Published in final edited form as:

Proc SPIE. 2012 ; 8313: . doi:10.1117/12.910791.

Overcoming x-ray tube small focal spot output limitations for high-resolution region of interest imaging

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Abstract

We investigate methods to increase x-ray tube output to enable improved quantum image quality with a higher generalized-NEQ (GNEQ) while maintaining a small focal-spot size for the new high-resolution Micro-angiographic Fluoroscope (MAF) Region of Interest (ROI) imaging system. Rather than using a larger focal spot to increase tube-loading capacity with degraded resolution, we evaluated separately or in combination three methods to increase tube output: 1) reducing the anode angle and lengthening the filament to maintain a constant effective small focal-spot size, 2) using the standard medium focal spot viewed from a direction on the anode side of the field and 3) increasing the frame rate (frames/second) in combination with temporal filter. The GNEQ was compared for the MAF for the small focal-spot at the central axis, and for the medium focal-spot with a higher output on the anode side as well as for the small focal spot with different temporal recursive filtering weights. A net output increase of about 4.0 times could be achieved with a 2-degree anode angle (without the added filtration) and a 4 times longer filament compared to that of the standard 8-degree target. The GNEQ was also increased for the medium focal-spot due to its higher output capacity and for the temporally filtered higher frame rate. Thus higher tube output, while maintaining a small effective focal-spot, should be achievable using one or more of the three methods described with only small modifications of standard x-ray tube geometry.

Keywords

NEQ; Focal spot; GNEQ; ROI; MAF; EIGI; MTF; NNPS; X-ray tube; CNR

1. INTRODUCTION

Stroke or neurovascular disease is the third greatest killer, and leading cause of disability in the US as well as the leading cause of death and disability worldwide.^[1,2] Besides the usual drug treatment, the primary therapy for neurovascular disease is endovascular image-guided interventions (EIGIs).^[3] Because of higher mortality, morbidity, and discomfort to the patient in surgical procedure that involves a major opening craniotomy, EIGIs have been replacing invasive surgical procedures. EIGIs commonly involve the insertion of a catheter into the femoral artery through a small incision in the skin; the catheter is then navigated through the vasculature to the target location using real-time imaging fluoroscopy for the treatment of different pathologies such as brain vessel hemorrhages, aneurysms, stenoses, tumor vessel beds, and arteriovenous malformations. EIGIs involve the use of medical devices such as stents, balloons, coils, snares etc., which are manipulated using such catheters. During EIGIs, low dose x-ray fluoroscopy is used for the guidance of the catheter and high dose x-ray imaging digital subtraction angiography (DSA) is used to evaluate the

progress of the advancement of the procedure as well as to see the vasculature and interventional devices at the site of intervention. In most cases, a large FOV detector is not needed during the procedure but rather superior image quality with higher resolution is needed only in a small ROI, where the interventional activity is occurring. Hence, region-of-interest (ROI) imaging using a small field-of-view (FOV) high-resolution ROI imager may generally be used in and around the region of intervention. The advantages of ROI imaging include reduction in scatter in the ROI and the reduction in the effective dose to the patient by about an order of magnitude because of the small FOV.^[4,5] Furthermore, use of the high-resolution MAF provides a substantial improvement in image quality. This small FOV ROI imager can be used with currently commercially available angiographic systems, and in conjunction with the standard large FOV flat panel image receptor. Hence, whenever a high resolution image with greater image quality is desired at a targeted ROI, the ROI detector can be deployed to fulfill these requirements.

Superior resolution is needed in the EIGIs for both real time fluoroscopic imaging during the treatment, and also for superior angiography to evaluate the advancement of the catheter tip throughout the procedure as well as for deciding optimally when the endpoint is reached. As the advancement of technologies leads towards the development of smaller and more sophisticated interventional devices, it is becoming a necessity to visualize interventional devices and vessel sizes that can be less than 100 microns. Additionally one must be careful to avoid the blockage of secondary vessels called perforators that can be as small as 50 microns. Blocking these can potentially impact brain function. For the initial parts of the EIGIs where fluoroscopy is used to guide insertion of catheters or wires optimal image quality may not be needed; however, higher image quality becomes extremely important during sensitive parts of the angiographic procedure such as during deployment of intervention devices or position verification after deployment. Usually the small focal spot instead of the medium focal spot is necessary to obtain the high resolution images for EIGIs. However, the tube output using the small focal spot is limited because of the smaller target area on the anode disk. Hence, the higher output of the medium focal spot may be required to be used even though resolution is sacrificed. Methods to maintain the small focal spot and hence spatial resolution while increasing the tube loading to values characteristic of the medium focal spot would thus be greatly beneficial.

To be able to quantitatively evaluate the proposed methods for increasing tube output we will compare maximum possible system Noise equivalent quanta (NEQ) values rather than detective quantum efficiency (DQE) values since we are interested in the final image SNR rather than a gauge of relative detector SNR efficiency. The NEQ gives the effective number of input quanta used by the system to produce an output image; that is, it gives the number of Poisson distributed quanta for an ideal detector that would provide the same signal-to-noise (SNR) as measured. However, the NEQ provides the intrinsic performance of a detector, not the total system performance in a realistic clinical situation that may involve scatter radiation due to the patient or focal spot blurring due to the finite size of the focal spot and object magnification. Whereas, the generalized NEQ (GNEQ) provides the statistical image quality information of a complete imaging system, including the effect from the finite focal spot size and the object magnification.^[6-8] In this paper, we do not include the effect of scatter radiation as we are interested here only in evaluating the effect of focal spot blurring on the image quality. The GNEQ using the small focal spot is limited by the lower tube output because of the smaller target area on the anode disk. Hence in this paper, we investigate possible methods for increasing the tube output for a small effective focal spot for a small field of view ROI detector, which we have developed for EIGI procedures.

2. METHODS AND MATERIALS

2.1 Micro-angiographic Fluoroscope (MAF)

The MAF is a ROI x-ray imaging detector with a small field of view ($3.6 \times 3.6 \text{ cm}^2$), low instrumentation noise, high sensitivity, and large variable gain.^[9,10] Figure 1 shows the schematic of the components inside the MAF and Figure 2 is the picture of the MAF. At the front end, a CsI(Tl) (Hamamatsu Corp., Bridgewater, NJ) imaging phosphor is coupled to a generation 2 dual light image intensifier (LII) (Model PP0410K, DEP Inc., Dwaziewegen 2, NL-9300 AB Roden, The Netherlands) through a fiber optic plate. The LII is coupled to a CCD camera (Model Pantera TF-1M30, Dalsa Corp., Waterloo, ON, Canada) through a 2.88:1 ratio fiber optic taper (FOT). The LII is the key component in the imaging chain that provides a large variable gain, and enables the MAF to be used both in high exposure x-ray imaging (angiography) and low exposure x-ray real time imaging (fluoroscopy). The MAF operates in the quantum noise limited region even at very low exposures because of the LII. Compared to the commercially available flat panel display (FPD), the MAF has higher spatial resolution because of its small active pixel size of 35 micron, while the thickness of its imaging phosphor is 300 microns.

2.2 Methods to increase the image quality

In the subsequent sections, we describe three methods to achieve higher tube output, hence image quality, while the resolution is maintained in the region of intervention for the MAF ROI imager.

2.2.1 Method 1: Design of an x-ray tube with higher tube output for a small focal spot—In EIGIs, the small focal spot is necessary to preserve the high resolution requirement; however, it is limited in tube output, hence the image quality is also limited due to the lower tube loading capacity. (The current Toshiba x-ray tube being used in our laboratory is limited to 160 mA for the small focal spot.) Herein, we propose a new geometrical design of x-ray tube insert with a change to the standard x-ray tube insert geometry that can be used for ROI imaging as well as for large FOV imaging. We evaluate the increase in the tube output obtained by increasing the small focal spot cathode filament length (keeping the diameter of the filament unchanged). The effective focal spot size at the central axis is maintained constant by decreasing the anode angle for a small FOV (5 cm) ROI imager such as the MAF. Since the area of heat deposition on the anode is increased, tube-loading capacity increases and the mA can be increased proportionally if the electron flux density at the target is unchanged. Consequently, high tube output, hence improved image quality can be achieved while maintaining the resolution using the standard Bremsstrahlung process. This newly designed x-ray tube could be used for both a small and large FOV imager, by angling the tube insert, and adjusting the filament length with central taps to allow a tradeoff between spatial resolution, tube loading and FOV. Although this new design would require constructing a new x-ray tube insert, the modification of the existing standard x-ray tube design does not appear to require radically new technology to achieve this increase in output.

As seen in Figure 3, at the central axis position at the detector plane, the effective focal spot size (viewing from the detector) is defined as $L \sin(\theta)$, where θ is the anode angle and L is the length of the electron beam projection onto the anode. The FOV in the direction parallel to the tube is calculated as AB , where $AB = 2 \times \text{SID} \times \tan(\theta)$; SID is the source to image detector distance, and is 105 cm for the MAF detector used on a Toshiba Infinix imaging system. The field of view for x-rays at the detector plane may be $AB \times AB$, assuming square collimation in the perpendicular direction to the tube axis. We used the existing x-ray tube

(model-E7854X, Toshiba Corporation, TOKYO JAPAN) with 8 degrees tungsten target anode as a reference for the calculations.

Figure 3 also shows the two different configurations of the x-ray tube insert in the schematic of a newly designed x-ray tube: large FOV imager (left) and small FOV ROI imager or the MAF (right). Large FOV imager can be used along with the small FOV imager with a little modification in the design of x-ray tube by which the tube insert would be rotated at a desired smaller angle; as a result, we get the increased spatial resolution due to the decrease in the effective anode angle, and increased tube output by selecting the large filament length. This also results in a small FOV at the 105 cm SID, but it is sufficient to cover the MAF or any ROI imager of 5 cm FOV. On the right side of Figure 3, the length of the usable filament is increased and the effective focal spot size is unchanged. To obtain the same effective focal spot size, the cathode or filament projection on the anode for the new smaller anode target angle is calculated by the following equation: $L_2 = (L_1 \times \sin(\theta_1)) / \sin(\theta_2)$, where θ_1 is the original anode angle and θ_2 is the new anode angle. The x-ray spectra for a tube with a tungsten anode were calculated using the software program SpekCalc for different anode angles and filtrations.^[11–13] To estimate the spectra exiting the skull in a neuroangiographic procedure, the materials of the uniform head equivalent phantom described in AAPM report-60 (3.2mm thick aluminum and 6 inches thick PMMA) were used in the x-ray beam path. Beam intensity was calculated at the central axis by integrating the energy fluences over all energies.

2.2.2 Method 2: Increase of the image quality and resolution for a medium focal spot

—An alternative method to increase tube output without degrading resolution unduly and without changing the x-ray tube design is to use the medium focal spot with its higher mA loading limits but using asymmetric collimation to shift the beam to the anode side so that we can utilize the Line-Focus-Principle. In this case without changing the actual tube design, we move the central beam axis to the anode side and effectively reduce the medium focal spot size in the anode-cathode direction to the size of the small focal spot on the central axis without reducing the medium focal spot tube loading limits. This method however does not reduce the size of the focal spot in the perpendicular direction and it will remain that of the larger medium focal spot.

The left and right sides of Figure 4 show the experimental setup with the MAF at the central axis position and at the anode side position, respectively. Since the maximum field size is limited to the size of the flat panel imager on our system, we could only move the MAF to the anode side by 6.4 cm from the central axis with an off axis angle of 3.5°. The anode angle is 8 degree in the central axis position, whereas the effective anode angle is 4.5° at the anode side. The MAF was at a distance of 105 cm from the x-ray tube. Small and medium focal spots of nominal sizes of 0.3 mm and 0.5 mm, respectively, were used in this study. An object magnification of 1.11, corresponding to the middle of the uniform head equivalent phantom with a fixed air-gap of 5 cm from the detector, was used in the generalized parameter calculations.

2.2.3 Method 3: Increase of the image quality for the existing small focal spot

—In this third method, we evaluate the increase in the image quality for a small focal spot without introducing a newly designed x-ray tube or without the use of the modified medium focal spot as described in the previous section. Considering that the current acquisition frame rate for angiography is sufficient at 3 frames/second, if we simply set the output per frame to near the limit and increase the frame rate while simultaneously temporally filtering, equivalent temporal resolution can be obtained with increased output.

Temporal filtering is the technique in which multiple images containing the signal are added (averaged) to reduce the effect of random noise (from both electronic and quantum statistical sources) while maintaining the signal. However, the disadvantage of temporal filtering is that temporal resolution decreases. This disadvantage can be mitigated if the acquisition frame rate is increased

A common temporal filtering technique is recursive temporal filtering, also called time-domain filtering. The equation used for a basic recursive temporal filtering is given by^[14,15]

$$Y(t) = \frac{1}{k} \times X(t) + \left(1 - \frac{1}{k}\right) Y(t-1) \quad (4)$$

where $Y(t)$ is the output of the filter, $X(t)$ is the current input signal at time or frame number t , $Y(t-1)$ is the previous frame's output signal, and k is the filter weight (lag or memory). The output of the filter is the weighted sum of its current input and previous outputs. The effect of filtering is to retain constant input signals, and suppress the effective noise through image integration. The recursive temporal filter generates a final image with an relative noise reduction of $1/(2k-1)$, if the noise is the same in all the images and number of images in the sequence is infinite.^[16] When the temporal filtering weight k increases, the effective noise in the image decreases, and, consequently, the image quality increases.

2.3 Measurement of Intrinsic Detector Performance Parameters

To evaluate the total system performance in terms of generalized parameters, we need to measure separately the MTFs for the focal spots and for the detector, and additionally the NNPS for the detector. Measurement of each of these metrics is described briefly in the following subsections.

2.3.1 Detector Modulation Transfer Function (MTF_D)—The detector MTF was measured using the slanted edge method described by Samei et. al..^[17] The edge response function of the detector was obtained from the flat field and offset corrected images of the edge and the first derivative of it was used to obtain the line spread function. Finally, a Fourier transform was performed to obtain the detector MTF from the line spread function. We obtained the two dimensional (2D) symmetric detector MTF by rotating the one dimensional (1D) MTF from 0 to 360°.^[9] An RQA 5 spectrum was used with a 74 kVp x-ray beam and added filtration of 21 mm of aluminum.

2.3.2 Focal spot MTF (MTF_F)—The focal spot MTF was measured using the standard pin-hole method, keeping the pin-hole close to the x-ray tube.^[18] X-ray images of the small focal spot with the MAF at the central axis, and the medium focal spot with the MAF at the central axis and the anode side were obtained using a 10 micron diameter pin-hole, and a higher magnification of 2.18 was used to minimize the effect of detector blur. We obtained the 2D PSF from the pin-hole images and took the Fourier transform of it to get the 2D focal spot MTF. For comparison purposes we also did radial averaging even for the asymmetric focal spots to calculate a 1D focal spot MTF from the 2D focal spot MTF. A scale factor based on the pin-hole magnification was used to get the images at unit focal spot magnification i.e. the focal spot plane.

2.3.3 Detector Normalized Noise Power Spectrum (NNPS_D)—The detector noise power spectrum (NPS) is defined as the square modulus of the Fourier Transformation of the differences between the mean value and individual pixel values in a flat field image, multiplied by the pixel area with units of mm².^[19,20] The 2D detector NPS was measured using the standard Fourier Transform method and normalized using the square of the

average pixel values to get normalized NPS (NNPS). As the 2D NNPS of the detectors is quite symmetric in the Fourier domain, a 1D NNPS was obtained by radial averaging of the 2D NNPS. An RQA 5 spectrum was used with a 74 kVp x-ray beam and added filtration of 21 mm of aluminum.

2.3.4 Generalized linear system parameters calculation—The generalized system parameter includes focal spot blurring due to the finite size of focal spot at the detector plane followed by the degradation due to the detector blur as part of the imaging chain. The Generalized MTF (GMTF) is given by combining the MTFs for the detector (MTF_D), and the focal spot (MTF_F) using the following equation:^[7,9]

$$GMTF(f, m) = MTF_F \left(\left(\frac{m-1}{m} \right) f \right) \times MTF_D \left(\frac{f}{m} \right) \quad (1)$$

Here f is the spatial frequency; m is the object magnification of features in a plane within the object onto the detector plane. GMTF is defined in the object plane, whereas intrinsic detector MTF is defined in the detector plane. Therefore, a scaling factor based on the object magnification was used for all the MTFs to transfer the spatial frequency in the detector plane to the spatial frequency in the object plane. The 1D GMTF was calculated from the 1D detector MTF and the 1D focal spot MTF (radially averaged). The 2D GMTF was calculated from the 2D symmetric detector MTF and the 2D asymmetric focal spot MTF.

Similarly, the Generalized Normalized Noise Power Spectrum (GNNPS) in the object plane is defined as

$$GNNPS(f, X, m) = \frac{NPS_D \left(\frac{f}{m}, X \right)}{m^2 d^2(X)} = \frac{NNPS_D \left(\frac{f}{m}, X \right)}{m^2} \quad (2)$$

Where $NPS_D(f/m, X)$ and $NNPS_D(f/m, X)$ are defined in the object plane and X is the x-ray exposure measured at the detector entrance surface. $d(X)$ in the denominator is the averaged pixel value of the flat field image, assuming the detector is linear or linearized.

The GNEQ gives the effective number of input quanta used by the system to produce an image; that is, it gives the number of Poisson distributed quanta for an ideal detector that would provide the same signal-to-noise (SNR) as measured. The GNEQ is the statistical measure of the image quality. Using the GMTF and GNNPS, we can define GNEQ as follow;

$$GNEQ(f, X, m) = \frac{GMTF^2(f, m)}{GNNPS(f, X, m)} \quad (3)$$

The 1D GNEQ was calculated from the 1D GMTF and the 1D NNPS (radially averaged). The 2D GNEQ was calculated from the 2D asymmetric GMTF and the 2D symmetric NNPS.

3. RESULTS AND DISCUSSIONS

3.1 Method 1: Design of an x-ray tube with higher tube output for a small focal spot

Table 1 shows the results of decreasing the anode angle from 8 to 2 degrees and increasing the filament length by a factor of 4 times to maintain a constant effective focal spot size.

Column 2 gives the relative beam intensity for the same mA, column 3 compares the maximum field-of-view dimensions at 105 cm SID, column 4 gives the increase in filament length for the same effective focal spot size, column 5 gives the relative tube output with constant electron density on the target, and column 6 gives the increase in tube current considering the current 8 degree limit is 160 mA. We can clearly see that the relative tube output can be increased 3.0 times by increasing the length of the filament projection on the target by 4 times after taking the intensity loss due to inherent filtrations at the anode disk into account due to the decrease of the anode angle from 8 degrees to 2 degrees. Also, tube output can be increased to about 4.0 times at 2 degrees by removing the 1.8 mm Al filtration that is added in the existing tube during angiographic procedures; removal of the added filtration is balance by the increased inherent filtration at 2 degrees due to the increased anode beam hardening. Keeping the electron flux density and heat distribution unchanged, the increase in tube output could be achieved by increasing the anode diameter from 132 mm (current Toshiba Infinix system) to 145 mm which is less than that of a commercial tube such as the replacement for GE CT scanners (Prospect, IAE X-RAY TUBES, Milano-Italy, web address: http://www.iae.it/serie-general-electric_47.html) of 165mm. The FOV for 2 degrees anode angle is sufficient to cover any 5 cm ROI imager such as the MAF. The HVL of the proposed x-ray tube at 75 kV_p, without a phantom and added filtration in the beam was calculated to be 3.5 mm Al; this is very close to that of the present x-ray tube (i.e. 3.6 mm) and is well above the minimum HVL requirement of the FDA.

Figure 5 and 6 show the x-ray spectra of a tungsten anode for different anode angles at 75kVp with and without a head phantom in the x-ray beam. We can see clearly that the average energy of the beam increases; however, the beam intensity decreases as the anode angle decreases. As compared to Figure 5, we can also see in Figure 6 that the beam intensity decreased at all the energies because of attenuation due to the phantom in the beam with almost complete absorption at the lower energies. Figures 7 and 8 show the x-ray spectra for a tungsten anode for 2 degrees anode angle (with and without added filtration) and 8 degrees anode angle (with the added filtration) at 75kV_p without and with the head phantom in the beam, respectively. We can see in Figure 7 that the x-ray spectrum for 2 degrees anode angle without added filtration is close to that of the 8 degrees anode angle with added filtration in terms of beam quantity and quality. Whereas, in Figure 8 with a head phantom in the beam, we can clearly see that they are almost equivalent in terms of quantity and quality. Figure 9 shows the x-ray beam intensity reduction at 105cm (position of the MAF) and the central axis position as the anode angle decreases due to the increased inherent filtration by the tungsten target itself with and without the phantom in the beam relative to that at 8 degrees. We can clearly see that the reduction in intensity is less with the phantom in the beam, because of the higher absorption of low energy photons. Figure 10 shows that the relative tube output increases as the anode angle decreases and the projected filament length increases assuming the effective focal spot size at the central axis is constant. For each anode angle, the corresponding increased projected filament length is shown in the Figure, relative to the filament length (L_1) of the current x-ray tube. Due to the heel effect, we get an intensity variation of 20% from edge to edge over the FOV for the MAF in the tube axis direction as compared to 16% for the 20×20 cm² Flat panel detector (FPD) imager.

3.2 Method 2: Increase of the image quality and resolution for a medium focal spot

Figure 11 shows the x-ray images of the small focal spot at the central axis, and medium focal spot at the central axis and at the anode side. In image c, we can clearly see the size of the medium focal spot is decreased in the direction parallel to the tube axis as compared to the medium focal spot at the center (image b), with no change in size in the perpendicular direction. The decrease in the effective focal spot size is due to the decrease in the effective

anode angle from 8 degrees to 4.5 degrees. Hence, the nominal size of the medium focal spot at the anode side in the tube axis direction is expected to be 0.28 mm. Figure 12 shows the 1D focal spot MTFs (radial average of the 2D MTFs) of the small focal spot at the center, and medium focal spot at the center and anode side. Here, the spatial frequency is defined at the focal spot plane. As the central axis at the anode side position is not perpendicular to the MAF imager plane, hence, there will be a small error of <1% in the spatial frequencies which we neglect in this paper. We can clearly see the degradation in MTF for the medium focal spot as compared to the small focal spot at the center. We also see the improvement in MTF for the medium focal spot at the anode side as compared to the medium focal spot at the center.

Figure 13 shows the 1D GNEQ of the small focal spot at the center, and medium focal spot at the center and anode position, and with different temporal filtering weights (TFWs). We use the 1D focal spot (radially averaged from the 2D asymmetric focal spot MTF) for the calculation of the 1D GNEQ. The GNEQ is calculated from the radial averaged. The medium focal spot is allowed to have higher tube output of 3.12 times as compared to the small focal spot at the central axis position; however, the tube output is 2.85 times higher at the anode side considering the 9% reduction in intensity due to the Heel effect, based on the maximum tube current available in our Toshiba Infinix system. The increased output for the MAF is justified since the effective dose is greatly reduced by the use of a small FOV compared to that of standard FPDs. We can clearly see that the GNEQ is higher almost at all the frequencies for the medium focal spot at the anode side than at the central axis, and the small focal spot at the central axis. Figure 14 shows the ratio of the 2D GNEQ of medium focal spot at the anode to small focal spot at the center. Here we use the 2D asymmetric focal spot MTF for the calculation of the 2D GNEQ. We can clearly see that the GNEQ ratio is higher at all the frequencies in the tube axis direction. As mentioned earlier, the collimation in the current x-ray tube output port is limited to the size of the larger imager. Hence, we are limited in moving our detector farther towards the anode side. If we could disable the collimation limit, then we would be able to get an even smaller effective focal spot size.

3.3 Method 3: Increase of the image quality for the existing small focal spot

Figure 15 shows the 1D GNEQ of a single image for different acquisition frame rates and temporal filtering weights using the existing small focal spot. Here an acquisition rate of 3 f/s (3 frames/sec) in the plot corresponds to the standard acquisition frame rate that is used in neuro-angiography in which images are displayed at 3 f/s, and we take this as a reference to compare with the other GNEQs. Here, we increase the acquisition frame rate by five times to 15 f/s for each case; however, the images are displayed at the same rate of 3 f/s where each image is obtained by integrating 5 consecutive images. As a result, the GNEQ increases by 5 times for all spatial frequencies since the total mAs for each resultant image is increased 5 times. We can achieve the similar increase of 5 times in the GNEQ by applying the recursive temporal filtering of weight 3 to the images obtained at 15f/s. The assumption that is made here is that the GNEQ of the image is obtained after an infinite number of images have been filtered. However, in the actual case, we used instead a finite image sequence of 15 images for the recursive filtering with an error of less than 1% in the GNEQ for $k=3$ (The signal of the new filtered image for the subsequent 15 images will be $1-(1-1/k)^{15} = 0.998$). Here, we get an increase in the image quality using the higher acquisition frame rate either with a running average of 5 frames or a recursion filter of weight 3. Additionally, we maintain the spatial and temporal resolution using the small focal spot and higher frame rate, respectively.

4. CONCLUSIONS

For neurovascular EIGIs where high resolution is essential over a small FOV or ROI imager, higher tube output required for high quality images can be achieved while maintaining the spatial resolution using longer filaments with reduced anode angle using the standard Bremsstrahlung process with only a small change to standard x-ray tube geometry. Also, the GNEQ can be increased for the MAF in the case of the medium focal spot using a view at the anode side with higher output or in the case of the small focal spot using increased frame rate with higher recursive temporal filtering. We show how the tube output and hence the image quality in terms of the GNEQ can be increased while maintaining the spatial resolution in the region of interventional interest without the limitations of the small focal spot with one or more of the above three methods. Increased dose can be acceptable for improved image quality when needed for the interventional task; in this case, the dose is over a much smaller FOV resulting in less effective dose to the patient. As a result of the increased image quality, a patient could be diagnosed and treated more efficaciously or the procedure could be completed more expeditiously. Moreover, the higher image quality can help endovascular device deployment and verification after deployment which in turn could increase the success probability of the procedure as well as improve the treatment especially in complex cases.

Acknowledgments

This work was supported in part by the NIH Grants R01-EB008425, R01-EB002873 and an equipment grant from Toshiba Medical Systems Corp.

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Figure 1.
Schematic of MAF

Figure 2.
Picture of MAF

Figure 3. Schematic of the same x-ray tube, but in two different configurations of tube insert: large FOV imager (left) and the MAF ROI imager (right).

Figure 4. Experimental set-up for the focal spot psf measurement with the MAF at the central axis (left) and at the anode side with an off-axis angle of 3.5° (effective anode angle of 4.5°) (right).

Figure 5.
X-ray spectra of tungsten anode for different anode angles without head phantom in the beam.

Figure 6.
X-ray spectra of tungsten anode for different anode angles with head phantom in the beam.

Figure 7.
X-ray spectra of tungsten anode with and without added filtration without phantom in the beam

Figure 8.
X-ray spectra of tungsten anode with and without added filtration with phantom in the beam

Figure 9.
Relative beam intensity at the central axis relative to that at 8 degree.

Figure 10.
Relative tube output vs. anode angles, keeping effective focal spot size as constant.

Figure 11.

Figure 1: X-ray pin-hole images of (a) small focal spot (nominal size 0.3 mm) at the center, (b) Medium focal spot (nominal size 0.5 mm) at the center, and (c) Medium focal spot at the anode side.

Figure 12.
1D focal spot MTFs of the small and medium focal spot.

Figure 13.
GNEQs of the small and medium focal spots for different outputs.

Figure 14.
GNEQ ratio of the medium focal spot (anode) to the small focal spot (center).

Figure 15.
GNEQs of the small focal spot for different acquisition frame rates.

Table-1

Imaging parameters at the image receptors without phantom in the beam

Anode angle (degree)	Relative beam Intensity (%)	Square Field of view dimensions (cm × cm)	Length of filament projection	Relative tube output	Maximum tube current
8	100	29 × 29	L	1.0	160 mA
2	76	7.4 × 7.4	4L	3.0	640 mA
2 (No AF)	97	7.4 × 7.4	4L	3.9	640 mA