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Focal Spot Deblurring for High Resolution Direct Conversion X-ray Detectors

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Abstract

Small pixel high resolution direct x-ray detectors have the advantage of higher spatial sampling and decreased blurring characteristic. The limiting factors for such systems becomes the degradation due to the focal spot size.

One solution is a smaller focal spot; however, this can limit the output of the x-ray tube. Here a software solution of deconvolving with an estimated focal spot blur is presented.

To simulate images from a direct detector affected with focal-spot blur, first a set of high-resolution stent images (FRED from Microvention, Inc., Tustin, CA) were acquired using a 75 μ m pixel size Dexela-Perkin-Elmer detector and frame averaged to reduce quantum noise. Then the averaged image was blurred with a known Gaussian blur. To add noise to the blurred image a flat-field image was multiplied with the blurred image. Both the ideal and the noisy-blurred images were then deconvolved with the known Gaussian function using either threshold-based inverse filtering or Weiner deconvolution.

The blur in the ideal image was removed and the details were recovered successfully. However, the inverse filtering deconvolution process is extremely susceptible to noise. The Weiner deconvolution process was able to recover more of the details of the stent from the noisy-blurred image, but for noisier images, stent details are still lost in the recovery process.

INTRODUCTION

X-ray detectors can be classified into two general categories direct and indirect detectors. In direct detection, the x-rays are absorbed in the detector layer and generate electron-hole pair signal charge, which is then read out by the electronics. Figure 1 shows the schematic of a direct detector. There is minimal spreading of the generated charge due to high electric field potentials across the detector layer. Amorphous Selenium based flat panel direct detectors have been used successfully in the medical imaging arena for a few years now [1].

In indirect detectors the x-ray photons are absorbed in a scintillator layer and gets converted to light photons. This is then converted to the signal (electrons) in the photodiode array structure, which is then read out using the electronics. Figure 2 shows the schematic of the indirect detector. In the indirect detector, there can be significant spreading of the generated light photons in the scintillator significantly contributing to the detector blur. Using a

structured scintillator can reduce this spreading, but it is still significantly higher compared to that for direct detectors. Cesium Iodide based flat panel indirect detectors are being widely used in commercial x-ray imaging systems [2]. The two general limiting factors for x-ray based imaging systems are the detector blurring and the focal spot blurring. The detector blurring is influenced by the different components involved in its construction. For an indirect detector the significant source of detector blurring can be the scintillator layer. As the pixel sizes get smaller the improvement in resolution is less significant due to blurring in the scintillator layer. However for a direct detector the limiting factor is the pixel size itself.

Due to the finite size of the focal spot, when the object is further away from the detector, the edges in the image of the object get blurred (Penumbra).

The total blur (degradation) in the final image is a combination of detector and focal spot blurring. CMOS based aSe detectors [3] have the advantages of higher spatial sampling due to smaller pixel sizes and decreased blurring characteristic of direct rather than indirect detection. For systems with such detectors, the limiting factor degrading image resolution then becomes the focal spot geometric un-sharpness. In this work we aim to investigate the removal of the effect of focal spot blur and thus improve the image quality and specifically obtain better visualization of endovascular devices.

METHODS

A set of high resolution images of a stent and flat-field images were acquired using a 75 μ m pixel size Dexela-Perkin-Elmer detector [4]. A set of 100 stent images were averaged to reduce quantum noise in image $I(x,y)$ shown in figure 3, while a set of 10 flat field images were averaged to generate a noise reference image $N(x,y)$.

To simulate images from a high resolution detector affected with focal spot blur, the averaged image was convolved with a Gaussian blur functions ($G(x,y)$) (figure 4) simulating the focal spot blurring, to generate $I_{blurred_ideal}(x,y)$ shown in figure 5.

$$i_{blurred_ideal}(u,v) = i(u,v) \times g(u,v) \quad (1)$$

where $i_{blurred_ideal}(u,v)$, $i(u,v)$, $g(u,v)$ are the Fourier transforms of $I_{blurred_ideal}(x,y)$, $I(x,y)$ and $G(x,y)$ respectively.

To add noise to the simulated image, the blurred image $I_{blurred_ideal}(x,y)$ was multiplied with the average flat-field image $F(x,y)$ to generate $I_{blurred_noisy}(x,y)$ figure 6.

$$i_{blurred_noisy}(x,y) = I_{blurred_ideal}(x,y) \times F(x,y) \quad (2)$$

In order to remove the effect of the Gaussian blur function two techniques based on deconvolution were employed: inverse filtering with thresholding and Wiener deconvolution

1) Inverse filtering with thresholding

The deblurred image ($I'(x,y)$) can be obtained by deconvolving the blurred image $I_{blurred_ideal}(x,y)$ with $G(x,y)$ as shown below [5].

$$i'(u,v) = \frac{i_{blurred_ideal}}{g(u,v)} \quad (3)$$

Figure 7 shows the Fourier transform of a Gaussian function $g(u,v)$. It can be observed that its value decreases as the frequency increases and can reach extremely low values. By using this as is, deconvolution can result in a noisy image, with much of the information buried in the noise. To avoid this, a threshold, α , is applied to the blurring function as shown in eqn. 3.

$$g(u,v) = \begin{cases} g(u,v), & g(u,v) \geq \alpha \\ \alpha, & g(u,v) < \alpha \end{cases} \quad (3)$$

From equation 1 if $G(x,y)$ is known or measured, then using the above technique on $I_{blurred_ideal}(x,y)$ results in a deblurred image, as shown in figure 8. However in a non-ideal case, even though the blurring function can be measured or estimated, the noise in the image can cause significant artifacts. Applying the above technique to the noisy image $I_{blurred_noisy}(x,y)$ results in a noisy deblurred image as shown in figure 9.

2) Wiener Deconvolution

When the image is blurred by a known low-pass filter, it is possible to recover the image by threshold-inverse filtering. However, inverse filtering is very sensitive to noise. The Wiener filtering executes an optimal tradeoff between inverse filtering and noise smoothing. It removes the noise and inverts the blurring simultaneously [6]. The Matlab implementation of the Wiener filter uses a built in function *deconvwnr*, the syntax of which is shown below [7].

$$I_{deblurred_noisy}(x,y) = deconvwnr(I_{blurred_noisy}(x,y), G(x,y), NSR)$$

where NSR is the noise to signal power ratio. When NSR is 0, the function is the same as inverse filtering. As the NSR increases the noise in the image is reduced and then inverse filtering is applied to remove the blurring effect. Applying the above technique with the NSR of 150 to the noisy image (figure 5) results in a noisy deblurred image shown in figure 10. Figure 11 shows a comparison of the line profile across the stent of the noisy blurred image and the deblurred image using both of the above techniques.

DISCUSSION

From Figure 8, it can be seen that the blur in the low noise image of Figure 5 was removed by deconvolving the image using the known or estimated Gaussian function with the inverse filtering technique. However the filtering technique is susceptible to noise in the image and can cause noisy artifacts as shown in Figure 9 where the attempt at deconvolution was done on the noisier image of Figure 6. Wiener deconvolution techniques resulting in Figure 10 offer a tradeoff between inverse filtering and noise smoothing. It removes the noise and inverts the blurring simultaneously. Comparing Figure 10 and 9, shows that results from Wiener deconvolution are better than from the inverse filtering technique. This is further demonstrated by the line profile comparison of Figure 11 which shows that the Weiner deconvolution produces improvement in the stent strut delineation compared to that in the blurred noisy image with better noise reduction than the inverse filtering with threshold technique.

Experimentally, focal spot blur at different magnifications can be measured using a pin-hole with a high resolution detector. This spread function can then be used to deblur the input images that are acquired at corresponding magnifications to correct for the focal spot blur. Similarly, if object magnification can be determined such correction may be applied in angiography applications depending upon the magnitude of the noise.

Acknowledgments

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Figure 1.
Schematic of a direct x-ray detector

Figure 2.
Schematic of an indirect x-ray detector

Figure 3.
High resolution stent image, obtained by averaging a set of 100 images obtained using a 75 um dexela detector

Figure 4.
Gaussian Blur function used to simulate focal spot blur

Figure 5.
Figure 3 blurred with the Gaussian blur shown in figure 4. The SNR ratio in the background is 2160.5

Figure 6.

Figure 5 multiplied with a flat field image to add quantum noise. The SNR ratio in the background is 627

Figure 7.
Fourier transform of the Gaussian blur function shown in figure 4

Figure 8.
Noise-free image of Figure 5 after deblurring using inverse filtering with thresholding

Figure 9.
Noisy image of Figure 6 after deblurring using inverse filtering with thresholding

Figure 10.

Noisy image of Figure 6 after deblurring using Wiener deconvolution technique

Figure 11.
Line Profiles across the stent (shown in figure 10) comparing Figures 6, 9, and 10