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Noise power characteristics of a micro computed tomography system

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Abstract

Objective—The aim of this study was to investigate the noise power properties of a micro computed tomography (micro-CT) system under different operating conditions.

Methods—A commercial micro-CT was utilized in the study that employed a flat panel detector with a 127 μm pixel pitch, and a micro focus x-ray tube. Conical tubes of various diameters were used under different acquisition conditions. Multidimensional noise power spectrums (NPS) were used as a metric to investigate the noise properties of the system. NPS was calculated from the

difference data generated by subtraction of two identical scans. The noise properties with respect to various parameters that include the impact of number of projections, x-ray spectra, mAs, slice location, object diameter, voxel size, geometric magnification (M), back projection filters and reconstruction magnification (M_{Recon}) were studied.

Results—At a same isocentric exposure rate of 270 mR/s, the noise power was much lower for the image reconstructed with 3672 views (122 s) as compared to the 511 views (17 s); while at a fixed isocentric exposure of 4600 mR, the noise power levels were almost similar. Image noise with a 50 kV beam was higher as compared to the 90 kV beam at a same isocentric exposure. Image noise from a 16 mm diameter conical tube was much lower as compared to the 28 mm and 56 mm tubes under identical isocentric exposures. The choice of back-projection filter influence NPS curves in terms of width and amplitudes. M_{Recon} applied during the reconstruction process increased the noise power at lower spatial frequencies while reduced the noise power at higher spatial frequencies. It can be established that for small details corresponding to high spatial frequencies, M_{Recon} can provide an improved SNR. At all spatial frequencies, the in-plane images had lower noise power levels as compared to the z-plane images.

Conclusion—The noise power properties investigated in this study provide important image quality references for refined cone beam system development, optimization and operations.

Keywords

Noise power spectrum; micro-CT; spatial frequency; cone beam

1. Introduction

Preclinical studies involving small animals are increasingly being used as models of human disease. These models have critical importance in the research of disease detection, new drug development and monitoring of treatment responses [1–2]. Micro computed tomography (micro-CT) is a noninvasive, high resolution cone beam imaging modality that has become an active area of research in the preclinical studies with significant potential for clinical applications. Numerous studies have reported the usefulness of micro CTs and cone beam scanners in both the preclinical studies and clinical environment [3–9].

For the development of refined and efficient systems, it is vital to have an in-depth characterization and understanding of the existing systems. As in other medical imaging systems, image quality metrics to characterize the image spatial resolution and noise are necessary to evaluate the micro CT system performance. The modulation transfer function (MTF) and noise power spectrum (NPS) are the well-established principal metrics to quantitatively evaluate the image quality performance of imaging systems by characterizing the spatial resolution and noise power properties. The capability of a medical imaging system to detect fine and low contrast structures can be strongly affected by the magnitude and frequency content of stochastic variation in the image signal. Although, the image variance value in cone beam systems provides a useful insight on the noise magnitude, it does not provide information to characterize the noise correlation induced by the detector blurring and reconstruction algorithm, as well as dependency on various object sizes [10–12]. NPS provides that information and describes the noise components at different spatial

frequencies [13]. In addition, the evaluation of NPS plays a crucial role in system characterization metrics such as detective quantum efficiency (DQE), noise-equivalent quanta (NEQ), signal to noise ratio (SNR) [14–17].

Comprehensive studies related to CT noise properties have been published since 1970s. Riederer et al. showed that the two-dimensional NPS of CT is proportional to $/H(f)^2/f$, where f is the radial spatial frequency and $H(f)$ is the kernel used for back projection filtering [14]. Hanson et al using the signal detection theory showed the detectability of large area, low-contrast objects to be chiefly dependent upon the low-frequency content of the noise power spectral density [15]. Kijewski and Judy derived an expression for the CT NPS images reconstructed with the discrete filtered back projection algorithm and studied the effects from sampling within the projection, rotation angle and reconstructed two-dimensional images [18]. Siewerdsen et al presented a methodological framework for experimental analysis of the NPS of multidimensional images that employed well-known properties of the n -dimensional (n D) Fourier transform and showed cone beam CT as a three-dimensional (3D) case [19]. Baek et al examined the nonstationary noise behavior of a cone beam CT system for local and entire volumes of the scans with FDK reconstruction [20]. In regards to the nonstationarity of CT, Zeng et al developed a method to estimate local NPS using only few fan beam scans [21]. Tang et al used NPS to characterize the imaging performance of a differential phase contrast CT [22]. Benitez et al and Yang et al used NPS extensively for the evaluation of cone beam breast CT scanners for their optimization, refinement and dose reduction [23,24]. To our best knowledge, there are no detailed studies on the noise power properties of cone beam micro-CT scanners that are used for small animal imaging. Using the established theory and methodology, the purpose of this study was to experimentally evaluate the quantum noise properties of an in vivo cone beam micro-CT system for various acquisition and geometric parameters.

2. Methods and Materials

2.2 Experimental Setup

The experiments were performed on a commercial in vivo based micro CT system (Quantum FX, Perkin Elmer, Massachusetts, USA). As shown in Fig.1, the system is equipped with an x-ray tube (L10101, Hamamatsu Photonics, Japan) of varying focal spot size ranging from 5 μ m to 30 μ m depending on the output power of the tube, and a flat panel detector (PaxScan 1313, Varian Medical Systems, California, USA) with a nominal pixel pitch of 127 μ m. The x-ray tube and the detector are placed opposite to each other on a rotating gantry around an animal bed. The source to image distance (SID) distance is fixed at 265 mm while the source to object distance (SOD) varies according to the value of geometric magnification (M). The system operates with three geometric magnification values ($M = SOD / SID$) of 1.72, 2.54 and 5.10. The x-ray tube has a tungsten (W) target and a beryllium (Be) output window with a thickness of 150 μ m. The tube has adjustable tube voltage ranging from 30 to 90 kV and adjustable tube current ranging from 20 to 200 μ A. The tube focal spot size varies from 5 μ m to 30 μ m depending on its output power while the focal spot to the output window distance is 6.8 mm. An inherent filtration of 100 μ m of aluminum (Al) and 60 μ m of copper (Cu) is used for beam hardening. A 2 mm thick bore cover made of acrylic is used that shields the

animal/ sample from falling on the x-ray source and detector. The pixel matrix and frame rate at 1×1 and 2×2 detector binning modes are 1024×1024 , 10 fps and 512×512 , 30 fps, respectively. The system design is symmetric with a total cone angle of 27.56° . The reconstruction and display matrix sizes are 512×512 , while the reconstruction algorithm is FDK filtered back projection. While linear ramp (Ram-Lak) is the intrinsic back projection filter used in the reconstruction process, shepp-logan and cosine filters are also available that can be used during the reconstruction process. Detailed specifications of the system can be found elsewhere [25]. The specifications associated with each field of view (FOV) used in this study are given in Table 1.

The reconstruction magnification (M_{Recon}) employed during the reconstruction process can be regarded as an interpolation process. Once a targeted voxel is projected onto the detector plane along a specific projection ray, an interpolation strategy can be used in the reconstruction process to achieve the value to be back projected onto this voxel depending on the selected field of view [26,27]. The 60 mm and 75 mm FOV scans are identical from the x-ray acquisition standpoint, as they both are acquired with 2×2 detector binning. In the 60 mm FOV, the raw sinogram is interpolated up by a factor of 1.25 to the displayed 512×512 matrix size, and thus a voxel size of $118 \mu\text{m}$ is achieved. Several interpolation techniques can be used for the M_{Recon} , i-e bilinear, nearest neighborhood, spline etc. Due to the vendor's proprietary software, the interpolation scheme applied to the raw sinogram data during the reconstruction process is not known.

In order to investigate the noise properties of the micro-CT, conical plastic tubes with various diameters (70, 56, 28 and 16 mm) filled with de-ionized water were used. The diameters of the conical tubes were 70 mm, 56 mm, 28 mm and 16 mm in order to simulate various sizes of the small animals that are frequently used in the in-vivo imaging. For each scan settings, the same phantom was scanned twice to generate two identical reconstructed data sets for the purpose of image subtraction. Table 2 lists the acquisition parameters used in the study. Two exposure times, 17 s and 122.4 s, were used for scanning the phantoms. With 17 s scan, 511 views are acquired while for the 122.4 s scan, 3672 views are acquired and 918 projections are used for the image reconstruction by accumulating 4 views together. All the conical tubes were placed on an animal bed that had the ability to translate in the horizontal and vertical directions to ensure that the phantom position is exactly at the isocenter and aligned with the axis of rotation.

2.2 Noise Power Spectrum (NPS) Computation

In this study, noise power spectra were calculated for multiple dimensions, including: 3D NPS, 2D NPS and 1D NPS. All these noise power spectra with different dimensions are closely related and are used in the evaluation and characterization of system performance with different purposes under different purposes. Fig.2 provides a block diagram of the 2D NPS computation and is explained as: (a) two identical scans of a conical tube filled with DI water are acquired; (b) the reconstruction algorithm reconstructs sequential slices of a sample in x - y , y - z and z - x planes; (c) extract an identical slice from the two scans; (d) create noise image by subtracting the two slices; (e) select an ROI of size 256×256 on the noise image and correct it for low-frequency trends by subtracting the mean ROI value from it to

form zero mean realizations; (f) the modulus of the 2D FFT is performed and the result is squared as according to Eq.1; (f) the process is repeated for each realization by creating partially overlapped ROIs; (g) the results are averaged; (h) if there is rotational symmetry, convert it to radial NPS (r). For any reconstructed imaging plane, the 2D digital NPS is calculated according to Eq.1 as [19,24]

$$\text{NPS}(u, v) = \frac{1}{N} \sum_{i=1}^N \frac{|\text{FFT}_{2D}[DI_i(x, y) - \overline{DI}_l]|^2}{2} \frac{\Delta x \Delta y}{N_x N_y} \quad (1)$$

where N is the total number of partially overlapped ROIs. N_x, N_y are the number of pixel elements for each ROI, and $\Delta x, \Delta y$ are the pixel sizes in each direction. $DI_i(x, y)$ is the ROI indexed with i and the mean value of this ROI is calculated as $\overline{DI}_l$. The process of subtraction of the CT data from two identical scans eliminates the structural noise in the uniform object scans which is produced by the photon scattering, beam hardening and streak artifacts caused from imperfect projection weighting during reconstruction process [18,22]. A factor of 2 is introduced in the equation to guarantee that the variance in the difference image is identical to that the two original images. For the noise only images, a total of sixteen partially overlapped 256×256 ROIs were used to determine the 2D NPS.

For calculating the 3D NPS, the difference data was generated from the volume data set with a matrix size of 512^3 . The volume data set was divided into 24 overlapped volumes of interest (VOIs) with a size of 256^3 . The 3D NPS for the micro-CT can be written as [24]

$$\text{NPS}(u, v, w) = \frac{1}{N} \sum_{i=1}^N \frac{|\text{FFT}_{3D}[DI_i(x, y, z) - \overline{DI}_l]|^2}{2} \frac{\Delta x \Delta y \Delta z}{N_x N_y N_z} \quad (2)$$

Radial averaging of the 2D NPS (u, v) can be used for the calculation of NPS (r) provided if the 2D NPS (u, v) is radially symmetric. The NPS (r) provides a straightforward graphic comparison of the NPS results with different acquisition and geometric parameters. The 1D NPS (r) is calculated by using the (u, v) coordinate system and its corresponding radial distance $r = (u^2 + v^2)^{1/2}$. All the NPS analyses and data processing were performed on a high level technical computing language software (MATLAB R2014b; The MathWorks, Inc. MA, USA) [28].

3. Results

3.1 Volumetric (3D) NPS

The volumetric NPS was analyzed using Eq. (2), and Fig.3 shows the three orthogonal views of the 3D NPS from the micro CT system acquired with a 40 mm FOV. A 3D missing cone was apparent from the 3D NPS data best shown in x-z plane. The size and shape of this missing cone tells about the insufficient sampling in Fourier space of the cone-beam acquisition [18,21]. Although the 3D NPS is the most intuitive approach for evaluating the noise power properties of micro CT systems, the 2D and 1D NPS are more useful for

making direct comparisons with each of the scan settings. In particular, the 2D NPS calculated from reconstructed in-plane (x-y) plane images has a direct impact on the imaging task since it is the native reconstructed plane that is parallel to the system rotation plane.

3.2 NPS in the In-plane (x-y)

The radial symmetry of the 3D NPS in the in-plane (x-y) can be well observed. It provides a validation for the utilization of radial averaging technique to convert the 2D NPS data into 1D NPS. Fig.4(a) provides a comparison between the horizontal and radial 1D NPS(r) using the 56 mm conical tube filled with water scanned at 90 kV, 200 μ A and 17s in 40 mm FOV. The horizontal 1D NPS was calculated with a total of 14 rows (7 rows above and 7 below, excluding the on-axis row) along the direction of the u axis [29]. The radial NPS is more intuitive, less noisy and smoother as compared to the horizontal or vertical NPS. For the rest of the study, the radial averaging technique was used for the conversion of the 2D NPS data into 1D NPS. Three back projection filters (ramp, cosine and shepp-logan) were used in the reconstruction process to investigate its impact on the image noise. Fig.4(b) illustrates the NPS (r) as a function of spatial frequency for the three back projection filters. Normally, the cosine filter is used for low noise and smoothed images while linear ramp filter is used for high resolution images. It is evident from the result that these two filters are on the extreme ends of noise levels, the noise peaks are smallest when using the cosine filter and largest when using the ramp filter. It is also clear that with the ramp filter, the width of NPS (r) is wider as compared to the two other filters. Since, linear ramp filter is the system's intrinsic back projection filter, the remaining study was performed with this filter for the reconstruction of phantom images.

The influence of number of projections and exposure time on the in-plane NPS was evaluated using a 56 mm conical tube scanned at 90 kV beam. From Fig.5(a), one can observe that the image reconstructed with 918 projections (3672 views) has less noise power as compared to image reconstructed with 511 projections, which produces a superior SNR. The isocentric entrance exposure rate and current for the 918 (122 s) and 511 (17 s) projection scans were fixed at 270 mR/s and 200 μ A. Thus, more photons were used for the 918 (122 s) projections scan and as expected, the image noise power is inversely proportional to the total radiation dose per scan. At a fixed isocentric entrance exposure level of 4600 mR, noise power peaks for the 17 s, 200 μ A and 122 s, 33 μ A scans were at 1755 HU^2mm^2 and 2190 HU^2mm^2 as shown in Fig.5(b). In this case, the isocentric entrance exposure rate for the 918 (122 s) projections scan was 37.5 mR/s, which is 7 times less than the 511 (17 s) projections scan value of 270 mR/s. It should be mentioned that the scan acquired with 918 (122 s) projections at high current values offers superior SNR and better spatial resolution as the number of projections and photons are directly related to the image quality.

3.3 Impact of Object Diameter and Voxel size

Next, the impact of object diameter, voxel size and magnification were investigated and the corresponding in-plane NPS results are shown in Fig.6. The conical tubes were all scanned at a maximum exposure level of 3.4 mAs (200 μ A $\times$ 17 s) with 90 kV. The peak noise power increased as the diameter (D) and geometric magnification (M) increased. With the

geometric magnification increasing, the resulting voxel size decreases and thus the average number of photons per voxel also decreases. This tends to increase the standard deviation, and a large standard deviation value results in a large noise power in the frequency domain. Similarly, as the sample size is reduced, more photons can penetrate the sample for the image formation and the signal detected on a flat panel detector has less relative noise.

These two trends are further explained in Fig.7(a), where peak noise powers are plotted against different values of M and D and fitted according to the second order polynomial equation $ax^2 + bx + c$, where a, b, c are constants and $\times$ represents M and D for the two curves. Although the NPS varies with different diameters and voxel sizes, the shape of the curves remain the same due to the employment of the same back projection filter during the reconstruction process. Due to the same reason, the noise power peaks occur at similar spatial frequencies for different diameters of conical tubes in a same FOV.

Fig.7(b) shows the image standard deviation (0-D NPS) plotted against the varying x-ray tube exposure (mAs) for three different voxel sizes of 118 μm , 80 μm and 40 μm . The 56 mm conical tube was scanned at 90 kV and 17 s with a tube current ranging from 20 μA to 200 μA , which corresponded to an x-ray tube exposure ranging from 0.34 to 3.4 mAs. The average standard deviation calculated from five ROIs was plotted against the tube exposure and fitted to the equation $\sigma_{\text{Total}}^2 = AE^{-1} + \sigma_{\text{System}}^2$, where σ_{Total}^2 is the measured noise in HU, E is the exposure in mAs, and A, σ_{System}^2 are constants. Noise level decreased by k/mAs as mAs increased, where k is constant of proportionality. One can see that the contribution of the system noise is not significantly large in the total noise and this analysis is consistent with the assumption that the overall noise is dominated by the quantum noise. In general, the instabilities of the rotating gantry, translational bed and system electronics are the major contributors to a large system noise. It is also evident that noise significantly drops from 20 μA to 160 μA ; further increase in the exposure will unlikely decrease the noise considerably.

3.4 Impact of Reconstruction Magnification (M_{Recon})

There are several FOVs where the reconstruction magnification (M_{Recon}) was used to make the voxel size smaller. From Fig.8, one can see that the application of interpolation in the reconstruction process increases the peak noise powers. For example, the peak noise power levels were at 2420 HU^2mm^2 and 1882 HU^2mm^2 in the 20 mm and 24 mm FOVs, respectively. This is an expected result; as smaller voxel sizes lead to increased noise. In the 20 mm FOV, $M_{\text{Recon}} = 1.25$ was applied to the raw sinogram to achieve a voxel size of 40 μm . Due to the vendor's proprietary software, it is not known which interpolation scheme was applied to the raw sinogram during the reconstruction process. One can see from the figures that not only the system has an excellent circular symmetry, it is less prone to noise aliasing, since the noise powers at the starting and ending spatial frequencies are approximately the same. It can be predicated that for smaller imaging details corresponding to large spatial frequencies, the FOVs where M_{Recon} was used, can give an improved SNR as compared to the FOVs where no M_{Recon} was applied.

3.5 Impact of Different X-ray Spectra

The impact of various x-ray spectra on NPS was investigated with the 40 μm in-plane images. First, simulated x-ray spectra of 50 kV, 70 kV and 90 kV were generated for a tungsten (W) anode with 0.06 mm Cu, 0.1 mm Al and 2 mm PMMA filter objects [30]. The output photon numbers were normalized to 1 as shown in Fig.9(a). Next, the 28 mm conical tube was scanned with 50 kV (200 μA), 70 kV (100 μA), and 90 kV (65 μA) at 17 s under a constant isocentric exposure of 4600 mR. The NPS results are shown in Fig.9(b) and one can see that the noise with 90 kV beam has lower power than the 70 kV and 50 kV beams. It is clear from the simulated spectra that there are high energy characteristic x-ray photons present in a 90 kV beam which has a direct impact on the NPS. With a 50 kV beam passing through the water phantom, more photons are absorbed and few photons depart for further generation of the digital signal as compared to a 90 kV beam. Furthermore, the low energy x-ray photons reaching the flat panel detector will certainly convert to less visible light photons that in turn will give a smaller digital signal resulting in more noise. The line profiles of CT numbers shown in Fig.10 also suggests that the 90 kV beam has the least fluctuations among the neighboring pixels. Thus, the result is small standard deviation (σ) and less noise.

3.6 NPS in z-planes and z-distance

NPS in z-direction (through-plane) was measured with a 40 μm voxel size using a 56 mm conical tube. As contrary to the in-plane (x-y), the z-planes are not rotationally symmetric and radial averaging cannot be performed on the 2D NPS curves. Thus, the NPS curves in z-direction were fitted to the sixth order polynomial equation for fair comparison. At lower spatial frequencies, the x-y plane demonstrated much lower noise power levels as compared to the z planes as shown in Fig.11(a). These properties are associated with the reconstruction algorithm employed by the system. With the increase in spatial frequency, the NPS curves in the three planes exhibit similar noise powers. For example, at 10 lp/mm, the noise powers for the x-z, y-z and x-y planes were 385 HU^2mm^2 , 331 HU^2mm^2 and 233 HU^2mm^2 . The noise power peak amplitudes were plotted in the x-y plane along the z-direction distance as seen in Fig.11(b). One can see that the peak noise powers follow a sinusoidal pattern along the z-distance. It has maxima points at $z = 2 \text{ mm}$, $z = 24 \text{ mm}$ and $z = 38 \text{ mm}$, and a well-defined minimum at $z = 20 \text{ mm}$. A likely explanation for this pattern is that the FPD slightly vibrates locally during the scan if it is not well aligned. This phenomenon is explained in detail in ref [23].

4. Discussion and Conclusion

Micro-CT imaging modality initially developed for industrial applications, is routinely used in small animal studies, but it has proven to be a valuable tool in various clinical applications such as in the characterization of breast cancer by performing micro CT imaging on human breast biopsy specimen, assessment of the 3D bone microstructure and bone mineral density (BMD) performed on human legs or on bone biopsy specimens and the imaging of tumor perfusion and angiogenesis. With the development of high power micro focus x-ray sources and smart FPDs, high resolution imaging (fraction of microns) of human subjects is not far away.

The objective of this study was to investigate the quantum noise characteristics of an in-vivo cone beam micro-CT system through multi-dimensional NPS measurements. For a cone beam system, in-plane (x-y) is the native reconstructed imaging plane as it is parallel to the rotational plane of the system and much of the emphasis in this paper was given to this plane. The effects on image noise power from different system parameters were investigated utilizing conical tubes of various diameters filled with deionized (DI) water. The volumetric results revealed that the cone-beam micro-CT noise power spectrum is rotationally symmetric within the in-plane, thus radial averaging can be used to convert the 2-D $NPS(u, v)$ to 1-D $NPS(r)$. The radial NPS is more intuitive, less noisy and smoother as compared to the horizontal or vertical NPS. Despite its symmetry in the x-y plane, NPS reliance on the z-direction is different. Theoretically, noise power must increase as the z distance increases; it does not follow this trend and has a semi-sinusoidal shape. In spite of having different amplitudes throughout the z-direction, the knowledge of the spatial frequency content of noise can be fundamental in developing noise reduction algorithms. It is common knowledge that the noise decreases as the total exposure increases. Thus, from the parameters investigated above, the pixel size, the exposure level, the kV and the number of projection acquired interfere with the intensity of the beam. The image reconstructed with 3672 views (122.4 s) resulted in much lower noise as compared to the image reconstructed with 511 views (17 s) at a same isocentric exposure rate (mR/s). With a fixed isocentric exposure (mR), the noise powers were comparable. For thicker samples, the 918 projection (3672 views) scan can be very useful due to the refinement in the angular sampling from 0.70° (511 projections) to 0.40° (918 projections) as well as the reduction in the scanning speed, which increases the number of views acquired and results in superior SNR. Our analysis confirmed that the total noise is dominated by the quantum noise and the contribution of the system noise is minimal. The image noise also depends on the x-ray spectrum and the object size when the entrance exposure levels were kept constant. For example, image noise with a 16 mm diameter conical tube was much lower than that of 56 mm diameter tube under identical isocentric exposure levels. The image noise with a 50 kV beam was higher as compared to the 90 kV beam because not only less x-ray photons reach the detector but those that reach, create fewer visible photons which creates more noise.

The process of reconstruction magnification M_{Recon} yields a smaller voxel size (particularly thinner slices), and the result is higher noise power peaks. For example, when M_{Recon} was applied in $M = 1.72$ (60 mm FOV), the resulting size was $118 \mu m$ and the NPS peaks increased from $395 \text{ HU}^2\text{mm}^2$ (75 mm FOV) to 471 mm^2 , respectively. Although, the total exposure per area remains constant, yet the smaller the pixel size is, the less energy it has to integrate per pixel which results in large noise peaks. However, the application of M_{Recon} (interpolation) reduces the noise at higher spatial frequencies corresponding to smaller imaging details which may improve the SNR provided that the spatial resolution remains the same. Interpolation schemes are routinely used in modern day CT scanners to produce thinner slices. In this study, we found that the width of NPS only changes for two cases namely, the filter type and the voxel size. In case of filter type, the width is expected to vary because different filters have different frequency response to the input signal. For ramp filter, the width and noise power peaks are largest because this filter enhances the high-frequency components of the signal, including the noise levels. At all spatial frequencies, the in-plane

(x-y) images had lower noise power levels as compared to the z-plane images with a significant difference at low spatial frequencies. The following parameters whose impact on noise properties were investigated must be selected prior to the scan are the pixel size of the FPD, exposure level, kV value, geometric magnification (M), object size and the number of projections acquired during the scan. The parameters that can be selected after the scan is performed are the M_{Recon} , choice of back projection filter and the reconstruction algorithm.

We were limited by the options available from the manufacturer in utilizing various cone angles, different reconstructed matrix sizes and reconstruction algorithms. Quantum noise along with the anatomical noise represents the total noise of a system. Uniform objects were utilized to characterize the quantum noise properties of the micro CT system and the structural noise only refers to the cupping artifacts. Future studies will be conducted to investigate the anatomical noise properties of small animals with tumors in an effort towards the complete noise property description of micro CT. We further intend to utilize reconstruction algorithms based on iterative methods and sparse theory and compare their image quality in terms of spatial resolution and noise with the existing micro CT data. In conclusion, the intention of this study was to illustrate the quantum noise properties of an in-vivo cone beam micro-CT system using uniform conical tubes. The investigation of noise power properties as well as spatial resolution properties provides guidelines for refined cone beam system development, optimization, operation and data reconstruction.

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Figure 1.
Schematics of the micro CT system illustrating a micro focus x-ray source and a flat panel detector attached to a rotating gantry.

Figure 2.

A functional block diagram illustrating the essential steps for the computation of 2D NPS. Noise image is the difference of two slices extracted from two scans.

Figure 3.
Volumetric NPS results in three orthogonal planes of micro-CT scan acquired with 90 kV, 200 μ A and 17s

Figure 4.

(a) Comparison between the horizontal and radial NPS; (b) NPS results with three back projection filters used in the reconstruction process.

(a)

(b)

Figure 5.

Comparison of the NPS results with different number of projections and exposure times (a)

NPS results with a constant entrance exposure rate of 270 mR/s for the two scan protocols.

(b) NPS results at a constant entrance exposure level of 4600 mR for the two scan protocols.

Figure 6.

(a) NPS results with 70 mm and 56 mm conical tubes in 60 mm FOV with a 118 μm voxel size (b) NPS results with 56 mm, 28 mm and 16 mm conical tubes in 20 mm FOV with a 40 μm voxel size.

(a)

(b)

Figure 7.

(a) Noise power peaks plotted against geometric magnifications (M) and object diameters;
 (b) The standard deviation plotted against the exposure (mAs) for three different voxel sizes of 118 μm , 80 μm and 40 μm using a 56 mm conical tube.

(a)

(b)

Figure 8.

The impact of interpolation (M_{Recon}) on in-plane NPS. (a) NPS results comparison between 73mm (148 μm) and 60mm (118 μm) FOVs using a 70mm conical tube; (b) NPS result comparison between 24mm (50 μm) and 20mm (40 μm) FOVs using a 56mm conical tube.

Figure 9.

(a) Simulated x-ray spectra for three x-ray beams generated with the parameters specific to the x-ray source of micro CT; (b) NPS results for the three different x-ray beams in the 20 mm (40 μ m) FOV.

Figure 10.
Line profiles of CT numbers with three x-ray beams.

Figure 11.

(a) NPS results in three reconstructed planes of the system. Polynomial curve fitting was applied on the z-plane NPS results; (b) Noise power peak amplitude plotted in the x-y plane along the z-direction distance in 40 mm FOV.

Table 1

The parameters associated with the micro-CT system

FOV (mm)	M	M _{Recon}	Binning	x-y pixel size (μm)	slice thickness (μm)	Nyquist freq (lp/mm)
75	1.72	1	2×2	148	148	3.37
60	1.72	1.25	2×2	118	118	4.23
40	2.54	1.25	2×2	80	80	6.25
24	5.10	1	2×2	50	50	10
20	5.10	1.25	2×2	40	40	12.5

Table 2

The acquisition parameters associated with the micro-CT system

Object Diameter (mm)	70, 56, 28, 16		
X-ray spectrum (kVp)	50	70	90
X-ray current (μ A)	200	100	33–200
Number of ROIs per slice	16		
Number of VOIs per volume	24		
Exposure Time,	17s, 512 projections		
Number of Projections	122.5s, 918 projections		