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Investigation of Noise and Contrast Sensitivity of an Electron Multiplying Charge-Coupled Device (EMCCD) based Cone Beam Micro-CT System

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Abstract

A small animal micro-CT system was built using an EMCCD detectors having complex pre-digitization amplification technology, high-resolution, high-sensitivity and low-noise. Noise in CBCT reconstructed images when using pre-digitization amplification behaves differently than commonly used detectors and warrants a detailed investigation. In this study, noise power and contrast sensitivity were estimated for the newly built system. Noise analysis was performed by scanning a water phantom. Tube voltage was lowered to minimum delivered by the tube (20 kVp and 0.5 mA) and detector gain was varied. Contrast sensitivity was analyzed by using a phantom containing different iodine contrast solutions (20% to 70%) filled in six different tubes. First, we scanned the phantom using various x-ray exposures at 40 kVp while changing the gain to maintain the background air value of the projection images constant. Next, the exposure was varied while the detector gain was maintained constant.

Radial NPS plots show that noise power level increases as gain increases. Contrast sensitivity was analyzed by calculating ratio of signal-to-noise ratios (SNR) for increased gain with those of low constant gain at each exposure. The SNR value at low constant gain was always lower than SNR of high detector gain at all x-ray settings and iodine contrast. The largest increase of SNR approached 1.3 for low contrast feature for an iodine concentration of 20%. Despite an increase in noise level as gain increases, the SNR improvement shows that signal level also increases because of the unique on-chip gain of the detector.

Keywords

EMCCD; Cone Beam CT; Micro-CT; Noise performance; Contrast Sensitivity; Noise Power Spectrum; Signal to Noise ratio

1. INTRODUCTION

Genetically engineered small animals have been instrumental in the research and development of new diagnostic/therapy methods and new drugs for human diseases. These small animals are imaged using many imaging modalities. One of the most used modalities for imaging small animals is micro-CT. Its applications for in vivo studies have increased

rapidly in the past few years^{1, 2, 3}. This is due to micro-CT being able to provide fast, accurate, and economically viable morphological and physiological measurements of scan samples.

Most micro-CT systems have used CCD or CMOS cameras as detectors. These systems have high spatial resolution but poor soft tissue contrast⁴. Additionally, specific to a standard CCD, there is an issue with noise introduced into the output signal due to the output digitizer. The sum of these problems with the selection of detector systems led to development of new systems. The advancement of X-ray detection technology spurred on by this development has led to an improvement in the spatial and temporal resolution that one can obtain in a micro-CT scan. One detector which takes advantage of these improving technologies is an electron multiplying charge coupled device (EMCCD) detector. This device uses an extra register which is placed before the output digitizer. This extra register is capable of amplifying the signal up to three orders of magnitude depending on the EMCCD gain parameter which is selected^{5, 16}. A schematic of the EMCCD can be found in Figure 1. The dependence of the gain of the EMCCD on the parameter setting follows an exponential function^{5, 6, 7, 8}. Through the electron multiplying process of the EMCCD, the effects of readout noise which plague a standard CCD are limited. The introduction of EMCCD detectors has widened the scope of research using micro-CT. To go along with the widening applications, the use and understanding of methods to assess the performance of the detector system is essential^{9, 10, 11, 12}. The fundamental method to describe the performance of any imaging system uses the system's signal and noise characteristics. These properties are well described by metrics in Fourier space such as Noise Power Spectrum (NPS), Modulation Transfer Function (MTF), and Detector Quantum Efficiency (DQE), as well as simpler metrics such as Signal-to-Noise Ratio (SNR) and Contrast-to-Noise Ratio (CNR)^{13, 14, 15}. It is through the assessment of these parameters that a more complete picture of the performance of an EMCCD detector can be gathered. Typically, these metrics are done using a consistent scan phantom so that the results are both reproducible and reportable.

2. MATERIALS AND METHODS

System description

We developed a Micro-CT system which uses an EMCCD (Photonic Sciences, East Sussex UK) for projection acquisition. The EMCCD is coupled to a 3:1 fiber optic taper and a 200 μm CsI phosphor. The EMCCD and optical ensemble was placed in a 3D printed casing (Figure 2, on the left) which ensured the protection of all of the optical components of the detector system. The system assembly was housed in a leaded enclosure, Figure 2. A turn table placed on a xyz motorized stage was used for positioning and object rotation. We used an Oxford XTG UltraBright MicroFocus X-ray source, and a home-made shutter system to control exposure times (Figure 1, on the right). The tube can deliver x-rays with energy between 20 to 80 kVp and power between 10 to 80 W. We used LabVIEW (Laboratory Virtual Instrumentation Engineering Workbench) to modulate the system and acquire images.

The projection images acquired in this study were acquired using 1° increments for a full 360° degrees. CT volumes were reconstructed using a separate program based on standard

FDK reconstruction algorithm. In all cases, the reconstruction plane is selected in the center of the projection image. The region to be imaged is adjusted to the middle of the projection images. The reconstructed region selection avoided dead pixels at the edges in the projection data. The turn table holds the phantom which is rotated to acquire projection images for every 1° rotation over a full 360° rotation.

The total gain of the system was converted into an arbitrary gain number with a 12 bit range, i.e ranging from 0-4095. The detector gain for our study was varied relative to a lower reference arbitrary gain number of 2000. The relative gain was calculated by taking the ratio of air value at the higher gain number to that of the air value at reference gain number. Air value is the detector signal obtained without an object in the beam.

Noise Characteristics

The noise characteristics were determined by imaging and reconstructing a cylindrical phantom filled with water with an external diameter of 10 mm. A picture of the phantom can be found in Figure 3a. The lower portion of the phantom is solid and acts as a holder. The top portion is hollow and is filled with water. Figure 3b, shows the top view of the phantom with an internal diameter of 8 mm. It also shows the hollow region where the water is held and imaged. This phantom was placed on the turntable of the micro-CT. The scan was performed using tube parameters of 20 kV and 10 W, and the EMCCD gain parameter of the detector was varied. Figure 3c shows a projection image which includes only the phantom region and not the holder of the phantom. The region of the phantom that is reconstructed is also indicated. The projection images were reconstructed with a voxel size of $20\ \mu\text{m}$. Figure 3d shows a reconstructed slice of the phantom. The region of interest selected using the red boundary was the region used to determine noise power spectrum plots. A LabVIEW code was used to calculate NPS for each reconstructed data set.

Contrast Sensitivity

We analyzed contrast sensitivity using a phantom containing six tubes filled with iodine contrast concentrations of 20%, 30%, 40%, 50%, 60% and 70% iodine and the remaining was filled with sodium chloride solution. Shown in Figure 4a, a 3D printed phantom of external diameter of 10 mm, containing six small tubes of 2 mm internal diameter was placed along the circumference of the external tube. Figure 4b shows the top view of iodine phantom. The individual tubes seen are filled with contrast solution. Different scans were performed with different current in the tube. The tube currents used for the scans were 0.5, 0.75, 1.25 and 1.5 mA with a constant voltage of 40 kVp. The scans were performed using two methods. First, the gain of the detector was adjusted between scans as the current of the x-ray tube was varied as to maintain the background air value of the projection image constant while keeping the voltage (kVp) constant. Second, the gain of the detector was maintained constant while the current was varied between scans. Figure 4c shows a projection image with all six different concentrations of iodine, and also the region for reconstruction. The axial slices were reconstructed using a voxel size of $20\ \mu\text{m}$. The reconstructed image containing different contrast solutions along with the percentage of iodine is shown in Figure 4d. The SNR was calculated for the sets of images with varied gain and with constant gain. The SNR at each tube in each set of images was calculated by

selecting a region of interest inside the tube and outside the tube, as shown in Figure 5d. SNR was calculated using the formula given in equation 1 below.

$$SNR = \frac{\text{mean}}{\text{std deviation}} = \frac{\mu}{\sigma} \quad (1)$$

where μ is the mean value of the signal in the selected region of interest, representing signal in the region and σ is the standard deviation of the selected background, representing noise in the image. The region of interest inside the tube was used to calculate the signal and the region of interest outside the tube was used to calculate noise for all the sets of reconstructed images. The ratio of SNRs from the scans which varied gain to those which kept gain constant was calculated using equation 2 below.

$$Factor(f) = \frac{SNR \text{ of images with varied gain}}{SNR \text{ of images with constant gain}} \quad (2)$$

3. RESULTS

Noise Characteristics

Reconstructed images of the water phantom are shown in Figure 5. The lowest level of EMCCD gain was used in the slice found in Figure 5a. The highest was used in the slice found in Figure 5f. The high detector gain noise power spectrum plot was greater than the low gain noise power spectrum plot. As the gain of the detector was increased the corresponding noise power plots also increased. NPS for the water phantom shown in Figure 6 shows that the noise level in the reconstructed image increases as the gain of the detector increases. The NPS plot taken at 31.25X relative gain was similar to the NPS plot in the Jaffray, et al⁹ paper on NPS in cone beam CT with a flat panel detector. The value of noise power at different line pairs per millimeter, such as 5, 10, 15, 20 and 25 lp/mm as shown in Figure 7, shows that the noise variance was highest at 10 lp/mm. In Figure 7, the 5 lp/mm plot is slightly higher than the 20 and 25 lp/mm due to the increase in noise at low frequencies.

Contrast Sensitivity

Reconstructed images of iodine contrast solutions are shown in Figure 8. Figures 8a to 8d are acquired at constant voltage, constant detector gain and with tube currents of 1.5 mA, 1.25 mA, 0.75 mA and 0.5 mA, respectively. Figures 8e to 8h were acquired for the above tube settings and varying the gain by maintaining the background air value of the projection image constant. These reconstructed images were not corrected for artifacts, as a result there are ring artifacts in the images.

SNR values for various tube settings and various concentrations of iodine, while keeping the detector gain constant at a setting of 4.29 relative gain, were plotted in Figure 9. The SNR

plot for various tube setting and varying gain of the detector was shown in Figure 10. The factor (f) calculated as the ratio of SNR with gain and without gain was plotted in Figure 11.

The largest increase in SNR was observed for low contrast features and low tube current; for example, with 20% iodine concentration and 0.5 mA, the SNR improvement was greater by a factor of 1.3. The 20% contrast feature had better improvement when compared to the 70% contrast as 20% contrast has higher signal since it is less attenuating than 70% contrast solution. This resulted in increasing the SNR factor to 1.3.

4. DISCUSSION

In this study, we analyzed the noise power performances and contrast sensitivity of a newly built EMCCD based Cone Beam Micro-CT system. The noise power performances of the EMCCD detector was analyzed by plotting noise power spectrum for reconstructed images of a water phantom. The noise power plots shows that the noise in the reconstructed images increases as the gain of the detector is increased. The contrast sensitivity of the detector was analyzed by calculating factor (f) which is the ratio of signal to noise ratio of images with varied gain to constant gain, as shown above. The largest increase in ratio of SNR was seen for 20% contrast at low tube current of 0.5mA. The SNR improvement was 1.3. These results shows that the gain of the detector plays a major role in this system, as the gain of the detector increases the signal in the image is amplified.

The amplification of signal due to gain of the detector plays a vital role and makes EMCCD a very useful detector for imaging small animals. EMCCD detectors have complex pre-digitization amplification technology, high-resolution, high-sensitivity, low-noise and most importantly small pixel size. These properties make EMCCD detectors better than CCD for small animal imaging.

5. CONCLUSION

In this study, one of the first implementations of an EMCCD-based Cone Beam Micro-CT system was presented. We used a phantom with water for evaluating noise characteristics in reconstructed images by plotting NPS and contrast sensitivity was obtained by calculating SNR for various iodine solution concentrations. The improvement of SNR with increased gain for reconstructed images of the iodine phantom was especially significant for low contrast features at low tube current. The unique on-chip gain feature is a substantial benefit allowing the use of the system at very low x-ray exposures.

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Figure 1.
Function block of an EMCCD.

Figure 2. The micro-CT system with EMCCD detector in lead housing. In our system, the tube and detector are stationary, while the scan object rotates on the turn table.

Figure 3.

a) Front view of a 3D printed phantom with holder, with an external diameter of around 10 mm. b) Top view of the phantom which is hollow, with internal diameter of 8 mm. Water fills the hollow part. c) Projection image of the water phantom, also indicating the region of reconstruction. d) Transverse section reconstructed image of water phantom showing the region of interest considered for NPS calculation.

Figure 4.

a) Front view of a 3D printed iodine phantom with 10mm external diameter. b) Top view of iodine phantom with six small tubes of 2mm internal diameter placed along the circumference of the external tube. c) Projection image containing six concentration solutions along with plane of reconstruction. d) Reconstructed image of iodine phantom showing different concentrations and also region of interest for calculating the signal level and noise level in the image.

Figure 5.

a–f are reconstructed images of water phantom acquired at low tube setting and increasing gain of the detector. The relative gain of the detector was increased from lower to higher value, they were- a) 4.29X b) 5.89X c) 8.31X d) 12.44X e) 19.13X and f) 31.25X.

Figure 6.

Plot of noise power versus spatial resolution (lp/mm) for various detector gain setting at 20kVp and 10W.

Figure 7. Plot of Noise Power readout versus different relative gain for various line pairs per millimeter.

Figure 8.

a–d are reconstructed images of contrast solutions for different tube current and constant gain. e–f are images of varied gain with decreasing tube current.

Figure 9. Plot of SNR value versus tube power setting, for various concentration of iodine (20% to 70%) and keeping the gain of the detector constant at a setting of 4.29 relative gain.

Figure 10.

Plot of SNR value versus tube power setting, for various concentration of iodine (20% to 70%) and varying gain of the detector in a relative manner.

Figure 11. Plot of factor (f) versus tube settings, for various iodine contrast agent concentrations (20% to 70%).