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Abstract. Screening for breast cancer with mammography has been very successful, resulting in part to a reduction of breast cancer mortality by approximately 39% since 1990. However, mammography still has limitations in performance, especially for women with dense breast tissue. Iodinated contrast-enhanced, dedicated breast CT (BCT) has been proposed to improve lesion analysis and the accuracy of diagnostic workup for patients suspected of having breast cancer. A mathematical analysis to explore the use of various x-ray filters for iodinated contrast-enhanced BCT is presented. To assess task-based performance, the ideal linear observer signal-to-noise ratio (SNR) is used as a figure-of-merit under the assumptions of a linear, shift-invariant imaging system. To estimate signal and noise propagation through the BCT detector, a parallel-cascade model was used. The lesion model was embedded into a structured background and included a realistic level of iodine uptake. SNR was computed for 84,000 different exposure settings by varying the kV setting, x-ray filter materials and thickness, breast size, and composition and radiation dose. It is shown that some x-ray filter material/thickness combinations can provide up to 75% improvement in the linear ideal observer SNR over a conventionally used x-ray filter for BCT. This improvement in SNR can be traded off for substantial reductions in mean glandular dose. © 2017 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: [10.1117/1.JMI.4.1.013504](https://doi.org/10.1117/1.JMI.4.1.013504)]

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1 Introduction

Screening for breast cancer using mammography has been successful in the effort to reduce breast cancer mortality and its use has in part resulted in the 39% reduction in breast cancer mortality observed since 1990.¹ However, diagnostic mammography, defined as a separate procedure from screening in which supplemental mammographic views are acquired to assist radiologists in the evaluation of an area of clinical or radiographic concerns, remains an area of breast imaging that is in need of improvement.

A recent publication by Carney et al.² studied the minimal acceptable interpretive performance for diagnostic workup of the breast. Using normative data from the Breast Cancer Surveillance Consortium, they reported that the median positive predictive value of patients recommended for biopsy based on diagnostic mammography following an abnormal screening mammogram was 24.7% and that for symptomatic patients was 38.7%. Currently, approximately 700,000 breast biopsies are performed annually in the U.S., and thus it is estimated that over a half-million breast biopsies annually turn out to be unnecessary. In addition to the financial cost, these false positive biopsies result in increased patient anxiety, pain, and other confounding factors. Maybe more problematic is that some women choose to forgo mammography screening altogether after experiencing a false positive study or biopsy.³

Differentiating malignant from benign lesions in women with dense breast tissue can be especially challenging as dense tissue can obscure all or part of the lesion. Breast density has

a well-documented negative effect in screening mammography but the dense breast is also challenging even in the diagnostic imaging stage. One of the primary limitations of digital mammography (DM) is that it is a two-dimensional (2-D) imaging modality, and thus the overlapping breast tissues on the resulting image can make diagnosis of breast cancer challenging. Two promising tomographic breast imaging methods have been developed to reduce breast structure overlap, namely digital breast tomosynthesis (DBT)^{4,5} and dedicated breast CT (BCT).^{6–11} However, visualizing lesions with these new modalities can still be challenging for women with dense breast tissue due to the minimal contrast for lesions surrounded by fibroglandular tissue.

In recent years, there has been renewed interest in improving lesion conspicuity in x-ray breast imaging (using both 2-D and three-dimensional methods) by administration of an iodinated contrast agent.^{12–24} Results are encouraging, especially for the dense breast, but there are some inherent difficulties in the approach for full-field DM and DBT that have possibly limited its acceptance. Breast compression used with DM and DBT generates external pressure on the breast that is greater than the venous pressure. Because of this, venous outflow is restricted and the contrast agent uptake by the interstitium is decreased. This reduced contrast enhancement while the tissue is under compression, as well as other problems that decrease signal-to-noise ratio (SNR), such as patient motion, ultimately penalize the temporal subtraction technique that subtracts a precontrast scan from a postcontrast scan (both acquired under compression) to eliminate background structure. Since dedicated BCT does not use breast compression, it is ideally suited

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for contrast-enhanced imaging. Furthermore, BCT inherently eliminates structural overlap, and thus there is really no need for subtracting the precontrast scan from the postcontrast scan (for quantitative purposes, a region of interest can be drawn on both the pre- and postcontrast images and from these measurements lesion uptake can be estimated). There have been a number of clinical studies performed to investigate CT imaging of the breast using conventional whole-body scanners.^{25–27} Recently, a few clinical studies with iodinated contrast-enhanced dedicated BCT have been performed and results have been very encouraging.^{24,28}

A number of previous studies^{29–34} have evaluated preferred kV settings and added filtration for BCT. Only a few of these studies considered optimal kV settings for iodinated contrast-enhanced BCT applications. Prionas et al.³³ and Weigel et al.³¹ studied performance of contrast-enhanced BCT using a select few filter selections, namely Al, Cu, and Sn, with the former study using a cadaveric breast specimen, and the latter study using simulations and experimental phantoms. In both of these studies, investigators used a dose-weighted contrast-to-noise ratio (CNRD) figure-of-merit.

The study conducted here builds and improves on these previous studies and uses a theoretical model of task performance to further explore a broad range of filter materials and exposure techniques for iodinated contrast-enhanced imaging with BCT systems. Unlike previous studies that used the CNRD as a figure-of-merit, a Fourier domain, ideal linear observer SNR ratio is used here to provide a task-based performance assessment of optimal acquisition parameters under the assumption of a linear, shift-invariant imaging system. This figure-of-merit is computed in the projection domain and is thus desirable because it estimates an upper bound on task performance given certain approximations, independent of the particular linear reconstruction method used. The linear ideal observer models important imaging factors that are not included in the CNRD, in particular the anatomical and quantum noise structures that have been shown to play an important part in the analysis of breast imaging systems.³⁵ A dedicated, cone-beam BCT system that employs a CsI, amorphous-Si-based detector similar to that used in many of the early BCT prototypes is modeled. A parallel-cascade model was used to estimate realistic signal and noise propagation through the detector, and a lesion model with iodine uptake was embedded into a structured breast background. Ideal observer performance was investigated for varying kV settings, filter materials, filter thickness, breast sizes, and composition and radiation dose. Although many of the x-ray filter materials studied here might not be feasible for routine clinical use because of x-ray tube heat loading considerations, a comprehensive theoretical evaluation of all possible materials and filter thicknesses is still valuable because it provides useful insight into filter selection and could provide motivation for improving BCT x-ray tubes to allow for use of these filter materials and thicknesses in the future.

2 Methods

The task modeled herein is the detection of an iodinated contrast-enhanced lesion embedded into a structured, cylindrical breast background. The BCT imaging system is assumed to be linear and shift-invariant with both quantum and anatomical noise having stationary Gaussian probability densities. Under these assumptions, the ideal observer SNR can be expressed as

$$\langle \text{SNR}^2 \rangle = N_{\text{views}} \iint_{u,v} \frac{G^2 T_o(u,v) T_5(u,v) |\Delta S(u,v)|^2}{NPS(u,v) + P_f(u,v)} du dv, \quad (1)$$

where u and v are 2-D Cartesian frequency coordinates, G represents the detector gain after passing through the parallel-cascade model, T_o and T_5 represent the stochastic blur of the detector modeled by the parallel-cascade process discussed below and the blur introduced by the pixel aperture, respectively, ΔS represents the Fourier transform of the lesion projected into the image plane, NPS represents the 2-D quantum noise power spectrum (NPS), P_f represents the anatomical NPS (using a power-law noise model), and N_{views} is the number of projection angles. Equation (1) was evaluated numerically. Each term in Eq. (1) was estimated based on an FDA-approved Koning (Rochester, New York) BCT system that currently resides at the University of Massachusetts Medical Center.

2.1 X-Ray Spectra, Breast Composition

Various x-ray spectra were investigated using a tungsten anode spectral model developed by Boone and Seibert.³⁶ Discrete spectra at 1 keV intervals were generated for 30, 40, 50, 60, 70, 80, and 100 kV settings. To model external x-ray filtration, Beer's law of exponential attenuation was used with mass attenuation coefficients obtained from the National Institute of Standards and Technology website.³⁷ X-ray filters with atomic number (Z) ranging from 10 to 90 with filter thickness of 0 to 2500 mg/cm² were studied. The x-ray spectrum (i.e., the number of x-rays emitted at each energy) was scaled so as to provide a mean glandular dose (MGD) to the breast of 6.0, 8.0, and 10.0 mGy for a 12-, 14-, and 16-cm diameter breast model, respectively, using Monte Carlo derived glandular dose coefficient as previously described.^{38–40} These MGD levels were selected to approximate that of a recent clinical trial studying BCT in a diagnostic population.⁴¹ Since it is more likely that BCT would be used for women with dense breast tissue, two dense breast tissue compositions were modeled: (1) 25% fibroglandular and 75% adipose tissue, and (2) 50% fibroglandular and 50% adipose tissue. The former composition was observed to be close to the third density quartile of a patient cohort entered into a recent clinical study evaluating BCT.⁴² In both cases, breast tissue composition was modeled with a statistically defined background structure using a power-law spectrum.

2.2 Lesion Model

The shape of the lesion modeled was described as a “designer nodule signal” by Burgess et al.⁴³ given as

$$s(\rho) = \text{rect}(\rho/2)(1 - \rho^2)^\kappa, \quad (2)$$

where ρ is the normalized distance (r/R) from the center of the lesion and R represents the lesion radius. In this analysis, $\kappa = 1.5$ was selected. The lesion was positioned at the center of a cylindrical breast model. This lesion model was converted to Cartesian coordinates, $s(x, y)$, and the Fourier domain signal in Eq. (1), ΔS , was computed as

$$|\Delta S(u, v)|^2 = |(\bar{q}_{\text{in}}^B - \bar{q}_{\text{in}}^L) \text{FT}[s(x, y)]|^2, \quad (3)$$

Fig. 1 Flowchart illustrating stages 0 to 5 of the parallel-cascade model used in estimate noise propagation through the CsI-based a-Si detector.

where $\bar{q}_{in}^B$ and $\bar{q}_{in}^L$ are defined as the mean x-ray fluence incident on the detector through the background and the background plus center of the lesion, respectively.

To model iodine uptake within the lesion, the attenuation coefficient of the lesion was modeled as having two components,^{44,45} one containing invasive ductal carcinoma and another with a mixture of iodine and blood

$$\mu_L = P_{IDC}\mu_{IDC} + (1 - P_{IDC})[P_I\mu_I + (1 - P_I)\mu_B], \quad (4)$$

where P_{IDC} is the fraction of tissue within the lesion that has x-ray attenuation properties of invasive ductal carcinoma. It is assumed that the other component in the lesion is made up of an iodine/blood mixture, where P_I represents the fraction of iodine within this mixture, and μ_I and μ_B represent the linear attenuation coefficient of iodine and blood, respectively. In this study, we use measurements reported by Ref. 46 to model μ_{IDC} and assume that $P_{IDC} = 0.6$. Based on a previous clinical study of iodinated contrast mammography,²³ as well as other studies evaluating contrast breast imaging,^{22,47,48} the density of iodine within the lesion was taken as .0042 g/ml. Assuming the density of blood (ρ_{blood}) is 1.06 g/ml,⁴⁵ then $P_I = .004$.

2.3 Quantum Noise Power Spectrum

To estimate the propagation of noise through the detector, a parallel-cascade model was used as described by Cunningham et al and others.^{49–51} A flowchart illustrating the first four stages of the parallel-cascade model is shown in Fig. 1. This model requires the estimate of a number of detector parameters including the energy-dependent quantum detection efficiency, the mean number of quanta produced at each of the three paths, the mean gain along each path, the probability of a K x-ray being reabsorbed with the scintillator, the fluorescent yield, and detector blurring models. The methods used to estimate

Table 1 Definition and values of parameters used in the parallel-cascade model.

Parameter	Value
CsI:TI thickness (μm)	491
Fluorescent yield (ω)	0.87
Probability of K reabsorption	0.66
Fill factor	0.57
Number of projection angles	300
Electronic noise	2000 e^-
Breast thickness (cm)	12, 14, 16
μ_a (adipose) and μ_g (glandular)	Johns and Yaffe ⁴⁶
C (breast composition)	25%/75% 50%/50%

the NPS follow closely from Ref. 29, and thus we will provide only a brief overview here.

The mean number of optical photons leaving stage 2 (see Fig. 1) can be expressed as

$$\bar{q}_2 = \bar{q}_{2a} + \bar{q}_{2b} + \bar{q}_{2c}, \quad (5)$$

where $\bar{q}_{2a} = \bar{q}_{in}\bar{g}_1(1 - \zeta\omega)\bar{m}_A$, $\bar{q}_{2b} = \bar{q}_{in}\bar{g}_1\zeta\omega\bar{m}_B$, and $\bar{q}_{2c} = \bar{q}_{in}\bar{g}_1\zeta\omega f_k\bar{m}_C$.

These variables are defined in Table 1 and further described in the parallel-cascade model flowchart of Fig. 1. ζ is the probability of a K-shell interaction relative to all possible interactions, ω is the fluorescent yield, f_k represents the probability of a K-shell characteristic x-ray being reabsorbed within the scintillator, and $\bar{m}_A$, $\bar{m}_B$ and $\bar{m}_C$ are the mean gains for paths A, B, and C as indicated in Fig. 1.

The NPS at the output of stage 2 can be expressed as

$$NPS_2 = NPS_{2A} + NPS_{2B} + NPS_{2C} + NPS_{BC,2} + NPS_{CB,2}, \quad (6)$$

where

$$NPS_{2A} = \bar{q}_{in}\bar{g}_1(1 - \zeta\omega)\bar{m}_A[\bar{m}_A + 1 + \epsilon_A], \quad (7)$$

$$NPS_{2B} = \bar{q}_{in}\bar{g}_1\zeta\omega\bar{m}_B[\bar{m}_B + 1 + \epsilon_B], \quad (8)$$

and

$$NPS_{2C} = \bar{q}_{in}\bar{g}_1\zeta\omega f_k\bar{m}_C[\bar{m}_C + 1 + \epsilon_C]. \quad (9)$$

The Poisson excess variables ϵ_A , ϵ_B , and ϵ_C for each pathway express the relative amount by which the variance of the detector gain process exceeds that obtained with Poisson amplification. Since the light generated in paths B and C is statistically correlated, there exists a cross-spectral density term that can be expressed as⁵⁰

$$NPS_{BC,2} + NPS_{CB,2} = 2\bar{q}_{in}\bar{g}_1\zeta\omega f_k\bar{m}_B\bar{m}_CT_k(u, v), \quad (10)$$

where T_k is the stochastic blur due to characteristic K x-ray reabsorption. Appropriate parameters in Eqs. (6)–(10) were

computed using the GEANT 3 Monte Carlo software package as described in Ref. 52. Stage 3 illustrated in the parallel-cascade flowchart of Fig. 1 represents the stochastic blur, $T_o(u, v)$ introduced by the spreading of optical photons generated from each of the three pathways as they travel toward the detector pixels. This function was estimated by weighing an empirically measured presampled modulation transfer function (MTF) as suggested by Ref. 50. Using this stochastic blur, the NPS at stage 3 is expressed as

$$\text{NPS}_3 = [\text{NPS}_2 - \bar{q}_2]T_o^2(u, v) + \bar{q}_2. \quad (11)$$

Stage 4 describes the absorption of optical quanta by the a-Si pixelized detector and is modeled as a stochastic gain stage that follows a binomial selection process. The NPS at stage 4 is given as

$$\text{NPS}_4 = \bar{g}_4^2 \text{NPS}_3 + \bar{q}_2 \bar{g}_4 (1 - \bar{g}_4), \quad (12)$$

with $\bar{g}_4$ being the optical collection efficiency representing coupling of optical quanta to the pixelized a-Si detector. Stage 5 models the integration of quanta over each pixel and is modeled as a deterministic blurring as

$$T_5(u, v) = |\text{sinc}(\pi a_{pd} u) \text{sinc}(\pi a_{pd} v)|, \quad (13)$$

where a_{pd} represents the sampling aperture, which is related to the pixel pitch (a_{pix}^2) by the fill factor (ff)

$$a_{pd}^2 = \text{ff}(a_{\text{pix}}^2). \quad (14)$$

The output NPS at stage 5 can then be expressed as

$$\text{NPS}_5 = a_{pd}^4 \text{NPS}_4 T_5^2(u, v). \quad (15)$$

By assuming the same noise propagation for primary and scattered x-rays, the effect of scattered radiation within the breast can be included in the analysis as

$$\text{NPS}_6 = (1 + \text{SPR})\text{NPS}_5, \quad (16)$$

where SPR is the scatter-to-primary ratio. The SPR used in Eq. (16) was estimated for each energy spectrum, x-ray filter, and breast size by using Monte Carlo simulation to track monoenergetic x-rays (10 to 100 keV) incident on varying diameter cylindrical breast models. Pixel sampling was then modeled as the convolution of Eq. (16) with the Fourier transform of the sampling grid that possibly introduces aliasing into the NPS

$$\text{NPS}_6^\dagger = \text{NPS}_6^{**} \sum_{m,n=-\infty}^{\infty} \delta(u - mu_s, v - nv_s), \quad (17)$$

where δ is a 2-D comb function with u_s and v_s representing the sampling frequencies in Cartesian coordinate frequency space. The last stage models electronic noise and thus the final output NPS can be expressed as

$$\text{NPS}_{\text{final}} = \text{NPS}_6^\dagger + \text{NPS}_{\text{add}}, \quad (18)$$

where the NPS for the additive electronic noise is modeled as

$$\text{NPS}_{\text{add}} = \frac{\sigma_{\text{add}}^2}{4u_{\text{nyq}}v_{\text{nyq}}}, \quad (19)$$

Fig. 2 T_k , T_5 , and T_k MTFs used in the parallel-cascade model.

and σ_{add}^2 represents the variance of the electronic noise (modeled as Gaussian white noise) and u_{nyq} and v_{nyq} represent the Nyquist frequency in the u and v frequency coordinates, respectively.

Table 1 shows pertinent parameter values used in computing SNR (Eq. 1) using the parallel-cascade model discussed above. These parameters were chosen to roughly model a commercial dedicated BCT system (Koning Corp., Rochester, New York) that employs a Varian 4030CB flat-panel detector. This detector uses a CsI scintillator with nominal thickness of 600 μm . We assumed a packing density of 81%, thus simulations performed herein used an estimated scintillator thickness of 491 μm . Estimates for the probability of K reabsorption and T_k were computed using the GEANT3 Monte Carlo simulation.⁵² Figure 2 shows the T_k , T_3 , and T_5 MTFs used in the analysis.

2.4 Anatomical Noise Power Spectrum

The anatomical NPS of BCT projections was assumed to follow an isotropic power-law given as

$$P_f(f) = \frac{A}{f^\beta}, \quad (20)$$

where f represents the radial frequency coordinate with $f^2 = u^2 + v^2$ and A is a constant and the units of the power spectrum are in $e^2 \text{ mm}^2$. Based on previous studies,^{35,53} the exponent β was assumed to be 3.0.

3 Results and Discussion

Figures 3 and 4 show the two basic measurements that provide insight into the ideal observer SNR results as computed by Eq. (1). Figure 3 shows the photon fluence incident on the detector face (q_{in}) through the center of the three sized phantoms (and through the lesion) as a function of monoenergetic energy. As expected, the photon fluence generally increases with increasing energy. Since the model assumes varying MGD for each size phantom, x-ray fluence does not necessarily decrease with increasing phantom thickness. Figure 4 shows the iodinated lesion contrast in a projection image as a function of energy for the three sized phantoms [contrast = $(\bar{q}_{\text{in}}^B - \bar{q}_{\text{in}}^L)/\bar{q}_{\text{in}}^B$]. As expected, there is a large spike in lesion contrast that occurs

Fig. 3 Photon fluence incident on the detector behind the center of the breast model versus energy. Photon fluence is shown for three different size breast models.

Fig. 4 Lesion contrast versus energy for three different size breast models.

at approximately 33 keV, the K-edge of iodine. Based on this graph, one would expect that x-ray spectra with predominant energies just above the K-edge would offer improved performance. Thus Figs. 3 and 4 clearly show that the selection of optimal spectra involves a tradeoff between maximizing photon fluence (i.e., reducing noise) that increases with x-ray energy and maximizing subject lesion contrast that generally decreases with increasing x-ray energy. Dose considerations also contribute to this tradeoff.

The ideal observer SNR [as expressed in Eq. (1)] is a task-based figure-of-merit that provides an upper bound on information content in the projection data. Figures 5–7 show color-coded 2-D SNR maps as a function of filter thickness and filter material atomic number (Z_{filter}) for three different

diameter breast phantoms, 12, 14, and 16 cm (Figs. 5, 6, and 7, respectively). Each figure shows two breast compositions, 25% and 50% fibroglandular tissue. Note that thickness is given in units of mg/cm^2 , thus material density is required to determine linear thickness. SNR maps are shown for multiple kVp settings from left to right columns including 30, 40, 50, 60, 70, 80, and 100 kVp. The relative color scale for the 2-D SNR maps color is displayed on the right side of the figure. The reader is reminded that the MGD used in computing SNR for phantom sizes of 12, 14, and 16 cm are 6, 8, and 10 mGy, respectively, as determined from a recent clinical BCT study.⁴¹ From these SNR maps, it can be observed that filter thickness and filter material can greatly affect SNR. In general, the best performing combinations of filter and thickness (yellow, orange, and red colors) occur for 40 to 70 kVp spectra. Observing various filter materials, it is observed in general that filters with $Z = 40$ to 55 perform very poorly (horizontal black and blue stripe in SNR maps of 40 to 80 kVp), with a strong increase in filter performance for filters with $Z = 55$ to 70. It was observed that increased breast composition decreased overall performance; however, the observed trends seemed to be similar for both densities.

Figures 5–7 show the SNR performance for a wide range of filter material/thickness combinations, some of which require aggressive x-ray filtration that would not be feasible in practice due to high x-ray heat loading. However, the data as presented here are still useful in providing an idealized view of how performance with iodinated contrast-enhanced BCT can be improved with use of filters. The clinical prototype system that currently resides at the University of Massachusetts (Koning Corp., Rochester, New York) uses a 3-phase pulsed M-1581 x-ray tube (Varian Medical Systems) with 1.5 mega-heat units heat loading capacity. A typical BCT scan with exposure technique of 49 kV and 125 mA would have power consumption of $1.35 \times 49 \text{ kV} \times 125 \text{ mA} \times 8 \text{ ms}$ (pulse width) $\times 300$ (angles) = 19.8 kHU. Thus, imaging with a tenth value layer filter thickness would be entirely feasible. Figure 8 shows SNR maps for the same filter materials and thicknesses as shown in the top row of Fig. 6 (i.e., 14-cm breast diameter), but with the SNR set to zero for all filters that transmit less than 10% of the x-ray beam. Although using this tenth value layer constraint on x-ray transmission removes many filter/thickness combinations, it is observed that many high performing filter/thickness combinations are still feasible.

The graph in Fig. 9 shows SNR values for a selected number of feasible filter/thickness combinations: (a) 1.6-mm Al, (b) 0.2-mm Cu, (c) 0.3-mm Gd, (d) 0.25-mm Er, and (e) 0.3-mm Ce. These were computed with various x-ray spectra (30 to 100 kV) for a 14-cm diameter breast at 8-mGy dose. The trends for the other breast thicknesses studied were similar. Figure 10 shows accompanied transmission fractions through selected filter/thickness combinations. One of these filters examined (1.6-mm Al) was selected to approximate that used in a clinical BCT system (Koning Inc., Rochester, New York). Another filter (0.2-mm Cu) was suggested in Prionas et al.³³ as an optimal selection using a 60-kV spectra. The choices of Gd and Er shown in Fig. 9 were selected because of demonstrated higher performance, availability, and ease of use. Of these filters examined, it is observed that some combinations can provide substantial improvement over the conventional 1.6-mm Al filter. The 0.2-mm Cu filter/thickness at 60 kV suggested as optimal by Prionas et al.³³ provided an improvement of 32%

Fig. 5 Ideal linear observer SNR maps for the 12-cm breast diameter. Each color-coded map shows SNR for materials with $Z = 10$ to 90 (y -axis) and thickness ranging from 0 to 2500 mg/cm^2 . The color bar on the right describes the relative SNR ranking (1.0 is highest SNR for the 12-cm breast diameter). SNR maps for the 25% and 50% fibroglandular composition phantoms are shown in rows 1 and 2, respectively. SNR maps are shown for multiple kVp settings 30, 40, 50, 60, 70, 80, and 100 kVp in columns from left to right.

Fig. 6 Ideal linear observer SNR maps for the 14-cm breast diameter. Each color-coded map shows SNR for materials with $Z = 10$ to 90 (y -axis) and thickness ranging from 0 to 2500 mg/cm^2 . The color bar on the right describes the relative SNR ranking (1.0 is highest SNR for the 14-cm breast diameter). SNR maps for the 25% and 50% fibroglandular composition phantoms are shown in rows 1 and 2, respectively. SNR maps are shown for multiple kVp settings 30, 40, 50, 60, 70, 80, and 100 kVp in columns from left to right.

over the conventional 1.6-mm Al filter, whereas the Gd and Er filters provided an even higher improvement of 70% and 75%, respectively.

Some insight into how spectral shaping can affect performance for iodinated contrast-enhanced BCT can be gained by observing the x-ray spectra for these filter/thickness combinations. Figure 11 shows x-ray spectra for (a) 1.6-mm Al, (b) 0.2-mm Cu, (c) 0.3-mm Gd, (d) 0.25-mm Er, and (e) 0.3-mm Ce. It is observed that the best performing filter/thickness combinations, 0.3 mm of Gd and 0.25 mm Er, both shape the x-ray spectra to provide the majority of x-rays with energy just above the K-edge of iodine. In particular, Gd is especially an appealing material in that it has a K-edge of approximately 50 keV, thus the spectra consist of x-rays with predominant

energies between 30 and 50 keV. The Ce filter, which has a K-edge of approximately 40 keV, has been shown to perform well for BCT without administration of iodine contrast agent;^{29,30} however, it is not as well suited for contrast imaging because the x-ray spectra after Ce filtering straddles the iodine K-edge of 33 keV.

Improvement in SNR achieved with certain filters can be traded off for decreased radiation dose without reduction in performance. Figure 12 shows the MGD for the different filter/thickness combinations that were plotted in Fig. 9, where for each case dose was adjusted until the SNR was equivalent to that achieved with the standard filter/thickness combination of Al 1.6 mm. The MGD with the standard Al 1.6-mm filter and 12-, 14-, and 16-cm phantoms was 6, 8, and 10 mGy,

Fig. 7 Ideal linear observer SNR maps for the 16-cm breast diameter. Each color-coded map shows SNR for materials with $Z = 10$ to 90 (y -axis) and thickness ranging from 0 to 2500 mg/cm^2 . The color bar on the right describes the relative SNR ranking (1.0 is highest SNR for the 16-cm breast diameter). SNR maps for the 25% and 50% fibroglandular composition phantoms are shown in rows 1 and 2, respectively. SNR maps are shown for multiple kVp settings 30, 40, 50, 60, 70, 80, and 100 kVp in columns from left to right.

Fig. 8 Ideal linear observer SNR maps for the 12-, 14-, and 16-cm diameter breast model with 25% fibroglandular composition. SNR values are set to zero for filter material/thickness combinations that transmit less than 10% of the x-ray beam.

respectively. Results shown in Fig. 12 demonstrate that substantial reductions in dose can be achieved by using selected filter/thickness combinations. For example, it is observed that using the Gd 0.3-mm or Er 0.25-mm filter can provide a reduction in MGD of 30% to 50%.

There are some limitations of this study. Based on previous studies,^{22,23,47,48} we assumed a representative lesion iodine uptake of 4.2 mg/ml . Iodine concentration in the lesion is

affected by many factors, including imaging time, lesion type, patient weight, amount of iodine injected, and wash-in/wash-out kinetics, and is thus difficult to model. Since the primary lesion contrast mechanism in contrast-enhanced BCT is taking advantage of the k -edge of iodine, we assume that lesion iodine concentration will have less of an impact on the trends observed herein. The analysis herein computed the ideal observer on the projection data, and thus the anatomical structure of the

Fig. 9 SNR versus kV for 14-cm diameter, 25% fibroglandular breast composition with 8-mGy dose. Results are shown for six selected filter material/thickness combinations, including Al 1.6 mm, Al 9.0 mm, Gd 0.30 mm, Er 0.25 mm, Ce 0.30 mm, and 0.20 mm Cu.

Fig. 10 The fraction of x-rays transmitted through the six selected filter material/thickness combinations is shown in Fig. 9 as a function of kV setting.

breast was modeled with power-law noise using a β of 3.0 as reported by Burgess et al.³⁵ This model assumes stationary noise, whereas a nonstationary noise model might provide different results.

4 Conclusions

In this study, a theoretical framework was used to assess possible performance improvements in iodinated contrast-enhanced BCT when using various x-ray filters to shape x-ray spectra. An ideal linear observer computed in the projection domain was used as the figure-of-merit, providing insight into the upper bound on achievable performance with various filters regardless of reconstruction filter. To compute this figure-of-merit, models were developed for iodinated lesion uptake, and signal and noise

Fig. 11 X-ray spectra for five filter material/thickness combinations.

Fig. 12 MGD for 60-kV spectra with five filter material/thickness combinations as a function of breast diameter. For each breast diameter, the SNR of each filter is equivalent (equal to that obtained with the Al 1.6-mm filter).

transport through a CsI-based flat-panel detector. To estimate noise transport, a linear parallel-cascade model was used with specific parameters estimated from Monte Carlo simulation. Ideal observer SNR for 84,000 different combinations of x-ray filter material, thickness, kV, and breast size were computed, and it was observed that many x-ray filter/thickness combinations can provide substantial improvement in performance over the standard 1.6-mm Al filter used for BCT, as well as the 0.2-mm Cu filter recommended for iodinated contrast-enhanced BCT in a previous study.³³ However, many of these combinations require aggressive filtering and are thus impractical with current x-ray generation technology due to x-ray tube heat loading. Nonetheless, there are a number of high performing filter material/thickness combinations with feasible heat loading characteristics. In particular, it was shown that using 0.3 mm of

Gd or 0.25-mm Er as an x-ray filter could provide up to 75% increase in ideal observer SNR without excessive heat loading (i.e., with greater than 10% of x-rays transmitting through the filter). This improvement in SNR can be traded off for decrease in MGD. It was estimated that 30% to 50% reduction in MGD could be achieved by selecting the appropriate filter/thickness combination. In the future, this reduction in dose could be important if iodinated contrast-enhanced BCT becomes a more routinely used tool for breast lesion analysis.

Disclosures

The authors have no financial interests or potential conflicts of interest to disclose.

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