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Towards Standardization of X-ray Beam Filters in Digital Mammography and Digital Breast Tomosynthesis: Monte Carlo simulations and analytical modelling

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Abstract

In digital breast tomosynthesis and digital mammography, the x-ray beam filter material and thickness vary between systems. Replacing K-edge filters with Al was investigated with the intent to reduce exposure duration and to simplify system design. Tungsten target x-ray spectra were simulated with K-edge filters (50 μ m Rh; 50 μ m Ag) and Al filters of varying thickness. Monte Carlo simulations were conducted to quantify the x-ray scatter from various filters alone, scatter-to-primary ratio (SPR) with compressed breasts, and to determine the radiation dose to the breast. These data were used to analytically compute the signal-difference-to-noise ratio (SDNR) at unit (1 mGy) mean glandular dose (MGD) for W/Rh and W/Ag spectra. At SDNR matched between K-edge and Al filtered spectra, the reductions in exposure duration and MGD were quantified for three strategies: (i) fixed Al thickness and matched tube potential in kilovolts (kV); (ii) fixed Al thickness and varying the kV to match the half-value layer (HVL) between Al and K-edge filtered spectra; and, (iii) matched kV and varying the Al thickness to match the HVL between Al and K-edge filtered spectra. Monte Carlo simulations indicate that the SPR with and without the breast were not different between Al and K-edge filters. Modelling for fixed Al thickness (700 μ m) and kV matched to K-edge filtered spectra, identical SDNR was achieved with 37–57% reduction in exposure duration and with 2–20% reduction in MGD, depending on breast thickness. Modelling for fixed Al thickness (700 μ m) and HVL matched by increasing the kV over [0,4] range, identical SDNR was achieved with 62–65% decrease in exposure duration and with 2–24% reduction in MGD, depending on breast thickness. For kV and HVL matched to K-edge filtered spectra by varying Al filter thickness over [700,880] μ m range, identical SDNR was achieved with 23–56% reduction in exposure duration and 2–20% reduction in MGD, depending on breast thickness. These simulations indicate that increased fluence with Al filter of fixed or variable thickness substantially decreases exposure duration while providing for similar image quality with moderate reduction in MGD.

1. INTRODUCTION

The x-ray beam filter is an essential component of a breast imaging system for the spectral shaping of the beam for a given target (anode), which affects image contrast and radiation

dose. Commonly used x-ray beam filters are molybdenum (Mo), rhodium (Rh), silver (Ag) and aluminium (Al). The x-ray beam filter material and its thickness vary among current mammography and digital breast tomosynthesis (DBT) systems (Sechopoulos, 2013, Vedantham et al., 2015). For DM, most systems use K-edge filtration such as Mo, Rh and Ag, and at least one system uses Al filter (Microdose, Koninklijke Philips N.V.). For DBT, several manufacturers use Al filtration (e.g., Hologic, Inc., Fujifilm Corp.), whereas other manufacturers (e.g., GE Healthcare, Siemens Healthcare, IMS s.r.l.) use K-edge filters. In general, during digital mammography (DM), the filter with the higher characteristic K-edge is used for the more attenuating breast and this approach is effective in maintaining image contrast at an acceptable dose.

Several studies have reported on the optimization of x-ray spectra for mammography (Desponds et al., 1991, Fahrig et al., 1996, Andre and Spivey, 1997, Dance et al., 2000b, Flynn et al., 2003, Bernhardt et al., 2006, Åslund et al., 2006, Toroi et al., 2007, Baldelli et al., 2010, Cunha et al., 2012, Salvagnini et al., 2013). These studies compared different target/filter combinations and some of these studies are specific to a particular mammography system, e.g., (Åslund et al., 2006, Bernhardt et al., 2006, Baldelli et al., 2010, Salvagnini et al., 2013). Almost all of the aforementioned studies show a benefit for using tungsten (W) target. Increasingly, systems designed for DM, and more so for DBT, utilize W targets due to its higher melting point that allows for higher instantaneous heat load (Johns and Cunningham, 1983). Hence, this study is restricted to systems employing W target x-ray tubes. The results from many of the aforementioned studies also indicate that the optimal K-edge filter varied with breast thickness, e.g., (Fahrig et al., 1996, Åslund et al., 2006, Cunha et al., 2012, Salvagnini et al., 2013).

Among the studies investigating system-specific optimization of anode-filter combination, at least three studies considered the influence of exposure duration. Fahrig et al (Fahrig et al., 1996) investigated x-ray beam filters over the atomic number (Z) range of $Z = [35, 50]$ for an amorphous selenium (a-Se) detector and at specified exposure duration limits based on x-ray tube heat-load. The study optimized the filter with W anode using signal-to-noise ratio (also referred to as signal-difference to noise ratio, SDNR in other studies) under the constraint of a fixed mean glandular dose (MGD) and specified limits for exposure duration. The study observed that W/Rh at applied x-ray tube potential (kV) of 25 kV and W/Sn at 32 kV, were optimal for 4 and 8 cm thick breasts, respectively. Flynn et al (Flynn et al., 2003) noted that the W anode could lead to substantial reduction in exposure duration. Subsequently, Åslund et al (Åslund et al., 2006) investigated the spectral conditions suitable for a photon-counting multi-slit mammography system. One of the approaches used for optimization considered a constant SDNR per unit time, referred to as iso-SDNR-rate. The study considered 0.5 mm Al filtration as the reference and investigated if switching to Ag, Cd and In K-edge filters would be of benefit. The study reported that the use of Ag, Cd and In K-edge filters resulted in a modest 4% or less improvement compared to an Al filter.

Since the improvement provided by K-edge filters over Al filter is modest (Åslund et al., 2006), and several studies (Fahrig et al., 1996, Åslund et al., 2006, Cunha et al., 2012, Salvagnini et al., 2013) report that the optimal K-edge filter is breast thickness dependent, this study investigates if Al filtration can be used for all breast thicknesses. Unlike the prior

studies, where the intent is to perform system-specific and breast thickness-specific optimization, the intent of this study is to investigate the potential of Al filters to reduce the exposure duration and the mean (fibro)glandular dose (MGD), without adversely affecting SDNR over the commonly observed range of compressed breast thickness. Since the Al filter is an order of magnitude thicker than K-edge filters, there is need to determine if this contributes to increased x-ray scatter originating from the filter. This study addresses these needs. Further, the clinical use of DBT warrants consideration of the exposure duration per projection, and hence the total duration of the DBT acquisition, and consequently the likelihood for patient motion. Additionally, for DBT systems employing continuous motion of the x-ray tube, the shorter exposure duration per projection can reduce the focal spot associated blur in the projections due to the extended focal spot dimensions along the x-ray tube movement direction (Zhou et al., 2007, Marshall and Bosmans, 2012).

The exposure duration in mammography is by far the longest in radiographic imaging, typically ranging from about 0.5 seconds to more than 2 seconds depending on breast thickness and composition. Modest increase in x-ray tube power by increasing the tube current capability from typically 100 mA to closer to 200 mA for more recent units represents an improvement, but the exposure duration has not reduced substantially. X-ray tubes can be developed to operate at much higher tube current, e.g., 300–400 mA, but maintaining a small focal spot would present technical challenges and the resulting cost of manufacturing might be too high for commercial adoption. In the past, optimization of x-ray filtration has been focused on contrast and radiation dose. In this study, we hypothesize that a single material for x-ray filter, e.g., aluminum, can increase x-ray fluence per mA, which can reduce the exposure duration without degrading image quality. This will represent a very inexpensive alternative to designing a new tube. The application of aluminum (Al) beam filtration can lead to shorter exposure duration and this can contribute to reducing the likelihood for motion effects and reducing patient discomfort from prolonged breast compression.

The potential advantages in converting from K-edge filtration to aluminum filtration are: (1) higher x-ray photon fluence per mA, resulting in shorter exposure duration for typical fixed mA systems; (2) simplification of system design; (3) consistency in spectra between DM and DBT acquisition modes; (4) standardization across different DM and DBT platforms; (5) simplification of instruments, and calibration requirements for instruments, used to measure x-ray tube potential; (6) low-cost wide availability of high purity aluminum; and, (7) Eventual elimination of the beryllium window requirement in mammography tubes and replacing it with lower cost aluminum or even thin glass.

Hence, in this study, we investigated the use of Al filtration of a single or variable thickness. The objective is to arrive at a range of Al filtration thickness that increases photon fluence per unit mA than the “standard” K-edge filters, while matching the average energy and the half-value layer (HVL) of the standard K-edge filters. The Al filtered spectrum cannot match the spectra provided by K-edge filters; however, reasonable matching of the mean energy and the HVL is achievable. In order to match the mean energy and the HVL of the spectra provided by K-edge filters, the Al filter needs to be thicker. Hence, the x-ray scatter contribution from Al and K-edge filters were quantified in terms of scatter-to-primary ratio

(SPR) and point-spread function of x-ray scatter (scatter PSF). To our knowledge, there have been no prior reports on the scattering properties of filters used in mammography systems; although, there have been studies on x-ray scatter from elemental filters in higher energy general radiography (Chan and Doi, 1985, Karellas et al., 1986).

The article is arranged as follows: (1) spectral simulations comparing K-edge filters (Rh and Ag) with Al filter for 3 different strategies (matched kV and fixed Al thickness; matched HVL and fixed Al thickness; and, matched kV and HVL) to show that Al filtration increases x-ray photon fluence and consequently the potential for reduced exposure duration; (2) Monte Carlo simulations to show that the scatter-to-primary ratio (SPR) and the scatter point spread function (PSF) arising from x-ray filters alone are not substantially different between Al and K-edge filters; (3) Monte Carlo simulations to show that the SPR with the breast in the x-ray beam is not substantially different between Al and K-edge filters; (4) Monte Carlo simulations to determine the dose-conversion coefficients (normalized glandular dose coefficients or DgM), which is necessary for computing the dose-dependent signal-difference to noise ratio (SDNR); and, (5) SDNR computations to show that the W/Al spectrum reduces the exposure duration and the MGD at SDNR matched to K-edge filters.

2. MATERIALS AND METHODS

Elemental composition for skin, adipose and fibroglandular tissues were obtained from Hammerstein et al (Hammerstein et al., 1979). Mass attenuation coefficients of breasts with varying fibroglandular weight content (expressed as percentage throughout the article) were determined by mixture rule. Mass attenuation coefficients for skin, adipose tissue, fibroglandular tissue, and the x-ray beam filters were obtained from National Institutes of Standards and Technology (Berger et al., 2011) and log-log cubic interpolation was used to match the energy bins from spectral simulations using the TASMIP model for mammography (Boone et al., 1997). The simulated spectra are in units of photons/mm² at fixed (constant) mAs.

2.1 Photon fluence with Al and K-edge filters

The x-ray target(s) and filters used in DM and DBT systems vary substantially. In terms of target, Mo, Rh, and W are used in clinical systems. Mo (25–30 μm), Rh (25–60 μm), Ag (50–75 μm) and Al (400–700 μm) filters are used with varying target combinations. Restricting to W target due to aforementioned reasons, the x-ray spectra encompassing the typical DM and DBT energy range of 24 to 34 kV were simulated. The attenuation due to the thin plastic mirror in the collimator housing was not included in the current simulation but if Al is to be adopted as a filter, the attenuation from the mirror would be minimal. The fluence per unit exposure, q_0/X , was computed using (Johns and Cunningham, 1983):

$$\frac{q_0}{X} = \int_{E=0}^{E=E_{\max}} \frac{k q_{\text{norm}}(E)}{E[\mu_{\text{en}}/\rho]_{\text{air}}} dE \quad (1)$$

where $[\mu_{en}/\rho]_{air}$ is the mass-energy absorption coefficient for air in cm^2/g , E is the energy in keV, $q_{norm}(E)$ is the simulated spectrum normalized to unit area, and $k = 5.45 \times 10^5$ keV/g/mR, is the unit conversion constant. Exposure was converted to air kerma by using the conversion factor of 1 mR=8.696 μGy .

In the United States, the normalized glandular dose conversion factors (DgN coefficients) used to determine the MGD are based on an assumed “average” composition of 50% fibroglandular content and the DgN coefficients based on the works of Wu et al. (Wu et al., 1991, Wu et al., 1994) are provided in the American College of Radiology (ACR) mammography quality control manual (ACR, 1999). The 2016 update to the ACR manual specific for digital mammography uses the “ kgs ” model by Dance et al (Dance et al., 2000a); however the g and c factors used to determine the MGD for an “average” breast correspond to 50% fibroglandular content (ACR, 2016). Hereon, the fibroglandular weight fraction is denoted by f_g and is expressed as percentage. For breast dosimetry in DBT, Dance et al (Dance et al., 2011) and the American Association of Physicists in Medicine (AAPM) Task Group 223 report (Sechopoulos et al., 2014) provide an additional DBT-system dependent factor to correct the DgN coefficients. Individual studies have provided DgN coefficients for mammography for breasts of various f_g and compressed thicknesses and additional correction factors for DBT, e.g., (Wu et al., 1991, Wu et al., 1994, Dance et al., 2000a, Boone, 2002, Sechopoulos et al., 2007a, Dance et al., 2009). Hence, although recent studies indicate that the average $f_g=15\%$ (Yaffe et al., 2009, Vedantham et al., 2012b), in this study, $f_g=50\%$ was considered as the “average” composition. Assuming the same technique factors, this choice of $f_g=50\%$ than $f_g=15\%$ to represent an “average” breast will result in lower estimates for the MGD that varies with compressed breast thickness.

The x-ray fluence (photons/ mm^2) and fluence per air kerma (photons/ $\text{mm}^2/\mu\text{Gy}$) incident on the detector were computed with each x-ray filter, and with and without the breast in the x-ray beam. In order to test the hypothesis that Al filter can replace K-edge filters (typically, Rh and Ag), three strategies were investigated:

- i. fixed Al thickness and matched kV, resulting in variation in HVL
- ii. fixed Al thickness and varying the kV such that the HVL from K-edge and Al filtered spectra are matched, and
- iii. varying the Al thickness at matched kV, such that the HVL from K-edge and Al filtered spectra are matched.

The percent increase in x-ray fluence with Al filtration was quantified with respect to K-edge filters as per Eqn. 2, where q_0^{Al} and q_0^K are the x-ray photon fluence at constant mAs with Al and K-edge filtration. All computations were performed using Matlab™ (The Mathworks, Inc., Natick, MA).

$$\% \text{Fluence increase} = \left(\frac{q_0^{Al} - q_0^K}{q_0^K} \right) \times 100 \quad (2)$$

2.2 Monte Carlo simulations

Four sets of Monte Carlo simulations were performed using the GEANT4 toolkit (Agostinelli et al., 2003). Two sets of these simulations were designed to quantify the x-ray scatter contribution arising from the x-ray filters alone and were intended to determine the scatter-to-primary ratio (SPR) and the point spread function (PSF) of x-ray scatter. The other two sets of Monte Carlo simulations correspond to the determination of the radiation dose to the breast and the determination of the SPR with the breast in the x-ray beam path. The Livermore low energy physics model was adopted for all simulations. All simulations used monoenergy x-ray photons emitted from a point focus. Physical processes for x-ray photon interactions included incoherent or Compton scattering, coherent scattering, and photoelectric interaction.

2.2.1 SPR of Al and K-edge filters alone—The proposed standardization of filters would require the use of Al filtration of a single or variable thickness that would be substantially thicker than Rh and Ag K-edge filters. Hence, there is a concern that the Al filter might generate substantially more x-ray scatter. One commercially available DBT/DM system uses either a 50 μm Rh filter or a 50 μm Ag filter during two-dimensional (2D) DM acquisition, and 700 μm Al filter for DBT acquisition (Sechopoulos, 2013, Vedantham et al., 2015). Hence, the scatter properties of 700 μm thick Al, 50 μm thick Rh and 50 μm thick Ag were quantified using Monte Carlo simulations. Monoenergy x-ray photons from 2 keV to 32 keV, in 2 keV intervals, were simulated. For each monoenergy beam, 100 million x-ray photons emitted from a point focal spot were tracked. The system geometry used for the Monte Carlo simulation is shown in Fig. 1 and emulated a commercially available DBT/DM system. The central-ray, defined as the shortest path-length from the focal spot to the detector, was centered laterally along the chest-wall edge of the detector and is co-incident with the z-axis (Fig. 1). The origin of the coordinate system is the intersection of central ray with the detector plane. Since the intent is to determine the scatter contribution from x-ray filters alone, the compression paddle, the breast, and the breast support plate were not included in these simulations. Different filters (700 μm Al, 50 μm Rh and 50 μm Ag) located at 5 cm from the focal spot and a 250 μm thick amorphous selenium (a-Se) detector of size 29 cm $\times$ 23 cm located at 70 cm from the focal spot were simulated. Since the x-ray scatter is a long-range effect and to limit the number of x-ray photons simulated and consequently the simulation time to acceptable levels while obtaining adequate statistics, the pixel size of the detector was set to 1 mm on each side. The emission angles (fan and cone angles) from the source were constrained to the detector dimensions. During the simulation, each photon incident on the detector was classified as incoherent scattering if it underwent at least one Compton interaction, as coherent scattering if it underwent Rayleigh interaction alone, and as primary photons if it did not undergo any interaction (Vedantham et al., 2014). The sum of incoherent and coherent scattered photons provided the total number of scattered x-ray photons. The spatial distribution of scattered photons $S(x, y)$ and primary photons $P(x, y)$ were obtained by weighting the results from monoenergetic simulations with x-ray photon energy E emitted from the focal spot with the normalized spectrum $q_{\text{norm}}(E)$ as:

$$\frac{S(x, y)}{P(x, y)} = \frac{\int S(x, y, E) q_{norm}(E) dE}{\int P(x, y, E) q_{norm}(E) dE} \quad (3)$$

In Eqn. 3, (x, y) refers to detector coordinates. The location-dependent scatter-to-primary ratio, $SPR(x, y)$ was computed as:

$$SPR(x, y) = \frac{S(x, y)}{P(x, y)} \quad (4)$$

For location-independent analysis, aimed to provide a measure of the total number of scattered photons relative to primary photons reaching the detector, the SPR was computed as, $\Sigma S(x, y) / \Sigma P(x, y)$.

2.2.2 Scatter point spread function from Al and K-edge filters—While the aforementioned Monte Carlo simulation setup provides an estimate of the magnitude of x-ray scatter relative to the primary photons, it is also important to quantify the spatial distribution of x-ray scatter arising from filters of varying composition and thickness. Hence, the point spread function of x-ray scatter, herein referred to as scatter PSF was determined. The imaging geometry was identical to Fig 1. The aforementioned Monte Carlo simulation setup was modified so that a pencil beam is emitted along the central ray (z -axis). For each energy bin, from 2 keV to 32 keV, in 2 keV intervals, 100 million photons were tracked. The monoenergetic scatter distribution recorded by the detector, $S(x, y)$ was weighted by the normalized spectrum as per Eqn. 3. $S(x, y)$ was radially averaged to provide $S(r)$, where $r = \sqrt{x^2 + y^2}$. The scatter PSF, $PSF(r)$, was determined as, $S(r)/S(0)$.

2.2.3 Radiation dose to the breast—In the aforementioned Monte Carlo simulations, the breast was not included in the x-ray beam as the intent was to determine the scatter contribution from x-ray filters alone. For the computation of MGD to the breast, Monte Carlo simulations were performed by modifying a previously described setup (Vedantham et al., 2012a). Specifically, the imaging geometry was modified to reflect a clinical DM/DBT system with a source to image-receptor distance (SID) of 70 cm as shown in Fig. 2. An air gap of 3 cm between the image-receptor and breast support plate was assumed. The breast support plate was modeled as carbon fiber of 1.5 mm thickness and the compression paddle was approximated as 2 mm thick polymethyl methacrylate (PMMA). The breast was modeled as a half-cylinder approximating the compressed breast shape for cranio-caudal (CC) view with homogenous mixture of adipose and fibroglandular tissue enveloped by a layer of skin on all sides except for the chest-wall. The radius (r , Fig. 2) of the semi-cylinder inclusive of the skin layer was 8.5 cm to match the geometry used in a prior study (Boone, 2002). Although recent studies from breast computed tomography (CT) indicate that the average skin thickness (t_s , Fig. 2) comprising the dermis and epidermis is approximately 1.45 mm (Huang et al., 2008, Shi et al., 2013), $t_s = 4$ mm was used to be consistent with prior studies on digital mammography (Wu et al., 1994, Wu et al., 1991, Boone, 1999) and

DBT (Sechopoulos et al., 2007a, Sechopoulos et al., 2014). In comparison, the ‘*kgs*’ model for DM (Dance et al., 2000a) and DBT (Dance et al., 2011) are based on a 5 mm thick adipose layer to represent the skin. The assumptions regarding skin thickness and composition impacts the glandular-dose conversion coefficients and consequently the MGD estimate. For the same skin composition, assuming a thicker skin layer would result in lower MGD estimate due to shielding (Huang et al., 2008). The compressed breast thickness was varied from 2 to 8 cm in 2 cm increments and with $f_g = \{1, 15, 25, 50, 75, 100\}\%$. Monoenergy x-ray photons from 3 to 35 keV in 1 keV intervals were simulated with each energy bin consisting of 10 million photons. The energy deposited within the breast was recorded and apportioned to fibroglandular tissue on a per-interaction basis (Wilkinson and Heggie, 2000), from which the monoenergy normalized glandular dose $DgN(E)$ coefficients were computed (Boone, 1999, Vedantham et al., 2012a) and weighted by the normalized polyenergetic spectrum to provide the spectrum-weighted DgN coefficients.

2.2.4 Scatter-to-primary ratio with breast—Comprehensive studies on x-ray scatter for DM (Boone et al., 2000) and DBT (Sechopoulos et al., 2007b) show that the SPR is mainly dependent on compressed breast thickness and minimally dependent on fibroglandular content. Importantly, these studies observed negligible dependence on x-ray photon energies over the energy range of interest for DM and DBT. Since the intent of this article is to investigate the effect of using Al filters instead of K-edge filters, it is sufficient to determine if the use of Al filter instead of a K-edge filter alters the SPR for a given breast size (radius), fibroglandular composition and compressed breast thickness. Hence, Monte Carlo simulations were performed for a 5 cm thick, $f_g = 50\%$ using the geometry in Fig 2 that included the breast support plate (Sechopoulos et al., 2007b). Monoenergy scatter and primary maps were obtained by tracking 10 million photons per energy bin in 2 keV intervals over the energy range of 2 to 34 keV, from which the spectrum-weighted SPR maps for 24 to 34 kV, 700 μm thick Al, 50 μm thick Rh, and 50 μm thick Ag filtered x-ray spectra were determined as per Eqn. (3). From each of the spectrum-weighted SPR map, the SPR at the projection of the center-of-mass was extracted to determine if the use of Al filter instead of K-edge filters altered the SPR.

2.3 Signal-difference to noise ratio computation

Let $q_{norm}(v, f)$ represent the normalized spectrum, where v and f represent the applied tube voltage (kV) and x-ray beam filter, respectively. Let t_b represent the thickness of the background and t_s represent the thickness of the signal object that is added to the background to provide a total thickness, $t = t_b + t_s$. For simplicity, let us assume that the background and the signal object are of the same composition, and hence, the linear attenuation coefficient $\mu_b = \mu_s = \mu$. Then, the transmitted fraction per incident photon can be determined from Beer-Lambert’s law under parallel-beam approximation as,

$$\begin{aligned} I_b &= \exp \left(- \int_0^v q_{norm}(v, f) \mu t_b dE \right) \\ I_s &= \exp \left(- \int_0^v q_{norm}(v, f) \mu t dE \right) \end{aligned} \quad (5)$$

In the above equations, the energy-dependence of $q_{norm}(v, f)$ and μ is implicit. For Poisson distributed x-ray photons, the SDNR per photon can be computed as,

$$\frac{SDNR(v, f, t)}{photon} = \frac{[I_b - I_s]}{\sqrt{[I_b + I_s]}} \quad (6)$$

From $q_{norm}(v, f)$, the photon fluence per unit air kerma $\frac{q_0(v, f)}{X}$, where X is air kerma (mGy) is computed as per Eqn. 1, and by using the conversion factor of 1 mR=8.696 μ Gy. From the DgN corresponding to $q_{norm}(v, f)$ and total thickness t , represented as $DgN(v, f, t)$, the entrance skin air kerma $ESAK(v, f, t)$ needed for the targeted MGD can be determined as,

$$ESAK(v, f, t) = \frac{MGD}{DgN(v, f, t)} \quad (7)$$

From which, the fluence incident on the breast $q_0(v, f, t)$ to provide the targeted MGD is computed as,

$$q_0(v, f, t) = ESAK(v, f, t) \left(\frac{q_0(v, f)}{X} \right) \quad (8)$$

Then for Poisson distributed x-ray photons, the signal-difference to noise ratio for the specified spectrum and the targeted MGD can be determined as,

$$SDNR(v, f, t) = \sqrt{q_0(v, f, t)} \left(\frac{SDNR(v, f, t)}{photon} \right) \quad (9)$$

Let us first consider a K-edge filter, e.g., $f = K$. Let us assume that the compressed breast thickness t , applied tube voltage v_K , and the MGD are specified. Then the SDNR can be determined from Eqns. 5–9. For the same compressed breast thickness t , the goal is determine the MGD and exposure duration reduction, when the SDNR from K-edge filtered spectra is matched to that with Al filtered spectrum $f = Al$, where the Al thickness and applied tube voltage v_{Al} depend on the three strategies stated in section 2.1, viz., (i) fixed Al thickness and matched kV; (ii) fixed Al thickness and matched HVL; and, (iii) matched kV and HVL. Thus, for the same compressed breast thickness t , the goal can be stated as $SDNR(v_K, K) = SDNR(v_{Al}, Al)$. Then the MGD ratio can be computed by substituting Eqns. 7–9 and rearranging (Appendix A) as,

$$\frac{MGD_{Al}}{MGD_K} = \left(\frac{DgN(v_{Al}, Al)}{DgN(v_K, K)} \right) \left\{ \frac{[q_0(v_K, K)/X]}{[q_0(v_{Al}, Al)/X]} \right\} \left\{ \frac{[SDNR(v_K, K)/Photon]^2}{[SDNR(v_{Al}, Al)/Photon]^2} \right\} \quad (10)$$

The percent reduction in MGD with Al filter compared to K-edge filter is reported as, $100 \times [1 - (MGD_{Al}/MGD_K)]$. The exposure duration ratio can be computed as shown in Appendix A as,

$$\frac{s_{Al}}{s_K} = \frac{\left\{ [SDNR(v_{Al}, Al)/Photon]^2 / q_0^T(v_{Al}, Al) \right\}}{\left\{ [SDNR(v_K, K)/Photon]^2 / q_0^T(v_K, K) \right\}} \quad (11)$$

where, $q_0^T(v_{Al}, Al)$ and $q_0^T(v_K, K)$ are the Al filtered and K-edge filtered spectra, respectively, as provided by the TASMIP model and prior to any normalization. The percent reduction in exposure duration with Al filter compared to K-edge filter is reported as, $100 \times [1 - (s_{Al}/s_K)]$.

For a given compressed breast thickness t , the appropriate K-edge filter needs to be determined. Hence, annual quality control audit of a W-anode DM/DBT system (Selenia® Dimensions®, Hologic Inc.) was reviewed. For 2 to 8 cm thick BR-12 tissue substitute material, in 2 cm steps, the kV/filter combination selected by the automatic exposure control (AEC, 'Auto-filter' mode) of the system were 25 kV/Rh, 27 kV/Rh, 30 kV/Rh and 32 kV/Ag for 2, 4, 6 and 8 cm thicknesses, respectively. Hence, Al filter was compared with Rh filter for 2 to 6 cm thickness and Ag filter for 8 cm thickness for each of the three aforementioned strategies. Since the aforementioned empirical data for system selected kV and filter were with BR-12, and it has been shown that the BR-12 approximates a 50% fibroglandular breast ($f_g = 50\%$) in terms of linear attenuation coefficient (Vedantham and Karellas, 2013), analytical computation of MGD reduction and exposure duration reduction were performed with $f_g = 50\%$ breast. For analytical computation, $t_b = 2$ to 8 cm thick, $f_g = 50\%$ breasts, in 2 cm steps for background, and $t_s = 4$ mm thick PMMA as signal object were used. The 4 mm thick PMMA for the signal object was chosen as it corresponds to that used for contrast-to-noise ratio (CNR) measurements traditionally used for mammography, performed as part of ACR accreditation requirements (ACR, 1999). For each compressed breast thickness t and simulated spectrum q_0^T , the $SDNR/photon$, the q_0/X and the DgN were computed, from which, the percent reduction in MGD and exposure duration were determined as per Eqns. (10) and (11), respectively.

3. RESULTS

3.1 Photon fluence

For fixed Al thickness of 700 μm and for kV matched between Al and K-edge filters, the photon fluence (photons/ mm^2) at fixed mAs from the TASMIP model, the photon fluence per unit air kerma (photons/ $\text{mm}^2/\mu\text{Gy}$), the mean energy (keV), the HVL (mm of Al), and the percent increase in fluence with Al filter compared to K-edge filters are summarized in Table 1.

It can be inferred from Eqn. 2, that a 100% increase in fluence corresponds to doubling of photon fluence. The aforementioned results are without the breast in the x-ray beam. The x-ray photon fluence after transmitting through 2, 4 and 6 cm thick compressed breasts of 50%

fibroglandular content are shown in Fig. 3. In Fig. 3 (a), the x-ray photons fluence without the breast (labeled as '0 cm') is shown. The 700 μm thick Al results in higher x-ray fluence incident on the detector compared to Rh and Ag filters of 50 μm thicknesses at matched kV and mAs.

From Table 1, at each kV considered, the HVL of 700 μm Al spectrum is lower than that of the Ag filtered spectrum. Also, the HVL of the Al filtered spectra is lower than that of the Rh filtered spectra for the 24–30 kV range. If matching of HVL is desired, then it can be achieved by, varying the kV with fixed Al thickness, varying the Al thickness and matching the kV, or a combination thereof. Results from the former approach are summarized in Table 2 and shows that with 700 μm Al, increasing the kV by the range [0, 3] and by [3, 4] matched the HVL of the 50 μm Rh and 50 μm Ag filtered spectra, respectively. The photon fluence with Al filtration doubled compared to Rh filtration (124–169% increase) and Ag filtration (119–135% increase). Although the HVL is matched between Al and K-edge filtered spectra, the kV differ and consequently the spectra would be different.

Table 3 summarizes the results for the alternative approach to match the HVL of Al and K-edge filtered spectra, where the thickness of Al was varied, while maintaining the same kV as that with K-edge filters. Al filter thickness range of [700, 900] μm and [800, 880] μm matched the HVL of 50 μm thick Rh and Ag filters, respectively. With Al filter, the photon fluence increased by approximately 75% (27–124% increase) compared to Rh, and by approximately 35% (range: 20–49%) compared to Ag filter. Fig. 4 shows that even though the kV and the HVL are matched between K-edge and Al filtered spectra, the spectrum provided by Al filter is different from that provided by K-edge filters.

Thus, the results show that for all three investigated strategies, *viz.*, fixed Al thickness and matched kV resulting in HVL variation, matched HVL with fixed Al thickness resulting in varying kV, and matched HVL and kV resulting in Al thickness variation, the photon fluence with Al filtration is increased with respect to K-edge filters. It is also possible to jointly vary the Al thickness and the kV to match the HVL of K-edge filtered spectra. It is important to note that all computations pertaining to the fluence increase provided by Al filters, with the exception of Fig. 3, were without the breast in the x-ray beam. For K-edge filters, when the kV (e.g., 24 kV with Rh and 26 kV with Ag) is selected to be above the K-edge of the filter (23.2 keV for Rh, 25.5 keV for Ag), the proportion of photons above the K-edge is small. In contrast, Al filters would have a higher proportion of counts at these energies as shown in Fig. 4. Hence, with the breast in the x-ray beam, and depending on the selected kV, the photon fluence incident on the detector with Al filter is likely to be even higher.

3.2 Monte Carlo results

Four sets of Monte Carlo simulations were performed, of which, two were without the breast in the x-ray beam path. These two simulations were intended to determine the scatter properties of the x-ray filters alone. To our knowledge, there have been no prior studies reporting on the x-ray scatter properties of spectral-shaping filters in the mammography energy range. However, there have been several studies reporting on Monte Carlo derived breast dosimetry and on x-ray scatter from breast in the mammography energy range. Hence, the results from our study for the two sets Monte Carlo simulations with the breast in the x-

ray beam are presented first, so that it can also serve as validation of our Monte Carlo simulations.

3.2.1 Radiation dose to the breast—The monoenergy $DgN(E)$ for 2 to 8 cm thick compressed breasts, in 2 cm intervals, were computed from Monte Carlo simulations. The spectrum-weighted DgN were computed for all three filters (50 μm Rh, 50 μm Ag, and 700 μm Al) and for 24 to 34 kV. The DgN results for 50% and 100% fibroglandular fraction from this study were compared with the DgN from the computer fit equations (Boone, 2002) and is shown in Fig. 5, along with the identity line. The differences between this study and that from computer fit equations (Boone, 2002) were 2.42% (Rh; $f_g = 50\%$), 2.36% (Rh; $f_g = 100\%$), 1.81% (Ag; $f_g = 50\%$), 1.91% (Ag; $f_g = 100\%$), 3.97% (Al; $f_g = 50\%$) and 2.00% (Al; $f_g = 100\%$).

3.2.2 SPR with breast—The monoenergy scatter and primary maps were obtained from Monte Carlo simulations, weighted by the normalized spectra from which the SPR were computed as per Eqns. (3) and (4). The SPR computed in this study for a 5 cm thick, $f_g = 50\%$ over the 24 to 34 kV range were compared with the fit equation by (Boone et al., 2000) and is shown in Fig. 6. In terms of the SPR, the results from this study shown in Fig. 6 as well as prior studies indicate minimal dependence with kV for a given compressed breast thickness (Boone et al., 2000, Sechopoulos et al., 2007b). Importantly, Fig. 6 shows that the SPR for Al and K-edge filtered spectra are not different for a given compressed breast thickness. The implication is that the degradation in signal-difference (contrast) due to SPR with Al and K-edge filtered spectra are similar. Hence, it is sufficient to consider the primary photons alone for comparing the SDNR for various spectra, and consequently the MGD reduction and the exposure duration reduction.

3.2.3 Scatter from filters alone—The results from the location-independent analysis of SPR, computed as $\Sigma S/\Sigma P$, are presented first. The intent of this analysis is to understand the energy-dependence of SPR. Fig. 7 shows the results from location-independent analysis of SPR for monoenergetic x-ray photons from 10 to 32 keV with 700 μm thick Al and 50 μm thick Rh and Ag filters. Over the investigated energy range, the maximum SPR was less than 0.008, and did not vary substantially among the filters. The coefficient of variation (CV) in SPR from three independent runs was used to quantify precision. For all 3 filters and at each x-ray photon energy the CV was less than 0.6%.

Table 4 summarizes the location-independent estimates of SPR ($\Sigma S/\Sigma P$), coherent scatter to primary ratio (CSPR; $\Sigma S^{coh}/\Sigma P$) and incoherent scatter to primary ratio (ISPR; $\Sigma S^{incoh}/\Sigma P$) for all 3 filters. Even though the Al filter is substantially thicker (700 μm) compared to 50 μm thick Rh and Ag filters, the SPR is similar for all 3 filters. Although not presented, our observation is that for Al thickness ranging from 650–800 μm , the SPR is similar. Table 4 shows that in the absence of the breast in the x-ray beam, the majority of the scattered photons incident on the detector for all 3 filters are due to coherent scatter. Since coherent scatter is forward-peaked, its contribution to the SPR is substantially more compared to incoherent scatter. Incoherent scatter was nearly negligible. This can be explained by the large distance from the filter to the detector (65 cm, Fig. 2). Since coherent scatter is the dominant contributor to the total scatter, the energy distribution of scattered x-

ray photons incident on the breast is similar to that of primary photons. Additionally, the amplitude of $\Sigma S/\Sigma P$ is small. These observations are of practical relevance during Monte Carlo simulations with the breast. It is sufficient to perform monoenergy simulations without the x-ray beam filter followed by weighting with the specified normalized spectrum as the number of scattered photons (relative to the primary photons) incident on the breast is small and the energy is largely unaltered.

The monoenergetic $S(x, y, E)$ and $P(x, y, E)$ can be weighted with any desired normalized spectrum, $q_{norm}(E)$ to provide the spectrum-weighted $S(x, y)$ and $P(x, y)$ as per Eqns. 3 and 4. The spectrum-weighted $S(x, y)$ without the breast in the x-ray beam for the 700 μm Al, the 50 μm Rh and the 50 μm Ag filters at 32 kV are shown in Fig. 8. Visually, $S(x, y)$ appears higher with the 700 μm Al than the 50 μm Rh and the 50 μm Ag filters. Hence, one dimensional estimates of $S(y)$ and $P(y)$ were obtained by integrating along the x -direction (chest-wall to nipple direction, Fig. 1). Fig. 9 confirms that with Al filtration, there is an increase in $S(y)$. However, there is a concomitant increase in $P(y)$, resulting in $SPR(y)$ that is marginally higher with Rh (50 μm) compared to Al (700 μm) and Ag (50 μm) filters. This observation is consistent with Table 4.

In the above results, the amplitudes of x-ray scatter and SPR without the breast in the x-ray beam were reported. The scatter PSF $S(r)$ $S(0)$ for all 3 filters considered at 32 kV are shown in Fig. 10. In Fig. 10(a) the scatter PSF is shown up to the radial distance $r=100$ mm, while in Fig. 10(b) it is shown up to $r=10$ mm. Since the forward-peaked coherent scatter is the dominant contributor, there is increased scatter contribution closer to the primary pencil beam that is incident at $r=0$. The flattening of the curve at $r>10$ mm is due to the large distance between the filter and the detector (65 cm, Fig. 2).

3.3 MGD and exposure duration reduction

The results from the analyses quantifying the MGD reduction and the exposure duration reduction at matched SDNR between Al and K-edge filtered spectra for each of the aforementioned 3 strategies are presented. For 2 to 6 cm thick BR-12 that is equivalent to $f_g = 50\%$ breast tissue in terms of attenuation (Vedantham and Karellas, 2013), the automatic exposure control (AEC; 'Auto-filter' mode) for one DBT/DM system (Selenia® Dimensions®, Hologic, Inc.) selected Rh filtration, whereas for 8 cm thick breast the system selected Ag filtration. Hence, for 2 to 6 cm thick breasts the comparisons between Al filtered spectra are with 50 μm Rh filtered spectra, and for 8 cm thick breast the comparisons are with 50 μm Ag filtered spectra. For K-edge filtered spectra ($f=K$) and applied tube voltage v_K , the MGD was set to 1 mGy ($MGD_K=1$ mGy) to determine the MGD ratio and the exposure duration ratio as per Eqns. 10 and 11. Alternatively stated, for a given compressed breast thickness t and applied tube voltage v_K for the K-edge filtered spectrum, $SDNR(v_K, K, t)$ is computed at MGD of 1 mGy, and for the Al filtered spectrum matched to this SDNR, the MGD ratio and the exposure duration ratio are calculated. Figures 11 through 13 show the results from these analyses for each of the three strategies.

In Fig. 11, the results from the analysis at matched KV between Al and K-edge filtered spectra and for fixed Al (700 μm) thickness are shown. The open circles in Fig. 11 indicate the thickness-dependent kV selected by the AEC for one DBT/DM system. Since the kV is

matched between Al and K-edge filtered spectra (Rh for 2–6 cm and Ag for 8 cm, based on AEC selection), the exposure duration reduction and the MGD reduction are quantified at kV corresponding to the open-circles in Fig. 11. With Al filtration, the exposure duration is reduced by 38%, 40%, 50% and the MGD is reduced by 5%, 2%, and 20%, for 2, 4, and 6 cm thicknesses, respectively, compared to Rh filtration. For 8 cm thickness, the Al filtered spectrum reduced the exposure duration by 37% and the MGD by 11%. For this strategy, while the MGD reduction is modest, except for 6 cm breast (20%), the 35% or more reduction in exposure duration is substantial.

In Fig. 12, the results from the analysis for fixed Al (700 μm) thickness and at matched HVL between Al and K-edge filtered, breast incident spectra are shown. The matching of the HVL between Al and K-edge filtered spectra was achieved by varying the kV with Al filtration. Thus, the kV for K-edge and Al filtered spectra are different. The x – axes in Fig. 12 correspond to the tube potential for Al filtration. For each kV with K-edge filters (Rh for 2 to 6 cm and Ag for 8 cm; AEC selection), the HVL was computed and the kV needed to match this HVL for Al filtration was determined from Table 2. For Al filtration, the HVLs of breast incident spectra are matched to the K-edge filtered spectra at 28 kV, 29.5 kV, 31 kV and 35 kV for 2, 4, 6 and 8 cm thicknesses, and are indicated as open circles in the figure. Practically, the 29.5 kV translates to either 29 or 30 kV. In terms of exposure duration reduction, this strategy indicates a relatively large 62% to 65% reduction for 2 to 8 cm thick breasts. The MGD reduction is 4% for 2 cm thick breast, and increases to approximately 14% to 24% for 4 to 8 cm thick breasts, which is substantial.

In Fig. 13, the results from the analysis for matched kV and matched HVL between Al and K-edge filtered, breast incident spectra are shown. For 2 to 8 cm thick breasts, the HVL is matched by varying the Al filter thickness, depending on the AEC-selected kV (open-circles, Fig. 13) and the K-edge filter (Rh for 2–6 cm and Ag for 8 cm). The thickness of Al filters are 900 μm , 830 μm , 750 μm and 800 μm for 2, 4, 6 and 8 cm breast thickness, respectively (Table 3). The analysis indicates that depending on breast thickness, the exposure duration is reduced by 23% to 56%, and the MGD is reduced by 5% to 20% with Al filters of varying thickness compared to K-edge filters.

4 DISCUSSION

The results of this study reveal potential benefits of aluminum filtration which have not been fully appreciated in the past. It can be inferred from Tables 1 through 3 that the use of Al filter of fixed or varying thickness provides for substantial increase in x-ray photon fluence compared to 50 μm Rh and 50 μm Ag filters. The fluence increase with the use of Al filter clearly indicates a concomitant decrease in the exposure time when used with automatic exposure control (AEC) for fixed mA mammography and DBT systems. When kV is selected to be above the K-edge of Rh and Ag, the Al filtered x-ray spectrum will not be identical to that of 50 μm Rh and 50 μm Ag filters even after matching of HVL and kV as shown in Fig. 4. The source data for the TASMIP model (Boone et al., 1997) are based on spectral measurements (Fewell and Shuping, 1978) from a W-anode mammography tube with target angle of 16.5° that is tilted by 6° with respect to detector plane, to provide an effective target angle of 22.5° (Jennings et al., 2001). That study also reported that the

measured spectra are normalized to 1 mAs (Jennings et al., 2001). For specific systems, the target angle may be different and needs to be factored during simulations.

The goal of the SDNR analysis is to show the potential of Al filters to reduce the exposure duration and to reduce the MGD, while providing for similar SDNR. When the SDNR is matched between Al and K-edge filtered spectra the results show that a significant reduction in exposure duration, as high as 62% – 65%, is achievable with Al filtration. Also, the results indicate that the Al filtration does not increase the MGD to provide the same SDNR as K-edge filtration, and showed 2% – 24% reduction in MGD, depending on the implemented strategy and the compressed breast thickness. The accuracy of the spectral model could influence the estimated reduction in exposure duration. For TASMIP model, over the 20 to 40 kV range, the difference between the measured spectra and the fitted model over all energies differed on average by less than 1% (Boone, 2002). Although the intent is not to compare the three strategies, a cursory analysis of the results indicates that the strategy of fixed Al thickness (700 μm in this study) and varying the kV of Al filtered spectra to match the HVL between Al and K-edge filtered spectra provided the largest reduction (62% – 65%) in exposure duration and a substantial reduction (14% – 24%, for 4 to 8 cm thick breasts) in MGD. Further studies are needed to determine the optimal Al filtration with various fibroglandular weight content breasts.

The study results indicate that the scatter contribution from the Al filter is minimal. Importantly, with the breast in the x-ray beam the SPR with Al filtered spectrum is not different from Rh and Ag filtered spectra. The magnitude of scatter from the filters is almost two orders of magnitude lower than that from the breast and hence is inconsequential. In addition, the scatter PSF for all three filters showed very little difference among the various filters.

In addition to the studies by (Fahrig and Yaffe, 1994) and (Åslund et al., 2006) discussed earlier, optimization of target/filter combinations have been reported in several studies, some of which are system-specific. Monte Carlo simulations showed that for a 2 cm thick breast Mo/Mo was preferable, whereas for other compressed breast thicknesses W/Rh or Rh/Al combined with increase in tube potential would provide for lower MGD (Dance et al., 2000b). Investigating two systems, one a computed radiography-based system and another using a CsI:TI-coupled amorphous silicon flat-panel detector (Vedantham et al., 2000, Suryanarayanan et al., 2004), both employing Mo/Mo, Mo/Rh and Rh/Rh target/filter combinations, it was reported that the optimal target/filter was Mo/Mo for breast thickness less than 45 mm and Rh/Rh for breast thickness of 45 mm and greater (Young et al., 2006). Comparing two similar amorphous selenium-based digital mammography systems from the same vendor, one with Mo/Mo and Mo/Rh target/filter combinations and another with W/Rh and W/Ag target/filter combinations, it was reported that the CNR^2/MGD was maximized with W/Rh for 4 cm to 6 cm thick breasts (Baldelli et al., 2010). A subsequent study investigated two amorphous selenium-based digital mammography systems, one with Mo/Mo, Mo/Rh and W/Rh target/filter combinations and another with W/Rh and W/Ag target/filter combinations, and using the detectability index from non-prewhitening observer model with eye filter (Salvagnini et al., 2013). For the system with Mo and W targets, Mo/Mo performed better for 2 cm and W/Rh for 4 cm and thicker PMMA. For the system

with W target, W/Rh performed better for 2 cm and 4 cm PMMA and W/Ag for 6 cm and thicker PMMA. Other investigators (Andre and Spivey, 1997, Flynn et al., 2003) have studied alternative K-edge filters such as Sn (which has higher K-edge than Ag) and concluded that it would be preferable for imaging thicker and/or denser breasts.

It may be argued that even with the use of a single K-edge filter, a combination of changing the kV and the filter thickness might increase photon fluence incident on the detector. Also, combining multiple thin layers of K-edge and/or other filters and with appropriate kV selection may generate comparable results to the proposed Al filtration. While the aforementioned alternatives are interesting, our objective was to reduce exposure duration with a simple approach. In this case, Al is clearly desirable.

This study prompts rethink of the need for K-edge filtration and opens up the discussion that the design of DM, DBT and DM-enabled DBT systems can be made simpler with the use of a single Al filter for both DM and DBT. This can significantly decrease the exposure time. Several other advantages were outlined earlier including better standardization across systems from different vendors that would enable NIST, or other applicable standards, - traceable spectra for calibrating testing equipment.

Our study has limitations. The analytical computation of SDNR, and consequently the MGD reduction and the exposure duration reduction were based solely on photon statistics. System and detector-specific effects were not modeled. It is possible that the broader spectrum from Al filters compared to K-edge filters may have an impact on detector-specific Swank noise. Future studies using the cascaded linear systems framework (Vedantham et al., 2004, Suryanarayanan et al., 2006, Zhou et al., 2007) to perform task-specific evaluation and optimization of Al thickness are being planned.

5 CONCLUSION

In summary, the potential for replacing K-edge filters with Al filtration in conjunction with W-target was investigated by using Monte Carlo simulations, spectral modelling and analytical computation of SDNR. It is shown that the x-ray scatter from the filter is predominantly coherent scatter and that the amplitude of SPR is approximately two orders of magnitude lower than that from the breast. Further, the SPR with and without the breast are not different between Al and K-edge filters and therefore the relative detrimental effect on contrast between Al and K-edge filters is negligible. This study indicates that there exists a potential to replace current K-edge filters used in mammography with Al, to almost double the fluence incident on the breast. Importantly, with the use of Al filtration there exist a potential to reduce the exposure duration by more than 60% and the MGD by approximately 15 to 20%, while maintaining image quality. The use of a single target/filter combination for digital mammography and digital breast tomosynthesis could simplify system design.

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APPENDIX A

Computation of MGD ratio and exposure duration ratio with Al filters compared to K-edge filters

For K-edge filter $f=K$, specified compressed breast thickness t , applied tube voltage v_K , and mean glandular dose MGD_K , the SDNR is computed as per Eqn. 9. When this SDNR is matched to that with Al filtered spectra for the same compressed breast thickness,

$$SDNR(v_K, K) = SDNR(v_{Al}, Al) \quad (a.1)$$

or, equivalently,

$$SDNR^2(v_K, K) = SDNR^2(v_{Al}, Al) \quad (a.2)$$

Substituting Eqn. 9,

$$q_0(v_K, K) \left[\frac{SDNR(v_K, K)}{\text{photon}} \right]^2 = q_0(v_{Al}, Al) \left[\frac{SDNR(v_{Al}, Al)}{\text{photon}} \right]^2 \quad (a.3)$$

Substituting Eqn. 8,

$$\begin{aligned} ESAK(v_K, K) \left[\frac{q_0(v_K, K)}{X} \right] \left[\frac{SDNR(v_K, K)}{\text{photon}} \right]^2 \\ = ESAK(v_{Al}, Al) \left[\frac{q_0(v_{Al}, Al)}{X} \right] \left[\frac{SDNR(v_{Al}, Al)}{\text{photon}} \right]^2 \end{aligned} \quad (a.4)$$

Substituting Eqn. 7,

$$\begin{aligned} \left[\frac{MGD_K}{DgN(v_K, K)} \right] \left[\frac{q_0(v_K, K)}{X} \right] \left[\frac{SDNR(v_K, K)}{\text{photon}} \right]^2 \\ = \left[\frac{MGD_{Al}}{DgN(v_{Al}, Al)} \right] \left[\frac{q_0(v_{Al}, Al)}{X} \right] \left[\frac{SDNR(v_{Al}, Al)}{\text{photon}} \right]^2 \end{aligned} \quad (a.5)$$

which can be rearranged to provide the MGD ratio as,

$$\frac{MGD_{Al}}{MGD_K} = \left(\frac{DgN(v_{Al}, Al)}{DgN(v_K, K)} \right) \left\{ \frac{[q_0(v_K, K)/X]}{[q_0(v_{Al}, Al)/X]} \right\} \left\{ \frac{[SDNR(v_K, K)/\text{Photon}]^2}{[SDNR(v_{Al}, Al)/\text{Photon}]^2} \right\} \quad (a.6)$$

For computing the exposure duration ratio, Eqn. (a.3) can be rearranged as,

$$\frac{q_0(v_{Al}, Al)}{q_0(v_K, K)} = \frac{[SDNR(v_{Al}, Al)/Photon]^2}{[SDNR(v_K, K)/Photon]^2} \quad (a.7)$$

Recalling that TASMIP model provides x-ray photon fluence for a fixed or a constant mAs, the mAs ratio can be determined as,

$$\frac{mAs_{Al}}{mAs_K} = \frac{[q_0(v_{Al}, Al)/q_0^T(v_{Al}, Al)]}{[q_0(v_K, K)/q_0^T(v_K, K)]} = \frac{\{[SDNR(v_{Al}, Al)/Photon]^2/q_0^T(v_{Al}, Al)\}}{\{[SDNR(v_K, K)/Photon]^2/q_0^T(v_K, K)\}} \quad (a.8)$$

where, q_0^T is the fluence as provided by TASMIP and prior to normalization. Since mammography systems operate at a fixed mA, the exposure duration ratio is,

$$\frac{s_{Al}}{s_K} = \frac{\{[SDNR(v_{Al}, Al)/Photon]^2/q_0^T(v_{Al}, Al)\}}{\{[SDNR(v_K, K)/Photon]^2/q_0^T(v_K, K)\}} \quad (a.9)$$

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Fig. 1. System geometry for Monte Carlo simulations used to quantify the x-ray scatter arising from x-ray beam filters alone.

Fig. 2. System geometry for Monte Carlo simulations used to quantify the radiation dose to the breast and the scatter-to-primary ratio incident on the detector.

Fig. 3. Photon fluence (photons/mm²) at constant mAs incident on the detector for different filters as a function of kV after transmitting through (a) 0 cm (without breast), (b) 2 cm, (c) 4 cm, and (d) 6 cm thick compressed breasts.

Fig. 4. Comparison of spectra provided by 50 μm thick K-edge filters and Al filters of varying thickness at matched kV and HVL. (a) 28 kV Rh filter with 28 kV Al filter (800 μm) [HVL = 0.49], (b) 32 kV Ag filter (50 μm) with 32 kV Al filter (800 μm) [HVL = 0.57].

Fig. 5.

Comparison of spectrum-weighted $D_g N$ determined in this study against those using computer-fit equations reported by (Boone, 2002) for (a) Rhodium, (b) Silver, and (c) Aluminum filtered spectra. Values in (.) and [.] indicate the filter thickness and the fibroglandular weight content of the breast, respectively.

Fig. 6. Comparison of Scatter to Primary Ratio (SPR) from the breast (5 cm thick, 50% fibroglandular weight content) for different filters with fit equation from (Boone et al., 2000). Over the kV range of interest for mammography and DBT, the SPR does not vary with x-ray beam filters considered.

Fig. 7.

Location-independent SPR ($\Sigma S/\Sigma P$) for monoenergetic x-ray photons from 10 to 32 keV for various filters. The SPR for all monoenergetic beams is less than 0.008 and are nearly the same for all the filters.

Fig. 8. Spatial distribution of $S(x, y)$ at 32 kV with different filters (a) 700 μm Al, (b) 50 μm Rh and (c) 50 μm Ag.

Fig. 9. Spatial distributions at 32 kV of (a) primary photons, (b) scattered photons, and (c) SPR along the y-axis which were obtained by integrating the distributions along the x-axis.

Fig. 10.

Scatter PSF, $S(r)/S(0)$, for 700 μm Al, 50 μm Rh and 50 μm Ag filters at 32 kV over a radial distance (a) $r=100$ mm and (b) $r=100$ mm.

Fig. 11.

At matched SDNR between Al and K-edge filtered spectra, the computed percent reductions in (a) the exposure duration and (b) the MGD for the strategy of fixed Al thickness (700 μm) and at kV matched to K-edge filtered spectra. For 2 to 6 cm thick breasts, the comparison between Al and K-edge spectra are with Rh (50 μm) filter and for 8 cm thick breast it is with Ag (50 μm) filter, based on system selection in AEC mode. The open circles indicate the system selected kV for the specified compressed breast (50% fibroglandular equivalent) thickness.

Fig. 12.

At matched SDNR between Al and K-edge filtered spectra, the computed percent reductions in (a) the exposure duration and (b) the MGD for the strategy of fixed Al thickness (700 μm) and at HVL matched between Al and K-edge filtered, breast incident spectra. The kV for K-edge and Al filtered spectra are different. The x-axes are the tube potential (kV) for Al filtered spectra. The open circles at 28 kV, 29.5 kV, 31 kV, and 35 kV for Al filtered spectra match the HVL of 25 kV/Rh, 27 kV/Rh, 30 kV/Rh and 32 kV/Ag spectra selected by one system in AEC mode for 2, 4, 6, and 8 cm thick 50% fibroglandular breasts, respectively.

Fig. 13.

At matched SDNR between Al and K-edge filtered spectra, the computed percent reductions in (a) the exposure duration and (b) the MGD for the strategy of matched kV and matched HVL between Al and K-edge filtered, breast incident spectra, by varying the Al filter thickness. The open circles at 25 kV, 27 kV, 30 kV and 32 kV, for 2 cm, 4 cm, 6 cm and 8 cm thick 50% fibroglandular breasts, respectively, correspond to a system selected kV in AEC mode.

Table 1

Photon fluence (photons/mm²) at constant mAs, photon fluence per unit air kerma (photons/[mm² µGy]), mean energy and HVL for 700 µm thick Al, 50 µm thick Rh and 50 µm thick Ag filters. The last column in the table shows the % increase in fluence when using 700 µm Al filter with respect to 50 µm Rh and 50 µm Ag filters (in the respective rows).

kV	Filter	Photon fluence ($\times 10^7$ photons/mm ²)	Photon fluence per unit air kerma ($\times 10$ photons/[mm ² µGy])	Mean energy (keV)	HVL (mm of Al)	% fluence increase
24	Al	5.29	4.52	17.48	0.36	
	Rh	3.14	4.82	18.09	0.43	68
	Ag	3.38	4.82	18.07	0.42	57
26	Al	7.52	5.09	18.42	0.40	
	Rh	4.27	5.11	18.59	0.46	76
	Ag	5.04	5.45	19.10	0.48	49
28	Al	10.0	5.69	19.36	0.44	
	Rh	5.25	5.30	18.90	0.49	91
	Ag	6.66	5.88	19.76	0.52	51
30	Al	12.7	6.29	20.25	0.48	
	Rh	6.13	5.49	19.19	0.50	108
	Ag	8.09	6.16	20.20	0.55	57
32	Al	15.7	6.91	21.13	0.52	
	Rh	6.98	5.73	19.54	0.52	124
	Ag	9.43	6.42	20.58	0.57	66
34	Al	18.9	7.55	22.00	0.56	
	Rh	7.88	6.04	19.97	0.54	140
	Ag	10.8	6.71	20.99	0.59	75

Table 2

Percentage increase in photon fluence with 700 μm Al filter with respect to 50 μm Rh and Ag filters at matched HVL that is achieved by a slight variation in kV.

Reference filter	kV (Reference filter)	kV (700 μm Al)	HVL (mm of Al)	% Fluence increase
50 μm Rh	25	28	0.45	169
	27	29.5	0.48	152
	28	30	0.49	142
	30	31	0.50	131
	32	32	0.52	124
50 μm Ag	28	32	0.52	135
	30	33	0.55	113
	32	35	0.57	119

Table 3

Percentage increase in photon fluence with Al filters of varying thickness with respect to 50 μm thick Rh and Ag filters at matched kV and HVL.

Reference filter	kV	Al thickness (μm)	HVL	% Fluence increase
50 μm Rh	25	900	0.45	27
	27	830	0.48	54
	28	800	0.49	68
	30	750	0.50	95
	32	700	0.52	124
50 μm Ag	28	880	0.52	20
	30	840	0.55	34
	32	800	0.57	49

Table 4

Scatter to Primary Ratio (SPR), Coherent Scatter to Primary Ratio (CSPR) and Incoherent Scatter to Primary Ratio (ISPR) with Al, Rh and Ag filters at 32 kV.

	Al (700 μm)	Rh (50 μm)	Ag (50 μm)
SPR	5.24×10^{-3}	5.62×10^{-3}	5.17×10^{-3}
CSPR	5.04×10^{-3}	5.59×10^{-3}	5.15×10^{-3}
ISPR	0.203×10^{-3}	0.031×10^{-3}	0.027×10^{-3}