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Abstract. Evaluation of flat-panel angiography equipment through conventional image quality metrics is limited by the scope of standard spatial-domain image quality metric(s), such as contrast-to-noise ratio and spatial resolution, or by restricted access to appropriate data to calculate Fourier domain measurements, such as modulation transfer function, noise power spectrum, and detective quantum efficiency. Observer models have been shown capable of overcoming these limitations and are able to comprehensively evaluate medical-imaging systems. We present a spatial domain-based channelized Hotelling observer model to calculate the detectability index (DI) of our different sized disks and compare the performance of different imaging conditions and angiography systems. When appropriate, changes in DIs were compared to expectations based on the classical Rose model of signal detection to assess linearity of the model with quantum signal-to-noise ratio (SNR) theory. For these experiments, the estimated uncertainty of the DIs was less than 3%, allowing for precise comparison of imaging systems or conditions. For most experimental variables, DI changes were linear with expectations based on quantum SNR theory. DIs calculated for the smallest objects demonstrated nonlinearity with quantum SNR theory due to system blur. Two angiography systems with different detector element sizes were shown to perform similarly across the majority of the detection tasks. © 2015 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: 10.1117/1.JMI.2.1.015503]

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1 Introduction

Comprehensive evaluation of flat-panel angiography equipment using conventional image quality measurements is challenging. Traditional spatial-domain metrics, including contrast, noise, contrast-to-noise ratio, and spatial resolution, provide independent assessments of critical image quality attributes; however, they cannot be readily combined to provide an overall assessment of system performance. Spatial frequency domain measurements including modulation transfer function and noise power spectrum can be combined into comprehensive metrics including noise equivalent quanta (of an image) and detective quantum efficiency (of a system). However, these frequency-space measurements cannot be readily performed on clinical angiography systems due to lack of access to the requisite unprocessed or “raw” image data. Observer models based on statistical decision theory have been proposed in part to overcome the challenges and limitations of these well-known spatial domain and spatial frequency-domain measurements.

Observer models have been applied to digital flat-panel x-ray detection systems in many past investigations.^{1–12} The objectives of these studies are to (1) assist computer aided diagnosis systems in chest radiography and mammography,^{1,2} (2) optimize dose,^{7,12} (3) compare model results with conventional Fourier domain metrics,^{3,6,10,11} and (4) evaluate the performance of

digital mammography systems in the spatial⁵ and Fourier domains.^{8,9} Many of these investigations relied on simulated data,^{4,6,7,10–12} or a Fourier-based metric,^{3,4,8,9} which requires assumptions that do not necessarily reflect actual digital x-ray systems.^{6,10,11} Additionally, application of linear observer models to fluoroscopic and angiographic imaging systems has been investigated.^{13–26} However, the vast majority of the reports feature investigations on simulated data.^{13–17,19,21–25} Although simulated data have significant merits for system design, data acquired from real imaging systems are required to assess real-world system performance. Only a few reports describe experiments using real images; however, these investigations did not implement channelized Hotelling observer (CHO) models as described below.^{18,20,26}

For hardware assessment, the Bayesian ideal observer or the ideal linear observer is desirable.²⁷ When considering object signal-to-noise ratio (SNR) as the model’s figure of merit (FOM), the Hotelling model is the optimal linear observer.²⁷ However, practical limitations of acquiring the requisite number of images and computational constraints limit practical application of the spatial-domain Hotelling model for evaluating imaging hardware in a clinical environment.²⁸ Therefore, the CHO model has been used extensively for practical application. The underlying theory of the CHO model is well described in numerous sources.^{29–32} In brief, the CHO model operates as a modified

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matched filter to calculate a test statistic, which forms the foundation for statistical based decisions. There are two main differences between a simple matched filter and the CHO model. One is that the CHO model operates in channel space. Real-space images are converted into channels or channel outputs by calculating the scalar product of the image with a selected channel profile (e.g., Gabor, Laguerre–Gauss, etc.). This procedure reduces the dimensionality of the dataset from the total number of pixels in the image to the total number of channels selected, and in many cases, it reduces the performance of the model, more closely matching human performance. The second major difference between the CHO model and a simple matched filter is that noise correlation is corrected in the CHO model. Specifically, noise is decorrelated by first determining the covariance matrix of the channelized image set and then applying the inverted covariance matrix to the channelized image dataset prior to applying the matching filter.

If appropriate channels are thoughtfully selected, CHO can well approximate the ideal linear observer.^{27,33} Specifically, in this study, a spatial-domain CHO model was developed to assess detection of disk objects, utilizing the detectability index (DI) as its FOM. This DI differs from conventional SNR measurements, which are obtained by dividing an average signal amplitude value in a region of interest by the standard deviation of the background (i.e., noise). The DI described, here includes not only signal amplitude and background noise magnitude, but also noise texture and resolution of the object, thereby combining several image quality parameters into a single metric. The purpose of this investigation was to develop a practical method to objectively evaluate hardware imaging components of x-ray angiography systems. Specifically, the primary aims of this work are to (1) develop a CHO model, (2) verify the linearity performance of the model, and (3) apply this model to evaluate x-ray angiography equipment. In this report, we describe the development of our model, including rationale behind the choice of key channelization parameters—a critical component of model development. Moreover, we apply the CHO model to angiography x-ray systems and assess the uncertainty of the results, demonstrate the linearity of the model's response with respect to quantum SNR theory, discuss the influence of blur on detectability as a function of object size, and directly compare two different angiography systems.

2 Model Development

A spatial-domain CHO model was created in MATLAB® (Mathworks, Natick, Massachusetts) to measure the DI of disk-shaped objects in angiography images. Generally, the channelized models described in the literature have been found useful within the constraints of the experiments for which they were developed. In a channelized observer model, test statistics are not calculated directly from the image data, but rather from a collection of channel output scalar values that result from the scalar product of a channel profile and the image data. As such, differences in test statistics describe differences in the interaction of the channel profiles with the images. Therefore, selection of an appropriate set of channel profiles should be considered an important step in the development of a channelized observer model including the selection of: type, number, frequency, rotation, bandwidth, and phase of the functions that used to channelize the image data.^{33,34} A straightforward description of the channel profiles used in this work is provided in the following paragraphs. Further explanation and rationale for the selection

of the specific set of Gabor channel profiles used are included in Sec. 5. Gabor channel profiles were calculated using the following equation:³⁵

$$G = \exp \left[-\frac{(x - x_0)^2 + (y - y_0)^2}{2\sigma^2} \right] * \cos \{ 2\pi f [(x - x_0) \cos \theta + (y - y_0) \sin \theta + \beta] \}, \quad (1)$$

where x and y are the spatial position values, x_0 and y_0 are the center position values, θ is the orientation angle, β is the phase offset, f is the central frequency, and $\sigma = 0.56/f$ (for 1 octave of frequency bandwidth).

Four rotation angles (θ) were chosen to sample the images over 360 deg: 0, 45, 90, and 135 deg. The selection of channel profile spatial frequencies was based on the range of disk object sizes used. In total, 12 spatial frequencies were chosen: $\sim 0.1 * [0.5, 1, 1.5, 2, 2.5, 3, 3.5, 4.5, 6, 7, 8]$ cycles/mm. The frequency bandwidth of the channel profile was set to one octave, and two phases (0 and π) were used for each rotation and center frequency combination.

The range of frequencies of the Gabor channel profiles described above was selected specifically to complement the range of disk object sizes (0.5 to 4 mm) used. Figures 1(a) and 1(b) serve to demonstrate the relationship between the channel profile's spatial frequency and disk's size. In Fig. 1(a), the lowest frequency Gabor channel profile is presented along with the largest (4 mm, with a magnification factor of 1.5) and smallest (0.5 mm) disk objects used in this work. As seen in Fig. 1(a), the 0.5-mm disk [inner ring in Fig. 1(a)] is much smaller than the signal variation within the lowest frequency Gabor channel profile. Consequently, the scalar product between the 0.5-mm disk image and the Gabor channel profile is negligibly different than the scalar product of the Gabor channel profile with a background-only image. Conversely, the lowest frequency Gabor channel profiles is well matched in size with the largest disk [outer ring in Fig. 1(a)] and the scalar product between the Gabor channel profile and the largest disk is much different than that of the background-only image. Figure 1(b) shows the highest frequency Gabor profile along with the smallest and largest disk images. This corresponding channel output is relatively insensitive to the presence of the largest disk [outer ring in Fig. 1(b)] as the entire appreciable nonzero channel profile is encompassed within the boundary of the largest disk. Consequently, there is little difference between the scalar product of the largest disk with this highest frequency Gabor channel profile and the scalar product of the background-only image with this Gabor channel profile. However, because the highest frequency Gabor profile size matches well with that of the smallest object [inner ring in Fig. 1(b)], there is a significant difference between the scalar product of the smallest disk and this Gabor channel profile versus the scalar product of the background-only image and this Gabor channel profile. The difference in the interaction between the channel profiles and the signal present versus background-only images provides the basis for measuring object detectability.

Figure 1(c) shows a one-dimensional (1-D) frequency-space representation of the 12 channel profile frequencies used. For each of the 12 channel profile frequencies (f), four rotation angles (θ) and two phases (β) were used. The composite spatial frequency content of the entire channel profile set is shown in Fig. 1(d). Here, the spatial 2-D frequency contributions (i.e., the fast Fourier transform of the Gabor channel profiles) from all 96

Fig. 1 (a and b) Sample images of Gabor channel profiles, including the (a) lowest and (b) the highest spatial frequencies. Shown as dashed circles are the blur-free, ideal sizes of the 4-mm disk images at 1.5 $\times$ and the solid circles indicate the blur-free, ideal sizes of the 0.5-mm disk in contact images. (c) Plot of normalized frequency gain from each of the different spatial frequency Gabor channel profiles. (d) Plot of the total frequency gain from the entire Gabor channel profile set. Each value was calculated from the integral of the spatial frequency contribution over all directions (i.e., 360 deg). The inset shows the composite frequency gain from all 96 channels in frequency space.

channel profiles were summed and this summation is shown in the inset of Fig. 1(d). The main body of Fig. 1(d) shows a 1-D frequency-space representation of the entire set of 2-D channel profiles and is provided here to demonstrate the relative magnitude and range of spatial frequencies contained in the Gabor channel profiles.

There are many well-established FOM to describe the performance of an observer model including percent correct in a multiple alternative forced choice test, area under a receiver operator characteristic curve, and DI. The DI was selected as an FOM for this work to support a desire for an FOM that responds linearly with respect to quantum SNR theory over a large range of object sizes and quantum fluence conditions. Under the condition that the test statistics are normally distributed, the DI is defined as the following:³²

$$d' = \frac{\bar{\lambda}_s - \bar{\lambda}_b}{\sigma}, \quad (2)$$

where $\bar{\lambda}_s$ and $\bar{\lambda}_b$ are the expectation test statistics from the signal present and signal absent conditions, respectively, and σ is the root mean square of the test statistic standard deviations under both conditions. The test statistics were calculated in the spatial domain as

$$\bar{\lambda}_i = \Delta \bar{g}' U^{-1} \bar{g}_i, \quad (3)$$

where $\bar{g}_i$ is the (channelized) expectation or average image from all images obtained from the i 'th condition (signal present or signal absent), U is the intraclass scatter matrix, which is the average of the covariance matrices from the two conditions, and $\Delta \bar{g}$ is simply the difference in expectation (channelized) images for the signal present and signal absent conditions

(i.e., $\bar{g}_s - \bar{g}_b$).^{30–32} The standard deviation term in Eq. (1) can be found explicitly from the test statistic distributions (i.e., $\sigma = \sqrt{(\sigma_s^2/2) + (\sigma_b^2/2)}$) or simply by the following equation:³²

$$\sigma^2 = \Delta \bar{g}' U^{-1} \Delta \bar{g}. \quad (4)$$

Preliminary experiments demonstrated that variance calculated by Eq. (4) or explicitly from the test statistic distributions were equal, and the variance was calculated using Eq. (4) for the entirety of this work.

3 Methods

3.1 Phantom Construction and Image Acquisition

A set of four disk-shaped object phantoms were constructed. The phantoms were composed of iodine doped epoxy cylinders (175 mg/ml iodine concentration) with 0.5-, 1-, 2-, and 4-mm diameter and were embedded in individual sheets of 1.6-mm thick polyethylene. Both the sizes and composition of the disks were selected to represent the size and contrast content of vessels commonly imaged with angiography systems. The cylindrical disk shape was chosen so that the model's performance could be compared with quantum SNR theory.

For each imaging experiment, the disk was placed upon two 5-cm thick Lucite slabs of area 30 $\times$ 30 cm². The table pad was removed and the Lucite slabs were raised 14 cm above the table to enable a minimum magnification factor of 1.04 with a source to image distance of 110 cm. The experimental setup is shown in Fig. 2(a), where the lateral plane system is in a working position and the anterior-posterior (AP) plane system is rotated 90 deg with respect to the working position.

Fig. 2 (a) Photo of the experimental setup. (b) Sample radiographic images of each of the disks acquired with a magnification factor of 1.5. Note to better visually appreciate the disk signal, the presented image of the 0.5-mm diameter disk is an average of two individual frames.

In total, 60 different image acquisition configurations were investigated and the relevant parameters of these configurations are shown in Table 1. For all imaging conditions, unique disk signal present and signal absent images were acquired for each disk size. Two different x-ray systems were employed, the AP and lateral (LAT) planes of a biplane angiographic system (Artis Zee, Siemens Medical Solutions USA, Inc., Malvern, Pennsylvania). Both the AP and LAT system detectors are indirect conversion type detectors with CsI scintillators. The AP system had detector element dimensions of $154\text{ }\mu\text{m} \times 154\text{ }\mu\text{m}$, whereas the LAT system had detector element dimensions of $184\text{ }\mu\text{m} \times 184\text{ }\mu\text{m}$. The x-ray tubes for each of these detector systems were nominally the same. Three geometric magnification factors (1.04, 1.25, 1.5) and two focal spot sizes (small 0.6 mm and large 1.0 mm) were investigated for each disk size and imaging system.

For all experiments described herein, image processing parameters accessible in service mode were turned off in accordance with recommendations of the x-ray system manufacturer. Specifically, all spatial and temporal image processing was

turned off. The lookup table, to convert from x-ray intensity to pixel value, was set to a linear response with automatically varying offset (it was not possible to fix the linear response offset of the clinical system). To acquire the images, the disk objects were placed in the visually determined center of the imaging receptor. It is suggested that the minimum number of images used should be at least 10 times the number of independent channels to provide a reliable estimate of the covariance matrix.¹⁹ With 96 unique channels, at least 960 images are required to meet this recommendation. The frame rate of the system was set to 30 fps and four series of 10 s were saved. Each 10 s series contained 301 images. Therefore, a total of 1204 signal present and signal absent images were acquired for each imaging condition and disk size.

For a given disk size and imaging condition, expectation images were calculated as the average of the 1204 images. The subpixel center of mass of the x-ray shadow of the disks was determined algorithmically from the signal present expectation image and was used to define the center of the Gabor channel profiles. For a given imaging condition and disk size, the same Gabor center was used for the signal absent as for the signal present images.

With the angiography systems investigated here and as is the case with most angiography systems, x-ray output is controlled by an automatic exposure rate control system. This feature was active during the acquisition of all images throughout these experiments. Consequently, during different acquisition configurations, x-ray tube output varied automatically within a small range. A specific experiment was conducted to investigate the influence of dose variations on DI. For the AP system with the small focal spot and a magnification factor of 1.5, the 2-mm diameter disk was imaged with different detector dose settings: 120, 170, and 240 nGy/frame. The DI associated with changes in detector target dose setting was assessed to demonstrate linearity of DI value changes compared to expectations based on quantum SNR theory.

3.2 Uncertainty Estimation

3.2.1 Finite sampling uncertainty

Standard statistical “bootstrapping” analysis of the test statistic distributions for each of the image acquisition configurations was performed to estimate the inherent random sampling uncertainty associated with a finite sample set. New test statistic distributions were generated by resampling the real test statistic

Table 1 List of image acquisition parameters.	
Acquisition parameter	Value (s)
Source to image	110 cm
Field of view	22 cm (AP); 25 cm (LAT)
Focal spot size	Small (0.6 mm); Large (1.0 mm)
Geometric magnification	1.5; 1.25; 1.04
Disk diameter	0.5; 1.0; 2.0; 4.0 mm
Detector dose	170 nGy/frame (120 to 240 nGy/frame)
kVp	66 kV
Frame rate	30 fps
Number of images	1204
Antiscatter grid	Yes
Scattering material	10 cm of Lucite

distributions. A new DI value was then generated from the resampled distribution. This process was repeated 10,000 times and the standard deviation of the 10,000 indices was determined for each of the image acquisition conditions and used as an estimate of uncertainty of the DI values.

3.2.2 Disk position and experimental uncertainty

As mentioned previously, the disk was placed in the visually determined center of the image. Consequently, the placement was neither exact nor identical for each disk. Further, any deviation from the center of the x-ray beam results in changes in disk-detector position as the magnification factor changes. A specific experiment was conducted to investigate the influence on the disk position on the detector. For a single configuration (small focal spot and 1.5 magnification factor), the 2-mm diameter disk was imaged after being placed in four different off-center locations. The disk was displaced ~ 1 cm toward the corners of the detector from the visually assessed center of the image. This 1 cm translation is more than double the maximum deviation (4.2 mm) from the center for any disk imaging scenario that occurred through visually centering the disk in the remaining experiments.

There is also experimental uncertainty owing to slight variation in experimental setup. The magnification factor for each configuration is dependent on positioning of both the table height and the detector location. The table and detector were repositioned for each image acquisition configuration. The position of table and detector is registered in increments of 1 cm, which yields a total possible deviation of relative table versus detector position of up to 2 cm.

3.3 Model Performance Verification

To verify linearity of model performance over a relevant range of experimental parameters, changes in the DI were compared with expected changes based on quantum SNR theory derived in the classic Rose model.^{36–38} Under a limited scope of ideal conditions, the Rose model is a good approximation of the ideal observer.^{39,40} Thus, for acquisition conditions that yield quasi-ideal images, classic quantum SNR theory provides a basis to verify the performance of the CHO model developed here.

More specifically, the data acquired using different acquisition parameters as described in Sec. 3.1 were analyzed in several different ways and compared with expectations from the classic Rose model of signal detection. DIs were examined as a function of: (1) detector target dose, (2) geometric magnification, and (3) disk diameter and these relationships were compared with expectations derived from quantum SNR object detection theory. All DIs were normalized to the output of the x-ray system to account for minor systematic variations in x-ray output generated under different acquisition conditions, which resulted from automatic exposure rate control. Specifically, the square root of the air kerma per frame at the interventional reference point was used to normalize the DI for each imaging configuration.

In the comparison of DI and disk diameter, images of the small disks cannot be considered ideal due to image blur. To better compare experimental results with expectations from the Rose model, the following simulation study was performed. Ideal disk images of similar size to the real disks were simulated. A second set of simulated images was generated by blurring the ideal disk images with a Gaussian filter. Filter parameters were chosen such that the blurred images resulted in a close visual

match with real images (which were affected by blur). Real background noise fields were taken from the images acquired under the disk signal-absent condition and added to both the ideal and blurred simulated disk images. DIs were calculated for both the simulated ideal and blurred disk images and compared with expectations from the Rose model and subjectively compared to results from real data.

3.4 Model Application

The model was applied to quantitate the differences in DI due to the focal spot blurring in images acquired with the large and small focal spot sizes for the three different magnification factors. Additionally, the performance of both the AP and LAT systems was compared across all image acquisition configurations. All DIs were normalized by the square root of the air kerma specific to each image as described above.

4 Results

4.1 Uncertainty Estimation

4.1.1 Finite sampling uncertainty

Bootstrapping experiments were conducted to assess inherent uncertainty in the DIs that stem from finite sampling of the pixel distributions. The results from these experiments demonstrated that the relative or percent uncertainty associated with each DI ranged between 1.2% and 2.2%, with the greatest uncertainty for the 0.5-mm disk (Tables 2 and 3).

4.1.2 Disk position and experimental uncertainty

Both the AP and LAT systems show insignificant variability of the DIs caused by small displacements from the center of the imaging detector. These results are shown in Fig. 3, in which the DIs relative to the maximum DI found in each system are plotted versus the disk's position relative to the center. Additionally, a standard homogeneity test was performed on the data.⁴¹ This test confirmed that with high probability, the DIs for each system arose from the same statistical distribution. In other words, small differences in the DI associated with modest spatial displacement of the disk are within the measurement variance associated with finite sampling and are not statistically significant.

There is also uncertainty associated with variability of precise localization of the table and detector. There is approximately 2% relative uncertainty in the magnification factor of the disk. This leads to a $\sim 2\%$ relative uncertainty in the DI (Sec. 4.2.2). The combination of finite sampling uncertainty and experimental uncertainty yields a total relative DI uncertainty of $\leq 3\%$.

4.2 Model Performance Verification

4.2.1 DI versus detector dose

Results from the dose dependency experiments described in Sec. 3.1 show that the DI is proportional to the square root of the detector target dose as shown in Fig. 4. These results confirm that the DIs change as predicted by quantum SNR theory. This enables straightforward normalization of DI values measured using different acquisition configurations with moderately variable x-ray tube output. Considering all experimental configurations, the normalization factors ranged from 1.0 to 1.2.

Table 2 Percent standard deviation of the detectability indices (DIs) found through bootstrapping data acquired with the anterior-posterior (AP) system.

Magnification factor	AP system							
	Large focal spot				Small focal spot			
	Disk diameter				Disk diameter			
	0.5 mm	1.0 mm	2.0 mm	4.0 mm	0.5 mm	1.0 mm	2.0 mm	4.0 mm
1.5	2.17	1.44	1.43	1.48	1.75	1.36	1.35	1.37
1.25	1.95	1.45	1.44	1.45	1.66	1.39	1.34	1.37
1	1.95	1.34	1.33	1.36	1.89	1.29	1.28	1.29

Table 3 Percent standard deviation of the DIs found through bootstrapping data acquired with the lateral (LAT) system.

Magnification factor	LAT system							
	Large focal spot				Small focal spot			
	Disk diameter				Disk diameter			
	0.5 mm	1.0 mm	2.0 mm	4.0 mm	0.5 mm	1.0 mm	2.0 mm	4.0 mm
1.5	1.99	1.32	1.36	1.29	1.52	1.27	1.23	1.28
1.25	1.81	1.31	1.36	1.32	1.65	1.24	1.25	1.26
1	1.94	1.30	1.34	1.25	1.82	1.23	1.27	1.18

4.2.2 DI versus magnification

DIs of the 4-mm disk images acquired with the small focal spot for both the AP and LAT systems demonstrate a linear relationship between DI and geometric magnification (Fig. 5). This linear relationship is consistent with expectation based on quantum SNR theory.

4.2.3 DI versus disk size

Results from both AP and LAT systems and across all acquisition configurations indicate that the DIs are linearly dependent

on the disk diameter for the 2- and 4-mm disks, whereas DI values deviate from linear relationship for the 1.0- and 0.5-mm disks (Figs. 6 and 7). We believe that this deviation from linearity is due to the greater influence of blur on the smaller disks. This hypothesis is supported by the simulation of the influence of blur on synthesized ideal disk objects embedded into real noise images as demonstrated in Fig. 8, which demonstrate similar changes in DI as disk size decreases. Blurring of the disk image reduces the detectability across all disk sizes with respect to corresponding ideal, nonblurred images. However, for the smaller disks, blur results in a reduction of signal amplitude throughout a significant portion of the disk area, if not the entire disk area. This causes a greater reduction in DI values for the

Fig. 3 Relative detectability indices (DIs) as a function of position relative to the center of the detector. The disk was displaced several centimeters from the center of the detector for each independent position. DIs from the anterior-posterior (AP) system are shown in blue circles and the DIs from the lateral (LAT) system are shown in red squares. The error bars correspond to the finite sampling error.

Fig. 4 DI as a function of dose for the 2-mm disk and AP system. The fit was constrained to a zero y-intercept.

Fig. 5 Plots of the DI versus magnification factor for both (a) AP and (b) LAT systems. The linear fits of the data were constrained to a zero-intercept.

Fig. 6 Plots of the DIs versus disk diameter for the AP system. (a–c) Data acquired with the large focus with magnification factors of: (d) 1.5, (e) 1.25, and (f) 1.04. (d–f) Data acquired with the small focus with magnification factors of: (d) 1.5, (e) 1.25, and (f) 1.04. The linear fits are based solely on the data from the 4-mm disk and are constrained to a zero-intercept. The error bars are shown but within the boundaries of the symbols.

0.5- and 1-mm disks compared to the 2- and 4-mm disks, for which the blurred edges represent a relatively smaller fraction of the overall disk area. As shown in Fig. 8(a), the disk DIs of the ideal (not blurred) disks are linearly proportional to the disk diameter for all disk sizes. By contrast, DIs from the blurred images are reduced with a greater reduction observed in the small disk values relative to the larger disks, thus causing these values to deviate from the linear fit [Fig. 8(b)]. For this limited simulated dataset, the magnitude of this deviation of the DI values for the 0.5- and 1-mm disks compared to the 2- and 4-mm disks is similar to that observed with the real x-ray images of the disks.

4.3 Model Application

4.3.1 DI versus focal spot size

Figure 9 shows both the small focal spot and large focal spot relative DIs as a function of disk diameter across the three different magnification factors. The DIs were normalized to the small

focal spot DI value to clearly show the relative changes in DI with focal spot size across all disk diameters. Analysis of the DIs as a function of focal spot size across all three magnification factors and disk sizes provides several findings. First, in the case of a magnification factor of 1.04, there are no significant differences between large and small focal spot DIs for both the AP and LAT systems. This is the expected result as there is minimal opportunity for focal spot penumbral blur when the disk is close to the detector. As the magnification factor increases, the large focal spot DIs decrease relative to the corresponding small focal spot values, particularly for smaller disks. For the 4-mm disk, the two focal spot sizes result in nearly equal DI values for all magnification factors, whereas there are significant and substantial DI differences between focal spots with the smaller disks at higher magnifications. The greatest relative changes in DIs are found for the 0.5-mm disk and the 1.5 magnification factor in which images acquired with the large focal spot size yield ~35% lower values than images acquired with the small focal spot size.

Fig. 7 Plots of the DIs versus disk diameter for the LAT system. (a–c) Data acquired with the large focus with magnification factors of: (d) 1.5, (e) 1.25, and (f) 1.04. (d–f) Data acquired with the small focus with magnification factors of: (d) 1.5, (e) 1.25, and (f) 1.04. The linear fits are based solely on the data from the 4-mm disk and are constrained to a zero-intercept. The error bars are shown but are within the boundaries of the symbols.

Fig. 8 Plots of DIs versus disk diameter for (a) ideal and (b) Gaussian blurred simulated data. The linear fits in both plots were both constrained with a forced zero intercept. The fit in (b) is based on data from the two larger disks only.

4.3.2 DI versus AP and LAT systems

DIs from the AP and LAT systems were compared and these results are shown in Fig. 10 where DI values for both the AP and LAT detectors were normalized by the DI for the AP detector to facilitate presentation. There are two trends in the data. First, for most configurations, the two systems yielded similar DIs; most differences were less than 5% and not significant. Second, for the 0.5-mm disk imaged at higher magnification factors with both focal spot sizes, the LAT system yielded DIs, which were 10% to 15% greater than with the AP system.

5 Discussion

The relationship between DIs and dose, magnification, and disk diameter demonstrates that the CHO model presented here responds linearly as expected based on classical quantum SNR object detection theory. Classical object detection theory

is found on empirical findings regarding the response of the human visual system. Originally, Blackwell charted the light source detection response of human volunteers as a function of object contrast and size.⁴² These results were used by Rose et al., who formulated the classical quantum-limited theory of signal detection. In this theory, object SNR is directly related to the number of photons passing through an object. Under ideal conditions, both the object's signal and variance are directly proportional to the number of photons passing through it. The object SNR is proportional to $\sqrt{N}$, where N is the number of photons.

Within a limited scope of image conditions, the Rose model is a good approximation of the ideal observer^{39,40} upon which the channelized Hotelling model is based. While keeping all other acquisition parameters constant, increasing the detector target dose (i.e., increasing the number photons passing through the target) should result in an increase in the DI, such that it increases in proportion to the square root of the detector

Fig. 9 Plots of the relative DIs versus disk diameter. DIs for each disk size are normalized to the DI from the small focal spot. Plots correspond to the following imaging conditions: (a–c) AP system; (a) 1.5, (b) 1.25, and (c) 1.04 magnification factors, and (d–f) LAT system; (d) 1.5, (e) 1.25, and (f) 1.04 magnification factors.

Fig. 10 Plots of comparisons of relative DIs for the AP and LAT systems acquired with (a–c) large focal spot; and (a) 1.5, (b) 1.25, and (c) 1.04 magnification factors, and (d–f) small focal spot; and (d) 1.5, (e) 1.25, and (f) 1.04 magnification factors. The data shown here are normalized to the corresponding DI for the AP system.

target dose. The results shown in Sec. 4.2.1 demonstrate this relationship.

As shown in Sec. 4.2.2, the large disk DIs increase linearly with magnification. This is an expected result, as corroborated by previous reports,⁴ and can be explained as a consequence of increasing quantum fluence through the object. To magnify the image, the disk is positioned closer to the x-ray source. As a result, an increasing number of photons are passing through the disk (as well described by the inverse square law). The number of photons intercepted by the object increases proportionally

with the square of the magnification factor. Thus, under ideal imaging conditions (i.e., absence of blur), the DI increases linearly with magnification factor. Images acquired of the 4-mm disk with the small focal spot size well-approximate ideal imaging conditions. The blurred edge of this large disk accounts for only a small fraction of the total signal and as such the blurred disk image can be considered approximately ideal (i.e., “pill-box” image profile). Accordingly, detectability of this disk is expected to increase linearly with magnification, as shown in Fig. 5.

Increasing the size of an object will increase the number of photons that interact with the object. There is a linear relationship between the number of interacting photons and object area. Hence, as Rose described, SNR of an ideal disk image increases in proportion to the square root of the disk area, or otherwise linearly with the diameter of the disk.³⁹ From the results shown in Figs. 6 and 7, it is evident that 0.5- and 1-mm diameter disks significantly deviate from the linear fits based on the 4-mm disk data. The angiography systems utilized in this investigation, like any real-imaging system, include blur. Because their edges account for a relatively small fraction of their total signal, the 2- and 4-mm disks are good approximations of ideal disk images in the presence of a relatively modest blur. As image blur becomes a more significant factor with the smaller 0.5- and 1-mm disks, the image of the disks becomes less ideal, transitioning from a pillbox profile to a nondescript feature, the shape of which is determined by the interaction of system blur with the disk. The signal amplitude of the majority or entire disk is greatly reduced, which results in a greater deviation from the otherwise expected linear relationship. This behavior is confirmed by the simulation results presented in Fig. 8. A similar nonlinear relationship between disk SNR and disk diameter was reported by Tapiovaara and Sandberg for small objects.²⁰

The CHO model used here provides a quantitative evaluation on the impact of focal spot size and magnification in the non-ideal imaging conditions. Qualitatively, the results are as expected. When image blur becomes a more significant component (i.e., increasing focal spot, increase magnification, and decreasing disk size), it is expected that detectability of the disk will be reduced. There are many factors that affect image quality under these different conditions. Increasing the magnification causes competing effects on the disk detectability. Increasing disk dose and the primary (x-ray)-to-scattering ratio (assuming larger air gap) with larger magnification serves to increase detectability; however, this effect is accompanied by increased focal spot penumbral blur, which serves to reduce the detectability. The CHO model accounts for all of these variables and combines them into a single quality metric. This unique capability of the CHO model may provide utility to optimize acquisition protocols in a previously unachievable manner with conventional image quality evaluation methodologies.

Last, this work indicated few instances in which DIs from the AP system significantly differed from values found from the LAT system. These systems are known to have two different detectors with different element sizes, whereas the x-ray sources are nominally the same. In this work, detector pixel pitch differences did not yield differences in DIs. When comparing the two systems, and assuming equal focal spot sizes, the effect of detector blur should have the greatest effect on DI for the contact imaging condition (i.e., magnification factor of 1.04). In these imaging scenarios, no significant differences were found between the two systems. By contrast, significant differences were found for the smallest disk and under higher magnification conditions. The root cause of these differences is not known, but may well be related to slight differences in the actual focal spot characteristics between the two systems.

In regards to the process of selecting the channel profiles of the model, the literature contains examples of the application of various channel profile types including Laguerre–Gauss, ratio and difference of Gaussians, wavelets, difference of mesa, and Gabor.³¹ For this work, we chose to implement Gabor function channels because individual channel profiles are asymmetric

and may have general applicability to future work using asymmetric objects, and because there is a developing body of literature which indicates their successful implementation.^{7,30,43,44} Given that the objects in this work were symmetric, it may be reasonable to assume that a symmetric channel profile could have been used instead. Other channel profile types were not investigated. As indicated in Eq. (1), the Gabor channel profiles have several parameters which affect size and shape and that may be expected to influence how they interact with images of objects. Therefore, selection of channel profile parameters can be expected to influence test statistic values and model response. As part of developing the model presented here, the spatial frequency range and number of channels were iteratively adjusted to determine a set of channel profiles which resulted in a stable model, i.e., a model for which reasonable changes to the channel set had inconsequential effects on the results. In our experience, the exact spatial frequency content of the channel profiles had a minor effect on the magnitude of DI or linearity of the CHO model responses provided that the spatial frequency content of the channel profiles broadly covered the range of disk object sizes used (Fig. 1). As described in Sec. 3, 12 spatial frequencies were selected such that the primary central band of the channel profiles ranged in size from smaller than the smallest object to larger than the largest object and the remaining 10 channel profiles were nominally uniformly spaced in the spatial frequency domain. In our experience, inclusion of smaller or larger channel profiles did not substantially influence model results. Consistent with the findings of others,³¹ small changes in the number of channel profiles within this spatial frequency range were found to have little influence on the results.^{30,31} The channel profile set described here provides a single observer model that is applicable to the range of disk object sizes used.

The results from these experiments suggest that this model, which yields results with $\sim 3\%$ uncertainty, could be a powerful tool to assess subtle angiographic system performance differences under a variety of imaging conditions. In its current form, this model could be used in conjunction with the presented disks or similar objects to investigate changes in object detectability of raw (or minimally processed) data in a multitude of applications. Such applications could include evaluation of: equipment from different vendors, different antiscatter grids, air gaps, spectral filters, and the influence of numerous acquisition parameters (frame rate, kVp, etc.). Also, the general applicability of this or a similar model in the presence of nonlinear imaging processes, such as processed image data, is a topic of interest for future work.

While this work supports practical implementation of the CHO model for x-ray angiography system performance assessment, it is important to recognize limitations of this implementation and results. The model presented herein is based on the relatively simplistic task of detecting a disk-shaped object on a uniform background. Only unprocessed images were used. As such, the work presented here provides some insight into the technical aspects of system performance. However, interpretation of the results of this work must recognize that clinical use of the system requires differentiation of tubular vessels, not disks, and that patient anatomy results in a very nonuniform image background. For example, the relative influence of blur on a tubular vessel can be expected to be different than on the disk-shaped object used here. Also, the influence of “anatomical” and spatially correlated noise versus a uniform exposure background is an area of active research. Further, image

processing is a major component of angiography systems. The final image generated for end user visualization will have likely undergone many different proprietary processing steps. Differences in image processing can have a significant effect on clinical images. As this investigation was designed for and applied to unprocessed images, it cannot provide insight into the influence of image processing on perceived image quality. The model presented here has been designed for practical application and to provide a straightforward method to evaluate overall angiographic system quality as natively produced by the hardware of the imaging system. As discussed above, there are many opportunities to explore enhancement of this approach to better approximate clinical application of angiographic systems.

6 Conclusions

We have developed a CHO model, verified linearly of performance with respect to quantum SNR theory, and applied the CHO model to evaluate the detectability of disk objects for two x-ray angiography systems. An important feature of the CHO model is that it simultaneously considers the influence of object size and contrast, system resolution (or blur), and noise to provide a single metric of system quality. The CHO model developed here demonstrates a strong correlation with classical, quantum-limited image quality theory, when “raw” or minimally processed images are used. This correlation serves to substantiate the validity of application of the CHO model to assess angiographic system hardware performance. Through application of the model, differences in image quality generated under different imaging conditions are well quantified. The model yields DIs with a total uncertainty of $\sim 3\%$, providing a high degree of sensitivity. Consequently, subtle differences in system performance are measurable.

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