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Study of the Generalized MTF and DQE for a New Microangiographic System

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Abstract

We study the properties of a new microangiographic system, consisting of a Region of Interest (ROI) microangiographic detector, x-ray source, and patient. The study was performed under conditions intended for clinical procedures such as neurological diagnostic angiograms as well as treatments of intracranial aneurysms, and vessel-stenoses. The study was performed in two steps; first a uniform head equivalent phantom was used as a “filter”. This allowed us to study the properties of the detector alone, under clinically relevant x-ray spectra. We report the detector MTF, NPS, NEQ, and DQE for beam energies ranging from 60–100kVp and for different detector entrance exposures. For the second step, the phantom was placed adjacent to the detector, allowing scatter to enter the detector and new measurements were obtained for the same beam energies and detector entrance exposures. Different radiation field sizes were studied, and the effects of different scatter amounts were investigated. The spatial distribution of scatter was studied using the edge-spread method and a generalized system MTF was obtained by combining the scatter MTF weighted by the scatter fraction with the detector MTF and focal spot unsharpness due to magnification. The NPS combined with the generalized MTF gave the generalized system NEQ and DQE. The generalized NEQ and the ideal object detectability were used to calculate the Dose Area Product to the patient for 75% object detection probability. This was used as a system optimization method.

Keywords

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

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1. INTRODUCTION

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. In this paper we attempt to create a unified system's approach to the characterization of an x-ray system. The properties that we intent to investigate are the scatter properties of a uniform phantom that simulates the human head, the detector properties like the signal transfer, the noise power spectrum or the signal to noise characteristics, and finally the effect of the focal spot and the beam energy on the image quality. In addition the requirements for the visualization and detection of specific objects of interest are also investigated. These properties are studied for clinically relevant conditions with the specific application to neurointerventional procedures^{1,2} where high resolution is necessary for angiography of small features such as perforator vessels and small interventional devices.

The effect of scatter on the image quality was studied in the past by various authors: Muntz³ studied the effect of the focal spot size, magnification and scatter on the image quality combining them in a comprehensive function. Sorenson⁴ and Krol⁵ studied the effect of scatter as a function of magnification, using the DQE and contrast improvement factors to optimize the x-ray system. Carlson⁶ 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 Doi⁷ and the focal spot unsharpness was included in the MTF and DQE by Shaw¹⁵.

The authors have previously unified⁸ 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, as a figure of merit to evaluate our system and to optimize it in order to image 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 reference(8). Other factors such as patient motion blur and image lag are not investigated here.

2. MATERIAL AND METHODS

2.1. X-ray system description and experimental setup

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.^{9,10,11} 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 $4.5 \text{ cm} \times 4.5 \text{ cm}$ 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 75 cm away from the x-ray tube. This is the maximum distance the detector can be placed from the x-ray tube because of the location of the image intensifier on the c-arm. 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 that restricts the radiation field, positioned at the x-ray tube plane, is used to reduce the amount of scatter reaching the detector, and it produces a square field. Other methods of scatter reduction like air gap and grids are not considered here. The first method is not desirable due to the effect of the geometric unsharpness. On the other hand, grids that will work in the energy range used in neuro-interventions with the required high line density for our detector pixel size, are currently not available.

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 12 inch $\times$ 12 inch square, 3.2mm thick aluminum, 6 inches thick PMMA and was clinically verified⁸ to be equivalent to an average human head. The phantom is placed on the x-ray table and positioned either 50 cm away from the detector, which defines the scatter free geometry, or 2.5 cm from the detector, which defines the clinically relevant configuration, which accepts x-ray scatter. For the scatter-free case the beam is collimated before and after the patient to produce a field size of 16 cm² at the image plane, to eliminate scatter as much as possible. The radiation field α , when the phantom is adjacent to the detector, can be varied from 16 to 100cm². For each measurement 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 exposure to the phantom was measured by removing the phantom, placing the ion chamber at the location of the entrance surface, and with the patient table moved close to the x-ray tube to avoid back scatter. The focal spot available to us was the 0.8 mm nominal. Smaller focal spots could not be used because of tube load limitations.

2.2. The generalized system quantities

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,^{12,13} 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 camera¹⁴. 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)⁸. The scatter MTF_S was introduced by Seibert¹⁵, Boone¹⁶ and Cooper¹⁷ in an effort to characterize the spatial distribution of scatter entering the detector, but also to use it as an image post-processing scatter removal filter. The authors⁸ 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.

First we consider scatter. Seibert¹⁵ and Boone¹⁶ have shown that the scatter MTF_S can be added to the detector MTF_D weighted by the scatter fraction ρ . The detector MTF_D will then filter the scatter entering the detector, so that the generalized MTF becomes:

$$\text{GMTF}(f, \rho) = (1 - \rho) \text{MTF}_D(f) + \rho \text{MTF}_D(f) \text{MTF}_S(f), \quad (1)$$

where f is the spatial frequency in mm^{-1} and $\rho = S/(S + P)$ is the scatter fraction. S and P are the scatter and primary components of the x-ray beam entering the detector. The effect of magnification on the Modulation Transfer Function, the Noise Power Spectrum (NPS), and the Detective Quantum Efficiency (DQE) was introduced by Shaw¹⁸. Focal spot unsharpness affects the primary beam and is a parallel process¹⁹ with scatter blur. Therefore it has a multiplicative effect on the MTF_D but an additive effect on the MTF_S. Combining the focal spot MTF_F, the scatter MTF_S and the detector MTF_D will give us the generalized system GMTF. The authors showed⁸ that the GMTF takes the following form

$$\text{GMTF}(f, \rho, m) = (1 - \rho) \text{MTF}_D\left(\frac{f}{m}\right) \text{MTF}_F\left(\frac{m-1}{m}f\right) + \rho \text{MTF}_D\left(\frac{f}{m}\right) \text{MTF}_S\left(\frac{f}{m}\right), \quad (2)$$

where m is the magnification factor. Notice that the focal spot MTF_F only modifies the first term. Also, in using this equation for this study, the scattering object is assumed to remain fixed with respect to the detector and the scatter fraction ρ is independent of m .

Magnification as used here is in reference to objects within the scattering object such as blood vessels or stents within the skull. In magnification radiology, the scatter fraction will be dependent on the air-gap between the patient and detector. In such a case, the scatter fraction as a function of the air-gap must be determined. Such studies have been performed in the past using the virtual scatter point⁴. Eq. (2) reduces to Eq. (1) for $m = 1$. Eq. (2) describes the signal transfer through the 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 , while the object MTF frequencies at the detector will be smaller by the same factor.

The detector Noise Power Spectrum (NPS_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 mm^2 . The detector Normalized NPS as a function of spatial frequency f , measured at a given detector entrance exposure X with a NPS_D is defined as

$$\text{NNPS}_D(f, X) = \frac{1}{d^2(X)} \text{NPS}_D(f, X) \quad (3)$$

where $d(X)$ is the detector output signal usually physically measured as the average pixel value of the flat field image, provided that the detector is linear or linearized. Following the reasoning used for extending the definition of the MTF to include the magnification effects, we can define the Normalized Noise Power Spectrum GNNPS(f, X, m) at the object plane with magnification m as

$$\text{GNNPS}(f, X, m) = \frac{1}{m^2 d^2(X)} \text{NPS}_D(f/m, X). \quad (4)$$

The factor $1/m$ multiplies the frequencies, because at the object plane the spatial frequencies will be reduced by a factor of m . On the other hand, from the object point of reference the pixel area that the NPS_D is measured, will be smaller by a factor of m^2 . The NPS_D is proportional to the detector input exposure^{19,20} plus the detector's electronic noise ($\text{NPS}_D(X) = aX + b$). For a linear detector with flat field and offset corrected images the normalization factor $d(X)$ is directly proportional to the detector input exposure ($d(X) \sim X$). Therefore the NNPS_D in Eq. (3) as a function of the entrance exposure X takes the form

$$\text{NNPS}_D(f, X) = \frac{a(f)}{X} + \frac{b(f)}{X^2}, \quad (5)$$

where $a(f)$ and $b(f)$ are constants dependent on the frequency and are obtained by fitting the measured NNPS_D at different entrance exposures X , as seen in Fig. 1(b). The Generalized NNPS is the noise output of our system. When scatter enters the detector, the detector entrance exposure includes the scatter plus primary components $X \sim (S+P)$, therefore scatter is now a part of the detector input signal. From Eqs. (4) and (5) we can write the GNNPS as:

$$\text{GNNPS}(f, \rho, X, m) = \frac{1}{m^2} \left(\frac{a(f/m, \rho)}{X} + \frac{b(f/m, \rho)}{X^2} \right), \quad (6)$$

where the exposure to the detector X is measured with the scattering material near the detector. a and b are the constants previously determined by Eq. (5) for different scatter configurations.

The Generalized system Noise Equivalent Quanta is defined as the squared value of the system output signal-to-noise ratio and follows from Eqs. (2) and (6)

$$\text{GNEQ}(f, \rho, X, m) = \frac{\text{GMTF}^2(f, \rho, m)}{\text{GNNPS}(f, \rho, X, m)} = m^2 \frac{\text{MTF}_D^2(f/m) \left((1-\rho) \text{MTF}_F\left(\frac{m-1}{m}f\right) + \rho \text{MTF}_S(f/m) \right)^2}{\frac{a(f/m, \rho)}{X} + \frac{b(f/m, \rho)}{X^2}}. \quad (7)$$

The GNEQ is the absolute measure of the image quality. The signal entering the detector will be filtered through the GMTF and the noise properties of the detector, and will provide us with information about the resulting image quality. A higher GNEQ corresponds to an image with lower noise.

The Generalized system Detective Quantum Efficiency (GDQE) is defined as the normalized system GNEQ by the number of photons q_o per unit area that have been attenuated, beam hardened through the scattering object and have entered the detector. q_o is the input signal to noise ratio squared. Let $q_{in}(X) = X\phi_{in}$ be the actual number of quanta that enter the detector, where X is the exposure to the detector, measured with the phantom near

the detector, and ϕ_{in} is the photon fluence per mR averaged over the x-ray spectrum entering the detector.^{21,22} For the system GDQE, the object of interest can be at a magnification m . At that magnification, the detector's unit area is smaller by a factor of m^2 , therefore the fluence must be multiplied by m^2 . We define $q_o(X, m) = m^2 q_{in}(X) = m^2 X \phi_{in}$ which is the photon fluence at a given magnification, for a given scatter fraction ρ , at a detector entrance exposure X . The GDQE can be defined by dividing the GNEQ with $q_o(\rho, X, m)$

$$GDQE(f, \rho, X, m) \equiv \frac{GNEQ(f, \rho, X, m)}{q_o(\rho, X, m)} = \frac{MTF_D^2(f/m) \left((1-\rho) MTF_F\left(\frac{m-1}{m}f\right) + \rho MTF_S\left(\frac{f}{m}\right) \right)^2}{\phi_{in} \left(a\left(\frac{f}{m}, \rho\right) + \frac{b\left(\frac{f}{m}, \rho\right)}{X} \right)}. \quad (8)$$

The zero frequency of the GDQE(0) is the same as the detector DQE_D(0), therefore the GDQE is consistent with the definition of the detector DQE_D. As the scatter fraction ρ goes to zero, only the primary radiation enters the detector, and as magnification m goes to 1, there are no effects from the focal spot, hence the detection efficiency is determined solely by the detector. We see that the final form of the GDQE is dependent on the detector entrance exposure, or equivalently the number of input quanta. This dependence though is not significant since the second term in the denominator is small compared to the first. The utility of a system GDQE is sometimes debatable since it is not an intuitive system parameter and it does not relate the risk to the patient. For a complete system, a more useful quantity would be a figure of merit that compares the image quality as a function of the stochastic risk to the patient, since the patient is an important part of our system. As we will see in the next section detectability could be used as a system figure of merit (FOM).

For the system under investigation the scatter fraction can be varied by adjusting the x-ray beam collimation (the radiation field size α). The authors have modeled⁸ the relation between the scatter fraction ρ and the radiation field size α for this system and found it to have the empirical functional form

$$\rho(\alpha) = \ln \left\{ (1 + c\alpha)^d \right\}, \quad (9)$$

where c and d are constants that depend on the system parameters, like the thickness of the patient, and the distance of the patient from the detector. It was shown^{8,23} that the scatter fraction does not depend on the tube voltage kVp for the range of radiation field sizes used in our experiment. Fig. 1(a) shows the relation between the scatter fraction and radiation field size. Using Eq. (9) we can express the generalized system functions with respect to the radiation field size α since it is a directly adjustable system parameter. The GMTF and GNNPS were measured for a range of tube voltages (kVp). The photon fluence also depends on the beam energy spectrum and it was simulated²² and modeled for our system at the range of kVp's we used. We can therefore express the generalized quantities with respect to all those variables that effect our system such as: the spatial frequency f , the magnification m , the collimated radiation field size α , the exposure to the detector X , and the tube voltage kVp:

$$\begin{aligned} & \text{GMTF}(f, m, \alpha, kVp), \quad \text{GNNPS}(f, m, \alpha, kVp, X), \\ & \text{GNEQ}(f, m, \alpha, kVp, X), \quad \text{GDQE}(f, m, \alpha, kVp, X) \end{aligned} \quad (10)$$

2.3. System optimization

The set of generalized functions of Eq. (10) can be used to optimize the system for the specific task of neuro-angiography. While detector properties such as MTF_D and DQE_D can describe the detector without regard to its clinical use, in a similar manner the generalized quantities, even though they take into account the focal spot unsharpness and scatter from the patient, cannot give information about detectability in clinical tasks. For example in Fig. 2 we show an optimization of the GMTF with respect to magnification. The shaded region shows the maximum magnification one could obtain for a scatter free geometry using the 0.8 mm focal spot without focal spot degradation. The thick curve inside the region shows the magnification which will maximize the GMTF at a given spatial frequency. Even though this graph is useful for optimizing the GMTF, it cannot give us information on whether detecting an object at a given magnification is possible, or not. A method that uses the system x-ray properties as well as the object of interest properties would be a more complete optimization technique.

2.4. Object detectability

Various methodologies have been used to determine those parameters to obtain optimum system performance. Cunningham and Shaw²⁴ used the NEQ and DQE, Dobbins *et. al.*²⁵ used the contrast to noise ratio, and Han²⁶ used the Detail to Noise ratio. The choice of detectability as a figure of merit for system optimization followed naturally after a perception study done by Ganguly *et. al.*⁹ for the specific system under investigation. Ganguly in her paper performed a two Alternative Forced Choice (2AFC) experiment where observers read images, obtained using the same system, of small vessels of different sizes, filled with different concentrations of iodine contrast solution. Ganguly then related the experiment with the human observers to the ideal observer. We will use this result to optimize our system while considering the dose to the patient and our system's capabilities.

By definition, the detectability by an ideal observer uses the detector output information to detect a specific object. The spatial frequencies of the object's x-ray transmission signal are used as an input to the system and filtered through the system's noise and signal transfer function. Using the generalized quantities defined in the previous section the detectability $\mathcal{D}$ can be defined:

$$\mathcal{D}^2(m, \alpha, X, kVp, G_O) = \frac{\Delta x \Delta y}{m^2} \int_0^{f_y} \int_0^{f_x} |O(f_x, f_y, G_O)|^2 \text{GNEQ}(\sqrt{f_x^2 + f_y^2}, m, \alpha, X, kVp) df_x df_y, \quad (11)$$

where $O(f_x, f_y, G_O)$ is the spatial frequency distribution of the object, and G_O is a set of variables that describe the attenuation properties, and size of the object. $f_x(f_y)$ and $x(y)$ are the spatial frequency and the detector pixel size in the $x(y)$ direction. For our detector the pixel size is 0.043 mm. The integral is multiplied by the pixel area over magnification to cancel out the units from the definition of the GNEQ. Notice also that since our object is two

dimensional the one dimensional GNEQ is radially rotated assuming radial symmetry. This is a good first approximation since there is radial symmetry in the NPS of our detector.^{8,9} This approximation does not take into account the change in the focal spot shape between the parallel and perpendicular anode to cathode directions. The effects of the focal spot⁷ and scatter to the imaged object are handled by the GNEQ. The detectability definition assumes that the number of quanta that are used to create the image of the object of interest is the same as the number of quanta entering the detector and produce an exposure X , which may include scatter from the patient. That amount of quanta will be filtered through the GMTF where only the quanta corresponding to the primary beam that contain information about the object will pass through the detector and focal spot MTFs and produce the output signal. That output signal, when integrated will give the proper definition of detectability.

We assume blood vessels of various radii r filled with $(1-w_I)$ part of blood of linear attenuation μ_B and with w_I part of iodine of linear attenuation μ_I . Another simplified assumption is that the vessel is straight with a fixed length L , and has a cylindrical form. Using these assumptions the attenuation of the blood vessel is:

$$O(x, y, kVp, w_I, r) = \left\{ \theta(y + \frac{L}{2}) - \theta(y - \frac{L}{2}) \right\} \left\{ 1 - \exp \left[- \{ \mu_I(kVp) w_I + \mu_B(kVp) \{ 1 - w_I \} \} z(x, r) \right] \right\}, \quad (12)$$

where $\theta(x)$ is the unit step function. The vessel profile in the z direction is defined as:

$$z(x, r) = 2 \left\{ \theta(x+r) - \theta(x-r) \right\} \sqrt{r^2 - x^2}, \quad (13)$$

Since the GNEQ is measured at the object plane magnification m does not modify the object. The frequency distribution of the object is obtained by

$$O(f_x, f_y, kVp, w_I, r) = \frac{2 \sin \left(\frac{f_y L}{2} \right)}{f_y L \sqrt{n_x n_y}} \left| \sum_{x=0}^{n_x-1} \sum_{y=0}^{n_y-1} \left[\left\{ 1 - e^{-\{ \mu_I(kVp) w_I + \mu_B(kVp) \{ 1 - w_I \} \} z(x, r)} \right\} e^{-\frac{2\pi i x y f_x f_y}{(n_x-1)(n_y-1)}} \right] \right| \quad (14)$$

where $n_x(n_y)$ is the number of data in the $x(y)$ direction which for our detector are 1024. The resulting object data are binned into 64 bins so as we retain the same sampling rate as for the GNEQ8. Fig. 3(a) shows the vessel x-ray transmission signal in real space, while (b) shows its frequency components and (c) shows the rotated GNEQ into two dimensions.

The GNNPS and GNEQ are defined in terms of the detector entrance exposure X and therefore the detectability $\mathcal{D}$ will also depend on it as we see from Eq. (11). The detector entrance exposure is not an intuitive system parameter, and relating it to the amount of dose received by the patient is not straightforward. Detect-ability may be expressed as a function of either the patient entrance exposure, or the mAs setting for the x-ray tube. These two parameters will allow us to optimize our system by reducing the dose to the patient. Also, we can relate detectability with the x-ray unit's mAs-kVp techniques. This could lead to an advanced form of "Automatic Exposure Control" (AEC) where the neurointerventionalist will indicate to the system the point of interest and the system will adjust the technique for the optimum detectability. Fig. 4(a) shows the calibration of the system as a function of the radiation field size and x-ray tube voltage. (a) shows the exposure to the patient per

exposure to the detector calibration and Fig. 4(b) shows the mAs per exposure to the detector calibration curve. Both data sets were used to determine the detectability as a function of the patient dose-area product (DAP) [*Dose in Air·Area* in (mGy·cm²)] (from the patient entrance exposure), and the mAs technique. The reason we chose to use the dose-area product is that for larger radiation field sizes the patient will receive a larger integral dose and therefore the DAP is a more direct indication of the stochastic risk to the patient.

3. RESULTS

The experimental procedures for the image acquisition and manipulation, as well as the details for the data analysis are described in detail in a different paper by the same authors⁸.

3.1. Generalized system functions

In this section we present the dependence of the generalized functions of Eq. (10) on the system properties. The MTF_D , MTF_F and MTF_S were measured for 60, 70, 80, 90, 100 kVp. The scatter fraction was measured using the standard beam stop method for square radiation field sizes of 0, 16, 36, 64 and 100 cm² (where 0 cm² defines the scatter free configuration), and for the same range of kVp's. The scatter fractions are plot in Fig. 1(a). For the range of beam energies and radiation field sizes we measured ρ was independent of the kVp. The GMTF was calculated using Eqs. (2) and (9) and is presented in Fig. 5. In (a), the GMTF is plotted for a range of magnifications for a scatter-free geometry at 80 kVp. We decided to use a log plot despite the usual presentation of the MTF, since a log plot enhances the separation of the curves that would otherwise overlap in a linear plot. This magnification range corresponds to objects within the head phantom ranging from the surface to the center, at about 12.5 cm from the detector. We observe that magnification degrades mostly the higher frequencies. In (b) the GMTF is plotted for different scattering configurations, and we see that the effect of scatter produces a low frequency drop.

Comparing the case for the maximum scatter at $m=1$ and the case where scatter is zero for $m=1.1$, the overall GMTF is worse when $m=1.1$ due to geometric unsharpness because of the relatively large focal spot that was used. In (c) we plot the GMTF for different x-ray tube kVp's. Even though a small range of frequencies was selected in order to exaggerate the differences in the GMTF, there is not a significant dependence on the kVp. The 60 kVp curve is slightly higher than the others due to the energy absorption dependence of the CsI module. The most common working conditions were chosen to be presented in (d) with the smallest possible collimation that doesn't affect the field of view and for the magnifications within the head phantom. We observe that with the smallest magnification the GMTF is at about 1% at 9.5 cycles/mm.

The detector NPS was measured at three different scattering conditions: $\rho=0$, 0.56, and 0.62 corresponding to the radiation field sizes $\alpha=0$, 64 and 100 cm² as seen in Fig. 1(a). The detector NPS measured with the same detector entrance exposure, at the different scattering configurations, was found to be independent of the scatter fraction⁸. This was expected since the NPS is a measure of the detector noise properties, and therefore is independent of whether the input signal is caused by scatter or not. For each scatter condition, the NPS was measured at three different detector entrance exposures X ranging from 0.6 to 2.8 mR and

for the range of kVp's. Using Eqs. (6), and (1) we constructed the generalized system GNNPS. Fig. 6 shows the GNNPS as a function of the kVp and spatial frequency. We see that GNNPS is independent of the kVp. The different surfaces in the plot correspond to different detector entrance exposures.

The GNEQ is plotted in Fig. 7, where we present selected frequencies as different surfaces, as a function of the radiation field size, and magnification. This way the dependence on the two quantities can be appreciated. Since the GNEQ has the information from both the GMTF and GNPS, their behavior is expected to be transferred to the GNEQ. As we see scatter mostly affects the lowest frequencies, while the magnification gives a significant degradation to the higher frequencies of the GNEQ.

The first two plots of the GDQE (a) and (b) in Fig. 8 have a similar functional behavior as the corresponding curves of the GMTF in Fig. 5. For each plot in Fig. 8, the top thick curve shows the ideal (no scatter, no magnification) detector DQE_D for comparison. We observe that the GDQE drops even when moderate amounts of scatter enter the detector, as can be seen in all plots of Fig. 8. The combination of the focal spot unsharpness and scatter has a severe effect on the GDQE as seen here. From graphs (c) and (d) we see that there is no significant dependence on either the detector entrance exposure or the x-ray tube voltage.

3.2. Threshold detectability and system optimization

The ideal detectability for this work was calibrated against the probability of object detection which was calculated for our system by Ganguly⁹ who performed a 2AFC experiment. We used a threshold detectability of $\mathcal{D}_T = 0.2$ for a 75% detection probability.²⁷ Using the threshold detectability we numerically solve Eq. (11) for the detector exposure while varying the different variables of our system. The conversion factors plotted in Fig. 4 were used to convert the detector exposure to the patient DAP and the mAs technique. Fig. 9 shows the DAP to the patient as a function of the tube voltage kVp for different vessel sizes all filled with 110 mg/ml iodine which corresponds to $w_I = 0.11$. The length of the vessels is always constant with $L = 20\text{mm}$. The measurements were done for $m = 1.1$ for the radiation field of 16 cm^2 . On each curve the mAs technique that must be used at the given kVp to obtain the threshold detectability is also displayed. We see that there is a minimum in the DAP to the patient at 70 kVp.

In Fig. 10 we plot the DAP to the patient as a function of the magnification m . The right vertical axis shows the mAs technique for the 70 kVp tube voltage that was used. Fig. 11 is the DAP to the patient a function of the radiation field size. Similarly the numbers on the curve are the mAs technique for the 70 kVp. As expected the more scatter enters the detector the higher the dose is to the patient, in order to maintain a 75% probability of detection, and similarly the mAs must increase. The data on both figures were obtained for actual clinical scenarios with realistic vessel sizes, iodine concentrations, x-ray techniques, and patient size.

4. DISCUSSION

From the GMTF, GNEQ and GDQE graphs (Figs. 5~8) we observe that the functional relation of the generalized quantities with the magnification is the same and it comes mostly from the explicit dependence of the GMTF on m . Generally the system suffers from focal spot unsharpness especially at the high frequencies. Scatter affects the system through the GMTF dependence since it filters the input signal to the detector and mostly the primary part of the radiation is transferred though. The system GDQE is degraded by the scatter through the GMTF because part of the input quanta to the detector is noise due to scatter. The GDQE for a complete system must be interpreted with caution, since depending on its definition it will provide us with different information. Here the GDQE is a measure of the system's detection efficiency of the primary signal exiting the patient as a function of kVp, magnification and radiation field size.

The dependence on the radiation field size does not suggest any form of optimization. Also our data did not show any significant energy dependence on the generalized quantities. The latter suggests that the detector has a good energy response. The object detectability though in Fig. 9, indicates that the range between 70 and 90 kVp's are to be preferred with regards to the dose to the patient. This is due to the iodine contrast, attenuation coefficient energy dependence, having a peak at 32.3 keV and producing a higher contrast image at the lower kVp's. The lowest kVp (60) though, is not the most desirable, because the high attenuation through the patient, forces a dramatic increase of the mAs, tube loading and patient dose. For the 60 kVp the highest exposure times are used, resulting in potentially increased motion artifacts in the image, something that must be avoided in angiography. Therefore the best solution would be to use a kVp in the mid-range like for instance the 70–80 kVp. It allows a reasonable technique to be used, (e.g. 400 mA and 60 ms for the 100 μ m radius vessel), while it does not increase the patient dose significantly.

In a similar manner, studies must be performed for other objects composed of different material, like steel and Nitinol, that are used for interventional devices⁹. For each material, we expect the results to be different as a function of the kVp. Even though one might expect that the magnification degradation of our system would prevent the detection of small objects, our clinical experience and the detectability calculations show the opposite! The optimization for the GMTF in Fig. 2 should be properly interpreted in combination with Fig. 3(b). The reason is that even a relatively small object has a significant amount of low frequency components. As shown in Fig. 3(b), a vessel of diameter of 200 μ m has most of its frequency content below 6 cycles/mm for $m=1.1$. When the object frequencies are multiplied by the GNEQ and integrated, the object is still detectable. This is shown in Fig. 10, where for the patient DAP is the lowest at $m=1.04$.

The graphs of the DAP to the patient vs m and α show that the GDQE must be used together with the object detectability in order to evaluate the efficiency of a complete system. Fig. 10 shows that for the ranges of magnifications inside the phantom, the object can still be detected with the same probability with very little increase of the patient DAP. For the same magnifications the GDQE suffers a significant degradation at the higher frequencies, therefore the relation between the GDQE degradation and object detectability with regards

to magnification is quite complex. For this case the object frequencies must also be considered. Fig. 11 on the other hand, shows that, in order to maintain 75% detection probability, the mAs must increase by 20% when the radiation field size increases from 16 cm² to 100 cm². Similarly, the GDQE decreases by 20% when going from 16 cm² to 100 cm² radiation field size for all the frequencies. Increasing the mAs technique will not change the GDQE (the system will not become more efficient) since it is independent of the detector entrance exposure, the object detectability on the other hand will certainly increase.

5. CONCLUSIONS

The generalized quantities helped us to obtain a thorough understanding of our system capabilities and its dependence on the different parameters that affect the physics of the image acquisition. They require careful interpretation though, in the sense that they can not be directly related to the object detection performance of our system. Instead they can be useful to compare different systems that have the same clinical functionality, in a similar manner that the MTF_D and DQE_D are used to compare different detector performances. The generalized Noise Equivalent Quanta with the ideal observer detectability can be used as an optimization tool for our system. It can be used to predict the size of objects that can be detected, it can be used to reduce the dose to the patient, and to properly select the system parameters that will provide optimum image quality.

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Fig. 1.

(a) Relation between the Scatter fraction as a function of the radiation field size at 80kVp.

There is no dependence on the kVp for the range of radiation field sizes for our experimental setup. The solid curve is the empirical model of Eq. (9). (b) NNPS_D the agreement of Eq. (5) with the measured data for a range of spatial frequencies (cycles/mm) shown under each curve.

Fig. 2.

The optimization of the GMTF for magnification for the 0.8 mm focal spot. The shaded region defines the range of magnification for a given spatial frequency where the GMTF is either equal or higher than the GMTF at $m=1$. The thick curve is the range of magnifications where the GMTF is maximized.

Fig. 3.

(a) is the x-ray transmission signal of the object. (b) is the Fourier frequencies of the object and (c) is the two dimensional GNEQ obtained by rotating the one dimensional GNEQ about the z axis. The graphs are obtained with $r=0.1$ mm, $m=1.1$, $\alpha=16$ cm², $w_I=0.1$ and 80 kVp.

Fig. 4.

(a) is the exposure to the patient per exposure to the detector as a function of the radiation field size and kVp. (b) is the mAs per exposure to the detector as a function of the radiation field size and kVp. The dots represent measured data points.

Fig. 5.

The generalized GMTF obtained for the 0.8 mm focal spot. (a) GMTF as a function of magnification obtained at 80 kVp, in a scatter free configuration. (b) is the GMTF as a function of the radiation field size, obtained at $m=1$ and 80 kVp. (c) is a close-up of the GMTF for different kVp's obtained at a scatter free geometry with $m=1$. (d) is the case with clinically relevant parameters where a collimation of 16cm^2 was used and different magnifications.

Fig. 6.
The GNNPS as a function of the kVp. The different surfaces correspond to different detector entrance exposures.

Fig. 7.

The GNEQ as a function of the radiation field size α and magnification m . The different surfaces correspond to different spatial frequencies.

Fig. 8.

The GDQE, (a) as a function of magnification, (b) as a function of the radiation field size (c) as a function of the kVp and (d) as a function of the detector entrance exposure. In each plot the thick top curve is the same detector DQE which represents the ideal case (for comparison), obtained at $m=1$, 80 kVp no scatter and at 1 mR .

Fig. 9.
DAP to patient ($\text{mGy}\cdot\text{cm}^2$) and mAs for constant probability of detecting the vessel filled with 110mg/ml Iodine, for different vessel diameters.

Fig. 10. Patient DAP in ($\text{mGy}\cdot\text{cm}^2$) as a function of magnification m . The right axis is the mAs technique for 70 kVp.

Fig. 11.
DAP to patient as a function of the radiation field size α . The numbers on the curve represent the mAs technique for the 70kVp tube voltage.