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Efficiency of the Human Observer Compared to an Ideal Observer Based on a Generalized NEQ Which Incorporates Scatter and Geometric Unsharpness: Evaluation with a 2AFC Experiment

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Abstract

Under certain assumptions the detectability of the ideal observer can be defined as the integral of the system Noise Equivalent Quanta multiplied by the squared object spatial frequency distribution. Using the detector Noise-Equivalent-Quanta (NEQ_D) for the calculation of detectability inadequately describes the performance of an x-ray imaging system because it does not take into account the effects of patient scatter and geometric unsharpness. As a result, the ideal detectability index is overestimated, and hence the efficiency of the human observer in detecting objects is underestimated. We define a Generalized-NEQ (GNEQ) for an x-ray system referenced at the object plane that incorporates the scatter fraction, the spatial distributions of scatter and focal spot, the detector MTF_D, and the detector Normalized-Noise-Power-Spectrum (NNPS_D). This GNEQ was used in the definition of the ideal detectability for the evaluation of the human observer efficiency during a two Alternative Forced Choice (2-AFC) experiment, and was compared with the case where only the NEQ_D was used in the detectability calculations. The 2-AFC experiment involved the detection of images of polyethylene tubes (diameters between 100–300 μm) filled with iodine contrast (concentrations between 0–120 mg/cm^3) placed onto a uniform head equivalent phantom placed near the surface of a microangiographic detector (43 μm pixel size). The resulting efficiency of the human observer without regarding the effects of scatter and geometric unsharpness was 30%. When these effects were considered the efficiency was increased to 70%. The ideal observer with the GNEQ can be a simple optimization method of a complete imaging system.

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Keywords

MTF; NPS; NEQ; DQE; SNR; system; generalized; detectability; observer; angiography; microrangiography

1. INTRODUCTION

The evaluation and optimization of medical imaging systems plays a major role in their evolution and advancement. Task specific assessment methods can lead to increased success in the diagnostic and therapeutic use of the imaging systems. At the same time, task-based optimization methods can promote the effective use of the radiation used for the image acquisition. In this paper we assess a prototype system used for x-ray angiographic imaging by comparing the human detection efficiency with a model ideal observer. The ideal observer model we use takes into account the detector signal and noise transfer characteristics, the x-ray source spatial properties, the patient scatter magnitude and spatial distribution, as well as the properties of the object of interest. The study of a clinical x-ray system usually involves the characterization of the detector, the scattering properties of the patient or a phantom and the characterization of the x-ray production, i.e. the properties of the x-ray tube. This paper is a follow-up from a paper presented last year by Kyprianou *et. al.* which presented a unified system's approach to the characterization of an x-ray system.¹ The properties investigated were the scatter properties of a uniform phantom that simulates the human head, the detector properties such as the signal transfer, the noise power spectrum, the signal to noise characteristics, and finally the effect of the focal spot and the beam energy on the image SNR.

In this work we compare the performance of the ideal observer described in ref.⁽¹⁾ with the performance of the human observer for a 2-Alternative Forced Choice (2-AFC) experiment performed by Ganguly *et. al.*² Furthermore, we estimate the variance of the ideal observer Signal to Noise Ratio (SNR), and the frequency dependent variance of the Normalized Noise Power Spectrum (NNPS) by a bootstrapping method. The requirements for the detection of objects of interest with the specific applications to neurointerventional procedures^{3, 4} where high resolution is necessary, such as the angiography of small features such as perforator vessels, are also investigated. Various efforts were made in the past to generalize metrics that describe the image quality from a system's point of view. In a series of papers, Wagner investigated the effects of the focal spot in magnification radiography⁵ and the effect of noise and scatter to the SNR,^{6, 7, 8} recognizing the necessity for a comprehensive approach⁹ to the imaging problem. Quantifications of scatter and its effect on the image quality were studied in the past by various authors. Muntz¹⁰ studied the effect of the focal spot size, magnification and scatter on the image quality combining them in a comprehensive function. Sorenson¹¹ and Krol¹² studied the effect of scatter as a function of magnification, using the DQE and contrast improvement factors to optimize the x-ray system. Rudin and Bednarek^{13, 14, 15} investigated the scatter rejection efficiency of optimized rotating scanning beams. Carlson¹⁶ generalized the contrast degradation and contrast improvement factors due to scatter by varying the magnification and the radiation field sizes. The effect of the focal spot on angiographic images was investigated in the past

by Doi¹⁷ and the focal spot unsharpness was included in the MTF and DQE by Shaw.³¹ Kyprianou previously unified^{1, 18} the study of the system properties by generalizing the concepts of the Modulation Transfer Function, the Noise Power Spectrum, the Noise Equivalent Quanta, and the Detective Quantum Efficiency which currently are only defined for the detector. These quantities were modified to include information about the scatter and the focal spot unsharpness of our system. In this paper we use the generalized system Noise Equivalent Quanta (GNEQ) in the determination of the ideal observer detectability for specific objects of interest. Here we also focus on the theoretical definitions of the generalized functions, as well as presenting the results obtained from the detectability simulations. The experimental methodology for obtaining and analyzing the data is rigorously presented in ref.⁽¹⁸⁾. Other factors such as patient motion blur and image lag are not investigated here.

2. MATERIAL AND METHODS

2.1. An ideal observer for small objects based on the imaging system signal transfer characteristics, noise properties, scatter distribution, and geometric unsharpness

Barrett *et. al.* and later^{19, 20} Albert and Maidment²¹ showed that the signal location averaged Signal to Noise Ratio (SNR) for a Signal Known Exactly/Background Known Exactly (SKE/BKE) detection task involving a discrete digital detector is:

$$\langle \text{SNR}^2 \rangle = G^2 \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} df_x df_y \left(\frac{\text{MTF}^2(f_x, f_y)}{\text{NPS}(f_x, f_y)} \right) |\Delta C(f_x, f_y)|^2 = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} df_x df_y \text{NEQ}(f_x, f_y) |\Delta C(f_x, f_y)|^2. \quad (1)$$

G is a constant proportional to the x-ray intensity (for a linear detector, G is the average graylevel of a flat field image), f_x and f_y are the spatial frequencies in the x and y directions, MTF is the presampled Modulation Transfer Function as defined by Fujita, Giger, and Doi.^{22, 23, 24, 25} The NPS is the Wiener spectrum as defined by various researchers in the past.^{26, 27, 28} C is the Fourier transformation of the difference between the signal image and the background. Using this definition the SNR is related to the ideal observer detectability.²⁰ For the evaluation of the integral the presampled MTF, as well as the NPS must be obtained beyond the Nyquist frequency f_{Nyquist} . Fujita²² showed a simple way to determine the MTF for frequencies beyond the Nyquist frequency. Since the NPS can only be measured from $-f_{\text{Nyquist}}$ to f_{Nyquist} its reflected replication can be used in the integral. This definition of the SNR can be used as an ideal observer for an object which is smaller than the detector area, and comparable or larger than the detector pixel size. The SNR can also be an evaluation of the efficacy of a detector system. In modeling a complete x-ray imaging system (see Fig. 1), accounting for the total imaging chain we will include the x-ray production (focal spot), attenuation, scatter and beam hardening through the patient/phantom, and a small object within the patient/phantom. The object should be relevant to the imaging task for which the imaging system is intended. For the purpose of obtaining an estimate of the SNR dependent on the system's properties such as scatter, focal spot geometric unsharpness, and detector properties, we defined a set of generalized system quantities that describe the imaging physics of our system.

The system generalized MTF is composed of three main components: the detector MTF_D , the focal spot MTF_F and the scatter MTF_S . The detector MTF_D is the standard presampled modulation transfer function as defined in the literature. The focal spot MTF_F is defined as the normalized modulus of the Fourier transformation of the magnified focal spot distribution calculated from its line spread function (LSF_F) as imaged through a slit camera.^{29, 30} The scatter MTF_S is defined as the frequency content of the spatial distribution of scatter entering the detector, and is calculated from the normalized modulus of the Fourier transformation of the derivative of the scatter Edge Spread Function (ESF_S).¹⁸ The scatter MTF_S was introduced by Seibert,³¹ Boone and Cooper³² in an effort to characterize the spatial distribution of scatter entering the detector, but also as an image post-processing scatter removal filter. The authors¹⁸ have used the scatter MTF_S and the focal spot MTF_F to generalize the definition of the MTF for the description of the complete x-ray imaging system. Combining the detector, the focal spot and the scatter spatial frequency information can provide us with important information about the behavior of our system.

The imaging chain starts with the x-ray production at the focal spot, whose spatial distribution is described by the focal spot $MTF_F(f')$. f' is the spatial frequency measured at the focal spot plane. For simplicity we use the radial frequency f assuming radial symmetry for the definitions of the various MTFs. The x-ray beam enters the patient, where a fraction of the beam ρ undergoes scatter whose spatial distribution is described by the scatter $MTF_S(f'')$. f'' is the radial spatial frequency at the detector plane, where the MTF_S is measured. The scatter MTF_S and the focal spot MTF_F are then referenced to the object plane. The object is assumed to be small enough to be considered uniplanar, inside the patient at magnification m . Scatter and focal spot unsharpness are two parallel processes³³ and can be added:

$$MTF_{F,S}(f, \rho, m) = (1 - \rho)MTF_F\left(\frac{m-1}{m}f\right) + \rho MTF_S(f/m), \quad (2)$$

where f without the prime is the radial spatial frequency in mm^{-1} referenced at the object plane, and $\rho = S/(S+P)$ is the scatter fraction. S and P are the scatter and primary intensities of the x-ray beam as measured at the detector (i.e. using the beam stop method). m is the magnification factor $m=H/(H-h)$ as can be seen in Fig. 1. $MTF_{F,S}(f)$ describes the signal transfer at the object plane, however without regarding the detector blur. Once this signal enters the detector it is convolved with the detector blur, hence we multiply the $MTF_{F,S}$ with the Modulation Transfer Function of the detector $MTF_D(f'')$, which is measured at the detector plane. The authors showed¹⁸ that the Generalized Modulation Transfer Function (GMTF) when expressed at the object plane takes the following form:

$$GMTF(f, \rho, m) = MTF_D(f/m) \left[(1 - \rho)MTF_F\left(\frac{m-1}{m}f\right) + \rho MTF_S(f/m) \right] \quad (3)$$

Notice that the scatter fraction ρ will be dependent on the phantom properties, the airgap between the detector and the phantom, as well as the radiation field size. Magnification as used here is in reference to objects within the scattering object such as blood vessels or stents within the skull. Eq. (3) reduces to the detector MTF for $m=1$ and $\rho=0$. Eq. (3) describes how the signal from a small object is transferred through the imaging system. The

further away the object of interest is from the detector, the size of its projected image is larger by a factor of m , therefore the detector spatial frequencies must be divided by the same factor. We chose to include the image degradation effect of scatter as part of the signal transfer function (in the Modulation Transfer Function) rather than to consider it as uniform noise in the linear imaging chain (in the Normalized Power Spectrum), as other authors did.³⁴ The reasoning behind this came from the empirical observation¹⁸ that the measured Normalized Noise Power Spectrum (NNPS) is independent from scatter when the exposure to the detector X is maintained constant. Unlike the NNPS however, the MTF is sensitive to scatter changes as Boone showed in the past, and a scatter spatial distribution can be measured.³⁵ The scatter spatial distribution is heavily dependent on the object, especially for non uniform objects.

The Generalized Normalized Noise Power Spectrum for a flat image can be modeled in terms of the detector electronic noise EN and the Poisson distributed quantum noise QN as follows:

$$\text{GNNPS}(f, m, \rho, P) = \frac{1}{m^2} \left((1 - \rho) \frac{\text{QN}(f/m)}{P} + (1 - \rho)^2 \frac{\text{EN}(f/m)}{P^2} \right). \quad (4)$$

Where ρ is the scatter fraction, and P is the primary radiation intensity. In certain circumstances is convenient to measure the detector NNPS_D with the scatter material in the beam. The GNNPS is therefore defined as:

$$\text{GNNPS}(f, m, X(\rho)) = \frac{1}{m^2} \text{NNPS}_D(f/m, X(\rho)). \quad (5)$$

$X(\rho)$ is the exposure to the detector measured with the scatter material near the detector. Notice that the GNNPS reduces to the detector NNPS_D when the scatter fraction ρ is zero, and magnification m is one. The detector Normalized Noise Power Spectrum (NNPS_D) is defined as the squared modulus of the Fourier transformation of the fluctuations about the mean of a flat field image, multiplied by the pixel area having units of mm². For the purpose of determining the frequency dependent variance of the NNPS_D, $\sigma^2_{\text{NNPS}(f_x, f_y)}$, we define a bootstrapped³⁶ replication of the NNPS_D as:

$$\text{NNPS}_D^b(f_x^u, f_y^v, X) = \frac{1}{\{G^b(X)\}^2} \frac{1}{N_{\text{ROI}}} \sum_{p=1}^{N_{\text{ROI}}} \frac{1}{N_{\text{I}}} \sum_{q=\text{rand}(1; N_{\text{I}}, b)} \left| \frac{\Delta x \Delta y}{N_x N_y} \sum_{i=1}^{N_x} \sum_{j=1}^{N_y} \left(\mathbf{I}_C^{b, q, p}(x^i, y^j) e^{-\frac{2\pi i x^i y^j f_x^u f_y^v}{N_x N_y}} \right) \right|^2, \quad (6)$$

where $\mathbf{I}_C$ is a flat field corrected uniform illuminated image with its mean value subtracted. $\text{rand}(1; N_{\text{I}}, b)$ is a function generating random integers in the range $[1, N_{\text{I}}]$, with a random seed having an index b . q is the index of the randomly selected image from the originally acquired image array of N_{I} images, and p is the index of a given Region of Interest (ROI) within the q th image (each image is divided into N_{ROI} non-overlapping ROIs). $\mathbf{I}_{\text{RAW}}$ is one of the N_{I} raw images obtained from the digital detector at an exposure X . $\mathbf{I}_C$ is calculated by:

$$\mathbf{I}_C^{b,q,p}(x^i, y^j) = \left[\frac{1}{N_x N_y} \sum_{i=1}^{N_x} \sum_{j=1}^{N_y} \left\{ \mathbf{I}_F^{b,p}(x^i, y^j) - \mathbf{I}_{DC}^p(x^i, y^j) \right\} \right] \left[\frac{\mathbf{I}_{RAW}^{q,p}(x^i, y^j) - \mathbf{I}_{DC}^p(x^i, y^j)}{\mathbf{I}_F^{b,p}(x^i, y^j) - \mathbf{I}_{DC}^p(x^i, y^j)} - 1 \right]. \quad (7)$$

i and j are the spatial indices. $\mathbf{I}_F$ is the bootstrapped realization of the flat field image of the p th ROI and is calculated as:

$$\mathbf{I}_F^{b,p}(x^i, y^j) = \frac{1}{N_I} \sum_{q=\text{rand}(1; N_I, b)} \mathbf{I}^{p,q}(x^i, y^j). \quad (8)$$

$\mathbf{I}_{DC}$ is the average dark field calculated from the N_{DC} acquired images:

$$\mathbf{I}_{DC}^p(x^i, y^j) = \frac{1}{N_{DC}} \sum_{k=1}^{N_{DC}} \mathbf{I}_{DC}^{p,k}(x^i, y^j). \quad (9)$$

G^b is the average gray-level value of each bootstrapped replicated image set, and is calculated as:

$$G^b(X) = \frac{1}{N_{ROI}} \sum_{p=1}^{N_{ROI}} \frac{1}{N_x N_y} \sum_{i=1}^{N_x} \sum_{j=1}^{N_y} \left\{ \mathbf{I}_F^{b,p}(x^i, y^j) - \mathbf{I}_{DC}^p(x^i, y^j) \right\} \quad (10)$$

x and y are the dimensions of the detector pixel in mm. The frequency bins will be $f_x = 1/(N_x - x)$ with similar expressions for the f_y . The mean bootstrapped NNPS_D and its variance are then defined as:

$$\text{NNPS}_{D,*}(f_x^u, f_y^v, X) = \frac{1}{N_b} \sum_{b=1}^{N_b} \text{NNPS}_D^b(f_x^u, f_y^v, X) \quad (11)$$

and

$$\sigma_{\text{NNPS}}^2(f_x^u, f_y^v, X) = \frac{1}{N_b - 1} \sum_{b=1}^{N_b} \left[\text{NNPS}_D^b(f_x^u, f_y^v, X) - \text{NNPS}_{D,*}(f_x^u, f_y^v, X) \right]^2. \quad (12)$$

Depending on the number of images N_I , N_b should be large enough so that the variance does not change as N_b increases. The bootstrapping method tries to recreate the statistical distribution of the measured quantity when only a limited number of observables can be obtained. However, this method of bootstrapping can be biased. For better estimates of the unbiased NNPS, methods such as the Bias Corrected and accelerated (BCa) bootstrap, proposed by Lee³⁷ can be used. Lee's method is also useful in reducing the number of bootstrap resamples required in order to determine the confidence intervals.

Tapiovaara and Wagner³⁸ identified the importance of an energy sensitive SNR definition, therefore the contrast C of the object of interest will be mathematically modeled as the line integral over energy of an object of thickness z :

$$\Delta C^{i,j}(kVp, L, r_{in}, r_{out}, w_I) = 1 - \frac{1}{\sum_{n=0}^{n_{kVp}} \varphi(E^n)} \sum_{n=0}^{n_{kVp}} \varphi(E^n) e^{-z^{i,j,n}(L, r_{in}, r_{out}, w_I)}, \quad (13)$$

where $\varphi(E)$ is the energy dependent x-ray fluence, and z is:

$$z^{i,j,n}(r_{in}, r_{out}, L, w_I) = 2 \text{rect}\left(y^j; L\right) \left[\text{rect}\left(x^i; 2r_{in}\right) \sqrt{r_{in}^2 + (x^i)^2} \{\mu_I(E^n) w_I + \mu_W(E^n) (1 - w_I) - \mu_V(E^n)\} + \text{rect}\left(x^i; 2r_{out}\right) \mu_V(E^n) \sqrt{r_{out}^2} \right] \quad (14)$$

z describes a hollow tube of length L , internal diameter r_{in} containing a solution of w_I part iodine with x-ray linear attenuation coefficient μ_I (in mm^{-1}), and $(1-w_I)$ part water with x-ray linear attenuation coefficient μ_W (in mm^{-1}). The tube has an external diameter r_{out} and is made of polyethylene with x-ray attenuation coefficient μ_V (in mm^{-1}). The attenuation coefficients are energy E dependent. i, j, n are the two spatial, and x-ray energy indices respectively. Even though Eq. (13) assumes parallel ray geometry Doi and Rossmann^{17, 39, 40} showed that the assumption is sufficient for small objects. They also showed that a small object can be assumed to be contained in a single plane, a property that allows the convolution of the focal spot MTF_F with the object spatial distribution in order to account for geometric unsharpness. Fig. 3 shows an example of the normalized C for the 100 μm vessel ($r_{in}=51 \mu\text{m}$, $r_{out}=203 \mu\text{m}$) with $w_I=0.1$. In the same plot the rect function of the pixel is also shown for comparison. The spatial distribution of the object's contrast is obtained by the Fourier transformation of Eq. (13),

$$\Delta C^{u,v}(kVp, L, r_{in}, r_{out}, w_I) = \sum_{i=-\infty}^{\infty} \sum_{j=-\infty}^{\infty} \partial x \partial y \Delta C^{i,j}(kVp, L, r_{in}, r_{out}, w_I) e^{-2\pi i x^i y^j f_x^u f_y^v}. \quad (15)$$

When evaluating Eq. (15), x and y are the x and y increments and should be much smaller than the pixel size.

Using the generalized quantities the location averaged, bootstrapped, DC suppressed SNR, and its variance σ_{SNR} can be defined as:

$$\langle \text{SNR}^2 \rangle_{\text{loc}*} = \frac{1}{N_b} \sum_{b=1}^{N_b} \langle \text{SNR}^2 \rangle_{\text{loc}|b} \quad \text{and} \quad \sigma_{\text{SNR}^2}^2 = \frac{1}{N_b-1} \sum_{b=1}^{N_b} \left[\langle \text{SNR}^2 \rangle_{\text{loc}|b} - \langle \text{SNR}^2 \rangle_{\text{loc}*} \right]^2. \quad (16)$$

where

$$\langle \text{SNR}^2 \rangle_{\text{loc}|b} = \frac{\Delta x \Delta y}{m^2} \sum_{u=-\infty}^{\infty} \sum_{v=-\infty}^{\infty} \left[\frac{(m^2 \Delta f_x \Delta f_y) |\Delta C^{u,v}(kVp, L, r_{in}, r_{out}, w_I)|^2 \text{GMTF}^2 \left(kVp, \rho, m \sqrt{(f_x^u)^2 + (f_y^v)^2} \right)}{\text{GNNPS}^{b,u,v}(kVp, X, \rho, m)} \right], \quad (17)$$

$f_x = 1/(N_x \cdot x)$ is the frequency increment in the x direction. A similar expression holds for the y direction. Eq. (17) assumes that the GMTF contribution to the SNR variance is significantly smaller than that of the GNNPS. Here we also assume radial symmetry for the MTF. The zero frequencies were omitted since it is very difficult to estimate the zero frequency NNPS experimentally. The SNR can also be expressed in terms of the Generalized system Noise Equivalent Quanta which is defined as:

$$\text{GNEQ}^{b,u,v}(kVp, \rho, X, m) = \frac{[\text{GMTF}^{u,v}(kVp, \rho, m)]^2}{\text{GNNPS}^{b,u,v}(kVp, X, \rho, m)}. \quad (18)$$

The signal entering the detector will be filtered through the GMTF and hidden by the detector and quantum noise, hence the GNEQ provides us with information about the resulting image quality.

For a low contrast object we can make some simplifications to the SNR which will allow us to study its dependence on some important system parameters. For $r_{in} \mu_{eff} \ll 1$, and an effective object attenuation coefficient that does not vary significantly with energy *i.e.* $\mu_{eff} \sim \text{const}$ then $\text{SNR}^2 \propto (\mu_{eff} r_{eff})^2 A_{eff} \propto w_I^2 r_{eff}^3 L$. r_{eff} is the effective radius of the vessel which takes into account the additive⁷ effect of the MTF blur, and the contribution from the inner and outer cylinders. μ_{eff} is the effective attenuation coefficient which is proportional to the percentage of iodine w_I in the vessel and A_{eff} is the effective projected area of the vessel. We can also study the effect of scatter to the SNR by assuming that the effect of the focal spot unsharpness is small, the electronic noise is small compared to the quantum noise, and that the scatter is uniformly distributed. Under those assumptions we see that $\text{SNR}^2 \propto P^2 (S + P)^{-1}$. Under the aforementioned assumptions we see that in order to double the SNR we must either: double the iodine concentration w_I , quadruple the primary fraction, or increase the vessel radius by 1.6.

2.2. The experimental setup

Fig. 1 shows a schematic of the x-ray imaging system experimental setup, simulating its clinical use. The microangiographic system consists of the x-ray source, the x-ray tube of the CAS 8000V (Toshiba America Medical Systems, Tustin, CA) angiography unit, the patient's head, or in our case a uniform head phantom, and the microangiography Region of Interest (ROI) detector.^{18, 41, 42} The detector is a full frame 12 bit, $1k \times 1k$ CCD camera, optically coupled to a CsI(Tl) scintillator via a fiber optic taper, with a $4.5 \times 4.5 \text{ cm}^2$ field of view, having an effective pixel size of 0.043 mm and can acquire images at 5 fps. In clinical use, the interventionalist will use the image intensifier to guide the catheter to the point of intervention and when high resolution image sequences at the region of interest are necessary, the microangiographic detector is positioned at the point of interest. When

acquiring images, the patient head must be as close to the detector as anatomically possible, in order to avoid the geometric unsharpness due to magnification. The detector is placed 100 cm away from the x-ray tube. The image intensifier of the angiographic unit is disabled and is not used during the high resolution image acquisition with the microangiographic detector.

A collimator is positioned at the x-ray tube plane reduces the amount of scatter reaching the detector by restricting the radiation field size, producing a square field of $5 \times 5 \text{ cm}^2$. For the purposes of our measurements a uniform, average human head equivalent phantom was constructed following the recommendations of AAPM Report 60. It consists of $30 \times 30 \text{ cm}^2$, 3.4 mm thick aluminum, 20 cm thick PMMA and was verified² to be equivalent to a human head for the Anterior Posterior position. The phantom is placed on the x-ray table and positioned 2.5 cm from the detector, which defines a clinically relevant configuration, allowing x-ray scatter to enter the detector. The detector entrance exposure was measured by replacing the detector with an ion chamber (Model 35050A dosimeter with model 96035B ionization chamber, Keithley Instruments Inc., Cleveland, OH). The focal spot available to us was the 0.8 mm nominal. Smaller focal spots could not be used because of tube load limitations. With this setup 30 images of a uniform, flat field were acquired with an exposure to the detector $X=0.8 \text{ mR}$ and 60 images of the dark current using the same exposure time, but without x-ray exposure to the detector. The detector, scatter, and focal spot MTFs were also measured for the same configuration and reported in the past;^{1, 18} these are shown in Fig. 3. The scatter fraction for this configuration is $\rho=0.44\pm0.01$. The Al -HVL of the bare x-ray beam was 2.85 mm, the Al -HVL after the phantom was 6.88 mm, and the Al equivalent of the x-ray table was 0.98 mm. These values were used for the determination of the x-ray spectrum $\phi(E)$ using the software from Report 78.⁴³

2.3. The 2-AFC perception study

Ganguly *et. al.*² performed the perception experiment as a way of evaluating the new microangiographic detector testing for small vessel detection. However, the ideal observer used to compare the human efficiency did not take into account for scatter or geometric unsharpness, even though Ganguly identified their effects on the MTF and DQE. Here we repeat the analysis of the human observer data in a slightly different way, accounting for inter-reader and case variability, and compare with the ideal observer defined in Eq. (16). The details of the experiment can be found in ref.⁽²⁾; here we briefly describe the procedure: Polyethylene hollow tubes of various inner diameters were filled with different iodine contrast solutions. The contrast agent used was OxilanTM (Cook Imaging, Bloomington, IN), with 62% Ioxilan solution, 300 mg/cc organically bound iodine. Table I shows the tube dimensions and iodine concentrations that were used. Six images were obtained per each configuration. The images were cropped down to 20 mm length as shown in Fig. 4, and the location of the display window was selected so that the vessel was randomly positioned either on the left (0) or the right (1) side of the frame. The window and level were predetermined and kept constant for the best possible contrast, and uniform presentation. The two possible locations for the vessel were marked on each frame using endpoint markers. Eleven observers (three neuroradiologists and six medical physicists) were trained with a small subset of the images ranging from very easy to very difficult to detect. The

required task is a 2-Alternative Forced Choice Experiment (2-AFC), where the observers were required to choose the correct location of the vessel out of two possible locations. Fig. 4, shows an easy and a difficult case. The experiment was performed in a dimly-lighted room on a high-end commercial CRT monitor with standard calibration. The images were presented to the observers using the freely available software ImageTool™ (University of Texas, Health Science Center, San Antonio, TX). The observers were not restricted in terms of time or positioning, however the images were presented in random sequence and the option of revising a decision was not available.

The results were recorded as 0 (left) and 1 (right) in a table that consists of the eleven readers, four vessel diameters, iodine concentrations (four or five different concentrations depending on the vessel size, see Table 1), and six realizations per each vessel diameter, and iodine concentration. For each case the correct location was also recorded (this allowed for reader left/right bias testing). The results were then divided per case, each case being a given vessel diameter for a given iodine concentration. For each case there were recorded responses of eleven readers, $N_r=11$, for the six realizations available, $N_c=6$. For a better estimation of the mean and variances of the detection probability p [or the Area Under the Receiver Operating Characteristic curve (AUC)] the results were resampled by bootstrapping.⁴⁴ To account for the inter-reader variability we created new arrays containing random selections of the readers, and to account for the case variability, the cases each reader read were also resampled randomly. From each bootstrapped replication of the data the AUC or p was calculated, and the process was repeated 5000 times. The results were tabulated and the mean p and its variance σ_p were then calculated.

For a two Alternative Forced Choice experiment the probability of detecting the object p can be related to the detectability d' and its variance σ_D^2 by^{20, 45}

$$d' = 2 \operatorname{erf}^{-1}[2p - 1], \quad \sigma_{d'}^2 = 4\pi e^{d'^2/2} \sigma_p^2. \quad (19)$$

The human detection efficiency η and its variance σ_η^2 are:

$$\eta = \frac{d'^2}{\operatorname{SNR}^2}, \quad \sigma_\eta^2 = 4\eta \frac{\sigma_{d'}^2}{d'^2}. \quad (20)$$

3. RESULTS

The detector, focal spot, and scatter MTF are shown in Fig. 3, reproduced from ref(18). Notice the low frequency dependence of the scatter MTF. Fig. 5 shows the two dimensional object C frequency distribution. The object is the same shown in Fig. 2, with $L=20$ mm, $w_I=0.1$, $r_{in}=0.51$ mm, and $r_{out}=0.203$ mm.

Fig. 6 presents: (a) the bootstrap estimate of the two dimensional $NNPS_D$, (b) the standard deviation associated with the NNPS as calculated from Eq. (11), and (c) the one dimensional $NNPS_D$ extracted from (a) with $f_y=0.09$ which is the first frequency bin, above the DC value. A value of $N_b=200$ bootstrap replications was determined to be adequate. The solid

curves indicate the band or 99% confidence interval determined by $3\sigma_{NNPS}$. It is important to note here that the log scale hides the fact that the standard deviation is increased at the lower frequencies.

The SNR and detectability d' were calculated for all the configurations, and are presented comprehensively in Fig. 6. The sum in Eq. (16) was carried out for continuously increasing values of f_x , and f_y until the SNR_b result did not change by more than $\varepsilon=10^{-5}$. The maximum values of f were never larger than $2f_{Nyquist}$ for this value of ε .

In Fig. 7a we compare d'^2 with SNR^2 for the ideal observer as described by Eq. (13), using $m=1.025$, and $\rho=0.44\pm0.1$, we obtain $\eta=0.7\pm0.3$. In Fig. 6b we compare d'^2 with SNR^2 for the ideal observer when $m=1$, and $\rho=0$, we obtain $\eta=0.3\pm0.1$. Notice that disregarding scatter and geometric unsharpness results in an overestimation of the ideal SNR, thereby underestimating the efficiency of the human observer.

Fig. 8 shows the SNR and d' as a function of w_I for the 100 and 150 μm vessels. As predicted in Section 2.1 the relationship between SNR or d' with w_I is linear with a determination coefficient $r^2=0.999$ for the SNR, and $r^2=0.77$ for d' . The linear fits allowed the determination of the contrast agent percentage w_I required to obtain an SNR or d' of 0.95 ($p=0.75$), as a function of the inner tube diameter $2r_{in}$. The results are presented in Fig. 9, with the fit $w_I(0.95)=a+b/r^{3/2}$. The first data point obtained from the human perception experiment was omitted from the fit since it shows a systematic deviation from the rest of the data points. A possible reason for this deviation is the large vessel wall thickness (see Table I), compared to the vessel's inner diameter which reduces the amount of additional iodine contrast that is required for threshold detection. This discrepancy could result in an effective diameter of the smallest vessel significantly larger than the inner vessel diameter when compared with the other vessel sizes.

4. DISCUSSION

From Fig. 3 we see that the detector MTF_D is nonzero at 10 cycles/mm. However the generalized MTF suffers a large degradation mostly due to scatter, as was presented in refs. (1, 18). In Fig. 5. we see that most of the frequency content of the smallest vessel (100 μm diameter) lies within frequencies smaller than 7 cycles/mm.

The bootstrapping method for estimating the $NNPS_D$ using a limited amount of images resulted in reduced variance, and a smoother frequency dependent $NNPS_D$ estimate, therefore no radial average was necessary. Furthermore this method made it possible for a frequency dependent estimate of the $NNPS_D$ uncertainty. We see in Fig. 6 that the $NNPS_D$ has a power law dependence on frequency, however for frequencies higher than 7 cycles/mm noise aliasing effects start becoming significant. Aliasing is not observed in the MTF_D since it becomes zero right after the Nyquist frequency, a characteristic of indirect detectors with small pixel sizes using CsI(Tl) as the scintillating phosphor.

The difference in SNR when scatter and geometric unsharpness are not considered is significant, resulting in underestimation of the average human detection efficiency. The SNR determination using the bootstrap method of Eq. (16) resulted in very small

uncertainties, something that allowed the validation of the linear relation (see Fig. 8) between the iodine concentration w_I and the inner radius of the vessel r_{in} predicted in Section 2.1, with coefficients of determination r^2 close to unity. The same linear model was used for the human observer detectability, however the large fluctuations of the data points resulted in much lower r^2 (0.77). The computational times required for the bootstrap SNR calculations totaled a few minutes on a standard high-end pc, making this method widely applicable.

The SNR data in Fig. 9 display a good agreement with the $1/r^{3/2}$ model for the iodine concentration w_I dependence, showing that the objects contrast ($\mu_{eff} r_{eff}$) is proportional to the inverse square of its effective projected area. Even though the same agreement cannot be evaluated for the human perception detectability d' results due to limited amount of data and the systematic deviation of the first point, the proximity of the data points to the model is remarkable.

5. CONCLUSIONS

We proposed an observer model that is based on measured properties of an x-ray imaging system that accounts for patient scatter, geometric unsharpness, the detector properties, and an analytical approximation of a small object of interest in neuro-angiography. The observer model correctly predicts the human observer detectability for an SKE/BKE detection task with an average efficiency of 70%. This methodology can be a simple and quick method for the evaluation and optimization of a complete imaging system, for a specific imaging task. Furthermore it can also be used for comparing two different systems designed for the same task with regards to their relative efficiencies. Further evaluation of the model observer is required in order to test some of the underlined assumptions, such as the scatter and focal spot unsharpness interaction terms in the definition of the SNR. For this purpose methods such the ones developed by Gagne,⁴⁶ Tapiovaara,⁴⁷ and Chakraborti⁴⁸ can be employed.

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References

1. Kyprianou, IS.; Rudin, S.; Bednarek, DR.; Hoffmann, KR. Study of the Generalized MTF and DQE for a New Microangiographic System. In: Yaffe, MJ.; Antonuk, LE., editors. Medical Imaging 2004: Physics of Medical Imaging; Proc. SPIE; 2004. p. 349-360.
2. Ganguly A, Rudin S, Bednarek DR, Hoffmann KR, Kyprianou IS. Micro-angiography for Neuro-vascular Imaging, Part 1: Experimental Measurements and Feasibility. Med Phys. 2003; 30(11): 3018–3028. [PubMed: 14655949]
3. Harrigan, MR.; Magnano, CR.; Paciorek, AM.; Nemes, B.; Hanel, RA.; Kyprianou, IS.; Rudin, S.; Guterman, LR.; Hopkins, LN. Experimental model of indirect revascularization using encephalomyosynangiosis and therapeutic cerebral angiogenesis. poster at The Seventh Joint Annual Meeting of the AANSCNS, Cerebrovascular Section and the ASITN; 2003.

4. Kyprianou, IS.; Wang, Z.; Hanel, RA.; Levy, E.; Nemes, B.; Rudin, S. Development of Methods for the use of the ROI microangiography in neurointerventional procedures (abstract). Scientific Program and 89th Scientific Assembly and Annual Meeting of RSNA, 2003, Chicago, Meeting Program; November, 2003; 2003. p. 172scientific exhibit 105PH-p
5. Wagner RF, Weaver KE, Denny EW, Bostrom RG. Toward a unified view of radiological imaging systems. *Med Phys.* 1974; 1(1):11–24. [PubMed: 4457746]
6. Wagner RF. Toward a unified view of radiological imaging systems. Part II: Noisy images. *Med Phys.* 1977; 4(4):279–296. [PubMed: 882063]
7. Wagner RF. Fast Fourier digital quantum mottle analysis with application to rare earth intensifying screen systems. *Med Phys.* 1977; 4(2):157–162. [PubMed: 850512]
8. Wagner RF, Barnes GT, Askins BS. Effect of reduced scatter on radiographics information content and patient exposure: A quantitative demonstration. *Med Phys.* 1980; 7(1):13–18. [PubMed: 7366537]
9. Sandrik JM, Wagner RF. Absolute measure of physical image quality: Measurement and application to radiographic magnification. *Med Phys.* 1982; 9(4):540–549. [PubMed: 7110085]
10. Muntz EP. Analysis of the significance of scattered radiation in reduced dose mammography, including magnification effects, scatter suppression, and focal spot and detector blurring. *Med Phys.* 1979; 6(2):110–117. [PubMed: 460060]
11. Sorenson JA, Floch J. Scatter rejection by air gaps: An empirical model. *Med Phys.* 1985; 12(3):308–316. [PubMed: 4010635]
12. Krol A, Bassano DA, Chamberlain CC, Prasad SC. Scatter reduction in mammography with air gap. *Med Phys.* 1996; 23(7):1263–1270. [PubMed: 8839422]
13. Rudin S, Bednarek DR. Improved contrast in special procedures using a rotating aperture wheel (RAW) device. *Radiology.* 1980; 137(2):505–510. [PubMed: 7433683]
14. Rudin S, Bednarek DR, Wong R. Design of rotating aperture cones for radiographic scatter reduction. *Med Phys.* 1982; 9(3):385–393. [PubMed: 7110067]
15. Bednarek, DR.; Rudin, S.; Wong, R.; Kaczmarek, R. Effect of a Multiple-Scanning Beam Device and Trough Filter on Scatter in Chest Radiography. *Medical Imaging and Instrumentation* 1985; Proc. SPIE; 1985. p. 13-18.
16. Carlsson GA, Nielsen B, Persliden J. Generalized use of contrast degradation and contrast improvement factors in diagnostic radiology. Application to vanishing contrast. *Phys Med Biol.* 1986; 31(7):737–749. [PubMed: 3749260]
17. Doi K, Rossmann K. The effect of radiographic magnification on blood vessel imaging with various screen–film systems. *Med Phys.* 1974; 1(5):257–261. [PubMed: 4449533]
18. Kyprianou I, Rudin S, Bednarek DR, Hoffmann KR. Generalizing the MTF and DQE to include x-ray scatter and focal spot unsharpness: Application to a new micro–angiographic system for clinical use. *Med Phys.* 2003; 32(2)
19. Barrett HH, Denny JL, Wagner RF, Myers KJ. Objective assessment of image quality. II. Fisher information, Fourier crosstalk, and figures of merit for task performance. *J Opt Soc Am A.* 1995; 12(5):834–852.
20. Barrett, HH.; Myers, KJ. *Foundations of imaging science.* Vol. 13. Wiley; New Jersey, NJ; 2004. Statistical decision theory; p. 813-911.
21. Albert M, Maidment ADA. Linear response theory for detectors consisting of discrete arrays. *Med Phys.* 2000; 27(10):2417–2434. [PubMed: 11099212]
22. Fujita H, Tsai D-Y, Itoh T, Doi K, Morishita J, Ueda K, Ohtsuka A. A simple method for determining the modulation transfer function in digital radiography. *IEEE Trans Med Img.* 1992; 11(1):34–39.
23. Dobbins JT III, Ergun DL, Rutz L, Hinshaw DA, Blume H, Clark DC. DQE(f) of four generations of computed radiography acquisition devices. *Med Phys.* 1995; 22(10):1581–1593. [PubMed: 8551982]
24. Giger ML, Doi K. Investigation of basic imaging properties in digital radiography. I. Modulation transfer function. *Med Phys.* 1984; 11(4):480–490. [PubMed: 6482844]
25. Fujita H, Doi K, Giger ML. Investigation of basic imaging properties in digital radiography. 6. MTFs of II--TV digital imaging systems. *Med Phys.* 1985; 12(6):713–720. [PubMed: 4079861]

26. Giger ML, Doi K, Metz CE. Investigation of basic imaging properties in digital radiography. 2. Noise Wiener spectrum. *Med Phys.* 1984; 11(6):797–805. [PubMed: 6513887]
27. Giger ML, Doi K, Fujita H. Investigation of basic imaging properties in digital radiography. 7. Noise Wiener spectra of II--TV digital imaging systems. *Med Phys.* 1986; 13(2):131–138. [PubMed: 3517612]
28. Dobbins JT III, Ergun DL, Rutz L, Hinshaw DA, Blume H, Clark DC. DQE(f) of four generations of computed radiography acquisition devices. *Med Phys.* 1995; 22(10):1581–1593. [PubMed: 8551982]
29. Doi K, Fromes B, Rossmann K. New device for accurate measurement of the x-ray intensity distribution of x-ray tube focal spots. *Med Phys.* 1975; 2(5):268–273. [PubMed: 1186633]
30. Shaw, CC.; Liu, X.; Lemacks, M.; Rong, JX.; Whitman, GJ. Optimization of MTF and DQE in magnification radiography - a theoretical analysis. In: Dobbins, J.; Boone, J., editors. *Physics of Medical Imaging*; Proc. SPIE; 2000. p. 466-475.
31. Seibert JA, Boone JM. X-ray scatter removal by deconvolution. *Med Phys.* 1988; 15(4):567–575. [PubMed: 3211049]
32. Cooper V, Boone JM, Seibert J, Pellot-Barakat C. An edge spread Technique for measurement of the scatter-to-primary ratio in mammography. *Med Phys.* 2000; 27(5):845–853. [PubMed: 10841386]
33. Cunningham, IA. *Handbook of Medical Imaging*. Beutel, J.; Kundel, HL.; Van Metter, RL., editors. Vol. 1. SPIE Press; Bellingham, WA: 2000. p. 79-162.
34. Burgess, AE. Evaluation of detection model performance in power-law noise. In: Krupinski, EA.; Chakraborty, DP., editors. *Medical Imaging 2001: Image Perception and Performance*; Proc. SPIE; 2001. p. 123-132.
35. Boone JM, Arnold BA, Seibert JA. Characterization of the point spread function and modulation transfer function of scattered radiation using a digital imaging system. *Med Phys.* 1986; 13(2): 254–256. [PubMed: 3702823]
36. Efron, B.; Tibshirani, RJ. *An introduction to the Bootstrap*. Chapman and Hall, Inc; New York, London: 1993.
37. Lee SMS. Nonparametric confidence intervals based on extreme bootstrap percentiles. *Statistica Sinica.* 2000; 10(2):475–496.
38. Tapiovaara MJ, Wagner R. SNR and DQE analysis of broad spectrum X-ray imaging. *Phys Med Biol.* 1985; 30:519–529.
39. Doi K, Rossmann K. X-ray images of small blood vessels in angiography: Question of isoplanatism. *Med Phys.* 1975; 2(6):201–206. [PubMed: 1143194]
40. Doi K, Rossmann K. Comparison of convolution and ray-tracing methods for computing small blood vessel images in angiography. *Med Phys.* 1977; 4(2):154–156. [PubMed: 850511]
41. Rudin, S.; Wu, Y.; Kyprianou, IS.; Ionita, CN.; Wang, Z.; Ganguly, A.; Bednarek, DR. Micro-angiographic detector with fluoroscopic capability. In: Antonuk, LE.; Yaffe, MJ., editors. *Medical Imaging 2002: Physics of Medical Imaging*; Proc. SPIE; 2002. p. 344-354.
42. Ganguly A, Rudin S, Bednarek DR, Hoffmann KR. Micro-angiography for neuro-vascular Imaging. II. Cascade model analysis. *Med Phys.* 2003; 30(11):3029–3039. [PubMed: 14655950]
43. Catalogue of diagnostic x-ray spectra and other data, *Inst. Phys. and Eng in Med. Report No. 78*, (1997).
44. Beiden SV, Wagner RF, Campbell G. Components-of-variance models and multiple-bootstrap experiments: an alternative method for random-effects, receiver operating characteristic analysis. *Acad Radiol.* 2000; 7(5):341–9. [PubMed: 10803614]
45. Burgess AE. Comparison between ROC and forced choice observer performance. *Med Phys.* 1995; 22(5):643–655. [PubMed: 7643805]
46. Gagne R, Chakraborti K, Thomas J, Gallas B, Myers K. Toward Objective Evaluation of Imaging Phantom Scores – ACR/MAP. AAPM 46th Annual Meeting, *Med Phys.* 2004; 31(6):1313–1343.
47. Tapiovaara, M. STUK Report: STUK-A196. Helsinki, Finland: 2002. Objective measurement of image quality in fluoroscopic x-ray equipment: FluoroQuality.

48. Chakraborty DP. Computer analysis of mammography phantom images (CAMPI): An application to the measurement of micro-calcification image quality of directly acquired digital images. *Med Phys.* 1997; 24(8):1269–1277. [PubMed: 9284251]

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Fig. 1.

The system setup: from the bottom of the image is the focal spot, phantom, with a vessel like object at distance h from the detector. The detector is at a distance H from the focal spot.

Fig. 2.

The three MTFs of the detector (solid), focal spot (dot), and scatter (dash) for comparison. The x-axis uses a logarithmic scale. [Graph reproduced from ref.⁽¹⁸⁾.]

Fig. 3.

(Top) The cross section of the $100\ \mu\text{m}$ inner diameter vessel/tube, filled $w_I=0.1$ iodine.

(Bottom) the normalized ΔC solid line and the pixel rect function for comparison.

Fig. 4.

The Fourier content of the object to be used in the SNR calculation. The long axis of the vessel is the x-axis in the image. The image on the right shows the central cross sections of the image on the right, in both the f_x and f_y directions.

Fig. 5. Sample images of vessels/tubes filled with different iodine contrast concentrations obtained with the microangiographic system. The images were presented to the observers and the probable locations of the vessel were indicated by using end markers.

Fig. 6.

(a) The two dimensional bootstrapped $NNPS_D$ (b) the associated standard deviation of (a), and a cross-section of (a) at $f_y=0.09$ cycles/mm, presented with a band marking the $3\sigma_{NNPS}$ confidence intervals.

Fig. 7.

The detectability D^2 vs. SNR^2 (a) using the GNEQ, accounting for scatter and geometric unsharpness, and (b) using the detector NEQ assuming zero scatter and no magnification. In both graphs the errors in the SNR are smaller than the point symbols.

Fig. 8. SNR and D vs. w_I % of iodine for the 100 μm vessel (top) and the 150 μm vessel (bottom). The straight lines are fits showing the linear relation between the quantities.

Fig. 9.

The threshold iodine w_I % (0.95) as a function of the vessel diameter. The curves are non linear fits to the model $w_I(0.95)=a+b/r^{3/2}$.

TABLE I

Vessel sizes and iodine concentrations used for detectability experiment. [reproduced from ref(2)]

No.	Vessel/Tubing specifications		Iodine concentration (mg/cc)
	Internal diameter (mm)	Wall thickness (mm)	
1.	0.102	0.152	30, 60, 75, 90, 105
2.	0.152	0.127	30, 60, 75, 90, 105
3.	0.203	0.102	0, 15, 30, 60
4.	0.280	0.165	0, 15, 30, 60