also investigated the relationships between MGD, phantom thickness, image quality and tube voltages. Their implications in mammography are discussed.

2.2. Experimental

2.2.1. Materials

The X-ray unit used for this work was a General Electric (Model No. 65447) three-phase Senographe 500T six-pulse X-ray generator. The rotating anode had a grounded molybdenum target with nominal, selectable focal spot sizes of 0.3 and 0.1 mm. The X-ray unit has an inherent filtration of 0.8 mm beryllium (an equivalent aluminium filtration of 0.028 mm Al) at 30 kVp and 0.03 mm molybdenum of added filtration. The breast substitute material used for the study was PTW (Physikalisch Technische Werkstätten) acrylic glass, SIB Mammographic Phantom (Type 42001). This phantom has attenuation properties similar to those of PMMA of similar thickness. The film and screen type used were AGFA Mamoray HDR (High Dynamic Range) and AGFA Mamoray (High Dynamic Screen) respectively. For film processing, an

AGFA Imaging Min/Med unit maintained at a temperature of 34.9°C with a processing cycle of 90 seconds was used. Film optical densities (OD) were measured with X-Rite (Model no: 331) densitometer. The densitometer has an accuracy of ± 0.02 OD and reproducibility of ± 0.01 OD. The X-ray tube output for the different exposure parameters was measured with a PTW-Diados mammography detector (Type T60005 0735) and PTW-Diados diagnostic dosimeter i.e. electrometer connected to a display unit (Type 11003-1129). The dosimeter and detector were calibrated by PTW at energy values 25 – 45 kVp with a precision of less than 0.5%. The calibration factor (N_K) for each radiation quality was 1. The detector showed a flat energy response over the mammography energy range and the dosimeter has reproducibility better than $\pm 0.5\%$ and a digital resolution of 1 nGy/s. A Gammex RMI Type 1100, 99.8% aluminium attenuator set was used for the attenuation and HVL measurements. The aluminium attenuators were plates of area 10 cm² and of thicknesses accurate to within $\pm 5\%$.

For the spectral analysis, a high purity intrinsic planar, liquid nitrogen cooled, germanium detector coupled to multi-channel analyser (MCA) manufactured by DSG Detector Systems (Model No. PGP 200-10) was used. The detector was operated at a bias voltage of –2500 V. The detector had a thickness of 10 mm, a 0.015 mm thick beryllium window and 200-mm² of sensitive area. The detector was coupled to a desktop computer fitted with Aptec Supervisor Automation software cards for data acquisition and processing.

2.2.2. Calibration of tube output

The PTW-Diados mammography-energy calibrated detector was supported in the X-ray beam at 5 cm above the cassette holder, 6 cm from the chest wall edge and centred laterally. A tube

current-exposure time product (tube loading) of 40 mAs was selected. For this study, the tube output was measured for tube voltages set at 25, 26, 27, 28, 29, 30 and 32 kVp. Corresponding HVL values, which were measured in the presence of the breast compression plate, are shown in Table 1. The measurements were all carried out with the breast compression plate in place. The tube output measurement has been expressed as air kerma per tube loading ($\mu\text{Gy/mAs}$) at 1 metre focus-detector distance.

2.2.3. Determination of tube loading

The tube loading (TL) measurement for ensuring the correct exposure of the standard phantom was determined following the procedure described by the European Commission (CEC, 1996). In this determination, the X-ray machine was set up for a cranio-caudal view, with the breast compression plate and a loaded cassette in place. The phantom was placed on the breast support table for a cranio-caudal position. The compression plate was brought down onto the phantom. The phantom was exposed using exposure parameters employed clinically for a standard-sized breast and the tube loading was recorded. The exposed film was processed with a dedicated mammography processor and the optical density (OD) measured. An OD including base and fog was then verified (recommended OD range given by the European Protocol i.e. 1.0 to 1.5 OD, (CEC, 1996)). The exposure settings were adjusted where necessary, and the procedure repeated until an OD of approximately 1.3 was achieved.

2.2.4. Measurement of X-ray tube output ESAK per tube loading

The PTW-Diados mammography detector was positioned at the reference point (phantom position) during the tube loading determinations above the cassette holder for the 20, 30, 40, 50 and 60 mm thicknesses of PMMA and 60 mm from the chest wall edge of the cassette holder. The radiation detector was exposed in the manual-mode, using the determined tube loadings and the dosimeter or electrometer reading (air kerma) was recorded for each. Corrections to ambient temperature, pressure and humidity values were made according to methods previously described (Assiamah et al. 2003a). The HVL for the selected tube loading was determined.

2.2.5. Determination of MGD from measured X-ray ESAK

The MGD was calculated from the ESAK (without backscatter), obtained from the output measurements and standard conversion factor tabulations and the equation:

$$MGD = Kgp, \quad (2.1)$$

where K is the ESAK at the specified HVL, g is the ESAK to MGD conversion factor. The p -factor converts air kerma for the PMMA breast substitute phantom to that for the model breast which it approximately simulates (Dance et al., 1999). Ambient temperature and pressure values were used to correct the dosimeter reading. All the MGD values presented are for a focal-spot to detector distance (FSD) of 650 mm. The conversion factors g and p , used for this work were those from Dance (1990) and were calculated assuming the ‘standard’ breast phantom composition. The g factors for the measured half value layers were derived by interpolation from the compiled data and gave a percent deviation of less than 5%.

2.2.6. Contrast measurements

Film contrasts were carried out using PMMA thicknesses and experimental alignments similar to those used for the tube loading measurements. A 0.2 mm thick aluminium sheet of area 10 x 10 mm² was centred on top of the PMMA. Film contrast was then obtained from the radiograph as the OD difference between the image of the aluminium square and the adjacent background (Desponds et al., 1994).

2.2.7. Detector calibration and determination of detector response

The detector was calibrated for energy scales, linearity checks, resolution and full energy peak efficiency using standard sources Am-241, Co-57 and Ba-133. An energy-channel conversion factor of 40 eV per channel with 0 keV corresponding to 0 channel number was set for the measurements. The detector system showed a linear energy response with a resolution of 500 eV and 610 eV at 17.78 and 59.4 keV respectively.

2.2.8. Spectral measurement and data correction techniques

For a good geometry and also to reduce pile-up effects, a lead-copper collimator of size 2 cm diameter and of thickness 1 mm each of lead and copper joined together was used to collimate the X-ray beam. An X-ray photon distribution was generated with and without collimation for each tube voltage. Three exposures were made at each tube setting. The measured photon distribution was corrected for detector efficiency and summed to obtain the number of photons for every 0.5 keV intervals. This keV interval was chosen to ensure that the K_{α} and K_{β} X-ray

peaks appeared at the correct energy interval (17.5 and 19.5 keV respectively). The details of the spectra correction techniques are described in Mavunda et al., (2004).

2.2.9. Determination of MGD using spectral data

The photon fluence (photons/mm²) obtained after correction for detector window and *K*-fluorescence escape was corrected for air attenuation as described by Assiamah et al. 2003a. The quality of each measured spectrum was modified by the addition of aluminium filters until the 1st HVL agreed with that of the 1st HVL from the ionisation chamber measurement with breast compression plate in place, but no added aluminium. The spectrum was normalized to correct for any variation caused by the addition of aluminium filters (Fewell, 1977). To determine the HVL that simulated the presence of the set up with the compression plate in the beam path, the normalized photon energy fluence for each tube voltage was attenuated with 3 mm PMMA thickness (the equivalent thickness of the compression plate). Using the normalized photon fluence remaining after PMMA attenuation $\phi_{i(corr)}$, exposure values, $X_i(E)_{T,P,H}$ in roentgen (R) (Johns and Cunningham, 1983) at absolute temperature T , barometric pressure P and humidity H was calculated employing equation (2).

$$X_i(E)_{T,P,H} = \frac{1.83 \times 10^{-11} \phi_{i(corr)} E_i (\mu_{en})_i^{air}}{\rho_{T,P,H}}. \quad (2.2)$$

The subscript i refers to the i^{th} photon energy interval E_i (keV), $(\mu_{en})_i^{air}$ is the energy absorption coefficient of air at s.t.p.; $\rho_{T,P,H}$ is the density of air at temperature T , pressure P and humidity H (Cember, 1996). The exposure values calculated were converted to air kerma

values using a factor 0.873 (Wolbarst, 1993). The MGD was calculated from the air kerma values using Eqn. (1) and the appropriate conversion factors.

2.2.10. Determination of attenuation curve and half value layer

Exposure values calculated as described previously were used to calculate attenuation points and the HVL of the spectral data. Points on the attenuation curve were found for each thickness of aluminium attenuator by calculating exposure $X_i(E)_{T,P,H}$ and summing over all the energy intervals. From the attenuation curve the HVL was calculated for each nominal tube voltage. To determine the HVL that simulated the presence of the compression plate in the beam path, the normalized photon energy fluence for each tube voltage was attenuated with 3 mm PMMA thickness. The mass attenuation coefficient data, mass energy-absorption coefficient data and the elemental composition of the PMMA phantom used for this work were obtained from the published results of Hubbell and Seltzer (1996). To facilitate the computation of the air kerma value, the mass energy-absorption coefficient $(\mu_{en})_i^{air}$ data by Hubbell and Seltzer, was fitted to an extension of a fifth degree polynomial equation (Tucker et al., 1991) in order to derive the parameters needed to generate the absorption energy "spectra". This was then used in computing $K(E)$. The extension to the Tucker et al. equation is found necessary as the published $(\mu_{en})_i^{air}$ data do not cover the full energy spectrum of interest (Assiamah et al., 2003b).

2.3. RESULTS AND DISCUSSION

The HVL values of the mammography X-ray machine used for this study are shown in Table 2.1. The variation in the tube output ($\mu\text{Gy/mAs}$) versus the nominal tube voltage (kVp) is presented in Figure 2.1. The comparison of ESAK (without backscatter) and the corresponding MGD calculated from both the direct and spectral data, for all the PMMA phantom thicknesses obtained at nominal tube voltages 25, 26, 27, 28, 29, 30 and 32 kVp are presented in Table 2.2. The measured and calculated attenuation curve measurements for the set tube voltages 26 and 28 kVp are shown in Figures 2.2 and 2.3 respectively for collimated X-ray beams and in Figures 2.4 and 2.5 for uncollimated X-ray beams. Figure 2.6 shows the contrast values as a function of phantom thickness and nominal tube voltages for an optical density of 1.3. In Figure 2.7 is presented the film contrast at 28 kVp with varying tube loadings for different PMMA phantom thicknesses for a range of optical densities. Figure 2.8 shows X-ray images of a 30 mm SIB phantom exposed at a nominal tube voltage of 28 kVp with tube loadings of 25 and 32 mAs respectively while Figure 2.9 presents X-ray images of 40 mm and 30 mm SIB phantom exposed with identical X-ray tube parameters (nominal tube voltage of 28 kVp and 40 mAs).

Although the targeted OD set for this work was 1.3 including base and fog, attaining this OD value was not always feasible at certain exposure parameters due to restrictions caused by the design of the X-ray machine. The tube output ($\mu\text{Gy/mAs}$) (Figure 2.1) values measured were found to be linear with the tube voltage settings, indicating that the mammography X-ray beam used was both reliable and consistent. For all the tube settings, the maximum deviation of each tube output from the mean was less than 5%.

Table 2.1: The HVL values of the mammography X-ray machine measured in the presence of the breast compression plate used for this study at the specified nominal tube voltages.

Nominal tube voltage (kVp)	HVL (mm Al)
25	0.32 ± 0.02
26	0.33 ± 0.02
27	0.34 ± 0.02
28	0.35 ± 0.02
29	0.36 ± 0.02
30	0.37 ± 0.02
32	0.39 ± 0.02

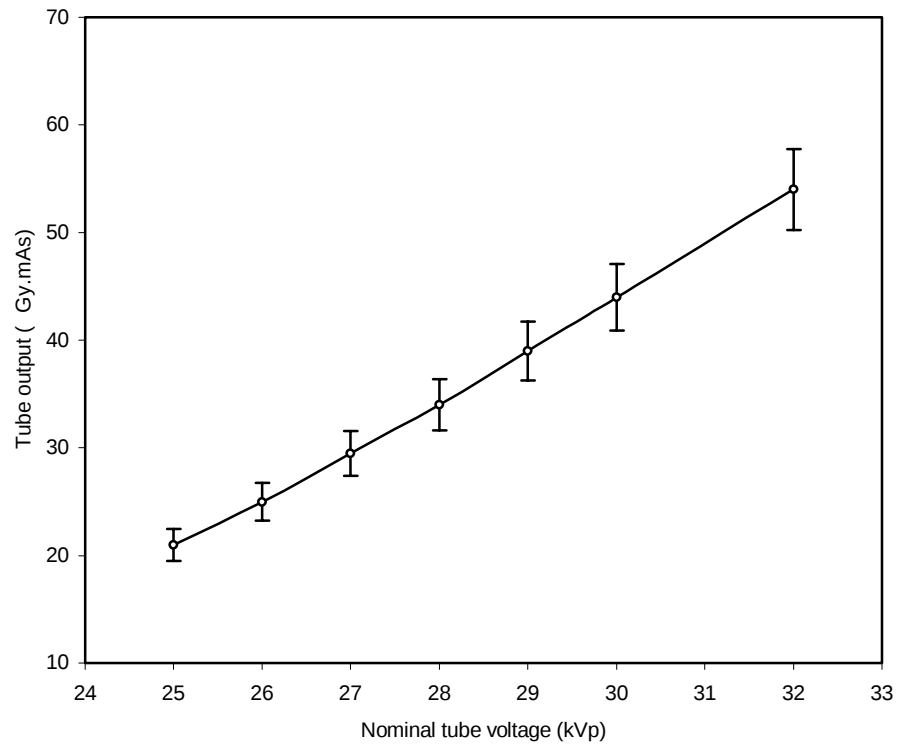


Figure 2.1: Tube output ($\mu\text{Gy}/\text{mAs}$) for the nominal tube voltages between 25 and 32 (kVp) used in this study.

The values of the ESAK and MGD (Table 2.2) from the measured air kerma and measured spectral data agreed generally to within $\pm 5\%$ at all the nominal tube voltages and for the phantom thicknesses studied. The percentage difference between the two techniques at few instances was about 10%. The generally good agreement between the results from the two techniques can be attributed to the collimating system that was used to reduce scatter radiation incident onto the germanium detector during spectral acquisition. It was observed in earlier studies, that larger components of scattered radiation incident on the germanium detector could lead to the calculated ESAK and MGD from the spectral measurements to differ at higher tube voltages and larger phantom thicknesses by as much as 100% from those of the direct measurements, even though the first HVL was the same (Assiamah et al., 2004). For the same phantom thickness, the different exposure parameters gave significantly different ESAK and MGD values with lower nominal tube voltages resulting in higher ESAK and MGD values when compared with the corresponding higher nominal kVps even though the optical densities were close. Young et al., (1996) made similar observation. The dose reduction in using a higher nominal kVp for a set PMMA phantom thickness was significant. The extent of the reduction however, was found to be dependent on phantom thicknesses. In particular, using nominal 26 kVp or 29 kVp instead of nominal 25 kVp for 20 mm PMMA phantom thickness, resulted in a MGD reduction of about 30%. The accompanying loss in contrast from using the higher nominal kVp was 10% and 15% respectively. For a phantom thickness of 30 mm, the loss in contrast in using nominal 26 and 27 kVps was 3%. The observed dose reductions were, in these cases, 3% and 6% respectively (Table 2.2). Using nominal values of 28 and 29 kVp to expose a 30 mm phantom resulted in a reduction in dose of 11.5% and 20% respectively. The corresponding contrast reductions were 8 and 11% respectively. In the case of the 40 mm

Table 2.2: Comparison of ESAK and MGD values in mGy calculated from measured air kerma values and from measured spectra at nominal tube voltages 25, 26, 27, 28, 29, 30 and 32 kVp. All values are for a focal-spot-detector distance of 650 mm.

Nominal tube voltage (kVp)	PMMA thickness (mm)	Direct measurements		Measured spectra	
		ESAK (mGy)	MGD (mGy)	ESAK (mGy)	MGD (mGy)
25	20	1.63 ± 0.11	0.71 ± 0.14	1.6 ± 0.32	0.7 ± 0.14
26		1.16 ± 0.08	0.52 ± 0.10	1.14 ± 0.23	0.51 ± 0.11
29		1.14 ± 0.08	0.54 ± 0.11	1.21 ± 0.24	0.58 ± 0.11
25	30	3.1 ± 0.22	0.96 ± 0.19	3.14 ± 0.63	0.97 ± 0.19
26		2.94 ± 0.21	0.93 ± 0.19	2.68 ± 0.5	0.78 ± 0.16
27		2.71 ± 0.19	0.9 ± 0.18	2.66 ± 0.53	0.88 ± 0.18
28		2.51 ± 0.18	0.85 ± 0.17	2.57 ± 0.51	0.87 ± 0.18
29		2.34 ± 0.13	0.78 ± 0.12	2.52 ± 0.5	0.85 ± 0.17
25	40	8.14 ± 0.57	1.91 ± 0.38	7.98 ± 1.6	1.87 ± 0.37
26		6.02 ± 0.42	1.45 ± 0.29	5.95 ± 1.19	1.43 ± 0.29
27		5.82 ± 0.41	1.47 ± 0.3	5.31 ± 1.06	1.34 ± 0.27
28		5.23 ± 0.37	1.35 ± 0.27	5.05 ± 1.0	1.31 ± 0.26
29		4.54 ± 0.32	1.17 ± 0.23	5.05 ± 1.0	1.31 ± 0.26
29	50	9.31 ± 0.65	1.91 ± 0.38	10 ± 2	2.27 ± 0.45
30		8.3 ± 0.58	1.8 ± 0.36	8.12 ± 1.62	1.76 ± 0.35
32		6.41 ± 0.45	1.44 ± 0.3	6.51 ± 1.3	1.48 ± 0.3
32	60	13.41 ± 0.94	2.62 ± 0.52	13.03 ± 2.61	2.57 ± 0.51

PMMA phantom thickness, using nominal kVp values of 27, 28 and 29 instead of nominal 25 kVp resulted in a MGD reduction of about 24%, 29% and 40% respectively. The losses in contrast in using the higher nominal kVp were respectively 6%, 9% and 14%. The dose reduction from using nominal 30 and 32 kVp values instead of nominal 29 kVp, for a PMMA phantom thickness of 50 mm was 6% and 25% with a contrast loss of 3% and 20% respectively. The 60 mm PMMA phantom had the lowest contrast and the highest MGD value. The low contrast value observed in larger phantom thickness was ascribed partly to the larger value of tube loading needed to expose the phantom resulting in longer exposure times with consequent reciprocity law failure¹.

The dose reduction associated with higher tube voltages is due to the fact that for the same phantom thickness, higher tube voltages contain higher energy X-rays that penetrate the phantom with less absorption. The lower tube voltages on the other hand contain, relatively, a significant number of lower energy X-rays that are more easily absorbed by the phantom, hence higher dose values result. For all PMMA phantom thicknesses, it was generally observed that contrast decreases with increasing tube voltage and that higher contrast was achieved only by using lower tube voltage.

Linear regression fits were conducted on plots of the logarithm of the relative attenuation values with varying aluminium filter thicknesses, obtained with (Figures 2.2 and 2.3) and

¹ To generate a stable latent image, sensitivity speck in a grain of silver halide must accumulate a “critical mass” of silver atoms. If exposure rate is too slow i.e. long exposure times and therefore low flux, some silver atoms could be freed and leak away from the sensitivity speck resulting in criticality not achieved. This behaviour is referred to as reciprocity law failure. In the case of mammography for a given kVp and mAs, films become lighter as a result of long exposure times (Wolbarst, 1993).

without collimation (Figures 2.4 and 2.5) for both the spectral and direct measurements. Deviations from the linear regression fits obtained from both the measured spectra and direct measurements for two tube voltage settings were compared. Goodness-of-fit tests (Walpole and Myers, 1993) at 95% confidence interval were conducted for the attenuation values. The attenuation data at 26 and 28 kVp had degrees of freedom of 12 and 10 respectively. The chi-squared distribution (χ^2) obtained from the goodness-of-fit for the spectral measurements without collimation were 50.6 and 37.3 for the nominal X-ray tube voltages of 26 and 28 kVp respectively while those from the direct measurements were 23.0 and 18.4 (Figures 2.4 and 2.5). The χ^2 values from the uncollimated spectral measurements at both voltages were much higher than the specified χ^2 values in published statistical tables (21.0 and 18.3 for 12 and 10 degrees of freedom respectively) (Walpole and Myers, 1993), signifying substantial deviation from linearity.

The χ^2 values of the direct measurements for the two voltage settings however, were comparable to the values specified for reasonable fits. Unlike the large deviation in χ^2 values obtained for the spectral measurements for uncollimated beams, the deviations from the linear regression fits in the case of the collimated beams were found to be comparable to the published χ^2 values for both direct and spectral measurements for the two X-ray beams. For the collimated X-ray beams at 26-kVp nominal voltage setting, the goodness-of-fit test conducted on the attenuation values gave χ^2 values of 23.0 and 20.5 for the direct and spectral measurements respectively (Figure 2.2), whilst the χ^2 values were found to be 18.4 and 17.3 for the direct and spectral measurement fits at the nominal voltage setting of 28 kVp (Figure 2.3).

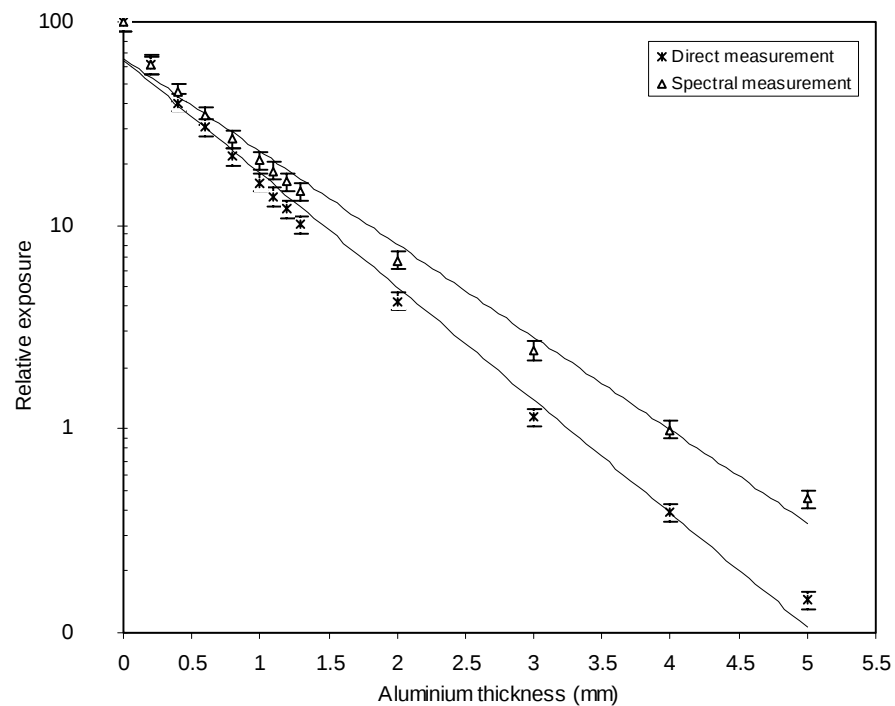


Figure 2.2: Comparison of relative attenuation curve calculated from measured spectrum of a collimated X-ray beam with the one measured directly for a tube voltage of 26 kVp.

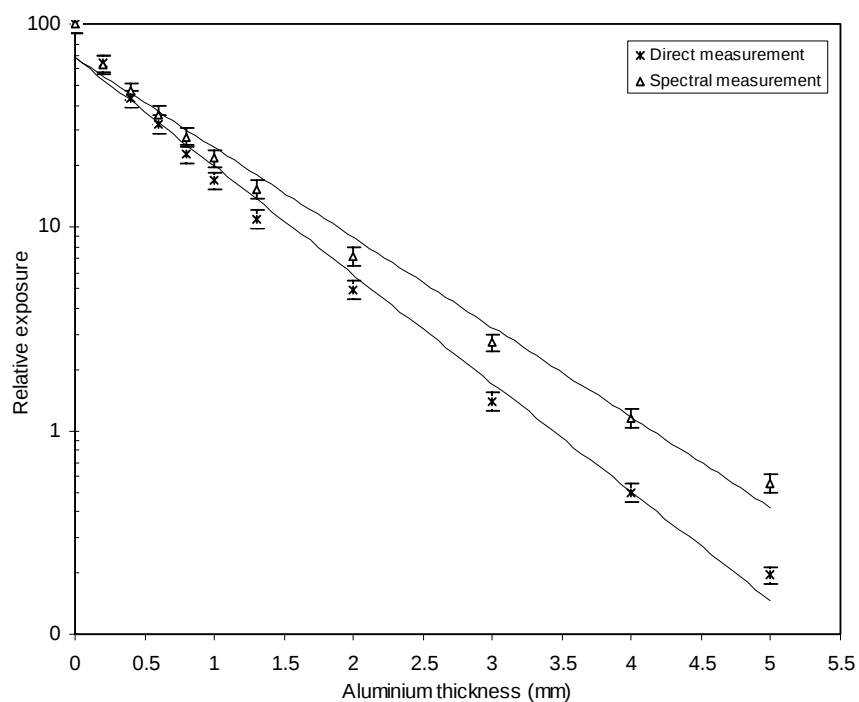


Figure 2.3: Comparison of relative attenuation curve calculated from measured spectrum of a collimated X-ray beam with the one measured directly for a tube voltage of 28 kVp.

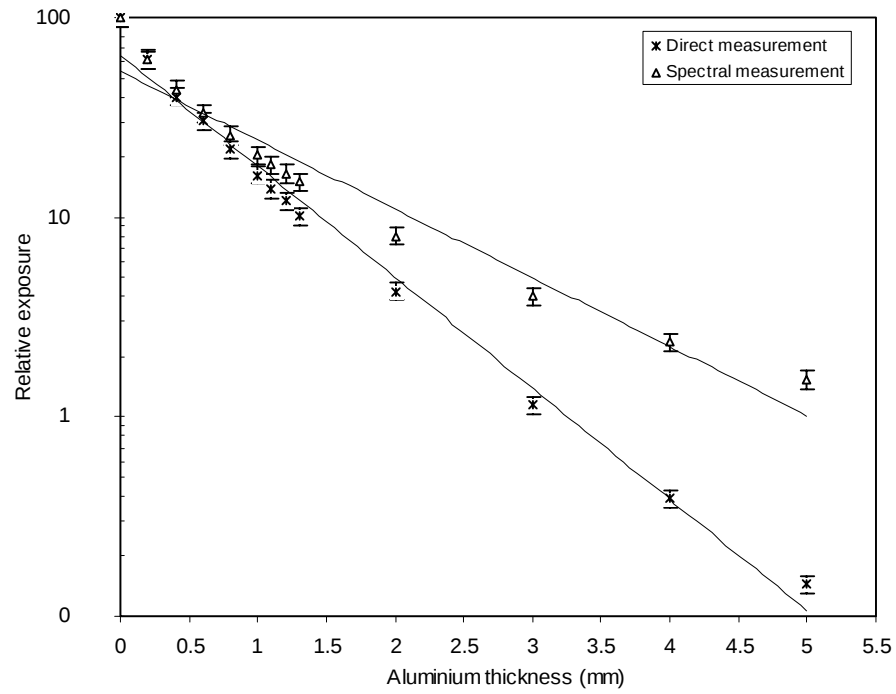


Figure 2.4: Comparison of the relative attenuation curve calculated from measured spectrum of an uncollimated X-ray beam with that measured directly for a nominal tube voltage of 26 kVp.

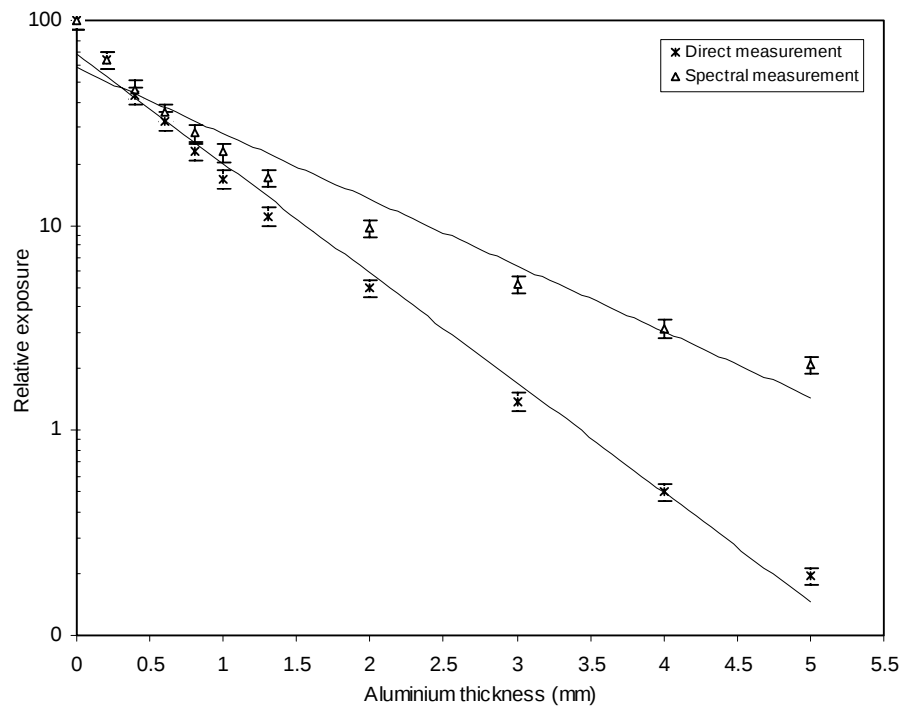


Figure 2.5: Comparison of the relative attenuation curve calculated from measured spectrum of an uncollimated X-ray beam with that measured directly for a nominal tube voltage of 28 kVp.

The significant reduction in deviations of the attenuation values of the spectral measurements from linearity fits when the X-ray beams were collimated lends further support to the observation that the spectral measurement method is sensitive to the presence of scatter radiation. The study has also shown that exposed film contrast is a function of tube voltages and phantom thickness (Figure 2.6); contrast decreasing with increasing tube voltage and phantom thickness. Figure 2.7 shows the film contrast at a nominal 28 kVp with varying tube loading for different PMMA phantom thicknesses for a range of optical densities. The optical density versus tube loading graphs of thinner phantoms had a larger gradient when compared to the values obtained from the larger thickness, implying that the selection of tube current for thinner phantoms should be done cautiously to achieve the right optical density and a good contrast. X-ray images of 30 mm phantom exposed at nominal tube voltage 28 kVp with 25 and 32 mAs (Figure 2.8) illustrates the effect of tube loading on contrast. For the same set tube voltage and tube loading, increasing phantom thickness decreases contrast as seen from Figure 2.9. This is because of the preferential removal of the lower energy X-rays with thicker phantoms and a consequent increase in the effective energy of the beam therefore.

For similar tube voltage and phantom thickness up to 40 mm, it was found that for every 10 mm increase in the phantom thickness, twice the mAs was needed to attain similar contrast. For larger phantom thicknesses however, more than twice the mAs was required for every 10 mm increase in order to achieve similar contrast. The observation suggests that the relation between contrast and phantom thickness has a steeper gradient for smaller thicknesses than for larger thicknesses.

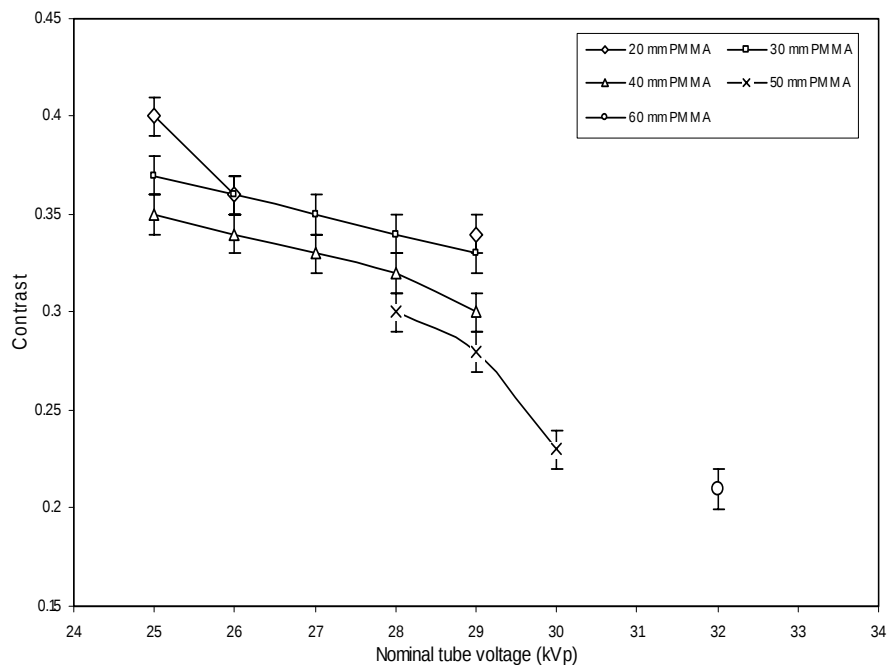


Figure 2.6: Contrast of 0.2 mm aluminium as a function of PMMA phantom thickness and tube voltage. The exposure setting was selected to give an optical density of 1.3.

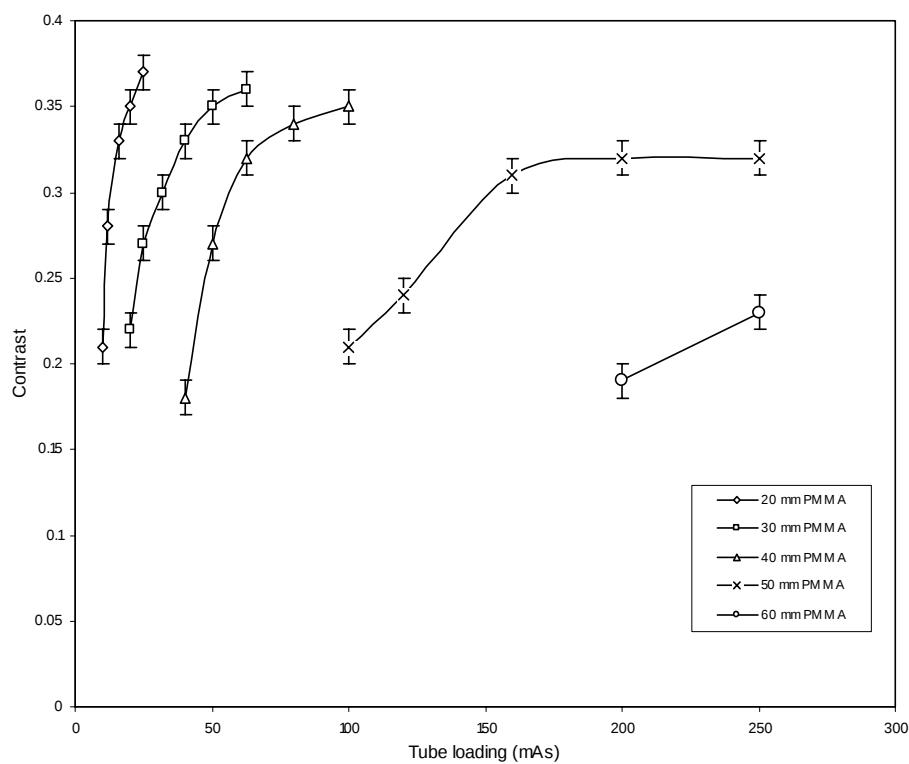


Figure 2.7: Variation of contrast for a range of optical density with tube loading for different thicknesses of PMMA phantom at a nominal tube voltage of 28 kVp.

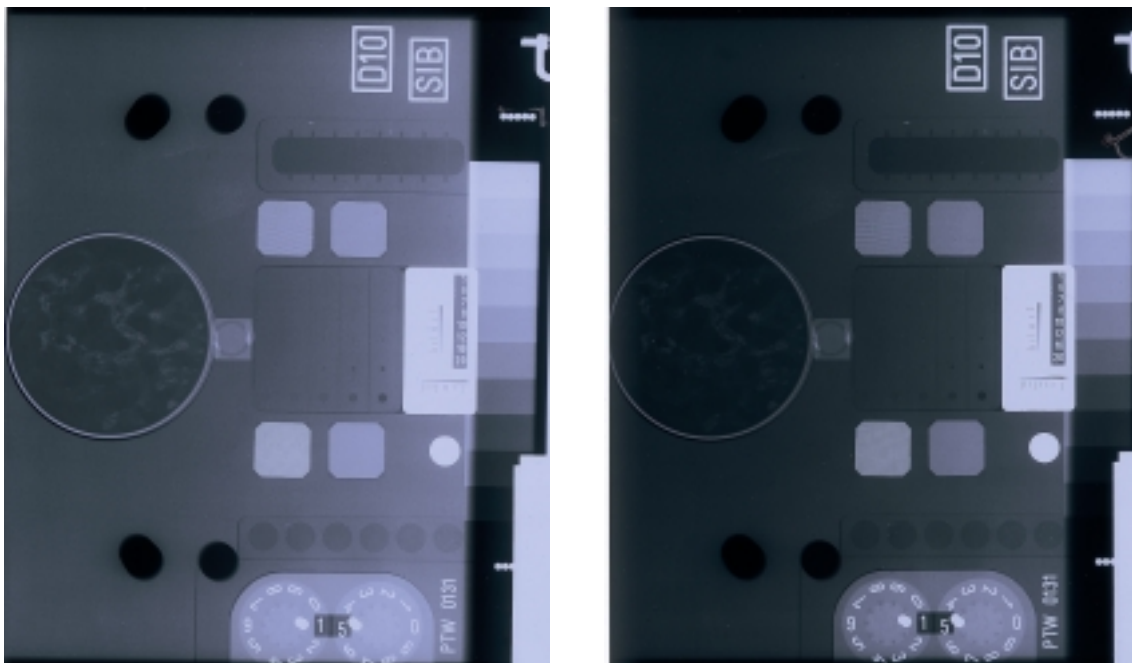


Figure2.8: X-ray images of 30 mm SIB phantom exposed at 28 kVp, 25 mAs (left) and at 28 kVp, 32 mAs (right).

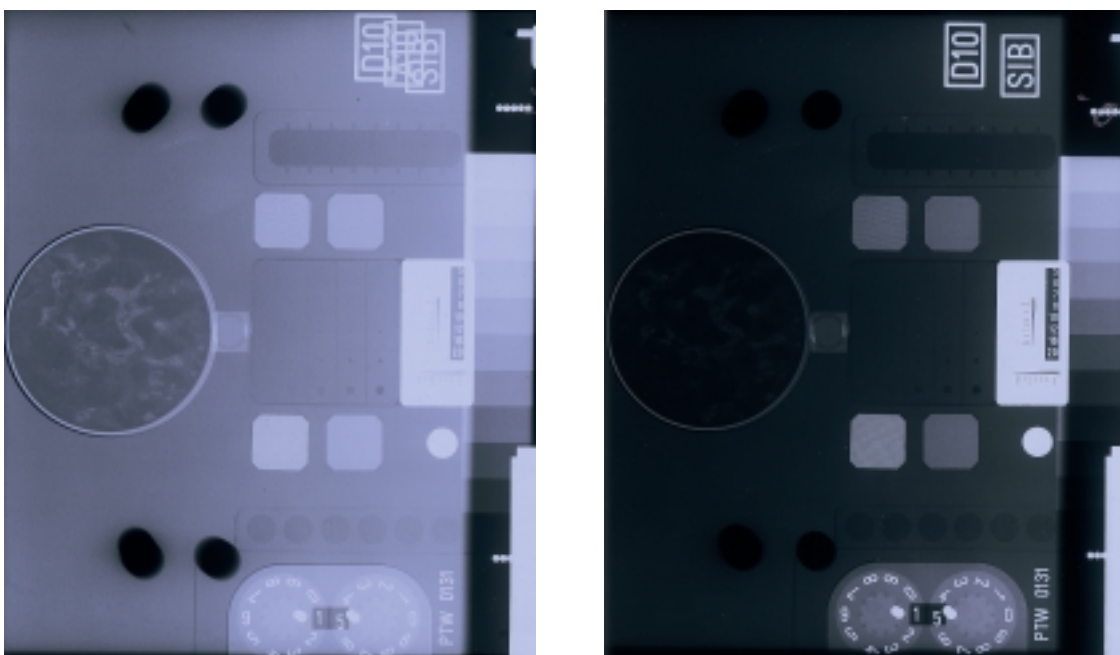


Figure 2.9: X-ray images of 40 mm (left) and 30 mm (right) SIB phantom exposed with identical X-ray tube parameters (28 kVp, 40 mAs).

2.5. CONCLUSION

The study has shown that both the direct and spectral measurement can be adopted to estimate the MGD. It can be concluded that the measured air kerma method can be as accurate as the measured spectral data method. It should however be borne in mind that values from the latter are based on large number of theoretically generated attenuation coefficient data that were required for the calculation. Highlighted in the studies is the sensitivity of the spectral measurement method to the presence of scatter radiation and the need for beam collimation and detector shielding in order to reduce radiation scattered onto the detector whilst acquiring spectral data. It can be concluded from the results of the study that films with similar optical density do not necessarily have similar contrast values. Optical density is also not a direct reflection of the deposited dose. The MGD values revealed that while the optical densities for some films were similar, their ESAK and corresponding MGD values were significantly different. It can further be concluded from the study that although using a higher tube voltage on thinner phantom thickness results in a lower MGD, the effect on contrast is significantly detrimental and cannot be ignored. For the same X-ray target material and same filter type, lower tube voltages should be used to expose thinner thicknesses and higher tube voltages for larger thicknesses. It has been found that for thinner phantoms, the best range of optical density is 0.9 – 2.0 and the equivalent contrast values are 0.25 – 0.38. For the thicker phantoms, the best optical density range is between 0.8 – 1.6 and the equivalent contrast values are 0.2 – 0.34. Optical densities outside these ranges could compromise contrast.

REFERENCES

- AAPM (American Association of Physicists in Medicine) (1990). Equipment requirements and quality control for mammography. AAPM Report No. 29, 45-61.
- American Cancer Society (2001). Surveillance research: breast cancer, facts and figures.
- Assiamah M., Mavunda D., Nam T.L., and Keddy R.J. (2003a). Effect of pressure, temperature and humidity in air on photon fluence and air kerma values at low photon energies. *Radiat. Phys. Chem.* 68 (5), 707-720.
- Assiamah M., Mavunda D., Nam T.L., and Keddy R.J. (2003b). Segmented multifit of polynomial function for mass attenuation and energy-absorption coefficient values. *Radiat. Phys. Chem.* 67 (1), 1-6.
- Assiamah M., Nam T.L., and Keddy R.J. (2004). Dosimetric techniques for mammography X-ray beams. Proceedings of the 9th International Symposium of Radiation Physics and Workshop on Radiation Based Analytical Techniques, 24-31 October 2003, Cape Town, South Africa. *Radiat. Phys. Chem.* 71, 957-958.
- Calicchia A., Gambaccini M., Indovina P.L., Mazzei F., and Puglianti L. (1996). Niobium/molybdenum K-edge filtration in mammography: contrast and dose evaluation. *Phys. Med. Biol.* 41, 1717-1726.
- CEC (European Commission) (1996). European protocol on dosimetry in mammography. European Commission, Luxembourg (EUR 16263).
- Cember H. (1996). Introduction to health physics (3rd ed.). McGraw-Hill, 169-179.
- Dance D. R. (1990). Monte Carlo calculation of conversion factors for the estimation of mean glandular breast dose. *Phys. Med. Biol.* 35, 1211-1219.

- Dance D. R., Skinner C. L., and Carlsson G. Alm. (1999). Breast dosimetry. Appl. Radiat. Iso. 50, 185-203.
- Desponds L., Klausz R., and Heidsieck R. (1994). Image quality and glandular dose for rhodium K-edge filter in mammography: performance comparison of the rhodium anode with molybdenum anode. Coolidge Sc. Review 1, 7-12.
- Fewell T. R. and Shuping R. E. (1977). Photon energy distribution of some typical diagnostic X-ray beams. Med. Phys. 4, pp. 187-196.
- GLOBOCAN (2000). Cancer incidence, mortality and prevalence worldwide, version 1.0, IARC Cancer Base No. 5. Lyon, IARC Press.
- Hubbell J. H. and Seltzer S. M. (1996). Tables of X-ray mass attenuation and mass energy absorption coefficients from 1 keV to 20 MeV for elements Z=1 to 92 and 48 additional substances of dosimetric interest. National Institute of Standards and Technology, U.S. Department of Commerce, Gaithersburg, MD 20899.
- ICRP (International Commission on Radiological Protection) (1987). Publication 52. Annals of the ICRP, (ICRP 17).
- IPSM (Institute of Physical Sciences in Medicine) (1989). The Commissioning and Routine Testing of Mammographic X-ray systems. IPSM Topical Group Report 59, IPSM, York.
- IPSM (Institute of Physical Sciences in Medicine) (1994). The Commissioning and Routine Testing of Mammographic X-ray systems. IPSM Topical Group Report 59 (2nd edition), IPSM, York.

- Johns H. E. and Cunningham J. R. (1983). The physics of radiology (4th edition). Thomas, Springfield, Illinois, 261-263.
- Klein R., Aichinger H., Dierker J., Jansen J. T. M., Joite-Barfub S., Sabël M., Schulz-Wendtland R., and Zoetelief J. (1997). Determination of average glandular dose with modern mammography units for two large groups of patients. Phys. Med. Biol. 42, 651-671.
- Mavunda R.D., Assiamah M., Nam T.L., and Keddy R.J. (2004). Bremsstrahlung spectra from diagnostic X-rays. Proceedings of the 9th International Symposium of Radiation Physics and Workshop on Radiation Based Analytical Techniques, Cape Town, South Africa, 24-31 October 2003. Radiat. Phys. Chem. 71, 991-992.
- NCRP (National Council on Radiation Protection and Measurements) (1986). Mammography— a user's guide. Bethesda, MD: National Council on Radiation Protection and Measurements, (NCRP Report 85), 40-56.
- National Cancer Institute (1999). Screening for breast cancer. Electronic text redistributed by University of Bonn, Medical Centre.
- Tucker D. M., Barnes G.T., and Charkraborty Dev P. (1991). Semi-empirical model for generating tungsten target X-ray spectra. Med. Phys. 18 (2), 211-218.
- Walpole R.E. and Myers R.H. (1993). Probability and statistics for engineers and scientists (5th edition), Prentice-Hall, Inc., Simon and Schuster Company, Englewood Cliffs, New Jersey 07632, 344-412.
- Wolbarst A. B. (1993). Physics of radiology (International edition). Appleton and Lange, Connecticut 06855 USA: 122-131, 185.

Wu X., Barnes G.T., and Tucker D.M. (1991). Spectral dependence of glandular tissue dose in screen-film mammography. *Radiology* 179, 143-148.

Young K.C., Ramsdale M.L., and Rust A. (1996). Dose and image quality in mammography with an automatic beam quality system. *The British Journal of Radiology* 69, 555-562.

