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Focal spot size reduction using asymmetric collimation to enable reduced anode angles with a conventional angiographic x-ray tube for use with high resolution detectors

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Abstract

The high-resolution requirements for neuro-endovascular image-guided interventions (EIGIs) necessitate the use of a small focal-spot size; however, the maximum tube output limits for such small focal-spot sizes may not enable sufficient x-ray fluence after attenuation through the human head to support the desired image quality. This may necessitate the use of a larger focal spot, thus contributing to the overall reduction in resolution. A method for creating a higher-output small effective focal spot based on the line-focus principle has been demonstrated and characterized. By tilting the C-arm gantry, the anode-side of the x-ray field-of-view is accessible using a detector placed off-axis. This tilted central axis diminishes the resultant focal spot size in the anode-cathode direction by the tangent of the effective anode angle, allowing a medium focal spot to be used in place of a small focal spot with minimal losses in resolution but with increased tube output. Images were acquired of two different objects at the central axis, and with the C-arm tilted away from the central axis at 1° increments from 0°–7°. With standard collimation settings, only 6° was accessible, but using asymmetric extended collimation a maximum of 7° was accessed for enhanced comparisons. All objects were positioned perpendicular to the anode-cathode direction and images were compared qualitatively. The increasing advantage of the off-axis focal spots was quantitatively evidenced at each subsequent angle using the Generalized Measured-Relative Object Detectability metric (GM-ROD). This anode-tilt method is a simple and robust way of increasing tube output for a small field-of-view detector without diminishing the overall apparent resolution for neuro-EIGIs.

Keywords

High Resolution Fluoroscope (HRF); Tilted Anode; Generalized Measured-Relative Object Detectability (GM-ROD); Line Focus Principle

INTRODUCTION

The ability of an x-ray detector to resolve fine device detail is paramount to the success of neuro-endovascular image-guided interventional procedures (neuro-EIGIs), provided the x-ray source requirements of sufficient fluence and minimal focal spot blurring are met. A

small focal spot is necessary for resolving the fine spatial detail contained in perforator vessels and vascular treatment devices with minimal focal spot blurring; however, using small focal spot sizes at their maximum output may not provide sufficient x-ray fluence to the detector to provide the image quality desired by interventionalists, and during long procedures the anode target is prone to become overheated, necessitating the use of a larger focal spot in lieu of the smallest focal spot for its increased tube output capabilities and increased heat capacity. However, this switch to larger focal spots causes an undesirable increase in focal spot unsharpness at magnifications greater than unity.

According to the line focus principle¹, the height of the electron beam incident on the anode is compressed by the tangent of the anode angle resulting in the effective focal spot size used for imaging. In this geometry, the effective focal spot size can be decreased in the anode-cathode direction if accessed such that the anode angle is reduced, as seen in Figure 1. Theoretical calculations have shown² a medium focal spot size on a C-arm gantry can be reduced to that of a small focal spot using a tilted anode to access the anode side of the source, effectively creating a high-output, small focal spot equivalent.

Using the experimental task-based Generalized Measured Relative Object Detectability (GM-ROD) metric, the resolution advantage of acquiring images on the anode-side was demonstrated and quantified, allowing for a comparison of the small focal spot at the central axis against the medium focal spot at the anode side.

MATERIALS AND METHODS

To access the anode side of the x-ray source, the Toshiba Infinix C-arm gantry (Toshiba Medical Systems Corp., Otawara, Tochigi Prefecture, Japan) was rotated caudally from 0°–7° in increments of 1° or 2°. As C-arm gantry angle increased, the effective angle of the anode was decreased. The x-ray imaging detector was mounted on a stand that rested on the patient table, rather than on the C-arm, to allow for the gantry to rotate free of the detector. To ensure that all aspects of the image remained unchanged except for focal spot size, the detector maintained constant alignment with the object for acquisition at all angles, and the detector was also aligned with the new perpendicular central ray after each incremental tilt, as shown in Figure 2. This action minimized x-ray obliquity across the detector field of view, which if left uncorrected could induce blurring. The new perpendicular ray was found using a vertical alignment system that was built in-house. After the C-arm was tilted to the desired angle, the patient table was translated under fluoroscopic guidance until vertical alignment between the detector, object, and x-ray source was achieved, indicating alignment of the new central ray and the center of the detector's field of view.

The x-ray beam was collimated to the active area of the detector for all exposures; however, having the detector located at the edge of the x-ray field allowed only three of the symmetrically controlled collimator blades to reach the detector field of view. At such angles, a thin lead block was manually placed on the x-ray source housing to collimate to the fourth side of the detector field of view.

The detector used for this study was the small field-of-view, high resolution CMOS-based HRF (high-resolution fluoroscopic) detector³ (Dexela 1207, Perkin-Elmer Inc., Waltham, MA). The HRF-CMOS₇₅ features 75 micron pixel pitch, an indirect detection CsI(Tl) phosphor layer to convert x-rays into a light image, and a low electronic noise CMOS readout. The small 7 cm × 12 cm region-of-interest field of view of the HRF-CMOS₇₅ allows the full detector face to receive x-ray fluence at the most severe tilt angle.

Two objects were imaged at each angle: a 5 mm long copper wire segment of diameter 113 microns, and a neurovascular pipeline stent. The objects were oriented perpendicular to the anode-cathode direction in every image. The technique parameters and other relevant parameters used in this study are shown in Figure 3. Images were acquired of both objects using Condition 1 and Condition 2. The tube current of 125 mA is the maximum allowed for a small focal spot (0.3 mm), as dictated by the x-ray system. A third set of images was acquired using the parameters described in Condition 3 to demonstrate the advantage of higher tube current afforded by the medium focal spot using a tilted C-arm setup. The tube current of 320 mA is the greatest current afforded by the medium focal spot (0.6 mm) that could be used without saturating the detector. The medium focal spot is capable of achieving a maximum x-ray tube current of 400 mA, and can achieve higher currents with reduced frame rates, all of which would serve to increase the advantage of the medium focal spot. To accentuate the effect of focal spot blurring, all images were acquired at a high magnification of 1.3.

To quantify the advantage of the smaller effective focal spot size, the Generalized Measured Relative Object Detectability (GM-ROD) metric⁴ was employed. The GM-ROD metric, as shown in Eq. 1, is based on ideal observer analysis metrics, and compares the relative performance of two systems and/or techniques in imaging a given measured object on the basis of the power of the Fourier transform of an image of that object ($OBJ(u, v)$), which already includes MTF information, divided by the normalized noise power spectrum (NNPS), and integrated from zero to the Nyquist frequency of the detector, then divided by the comparable integral for a different detector. The zero-frequency value is excluded from all calculations by instead integrating from 0.1 cycles/mm to the Nyquist, and similarly for the negative spatial frequencies, as the zero-frequency value is purely interpolated during calculation of the NNPS. In this experiment, the same detector was used to image the object, and the object function was variable to compare the off-axis imaging performance of the medium focal spot to that of the small focal spot at the central axis.

$$GM-ROD = \frac{\int \frac{|OBJ_1(u,v)|^2}{NNPS_1(u,v)} du dv}{\int \frac{|OBJ_2(u,v)|^2}{NNPS_2(u,v)} du dv} \quad (1)$$

Using a 10 micron diameter gold-platinum alloy pinhole, images of the small and medium focal spot were acquired at each off-axis angle to measure the relative size of the focal spot with increasing C-arm gantry angle.

RESULTS

Images acquired from this study are shown for the stent in Figure 4 for only the central axis and the most extreme tilt angle, 7° , corresponding to the smallest anode angle. This set of images shows the stent acquired with the medium focal spot at the central axis (Figure 4a.), the small focal spot at the central axis (Figure 4b.), and the medium focal spot at an 7° tilt angle (Figure 4c.). For the stent, the images taken using the medium focal spot at the off-axis location exemplify the diminished focal spot blurring by having comparable resolution to those images taken with the small focal spot at the central axis. The same result was evidenced in the images of the wire (not shown here), and these images were used for the quantitative GM-ROD results shown below.

The principal result of this study is the quantitative comparison between the images acquired off-axis using the medium focal spot and those acquired at the central axis under the small focal spot, both for the maximum small focal spot technique parameters and images using the increased medium focal spot parameters. The GM-ROD results of this comparison are shown in Figure 5 (left). The increase in exposure to the maximum technique parameters for the medium focal spot, and its subsequent change in the normalized noise power spectrum, causes a jump in GM-ROD value for these comparisons seen in Figure 5 (right). The error in these measurements was determined from three trials of each acquisition at every angle.

The C-arm tilt angle readout was limited to integer resolution, making accurate positioning of the C-arm difficult. During initial tests, a C-arm tilt angle of 8° was attempted. Slight differences in C-arm position contributed to some variation in collimator blade visibility in the images at this angle, which introduced greater uncertainty into the data acquired. This data was subsequently excluded from further analysis.

Further analysis of the images was performed to obtain information on the change or improvement in resolution at specific spatial frequencies. The GM-ROD metric can be altered to calculate the performance advantage at specific frequency ranges by integrating over a different frequency range rather than over all spatial frequencies up to the Nyquist frequency. In Figure 6 (left and right), the GM-ROD results for both objects are shown as a function of increasing lower bound of integration, with each subsequent bound putting greater emphasis on higher spatial frequencies, and excluding the lower frequencies. The results for both the stent and the wire indicate similar change with increasing angle. The GM-ROD results calculated with 1, 2, and 3 cycles/mm as the lower bound of integration initially have worse performance than the original GM-ROD values (plotted in black), but begin to increase in value at a C-arm tilt angle of 2 degrees. In contrast, the results calculated with 4, 5, and 6 cycles/mm as the lower bound of integration have initial values near that of the original GM-ROD values, and do not indicate high frequency improvement of the medium focal spot until the C-arm has been tilted to 6° .

The GM-ROD measurements can be separated into the object function and the noise power spectrum to determine the contribution to the GM-ROD values from each component in the definition, the noise and the object function. This is done by defining either the NNPS or the

Fourier transform of the object function to be a matrix containing only ones, so that only one part of the quotient contributes to the altered GM-ROD value.

The noiseless, object function only (NNPS is defined to be a matrix containing all ones) results from the stent images are shown in Figure 7, calculated for increasing lower bounds of integration. The GM-ROD result calculated for all spatial frequencies shows no improvement with increasing C-arm tilt. Conversely, results using all other bounds of integration show improvement beginning at 2° or 4° tilt, and continuing to increase at a tilt of 7°, with the plot emphasizing only the highest spatial frequencies (lower bound of 6 cycles/mm) showing the greatest improvement over images of the stent imaged using the small focal spot at the central axis.

The noise-only GM-ROD values (the Fourier transform of the object function is defined to be a matrix containing only ones) were calculated using the normalized noise power spectra acquired from images obtained using all three x-ray conditions. The results comparing the NNPS of images acquired using Condition 2 to images acquired using Condition 1 are shown in Figure 8 (left) for all bounds of integration. This result indicates little variation in the NNPS for all spatial frequencies between the medium and small focal spots for C-arm tilt angles 0°–4°. The results for 6° and 7° show a slight decrease, indicating an increase in noise seen in images acquired at these angles. This loss of exposure with increasing C-arm angle is due to the Heel effect.

Figure 8 (right) shows the noise-only results from the NNPS calculated from images acquired using Conditions 1 and 3. As expected, a large increase in GM-ROD value is seen as a result of the increased exposure used in Condition 3 as compared to Condition 2. The results as a function of increasing C-arm tilt angle show similar tendencies to those seen in Figure 8 (left). There is little fluctuation between 0°–4°, with a slight decrease seen for 6° and 7° due to Heel effect.

The pinhole images of the small and medium focal spot, shown in Figure 9, exemplify the additional compression of the effective focal spot size through the simple action of tilting the C-arm gantry. The focal spot images were acquired using technique parameters chosen to prevent focal spot blooming, and were selected to be 60 kVp, 125 mA, and 16 ms. The dimension of the focal spot perpendicular to the anode-cathode direction clearly differs between the small and medium focal spots, indicating the occurrence of blurring in this direction. Ideally, this dimension would be identical for small and medium focal spots such that a small focal spot equivalent could be readily created using the tilted C-arm method.

Line profiles exhibiting the increasingly compressed focal spot size of the medium focal spot are shown in Figure 10 (left). The decreasing maximum intensity of the line profiles also makes clear the presence of the Heel effect as the images are acquired increasingly further on the anode side of the field. Figure 10 (right) shows the profiles of the small and medium focal spot at the central axis, and the compressed medium focal spot profile imaged at 4°, normalized to show each curve with the same maximum intensity. The change in effective focal spot size is apparent, and a tilt angle of just 4° provides a full width, half maximum slightly larger than that of the small focal spot.

DISCUSSION

Though 320 mA was used in Condition 3 to show the utility of increased output of the medium focal spot in this study, greater x-ray tube currents are available from the medium focal spot that could provide more sufficient fluence to form an image during an intervention, adding to the usefulness of this technique. In addition, past studies by this group have shown² that removal of the added filtration when tilting the C-arm gantry could eliminate x-ray fluence losses experienced at low anode angles due to increased inherent self-filtration, or Heel effect, and could increase x-ray fluence by a factor of four while maintaining the beam quality. Future experiments will make use of this technique to allow for fairer comparisons of images acquired at the central axis and off-axis locations.

CONCLUSIONS

The decrease in focal spot size due to anode tilting was clearly demonstrated both qualitatively, as seen in the images used for comparison, and quantitatively using the task-based GM-ROD metric, and the similar performance of the small focal spot at the central axis and the medium focal spot off-axis was evidenced. The improved resolution provided by the compressed anode was exhibited in the noise-less, object only GM-ROD values and exhibited that the performance of the medium focal spot becomes comparable to the small focal spot at the most extreme C-arm tilt angle. With the larger medium focal spot size, the output of the maximum tube loading can be increased without suffering a substantial loss of resolution during neuro-endovascular image-guided interventions where therapeutic vascular devices can be difficult to resolve. This tilted C-arm setup appears simple, and can be implemented in a straightforward way in angiographic suites for real time interventional procedures. This method was designed to mitigate this effect for reduced field of view, high-resolution detectors, including the suite of high-resolution fluoroscopic detectors.

Acknowledgments

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Figure 1.

(Not to scale) A large effective focal spot (left) can be made into a smaller effective focal spot by simply tilting the anode to have a smaller projected focal spot size. The method depicted here reduces the angle on the anode itself, but the method proposed in this paper uses a tilt of the C-arm to change the effective anode angle.

Figure 2.

The C-arm is tilted caudally to access the anode side of the x-ray field of view, and with each angulation the new central ray must be determined.

Imaging Condition	Focal Spot	kVp	mA	ms	Magnification
Condition 1	Small	70	125	16	1.3
Condition 2	Medium	70	125	16	1.3
Condition 3	Medium	70	320	16	1.3

Figure 3.

Each of the three sets of x-ray technique parameters used for acquiring images at each angle.

Figure 4.

Images of a pipeline stent acquired for a. and b. at the central axis using the medium and small focal spot (Condition 2 and 1), respectively. The stent in image c. was acquired using the medium focal spot (Condition 2) with the C-arm tilted to 7 degrees.

Figure 5.

(Left) The GM-ROD results for the pipeline stent and copper wire, comparing images acquired at Condition 2 to Condition 1. (Right) The GM-ROD results for the pipeline stent and copper wire comparing Condition 3 to Condition 1. This jump in GM-ROD value from the result on the left is from the decrease in NNPS due to increased exposure.

Figure 6.

The GM-ROD results for the stent (left) and wire (right) calculated to show the change in resolution of the medium focal spot with increasing lower bounds of integration, relative to the small focal spot at the central axis.

Figure 7.
The GM-ROD results considering only the stent object function.

Figure 8.

The GM-ROD results considering only the normalized noise power spectra for Conditions 1 and 2 (left) and Conditions 1 and 3 (right). The x-ray technique parameters were kept constant for each acquisition.

Figure 9.

(Left) Pinhole images of the small focal spot at the central axis, and the medium focal spot at a range of angles. The compression of the medium focal spot size is apparent after a tilt of only 2 degrees, and continues to be compressed with increasing C-arm tilt. (Right) Normalized line profiles of the small focal spot and medium focal spot at the central axis along with the medium focal spot profile at a C-arm tilt angle of 4°. The compression of the medium focal spot is apparent after a tilt of just 4°.

Figure 10.

Line profiles drawn from the pinhole focal spot images. As the C-arm caudal tilt increases the full width-half maximum of each curve decreases, eventually becoming less than that of the small focal spot at 6 degrees.