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The effects of x-ray beam hardening on detective quantum efficiency and radiation dose

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Abstract

The goal of this preliminary study was to investigate the effects of x-ray beam hardening on the detective quantum efficiency (DQE) and the radiation dose of an inline x-ray imaging system. The DQE and the average glandular dose were both calculated under the same experimental conditions for a range of beam hardening levels, corresponding to no added beam hardening and two thicknesses each of Rhodium (Rh) and Molybdenum (Mo) filters. The dose calculation results demonstrate a reduction of 15% to 24% for the range of beam hardening levels. The comparison of all quantities comprising the DQE exhibit very close correlation between the results obtained without added beam hardening to the results corresponding to the range of beam hardening levels. For the specific experimental conditions utilized in this preliminary study, the results are an indication that the use of beam hardening holds the potential to reduce the radiation dose without decreasing the performance of the system. Future studies will seek to apply this method in a clinical environment and perform a comprehensive image quality evaluation, in an effort to further evaluate the potential of beam hardening to balance the tradeoff between dose and image quality.

Keywords

diagnostic x-ray imaging; beam hardening; radiation dose; detective quantum efficiency (DQE); noise equivalent quanta (NEQ); modulation transfer function (MTF); noise power spectrum (NPS)

1. Introduction

Throughout the history of mammography, the technology and methods have evolved with a consistent goal of improving the tradeoff between the risk of harmful radiation dose to the patient and the benefit of disease detection provided by improved image quality [1–4]. However, the physical formulation of the images, which relies solely on attenuation contrast, has remained the same. The similar composition of benign and malignant breast tissue produces very low attenuation contrast [5–7], which presents a considerable challenge for cancer detection in mammography. Unfortunately, current techniques utilized in

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mammography to improve the image quality also deliver an increased radiation dose. As a result, the challenge remains to more effectively optimize the tradeoff between image quality and radiation dose.

2. Background

Breast cancer typically arises in the glandular tissue [5, 8]; therefore, the average glandular dose has been established as the standard measurement of radiation dose in mammography, and guidelines have been created by numerous national and international councils for its calculation and supervision in clinical environments [8–12]. The formula for calculation of the average glandular dose D_g is as follows [5, 13–14]:

$$D_g = D_{gN} \cdot X_{ESE}, \quad (1)$$

where D_{gN} is the normalized average glandular dose coefficient and X_{ESE} is the object entrance exposure. D_{gN} is determined by experimental and computer simulation methods based on the following factors: radiation quality (x-ray energy or half value layer), x-ray tube target material, filter material, breast thickness and breast tissue composition [5]. Due to the complicated nature of the calculations, as well as the small range of values in mammography for each of the D_{gN} calculation parameters, numerous studies have published tables of D_{gN} values that are widely used in clinical and research environments [13–16]. However, the values must typically be calculated directly in investigational studies, due to the use of parameters outside the standard values. This study will therefore apply a Monte Carlo process described previously [14, 17–18].

A comprehensive quantitative determination of the performance of an x-ray imaging system is the detective quantum efficiency (DQE) [5, 19–20], as it represents the transfer function of the SNR of the system, which is demonstrated by the following equation [5, 19, 21–24]:

$$DQE(f) = \frac{SNR_{OUT}^2}{SNR_{IN}^2} = \frac{S(0)^2 \cdot MTF(f)^2}{NPS(f) \cdot q} = \frac{NEQ(f)}{q}, \quad (2)$$

where the modulation transfer function (MTF) is a measurement of the frequency response of the system [5, 19, 23, 25–26], the noise power spectrum (NPS) is a determination of the noise processed by the system [5, 19, 23, 27–29], and $S(0)$ is the mean intensity of a large area signal, assuming a uniform x-ray beam absorbed by the detector. The noise equivalent quanta (NEQ) is a combination of the three quantities and describes the number of quanta recorded at each spatial frequency, which provides information regarding the maximum attainable value of the output SNR [22, 30–33]. The photon fluence (q) is the number of incident photons per square millimeter, which is equal to the square of the input SNR for x-ray photons following a Poisson distribution [5]. In this preliminary study, comparisons were made of the DQE as well as the individual quantities, in an effort to provide a comprehensive investigation of the effects of beam hardening on the imaging performance.

Beam hardening is the removal of low energy photons from the x-ray beam before reaching the object, through the use of filtration or other objects placed in the path of the x-ray beam [5]. Low energy photons are absorbed more readily due to lower penetration ability [5];

therefore, removing them prior to x-ray exposure has the potential to reduce the patient dose without considerably decreasing the image quality, which would benefit the field of mammography by delivering a more effective balance between image quality and radiation dose. The goal of this study was therefore to provide an initial investigation of the effects of a range of beam hardening levels on the DQE and the average glandular dose.

3. Materials and Methods

3.1. Beam hardening

Beam hardening was provided through the use of the commonly used mammography filters Mo and Rh [3–5]. The typical filter thicknesses utilized are 25 μm of Rh and 30 μm of Mo [5]. This study thus evaluated four filter thicknesses to provide a thorough comparison: 25 μm and 50 μm of Rh, and 30 μm and 60 μm of Mo. In addition, the investigation included a fifth mode without added filtration to provide a baseline for the comparison.

3.2. Inline x-ray imaging system and measurement components

The inline x-ray imaging system prototype utilized in this study consists of the imaging and measurement components mounted along an optical rail, which allows the use of a precise laser alignment process detailed in a previous study [34]. This also provides the ability to adjust the x-ray source-to-object (R1) and object-to-detector (R2) distances to facilitate the x-ray configuration utilized in this study, which is given in Fig. 1. In this study, R1 and R2 were both 91.44 cm, which produced a source-to-detector distance (R1+R2) of 182.88 cm. A longer distance than typical clinical mammography values was utilized to target future applications, as this configuration has been used for preliminary studies in inline phase contrast x-ray imaging indicating the potential to reduce the dose in mammography without negatively affecting the detection capability [7, 35–39]. In addition, the air gap in this configuration provides a comparable amount of scatter rejection and resultant image quality improvement as the grid used in clinical systems [6, 40–41]. The object was simulated by a BR12 phantom with a thickness of 5 cm. BR12 phantoms were developed to simulate a 50% glandular, 50% adipose tissue composition [42], and thus are used extensively for dose estimation purposes in clinical and research applications [2]. For comparison purposes, a second set of measurements were performed without the object, in an effort to evaluate the effect of the object on the measurements. The prototype system employs an x-ray tube with a Mo anode and a Be output window (UltraBright Microfocus Source, Oxford Instruments, Scotts Valley, California, USA). The tube operates between x-ray energies of 20 and 60 kV with an output power range of 10 to 60 W. The focal spot diameter varies with power output, thus a constant power output of 20 W was maintained in this study to ensure a constant focal spot diameter of 20 μm , which corresponds to a tube current of 0.33 milliamperes (mA). The image detection system employed in this study is a computed radiography system with mammography plates (Regius 190, Konica Minolta Medical Imaging, Wayne, New Jersey, USA), which provides a pixel pitch of 43.75 μm . Data linearization was performed through a process presented previously [26, 31, 43] to allow quantitative DQE analysis of the image data. The entrance exposure level was measured with a calibrated ionization chamber (10X9-180 ionization chamber, Model 9095 measurement system, Radcal Corporation, Monrovia, California, USA). Results from a

previous study [24] indicated that utilizing the mean of numerous measurements reduces the error; therefore five measurements at each mode were acquired. The x-ray spectrum was measured through the use of an x-ray spectrometer with a $3 \times 3 \times 1 \text{ mm}^3$ CdTe detector (Amptek Incorporated, Bedford, Massachusetts, USA), using a pair of tungsten collimators provided by the manufacturer.

3.3. Imaging parameters

An x-ray energy of 60 kV was utilized for the measurements acquired in this study. A higher x-ray energy than current clinical values was utilized in this preliminary study in an effort to target future clinical applications in phase contrast imaging, as initial investigations have indicated the potential of increasing the x-ray energy to reduce the dose without impacting the image quality [6, 39, 44–46]. Subsequent studies will further evaluate the feasibility of beam hardening through comparisons with a range of diagnostic energies, including current clinical mammography values. The exposure time for each filter/object combination was selected to provide a constant detector entrance exposure of 10 mR, in an effort to provide a comparison based on similar radiation exposure for image acquisition. Once the exposure time was determined for each filter/object combination, the corresponding object entrance exposure was measured for calculation of the average glandular dose. Due to the constant detector entrance exposure for all combinations, the differences among the DQE and dose values was a result of only the factors controlled by this study.

3.4. Average glandular dose measurements

As demonstrated in Eq. (1), the average glandular dose D_g is based on the object entrance exposure (X_{ESE}) and the average glandular dose coefficient (D_{gN}). The object entrance exposure measurements were detailed previously, and the D_{gN} values for each mode were estimated with Monte Carlo simulations through a process detailed in previous studies [14, 17–18]. The simulations assume the presence of an object with 50% adipose and 50% glandular tissue composition in the path of the x-ray beam. Therefore, the D_{gN} values were calculated only for the spectra acquired when the object is utilized.

3.5. Detective quantum efficiency measurements

The DQE calculation given in Eq. (2) consists of measurements of the MTF, NPS, $S(0)$ and q . First, the MTF was calculated through acquiring one image of a $10 \mu\text{m}$ wide slit camera (Nuclear Associates, Carle Place, NY, USA) for each filter/object combination. The MTF of each image was determined according to previous methods [25, 47–50] by normalizing the absolute value of the line spread function (LSF), which is the response of the imaging system to a line stimulus [5]. Next, the noise power spectrum was measured through acquisition of 20 images for each mode in the absence of any additional objects in the path of the x-ray beam, which is referred to as a noise-only image. Due to the stochastic nature of noise in x-ray images, the experimental calculation of the two-dimensional (2D) NPS typically involves separating the noise-only image into numerous smaller regions and averaging the NPS values calculated from each region [22, 27, 29, 31, 43, 51]. The one-dimensional (1D) NPS is then determined through the use of a slice consisting of four data lines parallel to and immediately adjacent to the axes [22, 43]. The large area signal $S(0)$

was determined by calculating the mean pixel intensity of the 20 noise-only images within the region of interest utilized for calculation of the noise power spectrum. The value of q is determined experimentally through multiplying the detector entrance exposure by the photon fluence per unit exposure, which can be calculated from a measured x-ray spectrum [24, 43, 52].

4. Results and Discussion

4.1. Average glandular dose

Comparisons of the average glandular dose calculated for each filter with the 5cm BR12 object are given in Table 1, which provides the measured object entrance exposure and the corresponding range of D_{gN} values calculated from the Monte Carlo estimations. The two values were multiplied together as defined by Eq. (1) to obtain the average glandular dose value (D_g). The last column presents the percentage of dose reduction provided by each filter as compared to the dose value obtained without filtration. The results demonstrate a notable dose reduction of 15% to 24% for the range of beam hardening levels. The filter corresponding to the maximum dose reduction was the 50 μm Rh filter. It is also interesting to note that the filter providing the second largest dose reduction was the 25 μm Rh filter, which provided a lower dose than both Mo filters.

4.2. Detective quantum efficiency

Comparisons of the x-ray spectra acquired for each beam hardening level are given in Fig. 2 (a) and (b), which correspond to the images acquired without an object and with the 5cm BR12, respectively. The comparison without the object demonstrates the effects of beam hardening in removing low energy photons from the x-ray beam, while the comparison with the object illustrates the further hardening effects of the object on the x-ray beam. Comparisons of the MTF values calculated for each beam hardening level are given in Fig. 3 (a) and (b), which correspond to the images acquired without an object and with the 5cm BR12, respectively. Next, comparisons of the NPS curves calculated for each beam hardening level are given in Fig. 4 (a) and (b), which correspond to the images acquired without an object and with the 5cm BR12, respectively.

Combining the MTF, NPS and $S(0)$, the NEQ comparisons of the beam hardening levels without an object and with the 5cm BR12 are provided in Fig. 5 (a) and (b), respectively. In the MTF, NPS and NEQ comparisons without the object, the curve calculated without added beam hardening is noticeably different from the curves corresponding to the filters. However, when the 5cm BR12 is utilized to simulate the human breast, the separation among the curves is much smaller, which is an encouraging indication that the use of filtration in a clinical environment may not negatively affect the imaging response of the system. Next, comparisons of the q values corresponding to the beam hardening levels are given in Tables 2 and 3, which correspond to the measurements acquired without an object and with the 5cm BR12, respectively, in the path of the x-ray beam. The results correlate with the D_{gN} coefficients, through exhibiting an increasing trend from the minimum value without added beam hardening to the maximum value with the 50 μm Rh filter, which also produced the largest D_{gN} value. This follows intuition, as the largest q value indicates the

highest concentration of photons in the same surface area, which would also correspond to the largest dose coefficient. Finally, comparisons of the DQE values for the beam hardening levels without an object and with the 5cm BR12 are provided in Fig. 6 (a) and (b), respectively. The results also exhibit much closer correlation between the curves acquired with the 5cm BR12 than the curves generated without the object. Furthermore, within the DQE curves corresponding to the object, once again only a small divergence is distinguishable between the curve acquired without beam hardening and the curves corresponding to the range of beam hardening levels.

5. Conclusion

For the specific experimental conditions utilized in this preliminary study, the results of the dose and DQE comparisons are an initial indication that the use of beam hardening holds the potential to reduce the dose without decreasing the performance of the system, which would benefit the fields of diagnostic x-ray imaging, especially mammography. Subsequent studies will further evaluate the feasibility of beam hardening through comparisons encompassing a range of diagnostic energies, including current clinical mammography values. In addition, a comprehensive image quality evaluation will be performed, in an effort to investigate the potential of beam hardening to balance the tradeoff between dose and image quality.

Acknowledgments

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Fig. 1.
Illustration of the x-ray configuration utilized in this study.

(a)

(b)

Fig. 2.

Comparison of the x-ray spectra corresponding to each beam hardening level. The spectra were acquired at 60 kV, 20 W (a) without an object in the path of the x-ray beam, and (b) with the 5cm BR12 object in the path.

(a)

(b)

Fig. 3.

Comparison of the MTF curves corresponding to the beam hardening levels. The curves were calculated from slit camera images acquired at 60 kV, 20 W (a) without an object in the path of the x-ray beam, and (b) with the 5cm BR12 object in the path.

(a)

Fig. 4. Comparison of the NPS curves corresponding to the beam hardening levels. The curves were calculated from noise-only images acquired at 60 kV, 20 W (a) without an object in the path of the x-ray beam, and (b) with the 5cm BR12 object in the path.

(a)

(b)

Fig. 5.

Comparison of the NEQ curves corresponding to the beam hardening levels. The curves were calculated from the MTF, NPS and $S(0)$, (a) without an object in the path of the x-ray beam, and (b) with the 5cm BR12 object in the path.

(a)

(b)

Fig. 6.

Comparison of the DQE curves corresponding to the range of beam hardening levels, (a) without an object in the path of the x-ray beam, and (b) with the 5cm BR12 object in the path.

TABLE 1

Average glandular dose calculations for an x-ray energy of 60 kV and 20 W with a 5cm BR12 phantom, assuming 50% adipose, 50% glandular tissue composition. The exposure time was selected for each filter to provide a constant detector entrance exposure of 10 mR. The results indicate a dose reduction for all filters.

Filtration	Object Exposure (R)	D_{gN} (mrad/R)	D_g (mrad)	Dose Reduction (%)
None	0.81	333.4	268.69	
25 μ m Rh	0.61	343.8	210.02	22
50 μ m Rh	0.56	366.6	203.82	24
30 μ m Mo	0.68	337.0	229.17	15
60 μ m Mo	0.63	356.0	222.86	17

TABLE 2

Comparison of the photon fluence (q) values for the beam hardening levels, corresponding to the measurements acquired without an object in the path of the x-ray beam.

Filtration	Photon Fluence (q) (1/mm²)
None	3.134×10^5
25 μm Rh	4.314×10^5
50 μm Rh	5.089×10^5
30 μm Mo	4.024×10^5
60 μm Mo	4.498×10^5

TABLE 3

Comparison of the photon fluence (q) values for the beam hardening levels, corresponding to the measurements acquired with the 5cm BR12 object in the path of the x-ray beam.

Filtration	Photon Fluence (q) (1/mm²)
None	7.087×10^5
25 μm Rh	7.636×10^5
50 μm Rh	8.288×10^5
30 μm Mo	7.618×10^5
60 μm Mo	8.243×10^5