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Improved contrast and spatial resolution with Single Photon Counting (SPC) for an area x-ray imager, the newly developed high-resolution Micro-Angiographic Fluoroscopic (MAF) detector

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Abstract

Although in radiological imaging, the prevailing mode of acquisition is the integration of the energy deposited by all x-rays absorbed in the imaging detector, much improvement in image spatial and contrast resolution could be achieved if each individual x-ray photon were detected and counted separately. In this work we compare the conventional energy integration (EI) mode with the new single photon counting (SPC) mode for a recently developed high-resolution Micro-Angiographic Fluoroscopic (MAF) detector, which is uniquely capable of both modes of operation. The MAF has 1024×1024 pixels of 35 microns effective size and is capable of real-time imaging at 30 fps. The large variable gain of its light image intensifier (LII) provides quantum limited operation with essentially no additive instrumentation noise and enables the MAF to operate in both EI and the very sensitive low-exposure SPC modes. We used high LII gain with very low exposure (<1 x-ray photon/pixel) per frame for SPC mode and higher exposure per frame with lower gain for EI mode. Multiple signal-thresholded frames were summed in SPC mode to provide an integrated frame with the same total exposure as EI mode. A heavily K-edge filtered x-ray beam (average energy of 31 keV) was used to provide a nearly monochromatic spectrum. The MTF measured using a standard slit method showed a dramatic improvement for the SPC mode over the EI mode at all frequencies. Images of a line pair phantom also showed improved spatial resolution with 12 lp/mm visible in SPC mode compared to only 8 lp/mm in EI mode. In SPC mode, images of human distal and middle phalanges showed the trabecular structures of the bone with far better contrast and detail. These improvements with the SPC mode should be advantageous for clinical applications where high resolution and/or high contrast are essential

such as in mammography and extremity imaging as well as for dual modality applications, which combine nuclear medicine and x-ray imaging using a single detector.

I. Introduction

X-ray imaging is one of the most powerful imaging techniques in the medical field. Normally in x-ray imaging, the common mode of image acquisition is energy integrating (EI) mode. In EI mode, the total energy deposited by photons is integrated to form the image. The digital number of any pixel in the image represents the amount of energy deposited at that particular location. An alternative mode of imaging is the Single Photon Counting (SPC) mode. In this mode an individual x-ray photon is counted and the digital number at any pixel in the image represents the number of x-ray photons absorbed at that location. In energy integrating mode, detectors show a considerable spatial resolution but often are limited by the spatial resolution of the fluorescent phosphor. In comparison with EI mode, SPC mode shows better spatial resolution because the position of the event can be localized and this minimizes the blur associated with conversion to secondary quanta. Others have also shown the improvement of spatial resolution with SPC mode [1], [2], [3].

In this study, we compared the performance of our newly developed highly sensitive, high resolution Micro-Angiographic Fluoroscopic (MAF) detector in the two different modes of operation: Energy Integrating and Single Photon Counting. The MTF was measured in each mode to quantify resolution differences. The improvements in resolution and contrast were investigated qualitatively using images of a mammographic line pair phantom and human phalanges bones.

II. Method And Materials

A. Micro-Angiographic Fluoroscope (MAF)

The Micro-Angiographic Fluoroscopic (MAF) detector was used for this study. The MAF is a region of interest x-ray imaging detector with large variable gain and low instrumentation noise [4]. It is capable of real-time imaging (30 fps) for both fluoroscopic and angiographic applications [5]. The effective pixel size of 35 microns enables very high spatial resolution. Fig. 1 shows a schematic of the MAF.

As shown in fig. 1, there is a CCD camera (Model Pantera TF-1M30, Dalsa Corp., Waterloo, ON, Canada) coupled with a generation 2 light image intensifier (LII) (Model PP0410K, DEP Inc., Dwaziewegen 2, NL-9300 AB Roden, The Netherlands) through a 2.88:1 ratio fiber optic taper (FOT). The LII is coupled to a 300 micron thick CsI(Tl) phosphor (Hamamatsu Corp., Bridgewater, NJ) through a fiber optic plate (FOP). A picture of the MAF is shown in fig. 2 and its specifications are given in Table I.

B. X-Rays Spectrum

For this study we used an approximately monochromatic x-ray spectrum achieved by selecting an x-ray tube output of 50 kVp which was filtered with 15 mm of iodine contrast

media (density 350 mg/ml). The shape of the spectrum is shown in the fig. 3 as calculated using the SRS-78 x-ray spectrum generation software [6].

C. Spatial resolution and contrast studies

For the spatial resolution comparison, the standard slit method for MTF determination [7] and a mammographic line pair phantom were used for both modes and for various threshold values for the SPC mode. Additionally, a qualitative comparison was made using human distal and middle phalanges bones of the pinky-finger imaged to show the comparative contrast between the two modes. Fig. 4 shows images of the mammography line-pair phantom and location of the human phalanges bones.

D. Image Processing for Single Photon Counting

Photon absorption events were identified by using a threshold so that only those events above the selected threshold were registered. This effectively truncated the tails of the phosphor blur function. Afterward the final image was formed by adding all events above the threshold from all the frames. In the thresholding technique [1], [2], all the digital numbers below a threshold were considered zero and all the digital numbers above that threshold were considered one (indicating a single event). For this technique, a threshold for the SPC mode was chosen so that the average fluence rate for accepted events corresponded to the estimated theoretical fluence rate determined using exposure measurements, the calculated x-ray spectrum, and absorption characteristics of the converting phosphor. This threshold level then provides an average of one count per x-ray photon absorbed. Fig. 5(a) and 5(b) give an example of the simple thresholding technique. After getting a large number of thresholded images, we stacked them together to get a final image. All the images for SPC mode for this study were processed similarly. For comparison, images were also obtained in EI mode using an exposure per frame equal to the total exposure for all SPC frames together. In this way, we compared images from the two modes with the same total detector entrance exposure.

III. Results

Table II shows some of the calculations used to determine the number of absorbed x-ray photons and the total counts expected. It shows that for the spectrum, the scintillator absorption efficiency and the exposure ($1.37 \mu\text{R}/\text{frame}$) used, we expected to get 0.15 absorbed x-ray photons per pixel per frame. We also have an example calculation for the slit image. Knowing the slit area, we calculate the number of x-ray photons per frame for the slit image. Once we know that number, we apply the appropriate threshold to get the approximate count number to be close to the calculated one and then we summed the frames and formed the SPC mode image.

Fig. 6(a) and 6(b) show the comparison of the slit images taken in energy integrating and single photon counting mode, respectively.

These slit images were used to calculate the modulation transfer function (MTF) using the standard slanted slit method. Fig. 7 shows a comparison of the MTF for both modes and clearly demonstrates a substantial improvement with the SPC mode.

The images of a mammographic line pair phantom (Fig. 8) give a visual demonstration of the improvement in spatial resolution for the SPC mode.

Fig. 9 shows images of the human phalanges bone taken with EI and SPC modes.

Fig. 10 shows normalized line profiles taken over the same selected region for the two different modes (indicated by the dotted arrows shown in fig. 9). For these phalanges bone images 2000 frames were acquired with the same exposure used for the slit images ($1.37 \mu\text{R}/\text{frame}$).

IV. Discussion

Fig. 6 shows a comparison between the slit images taken with different modes. The slit image taken with SPC mode looks narrower than the image taken in EI mode. The experimentally derived MTF shows a dramatic improvement for the SPC mode over the EI mode at all frequencies as indicated in fig. 7. Improvement in MTF occurs because of the improved localization of the event enabled by reducing the effect of light spreading and limiting detection to a single pixel only. Images of the mammography line-pair phantom shown in Fig. 8 also demonstrate the resolution improvement in SPC mode. In EI mode we are able to see 8 lp/mm, while in SPC mode we were able to see 12 lp/mm. It is also evident that visualization in SPC mode improved for lower values of line pairs per mm.

Another demonstration related to contrast improvement. Fig. 9 shows images of the distal and intermediate human phalanges bones for both the modes taken at the same effective exposure. The first image is taken with EI mode, while the second one is taken with the SPC mode. For the contrast detail comparison, we took a line profile of the same region in both images and normalized it to the background levels. In the comparison of the contrast detail in the line profiles of Fig. 10, we observe higher contrast for the finer details with SPC mode.

In this study, we used monochromatic spectra because we were using a simple threshold technique to detect events. Better and more complex techniques (such as centroid detection) can be used to detect the event more accurately while retaining the energy information [1–3]. With those techniques, polychromatic spectra can be used.

Another issue of concern can be acquisition time. For this study, we used only 20 fps for acquisition; the time required for acquisition can be made shorter, if we could use a higher fluence rate enabled by the combination of a higher frame rate and/or smaller pixels.

V. Conclusion

The work presented here clearly demonstrates the unique imaging capability of the MAF in both EI and SPC modes; however the operation of the MAF in SPC mode provides higher spatial resolution. The SPC mode also provides better contrast that can be advantageous for demanding applications.

The MAF with SPC mode provides a considerable increase in contrast and spatial resolution over EI mode. Successful operation of the MAF in both modes may also provide a potentially attractive detector for dual imaging applications such as combined nuclear medicine and x-ray imaging with one detector.

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References

1. Beekman FJ, de Vree GA. Photon-counting versus an integrating CCD-based gamma camera: important consequences for spatial resolution. *Phys. Med. Biol.* 2005; 50:N109–N119. [PubMed: 15930598]
2. Miller BW, Barber HB, Barrett HH. Photon-counting gamma camera based on columnar CsI(Tl) optically coupled to a back illuminated CCD. *Proc. SPIE.* 2007; 6510
3. Miller, BW.; Barber, HB.; Barrett, HH.; Wilson, DW.; Chen, L. A low-cost approach to high resolution, single-photon imaging using columnar scintillators and image intensifiers. Presented at IEEE 2006 NSS/MIC; October 29–November 4; 2006.
4. Yadava GK, Kuhls-Gilcrist AT, Rudin S, Patel VK, Hoffmann KR, Bednarek DR. A practical exposure-equivalent metric for instrumentation noise in x-ray imaging systems. *Phys. Med. Biol.* 2008; 53:5107–5121. [PubMed: 18723932]
5. Ionita CN, et al. Implementation of a high-sensitivity Micro-Angiographic Fluoroscope (HS-MAF) for in-vivo endovascular image guided interventions (EIGI) and region-of-interest computed tomography (ROI-CT). *Proc Soc SPIE.* 2008:6918.
6. Cranley K, et al. Catalogue of diagnostic X-ray spectra and other data. *The Inst. Of Phys. And Eng. In Med.* 1997 *Report No.* 78.
7. Fujita H, Tsai DY, Itoh T, Doi K, Mrishita J, Ueda K, Ohtsuka A. A simple method for determining the modulation transfer function in digital radiography. *IEEE Trans. Med. Img.* 1992; 11:34–39.

Fig. 1.
Schematic diagram of the MAF.

Fig. 2.
Micro-Angiographic Fluoroscopic (MAF) detector.

Fig. 3.
Calculated x-ray spectrum used. The spectrum was generated using 50 kVp x-rays filtered with iodine (density*thickness = 350 mg/cm²).

Fig. 4.
Picture of mammography line-pair phantom and illustration showing human phalanges bones.

Fig. 5.
Demonstration of simple thresholding technique. 5(a) shows a single frame slit image before thresholding and 5(b) shows the image after thresholding.

Fig. 6.
(a) Slit image in EI mode and 6(b) slit image in SPC mode.

Fig. 7.
MTF comparison between EI and SPC mode.

Fig. 8.
Comparison of mammographic line-pair phantom. The first image (a) was taken with EI mode and second (b) with SPC mode.

Fig. 9. Human phalanges bone images for both the modes. The first image (a) was taken with EI mode and second image (b) was taken in SPC mode.

Fig. 10.

Comparison of contrast details in human phalanges bone images. Normalized line profiles over the same selected region for the two different modes indicating the superiority of the SPC mode.

Table I**MAF Specifications**

Scintillator	CsI(Tl)
CsI Thickness	300 micron
Field of view (FOV)	3.6 cm $\times$ 3.6 cm
FOT Ratio	2.88:1
Pixel Width	35 micron
Image Matrix	1024 $\times$ 1024
Light Image Intensifier	GEN 2 Dual MCP
Frame rate	30 fps

Table II**X-ray photon Calculations**

No. of incident x-ray ph /area /exposure	= 137 xph/mm ² /μR
No. of incident x-ray ph /pixel /exposure	= 0.17 xph/pix/μR
Exposure	= 1.37 μR/frame
No. of incident x-ray ph/pix/frame	= 0.23 xph/pix/frame
Absorption efficiency of CsI	= 0.67
No. of absorbed x-ray ph/pix/frame	= 0.15 xph/pix/frame
Slit Length	= 244 pixel
Slit Width	= 1/3 pixel
No. of xph per frame for slit area	= 10.98 xph/frame
Frame used for slit image	= 1200
Frame rate	= 20 fps
Time required	= 60 sec
Total counts expected	= 13176 counts