Effect of radiographic techniques (kVp and mAs) on image quality and patient doses in digital subtraction angiography

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We investigated how varying the x-ray tube voltage and image receptor input exposure affected image quality and patient radiation doses in interventional neuroradiologic imaging. Digital subtraction angiography (DSA) images were obtained of a phantom with 1 mm diameter vessels containing iodine at concentrations between 4.5 and 50 mg/cc. The detection threshold concentration of iodine was determined by inspecting DSA images obtained at a range of x-ray tube voltages and input exposure levels. Surface doses were obtained from measured x-ray tube output data, and corresponding values of energy imparted were determined using the exposure-area product incident on the phantom. In one series of experiments, the air kerma at the image intensifier (X) was varied between 0.44 μGy per frame and 8.8 μGy per frame at a constant x-ray tube voltage of 70 kVp. In a second series of experiments, the tube voltage was varied between 50 and 100 kVp, and the mAs adjusted to maintain a constant exposure level at the input of the image intensifier. At a constant x-ray tube voltage, the surface dose and energy imparted were directly proportional to the input exposure per frame used to acquire the DSA images. On our DSA system operated below 2.2 μGy per frame, the threshold iodine concentration was found to be proportional to $X^{-0.37}$, which is in reasonable agreement with the theoretical prediction for a quantum noise limited imaging system. Above 2.2 μGy per frame, however, the threshold iodine concentration was proportional to $X^{-0.26}$, indicating that increasing the input exposure above this value will only achieve modest improvements in image quality. At a constant image intensifier input exposure level, increasing the x-ray tube voltage from 50 kVp to 100 kVp reduced the surface dose by a factor of 6.1, and the energy imparted by a factor of 3.5. The detection threshold iodine concentration was found to be proportional to $kVp^4$, where n was 2.1 at 1.1 μGy per frame, and 1.6 at 3.9 μGy per frame. For clinical situations that can be modeled by a uniform phantom, reducing the x-ray tube voltage rather than increasing the exposure level would best achieve improvements on our DSA imaging system performance. © 2002 American Association of Physicists in Medicine. [DOI: 10.1118/1.1493213]

Key words: image quality, dosimetry, digital subtraction angiography, tube voltage, x-ray techniques

INTRODUCTION

Interventional neuroradiology involves studies of the vasculature and blood kinetics of the brain by means of catheterization performed with the transfemoral artery technique. In these interventional neuroradiologic procedures, image quality is critical with neuro-vascular instruments being as small as 90 μm, and even smaller vessel sizes. The continuous branching of the cerebral vessels leads to a reduction of the vessel diameter being imaged and thus the amount of iodine present in the vessel. A higher iodine concentration may sometimes be required to image distant vessels, especially during the capillary phase. Iodine concentration, however, is generally limited by direct toxicity at higher concentrations. An alternative to increasing the iodine concentration would be to improve the acquired signal-to-noise ratio. Practical ways to improve detection of small vessels during interventional neuroradiologic procedures include the variation of x-ray tube voltage and x-ray beam intensity used to produce the radiographic image.

Interventional neuroradiologic procedures often require long fluoroscopic exposure times as well as the acquisition of a large number of radiographic images. As a result, the amount of radiation received by the patient is a matter of major concern. The surface (skin) dose predicts the possibility of inducing deterministic injuries, and is the key dose parameter whenever the threshold dose for the indication of such injuries is exceeded. It is desirable to keep surface doses below threshold doses for the induction of deterministic effects, which is often taken to be ~2 Gy. For radiation doses below the deterministic threshold dose, the patient risk consists of the stochastic risk of carcinogenesis and the induction of genetic effects. For a given type of radiologic examination such as interventional neuroradiologic proce-
dures, the stochastic risk may be taken to be directly proportional to the total energy imparted to the patient. Reducing the total energy imparted to the patient will minimize the stochastic risk, and help to ensure that the patient dose is as low as reasonably achievable (ALARA).

DSA image quality needs to be adequate for the specified imaging task, and to ensure that there are no adverse clinical consequences as a result of inadequate visualization of catheters or vasculature. Changing radiographic technique factors for DSA procedures will impact both the surface dose and energy imparted. Any attempt to improve image quality by modification of radiographic technique factors should also require that the corresponding patient radiation dose be minimized. A quantitative understanding of the relationship between image quality and patient dose is therefore an important step towards the development of imaging protocols that maintain adequate diagnostic imaging performance whilst ensuring that patient radiation dose is kept ALARA. In this study, we investigated how changing x-ray tube voltage and input exposure to the image receptor affects radiation dose and image quality in DSA phantom studies that simulate interventional neuroradiologic procedures.

**METHOD**

**Image quality phantom**

The phantom used to assess image quality consisted of thirteen stacked acrylic blocks with dimensions of 30 cm × 30 cm × 1.3 cm. The acrylic vessel insert measured 30 cm × 9.0 cm × 1.3 cm and had 30 cylindrical vessels 1.0 mm in diameter and 35 mm in length drilled along its midplane at intervals of 8.0 mm apart. The vessels were filled with iodinated contrast prepared from Ultravist (Berlex Laboratories, Wayne, NJ) 300 iopromide solution diluted in heparin solution to accurately prepare very low concentrations of iodine. Heparin was selected as the dilution medium because this is what is used clinically to deliver iodine contrast into the blood stream. The contrast medium was mixed thoroughly before injected into the 1.0 mm vessels of the DSA phantom. To avoid settling of the iodine contrast at the bottom of the vessel walls, the vessel insert was positioned with the vessels oriented horizontally and exposures were taken shortly after the mixing was done with the vessel insert shaken in between exposures. The iodine concentrations of the contrast medium used to fill each vessel ranged from 4.5 mg/cc of iodine to 50 mg/cc, with the ratio of iodine concentration in any two adjacent vessels being 0.92. The acrylic phantom with a vessel insert appropriate for DSA is shown in Fig. 1.

In this study, an image quality index was defined as the lowest concentration of iodine in a 1.0 mm diameter vessel that may be visually detected in a DSA image. An example of DSA images obtained for this study is shown in Fig. 2. The five-point scale used to evaluate the DSA images is summarized in Table I. An observer was trained to score images of different iodine concentration using a representative series of ten DSA images of the phantom obtained during preliminary experimental work. The observer was asked to maintain a consistent criterion in using the ranking scheme listed in Table I. After training, images were presented in a random order. The concentration in the vessel that corresponded to a score of three (see Table I) was taken as the iodine detection

![Digital Subtraction Angiography Phantom](image)

Fig. 1. Schematic diagram of the acrylic phantom with the vessel and blank inserts used to simulate small vessels for the purpose of evaluating image quality in neuroradiology.

![Fig. 2. Digitally subtracted angiographic image of the DSA acrylic phantom with the 1.0 mm vessel insert.](image)

Fig. 2. Digitally subtracted angiographic image of the DSA acrylic phantom with the 1.0 mm vessel insert.

**TABLE I. Scoring scheme used to assess the visibility of vessel with varying concentrations of iodine contrast material.**

<table>
<thead>
<tr>
<th>Score</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Vessel not seen</td>
</tr>
<tr>
<td>2</td>
<td>Vessel poorly seen (&lt;50%)</td>
</tr>
<tr>
<td>3</td>
<td>50% of vessel delineated</td>
</tr>
<tr>
<td>4</td>
<td>Most of vessel visible (&gt;50%)</td>
</tr>
<tr>
<td>5</td>
<td>Vessel perfectly visible</td>
</tr>
</tbody>
</table>
threshold in this study, with linear interpolation used where necessary.

**DSA images**

All DSA acquisitions were performed using a biplane Toshiba (Toshiba America Medical Systems, Tustin, CA) KXO-80 high voltage diagnostic x-ray generator and the Toshiba DFP-2000/A3 digital fluorography system configured for interventional neuroradiologic procedures. The TV camera used on the Toshiba imaging system was a MTV-500A, high definition (1024×1024), low noise CCD camera. The x-ray source-to-image receptor distance (SID) was set to the maximum value of 105 cm, with the acrylic phantom positioned so that the geometric magnification of the vessel insert was 1.2. DSA acquisitions were obtained using the 0.6 mm focal spot size and a 23 cm diameter image intensifier mode. The x-ray tube filtration in the frontal and lateral imaging planes was equivalent to 3.0 mm of aluminum (tube inherent filtration). An additional 3.0 mm aluminum equivalent filtration was added in the frontal plane by the x-ray table. The generator was set to manual techniques allowing fine adjustments of the tube voltage (kVp), tube current (mA), exposure time (ms) and optical gain (TV iris). The x-ray tube voltage was monitored using a Machlett Dyanalys III (Greenwich Instrument CO., Inc., Greenwich, CT). A 10×5.60 (the 10×5.60 ionization chamber has a gain of ×10 for better resolution) ionization chamber of a MDH 1015C exposure meter was positioned behind the grid of the image intensifier to record the input exposure to the image receptor. An exposure of 2.58×10^-4 C/kg (1 Roentgen) was taken to correspond to an air kerma of 8.73 mGy.21

During digital subtraction acquisitions, a mask frame was acquired of the blank phantom insert. The blank insert was replaced by the vessel insert and twenty additional frames were acquired (3 frames/s), with the tenth frame of each DSA image acquisition used for the observer study. The window and level of the displayed images were adjusted to optimize signal detection during each DSA image acquisition. The optical gain was controlled by electronic adjustments of the iris diameter to produce a constant pixel value between 2000 and 2100, which corresponds to the target values used by the biplane imaging system for routine patient examinations.

In the first experiment, a series of measurements were performed where the air kerma at the input of the image intensifier was maintained at either 1.1 μGy/frame or 3.9 μGy/frame, and with the x-ray tube voltage varied between 50 and 100 kVp. As the x-ray tube voltage was increased, the corresponding mAs value was reduced to maintain a constant air kerma level at the image intensifier input, which was monitored with an ionization chamber. Table II summarizes the technique factors used in these experiments (i.e., kVp and mAs) together with corresponding measured pixel values. The data in Table II demonstrate the pixel value was kept relatively constant in these experiments, with average values (±standard deviation) of 2068±34 at 1.1 μGy/frame and 2039±56 at 3.9 μGy/frame. In the second experiment, another series of measurements were obtained at a constant tube voltage of 70 kVp, and with the input air kerma to the image intensifier varied from 0.44 μGy/frame to 8.8 μGy/frame in thirteen steps. The optical gain was adjusted to maintain constant pixel value as the input air kerma was modified, with an average pixel value for the thirteen measurements of 2051±28.

**Radiation dosimetry**

The entrance exposure to the acrylic phantom was recorded by a 10×5.6 ionization chamber of a MDH 1015C exposure meter attached to the beam entrance surface of the phantom during each DSA acquisition. The ionization chamber was positioned to avoid any overlap with the iodinated vessels of the vessel insert, and the measured exposure included backscatter radiation coming from the acrylic phantom. Frame entrance exposures were obtained from the integral exposure measured for each image acquisition sequence divided by the number of acquired frames. A conversion factor of 2.58×10^-4 C/kg (1R) corresponding to an absorbed dose of 9.3 mGy for muscle tissue was used to convert the beam entrance exposure to surface dose.21

The energy imparted was computed using the method described in detail elsewhere.22,23 The total phantom thickness was 16.9 cm of acrylic, which was taken to be equivalent to 20 cm of water for computing values of energy imparted, given that the density of acrylic is 1.19 g/cm^3.24 Since the exposure was obtained from direct exposure measurements during each image acquisition sequence, it included backscatter. The backscatter radiation fraction was measured for the corresponding phantom at each applied tube voltage and was subtracted from the measured exposures to obtain the free-in-air exposure.20 The exposure area at the beam entrance plane of the phantom was computed from geometry assuming a circular beam area of 23 cm diameter at the image intensifier plane. Values of energy imparted for head x-ray examinations were converted into the corresponding adult effective doses using E/e (mSv/J) conversion factors averaged for the posteroanterior and lateral views and ranged between 4.1 and 5.9 mSv/J depending on the selected x-ray tube voltage.25

**Table II. Imaging techniques during tube voltage experiments.**

<table>
<thead>
<tr>
<th>kVp</th>
<th>1.1 μGy/frame</th>
<th>3.9 μGy/frame</th>
</tr>
</thead>
<tbody>
<tr>
<td>mAs</td>
<td>Video level</td>
<td>mAs</td>
</tr>
<tr>
<td>50</td>
<td>28.0</td>
<td>2075</td>
</tr>
<tr>
<td>56</td>
<td>14.4</td>
<td>2018</td>
</tr>
<tr>
<td>60</td>
<td>9.28</td>
<td>2104</td>
</tr>
<tr>
<td>66</td>
<td>5.40</td>
<td>2028</td>
</tr>
<tr>
<td>70</td>
<td>4.00</td>
<td>2068</td>
</tr>
<tr>
<td>80</td>
<td>2.24</td>
<td>2054</td>
</tr>
<tr>
<td>90</td>
<td>1.50</td>
<td>2117</td>
</tr>
<tr>
<td>100</td>
<td>0.960</td>
<td>2079</td>
</tr>
</tbody>
</table>

*Baseline techniques.
RESULTS

Measurement precision

Five digital subtraction acquisition sequences, obtained at 70 kVp, were generated at different times over a one-day period. These five sequences were used to determine the radiation dosimetry precision and reader performance precision. The average entrance exposure measured was 0.852 ± 0.006 mGy/frame. The average of five observer readings for the threshold contrast concentration at 70 kVp and 2.6 μGy/frame was 8.0 ± 0.96 mg/cc. As expected, the precision of radiation exposure measurements (~1%) is much lower than the reproducibility of the detection threshold value (12%), given that the latter is based on the subjective assessment of a human observer.  

X-ray tube voltage

Figure 3 shows the threshold iodine concentration as a function of tube voltage. The solid and dashed lines shown in Fig. 3 are least squares fits to the experimental data. At 1.1 μGy/frame, the threshold iodine concentration was proportional to the $kVp^{2.1}$, where the coefficient of determination of the least squares fit ($r^2$) was 0.92. At this input exposure, a 10 kVp increase of the tube voltage from the baseline (70 kVp) increased the threshold iodine concentration by 32%. Reducing the x-ray tube voltage from 70 kVp to 60 kVp reduced the detection threshold concentration by 27%. At 3.9 μGy/frame, the threshold concentration was proportional to $kVp^{1.6}$, where the coefficient of determination of the least squares fit ($r^2$) was 0.97. At an exposure level of 3.9 μGy/frame, increasing the tube voltage from 70 kVp to 80 kVp increased the threshold concentration by 23%, whereas reducing the x-ray tube voltage from 70 kVp to 60 kVp reduced the detection threshold concentration by 21%. Over the x-ray tube voltage range investigated in this study (50–100 kVp), the average increase in detection threshold as a result of reducing the exposure level from 3.9 to 1.1 μGy/frame was 1.7 ± 0.3.

Figure 4 shows the relative behavior of surface dose and energy imparted as a function of tube voltage. The data of Fig. 4 have been fitted to power functions with resulting coefficient of determination ($r^2$) between 0.95 and 0.98. Table III provides a summary of the key dosimetry parameters (i.e., surface dose, energy imparted and effective dose) obtained at the two exposure levels investigated. As expected, relative changes of both surface dose and energy imparted as a function of x-ray tube voltage were similar at

Table III. DSA dosimetry summary.

<table>
<thead>
<tr>
<th>kVp</th>
<th>Surface dose (μGy/frame)</th>
<th>Energy imparted (mJ/frame)</th>
<th>Effective dose (mSv/frame)</th>
</tr>
</thead>
<tbody>
<tr>
<td>50</td>
<td>1.2[3.9]$^a$</td>
<td>1.6[5.0]</td>
<td>6.5[20]</td>
</tr>
<tr>
<td>60</td>
<td>0.62[2.4]</td>
<td>0.95[3.7]</td>
<td>4.4[17]</td>
</tr>
<tr>
<td>70</td>
<td>0.38[1.3]</td>
<td>0.66[2.3]</td>
<td>3.3[11]</td>
</tr>
<tr>
<td>80</td>
<td>0.28[0.96]</td>
<td>0.54[1.9]</td>
<td>2.9[10]</td>
</tr>
<tr>
<td>90</td>
<td>0.24[0.65]</td>
<td>0.50[1.4]</td>
<td>2.8[7.9]</td>
</tr>
<tr>
<td>100</td>
<td>0.20[0.66]</td>
<td>0.44[1.5]</td>
<td>2.6[8.8]</td>
</tr>
</tbody>
</table>

$^a$1.1 μGy/frame input air kerma.
$^b$3.9 μGy/frame input air kerma.
exposure levels of 1.1 μGy/frame and 3.9 μGy/frame. Increasing the x-ray tube voltage from 50 kVp to 100 kVp reduced the surface dose by a factor of 6.1, and reduced the energy imparted by a factor of 3.5.

Input air kerma

Figure 5 shows the threshold iodine concentration as a function of image intensifier input air kerma at a constant x-ray tube voltage of 70 kVp where both the ordinate and abscissa are plotted using a logarithmic scale. Data below and above 2.2 μGy/frame were fitted separately to a least squares fit to a line. Below 2.2 μGy/frame, the detection threshold concentration was found to be proportional to $X^{-0.57}$ where the coefficient of determination ($r^2$) was 0.98. Above 2.2 μGy/frame, the detection threshold concentration was found to be proportional to $X^{-0.26}$ where the coefficient of determination ($r^2$) was 0.96.

For our DSA imaging system, increasing the image intensifier input air kerma from 1.1 μGy/frame to 2.2 μGy/frame reduced the detection threshold concentration by 34%, whereas increasing the input air kerma from 3.9 μGy/frame to 7.8 μGy/frame reduced the detection threshold concentration by only 16%. The empirical data presented in Fig. 5 indicate that above an input air kerma of about 2.2 μGy/frame, there is only modest improvement in the detection threshold concentration of 1.0 mm diameter vessels with increasing input air kerma. These results are only applicable to the imaging system investigated in this study, and could be different for other types of DSA imaging system.

Radiation dose versus image quality

Figures 6 and 7 show the relative increases in surface dose and energy imparted as the iodine detection threshold-concentration is reduced by changing either the image intensifier input air kerma or the x-ray tube voltage. The data generated in Figs. 6 and 7 are based on the least squares curves fitted to the empirical data in Figs. 3–5.

At 70 kVp and 1.1 μGy/frame, a 25% reduction in detection threshold concentration may be achieved by either lowering the tube voltage to 61 kVp, or by increasing the input air kerma by 63%. At 3.9 μGy/frame, a 25% reduction in detection threshold concentration may be achieved by either lowering the tube voltage to 59 kVp, or by increasing the input air kerma by a factor of 200%. The increased surface dose and energy imparted required to achieve this 25% reduction on the iodine threshold concentration are summarized in Table IV. Improvements in image quality by reducing the x-ray tube voltage always resulted in lower doses than could be achieved by increasing the x-ray intensity level to the image receptor at a constant x-ray tube voltage.

DISCUSSION

For a film-screen imaging system, the exposure level is effectively fixed by the need to produce an optimum optical density. Any variations in the exposure level in screen-film radiography will produce over (or under) exposed films, with a concomitant drop in image contrast. In digital imaging systems, however, the range of receptor input exposure can vary by over an order of magnitude (e.g., Fig. 5), yet still produce diagnostic quality images. The advent of relatively sophisticated digital imaging equipment thus offers considerable
flexibility in acquiring x-ray image data.27 The wide range of input radiation exposures, as well as variations in the selected x-ray tube voltage, could play a significant role in helping to optimize patient dose and image quality. A quantitative understanding of the relationship between dose and image quality is therefore important to help operators make the best use of digital imaging equipment. Improvements in imaging performance should require the modification of those radiographic technique factors that minimize the patient radiation risk. Conversely, if it is possible to reduce image quality without adversely affecting diagnostic performance, technique factors should be modified to maximize patient dose savings.

In screen-film radiography, it is well established that increasing x-ray tube voltage or x-ray beam quality degrades image quality.18,28–30 The data presented in Fig. 3 shows that tube voltage has a significant effect on the iodine detection threshold concentration in DSA, as well. Changing the x-ray tube voltage will predominantly affect the contrast of the signal detected by the imaging system if the absolute x-ray intensity level remains constant. Reducing the x-ray tube voltage lowers the average photon energy and increases x-ray absorption because of the rapid rise in photoelectric absorption as the average photon energy approaches the iodine 33 keV K edge. An additional factor affecting image quality at different x-ray tube voltages is the presence of scatter radiation. Scatter radiation generally increases with increased x-ray tube voltage, and will thereby reduce image contrast.31

The data in Fig. 3 may be compared with theoretical changes in x-ray spectral energies and x-ray absorption coefficients for iodine. For the theoretical calculation, average incident photon energies and half-value layers (HVL) were computed for a 3 mm Al filtered x-ray tube running at constant voltage.32 The exit mean energy and HVL were obtained by filtering the x-ray spectra with 20 cm of water. The absorption coefficient of iodine was determined using the exit mean photon energies.21,33 Table V shows the computed values of incident photon energy and exit photon energy, together with the corresponding iodine mass attenuation coefficient.

The relationship between selected kVp and mass attenuation coefficient in Table V was fitted to a power law, i.e., \( \mu \propto kVp^n \), with the theoretical value of \( n \) determined to be 1.7. This theoretical prediction may be compared to the experimental data obtained at doses where quantum mottle is dominant—this corresponds to a slope of 2.1 obtained at an exposure rate of 1.1 \( \mu \text{Gy/frame} \) (see Fig. 3). The differences between the measured and theoretical values may be due to uncertainties in

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**Table IV. Changes in surface dose and energy imparted required to reduce the iodine threshold concentration by 25% by reducing the x-ray tube potential, or by increasing the input air kerma to the image intensifier.**

<table>
<thead>
<tr>
<th>Parameter</th>
<th>1.1 ( \mu \text{Gy/frame} )</th>
<th>3.9 ( \mu \text{Gy/frame} )</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>X-ray tube potential</td>
<td>Exposure level</td>
</tr>
<tr>
<td>Surface dose</td>
<td>44%</td>
<td>63%</td>
</tr>
<tr>
<td>Energy imparted</td>
<td>29%</td>
<td>63%</td>
</tr>
</tbody>
</table>

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**Table V. Computed mean x-ray energies of x-ray spectra, half-value layers (mm Al), and iodine attenuation coefficients as a function of x-ray tube voltage.**

<table>
<thead>
<tr>
<th>kVp</th>
<th>Incident mean energy and HVL</th>
<th>Exit mean energy and HVL</th>
<th>Mass attenuation coefficient of I (cm(^2)/g)(^c)</th>
</tr>
</thead>
<tbody>
<tr>
<td>50</td>
<td>32.6±[1.8](^b)</td>
<td>39.1±[2.2]</td>
<td>23.4</td>
</tr>
<tr>
<td>60</td>
<td>36.3±[2.2]</td>
<td>44.6±[3.9]</td>
<td>16.6</td>
</tr>
<tr>
<td>70</td>
<td>39.7±[2.5]</td>
<td>49.4±[5.2]</td>
<td>12.7</td>
</tr>
<tr>
<td>80</td>
<td>43.1±[2.8]</td>
<td>54.0±[6.26]</td>
<td>10.0</td>
</tr>
<tr>
<td>90</td>
<td>46.2±[3.2]</td>
<td>58.1±[7.2]</td>
<td>8.26</td>
</tr>
<tr>
<td>100</td>
<td>49.0±[3.6]</td>
<td>61.7±[7.9]</td>
<td>7.03</td>
</tr>
</tbody>
</table>

\( ^a \) Incident mean energy in keV.
\( ^b \) Half-value layer in mm Al.
\( ^c \) Mass attenuation coefficient is determined at the mean x-ray energy.

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**Fig. 7.** Increased energy imparted resulting from reducing the x-ray tube potential or increasing the exposure level to achieve the specified improvement in threshold detection concentration of iodine.
the x-ray spectra as well as changes in quantum noise as a function of kVp which were neglected in our computations.

The primary source of noise in (digital) x-ray imaging is usually quantum mottle, which corresponds to random spatial fluctuations of the distribution of x-ray quanta absorbed by the detector.\textsuperscript{8,34} Since the production and attenuation of x-rays are governed by Poisson statistics, quantum mottle is inversely proportional to the square root of the radiation exposure level used to generate an image. The results shown in Fig. 5 demonstrate that for input air kerma smaller than 2.2 \( \mu \text{Gy/frame} \), this specific DSA imaging system is quantum noise limited, since the observed performance shows proportionality close to the value of to \( X^{-0.5} \) expected for a quantum noise limited system. A significant performance gain can be achieved by increasing the input air kerma at these lower exposure levels.

At exposure levels greater than 2.2 \( \mu \text{Gy/frame} \), the measured iodine detection threshold concentration is proportional to \( X^{-0.20} \), demonstrating that imaging performance is no longer limited by quantum mottle alone. At these higher exposure levels, additional noise factors become significant and would need to be included in any analysis of system imaging performance. These noise sources include factors such as electronic noise, time jitter and structure noise.\textsuperscript{8,34–37}

The relative importance of these additional noise sources is also the reason for the differences in the slopes observed in Fig. 3. For a quantum noise limited system, the slope of the curve plotting threshold concentration and kVp would be independent of the radiation exposure level. System noise will set an upper limit to the x-ray flux, which may be taken to occur when the quantum and electrical noise components are equal.\textsuperscript{38}

One limitation of our study is that the phantom used had a constant thickness which is rarely achieved in clinical practice. For a non-uniform patient, the highest exposure level occurs in the thinnest body region and this should be mapped to the maximum signal level of the video camera.\textsuperscript{38} The resultant signal to noise ratio of the vessel containing the iodine will depend on the transmitted x-ray intensity in the region of interest. This limitation is of much greater importance in nonuniform areas such as the lung or extremities, than for abdominal or head imaging. Bright spots in DSA imaging should be minimized by appropriate bolusing to attempt to ensure that all regions of the image are obtained at maximum signal levels.\textsuperscript{31}

The data presented in Figs. 6 and 7 indicate how modifying radiographic technique factors affect imaging performance and patient dose, given a detection task of identifying a circular vessel (1.0 mm diameter) containing iodine contrast media. It is evident that achieving improved imaging performance by reduction in x-ray tube voltage is the preferred option, since the patient doses are significantly lower than when the same level of imaging performance is achieved by increasing the input radiation exposure level to the image receptor. Our results also show that increasing the image intensifier input air kerma beyond 2.2 \( \mu \text{Gy/frame} \) for our specific DSA system provided relatively little improvement in measured imaging performance. Use of such higher radiation exposure levels would require explicit justification in terms of the anticipated benefit to the patients to offset risk of deterministic effects and the higher stochastic risks.

At our institution, the typical interventional neuroradiologic DSA x-ray tube voltages are currently 76 to 80 kVp for the frontal plane and 66 to 70 kVp for the lateral plane, and with corresponding air kerma values at the input of the image intensifier set at 4.4 and 6.1 \( \mu \text{Gy/frame} \). A typical diagnostic examination can result in 400 DSA images in each of the two imaging projection planes. As a result, patient surface doses from DSA imaging could easily be of the order of 0.65 Gy, which would need to be added to any dose associated with fluoroscopic imaging, typically 0.25 Gy. Such data clearly demonstrate the potential for inducing deterministic effects in DSA studies, particularly in studies that require an increased number of radiographic images. Typical value of energy imparted to a patient undergoing an interventional neuroradiologic procedure is 6 J, and the corresponding patient effective dose would therefore be of the order of 30 mSv.\textsuperscript{20} These effective doses are much higher than is normally encountered in conventional radiology.\textsuperscript{39} DSA patient effective doses are at the upper end of the range of patient doses in radiology, and emphasize the need to pay careful attention to minimizing patient doses.

**CONCLUSION**

The results obtained in this study demonstrate that for improving low contrast detection on our DSA imaging system, reducing x-ray tube voltage (i.e., kVp) was a superior strategy to that of increasing the x-ray beam intensity level (i.e., mAs). For a given x-ray DSA imaging system, there is an exposure level at which quantum mottle will cease to be the dominant source of image noise. For the DSA system investigated in this work, this exposure level was found to be about 2.2 \( \mu \text{Gy} \) per frame at the input to the image intensifier. For this specific DSA system, increasing the image intensifier input exposure level beyond 2.2 \( \mu \text{Gy} \) per frame would be of limited benefit given that the improvement in low contrast detection performance is relatively modest.

**ACKNOWLEDGMENTS**

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