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Investigation of spatial resolution characteristics of an in vivo micro computed tomography system

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Abstract

The spatial resolution characteristics of an in vivo micro computed tomography (CT) system was investigated in the in-plane (x-y), cross plane (z) and projection imaging modes. The micro CT system utilized in this study employs a flat panel detector with a 127 μm pixel pitch, a micro focus x-ray tube with a focal spot size ranging from 5-30 μm , and accommodates three geometric magnifications (M) of 1.72, 2.54 and 5.10. The in-plane modulation transfer function (MTF) curves were measured as a function of the number of projections, geometric magnification (M), detector binning and reconstruction magnification (M_{Recon}). The in plane cutoff frequency (10% MTF) ranged from 2.31 lp/mm (M=1.72, 2 $\times$ 2 binning) to 12.56 lp/mm (M=5.10, 1 $\times$ 1 binning) and a bar pattern phantom validated those measurements. A slight degradation in the spatial resolution was observed when comparing the image reconstruction with 511 and 918 projections, whose effect was visible at the lower frequencies. Small value of M_{Recon} has little or no impact on the in-plane spatial resolution owing to a stable system. Large value of M_{Recon} has implications on the spatial resolution and it was evident when comparing the bar pattern images reconstructed with M_{Recon} =1.25 and 2.5. The cross plane MTF curves showed that the spatial resolution increased as the slice thickness decreased. The cutoff frequencies in the projection imaging mode yielded slightly higher values as compared to the in-plane and cross plane modes at all the geometric magnifications (M). At M=5.10, the cutoff resolution of the projection and cross plane on an ultra-high contrast resolution bar chip phantom were 14.9 lp/mm and 13-13.5 lp/mm. Due to the finite focal spot size of the x-ray tube, the detector blur and the reconstruction kernel functions, the system's spatial resolution does not reach the limiting spatial resolution as defined by the Nyquist's detector criteria with an ideal point source. The geometric magnification employed in the micro CTs provide a tradeoff between field of view and spatial resolution for a wide range of applications.

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Keywords

Spatial resolution; micro CT; MTF; geometric magnification; x-ray; interpolation

1. Introduction

With the evolution of computed tomography (CT), there have been major advancements in the technology which has led to the development of refined systems. Micro computed tomography (micro-CT) is a noninvasive, high resolution imaging modality that is becoming an active area of research for small animal imaging in preclinical studies with a significant potential for the clinical applications. In vivo imaging studies with the micro-CT are largely driven by the motivation of new drug development, tumor detection and monitoring, and investigations on the effectiveness of drugs in the disease treatment [1, 4], while in vitro imaging methods are used for the analysis of tissue specimens [5, 6]. Numerous studies have reported the value of the micro-CT in the clinical world regarding the imaging of human breast specimens [7-9]. There are many steps involved during the image acquisition and image reconstruction. Initially, the detector elements sample the emerging x-ray data from an object. The reconstruction algorithm back projects the projection data to a matrix of discrete pitch. The reconstruction data is then interpolated up or down to a displayed matrix, depending on the selected displayed field of view [10]. The reconstructed micro-CT image matrix is a group of discrete samples of the continuous signal produced from the back projection of the filtered projection images [11, 12]. Spatial resolution is one of the most important parameters and quality assurance tests for the micro-CT should include some criteria for its assessment. Previous studies have presented the modulation transfer functions (MTFs) of the reconstructed images of the cone-beam CT systems [10, 14-18]. The goal of this study was to investigate the spatial resolution properties of a cone beam in-vivo micro-CT system. The spatial resolution of the system was evaluated under different acquisition conditions in the in-plane (x-y), cross plane (z-direction) and projection imaging modes using several different phantoms. Three different phantoms were used for the quantitative assessment, while two different phantoms were used for the qualitative assessment of the spatial resolution in the three imaging planes.

2. Materials and Methods

2.1 Micro CT system

A commercial micro-CT system (Quantum FX, Perkin Elmer, Massachusetts, USA) was used in this study. It is equipped with a micro focus x-ray tube (L10101, Hamamatsu Photonics, Japan) and a flat panel detector (PaxScan 1313, Varian Medical Systems, California, USA). The x-ray tube and the detector are placed opposite to each other on a rotating gantry around an animal bed at a distance of 265 mm. The system operates at three geometric magnifications (M) of 1.72, 2.54 and 5.10. The x-ray tube has a tungsten target, beryllium output window with a thickness of 150 μm , adjustable tube voltage ranging from 30 to 90 kV and adjustable tube current ranging from 20 to 200 μA . The focal spot size of the x-ray tube varies from 5 μm to 30 μm depending on its output power while the focal spot to the output window distance is 6.8 mm. An inherent filtration of 100 μm of aluminum (Al)

and 60 μm of copper (Cu) is used for beam hardening [19]. The x-ray detector is a $130 \times 130 \text{ mm}^2$ amorphous silicon flat panel with a 127 μm pixel pitch and 14 bit digital output. 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. During the CT scan with 1×1 binning, a partial read out of 512×512 is made and thus making the frame rate of 20 fps. The system design is symmetric with a total cone angle of 27.56° ($2 \times \tan^{-1} 65\text{mm} / 265\text{mm}$). The reconstruction and display matrix sizes are 512×512 , while the reconstruction algorithm is FDK filtered back projection [20], with a standard Ram-Lak (ramp) filter. The system has an isotropic pixel size for every field of view (FOV) and the complete configuration associated with each FOV is given in Table 1. For the 2 min (122.4s) scan in 2×2 binning, 3672 frames are acquired and 918 projections are used for the CT reconstruction by accumulating 4 frames together. For the 3 min (180.4s) scan in 1×1 binning, 3608 frames are acquired and 902 projections are used for the CT reconstruction by accumulating 4 frames together. For the 2 min and 3 min scans, the acquisition angle for each accumulated projection is 0.40° , while for the 17 s and 26 s scans, it is 0.70° respectively.

The reconstruction magnification (M_{Recon}) employed in several FOVs can be thought of 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 to determine the value to be back projected onto this voxel depending on the selected field of view [10, 21]. For example, in the 75mm FOV, no interpolation scheme is applied to the raw sinogram as the projection data and the displayed matrix sizes are the same. In the 60mm FOV, the raw sinogram is interpolated up by a factor of 1.25 to the displayed 512×512 matrix size and voxel size of $118 \mu\text{m}$ is achieved. The 60mm and 75mm scans are identical from the x-ray acquisition standpoint, they both are acquired at 2×2 detector binning. There are several interpolation techniques that 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 this study we used 90 kV as the x-ray tube voltage and $160 \mu\text{A}$ as the tube current. The focal spot size corresponding to 14.4 W ($90 \text{ kV} \times 160 \mu\text{A}$) was approximately $21 \mu\text{m}$ according to the chart provided by the x-ray tube manufacturer.

2.2 Phantoms and Performance Evaluation

2.2.1 Phantoms for Quantitative Assessment of the Spatial Resolution—The modulation transfer function (MTF) is a well-established metric for the quantitative characterization of the spatial response of an imaging system. A custom designed tungsten wire phantom was used to measure the in-plane MTF curves for the spatial resolution measurements over a range of frequencies across three geometric magnifications of $M=1.72$, 2.54 and 5.10. As shown in Fig. 1(a), a tungsten wire was inserted in small sewing needles which were then implanted in a 25.6 mm thick Ethylene-Vinyl Acetate (EVA) foam. The needles were stretched so that the wire had enough tension to remain straight. The EVA foam provided the flexibility to adjust the needles. Two tungsten wires of $14 \mu\text{m}$ and $18 \mu\text{m}$ diameters were used to accommodate all the geometric magnifications (M) for the MTF measurements. The $18 \mu\text{m}$ tungsten wire was used in $M = 1.72$ and 2.54 scans, while the $14 \mu\text{m}$ tungsten wire was used in $M = 5.10$ scans. The reconstructed in-plane image of the

tungsten wire can be regarded as the point spread function (PSF), and the in-plane MTF was calculated from 5 adjacent slices whose line spread functions (LSFs) were averaged to reduce the noise. The base of the LSF was normalized to zero and then MTF was determined by computing the one dimensional fast Fourier transformation of the normalized LSF as [15,16, 22]

$$MTF(f) = |FFT(LSF(x))| \quad (1)$$

The FOVs where M_{Recon} was applied during the reconstruction process resulted in a smaller voxel size and in particular, thinner slice thickness. Although the pixel size in x-y plane becomes smaller, the in-plane (x-y) spatial resolution should remain the same as compared to the FOVs where no M_{Recon} was applied during the reconstruction. Once a projection dataset was collected, the best spatial resolution is what was collected from the detector. If the in-plane spatial resolution is degrading too much after the application of M_{Recon} , that particular FOV may not be regarded as stable. A micro-CT slice sensitivity profile (SSP) phantom (QRM, GmbH, Möhrendorf, Germany) was used to determine the cross-plane (z-direction) spatial resolution of the system. This phantom has been used in previous studies to measure the cross-plane resolution [23, 24]. The resin made phantom has an inserted gold platelet of 1 mm diameter and 10 μ m thickness as shown in Fig. 1(b). In standard imaging science dialect, SSP is the LSF along the z-direction of the CT scanner [23]. Similar to most of the in vivo based micro CT systems, this system also come with projection or live mode imaging for different purposes. Therefore, for the assessment of projection imaging spatial resolution, a slit camera phantom (07-624-1000, Nuclear Associates, USA) with 10 μ m width was used to collimate the x-ray beam into a line input for measurement of the MTF [25, 26]. All the MTF analysis were performed on a high level technical computing language software (MATLAB R2014b; The MathWorks, Inc. MA, USA).

2.2.2 Phantoms for Qualitative Assessment of the Spatial Resolution—A micro-CT resolution bar pattern phantom (QRM, GmbH, Möhrendorf, Germany) was used to qualitatively evaluate the in-plane spatial resolution of the system. The phantom consists of two 5×5 mm² silicon chips with bar and point patterns having diameters ranging from 5-150 μ m line/point thickness as shown in Fig.1 (c). An ultra-high contrast resolution bar chip phantom (016B, CIRS, Virginia, USA) was utilized to further compare the projection and the cross-plane imaging spatial resolutions. The phantom has a 17.5 μ m thick gold-nickel alloy bar pattern which has 18 segments ranging from 5 lp/mm to 28 lp/mm. All the phantoms 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.

3. Results

3.1 In-plane Spatial Resolution Evaluation

The projection and reconstructed images of the tungsten wire phantom are shown in Fig.2 (a) and (b), respectively. From the projection image, one can see that the inserted tungsten wire is straight and the embedded needles on the EVA foam can be adjusted easily to create a tilt for the measurement of the over-sampled LSF. As shown in Fig.2 (b), an ROI around

the reconstructed tungsten wire point was integrated in the mentioned direction to produce the LSF for the MTF calculation.

First, the influence of the number of projections on the in-plane spatial resolution was evaluated. From Fig.3, one can observe that the in-plane MTF curve for 918 projections is slightly better as compared to 511 projections. Although the effect can be prominently seen for the lower spatial frequencies on the MTF curve, it is not that obvious at high spatial frequencies and thus the cutoff frequencies almost remain the same. For the rest of this study, 902 and 918 projections were used for the 1×1 and 2×2 binning modes.

The effect of geometric magnification (M) on the spatial resolution was then investigated. From Fig.4, one can observe that the cutoff frequencies (10% MTF) and thus the spatial resolution increases with the three investigated geometric magnifications (M). The cutoff frequencies at 2×2 detector binning for 75mm, 40mm and 24mm FOVs were 2.31 lp/mm, 4.25 lp/mm and 7.25 lp/mm, which correspond to a detectable detail of 216µm, 122µm and 69µm, respectively. The cutoff frequencies at 1×1 detector binning for the 30mm and 10mm FOVs were 4.85 lp/mm and 12.56 lp/mm, respectively, which correspond to a detectable detail of 98 µm and 39 µm. Obviously, the detector binning has a direct impact on the spatial resolution and 1×1 binning resulted in higher cutoff frequencies as compared to the 2×2 binning. The improvement in spatial resolution with 1×1 binning comes at the expense of noise [27]. The cutoff frequencies for the three geometric magnifications in two data binning modes corresponding to all the FOVs are given in Table 2.

The application of $M_{\text{Recon}} = 1.25$ does not have a major impact on the in-plane spatial resolution owing to a stable system. This result is obvious when comparing the cutoff frequencies on the MTF curves for 75 mm and 60 mm FOVs. For the 75 mm FOV when no M_{Recon} was applied, the cutoff frequency was 2.31 lp/mm; whereas for the 60 mm FOV when $M_{\text{Recon}} = 1.25$ was applied, the cutoff frequency was 2.32 lp/mm. Similar results can be seen when comparing the FOVs of 24 mm and 20 mm where the cutoff frequencies were 7.25 lp/mm and 7.20 lp/mm.

For the validation of the MTF results, the in-plane spatial resolution was then qualitatively measured by visual inspection of the micro-CT resolution bar pattern phantom. The in-plane images of the bar pattern phantom for the 5 mm, 10 mm, 30 mm and 40 mm FOVs are given in Fig. 5, which provides a visual comparison of the spatial resolution to the measured MTF curves. For the 5 mm and 10 mm FOVs, the 30 µm bar lines can be distinguished whereas the 100 µm and 150 µm bar lines are distinguishable in 30 mm and 40 mm FOVs respectively.

Although, the 30 µm bar lines are not completely differentiated, the dotted line contrast profiles suggest that the 30 µm bar lines in the 10 mm FOV yield a higher contrast as compared to the bar lines in the 5 mm FOV. For $M_{\text{Recon}} = 1.25$, one data point is introduced for every four pixels, while for $M_{\text{Recon}} = 2.5$, three data points are introduced for every two pixels to fit the cropped raw sinogram data into the 512×512 displayed matrix. The more data points introduced into the matrix, the more sharpness may be lost in the in-plane

images, whose effect could be visible on the smaller objects. This argument can be seen when comparing the 5 mm and 10 mm FOV dotted line contrast profile.

3.2 Cross-plane (Z-direction) Spatial Resolution Evaluation

The MTF in the z-direction was next measured using the micro-CT SSP phantom. The MTF curves indicate that the cutoff frequency increases with a decrease in the slice thickness. From the MTF curves in Fig.6, the cutoff frequencies (10% MTF) for the 10 μm , 40 μm slice thicknesses corresponding to the 5 mm and 20 mm FOVs were 12.95 lp/mm, 7.52 lp/mm.

The cross-plane spatial resolution was then qualitatively measured by using the ultra-high contrast resolution bar chip phantom. The phantom images for the 5 mm and 20 mm FOVs are given in Fig.7. The resolution on the phantom for the 5 mm and 20 mm FOVs was in between 13-13.5 lp/mm and 8 lp/mm respectively. The contrast profile corresponding to the dotted line on the 20 mm FOV image suggests that the 8 lp/mm bar lines are at most detected.

3.3 Projection Spatial Resolution Evaluation

In vivo based micro-CT systems typically provide a projection imaging mode for various 2-D imaging purposes, thus the spatial resolution for the projection imaging mode was evaluated. As shown in Fig.8, the cutoff frequency at $M=5.10$ with 1×1 data binning was 14.11 lp/mm while at $M=5.10$ with 2×2 binning it is 7.95 lp/mm. The projection imaging spatial resolution is slightly better than the in-plane and the z-direction spatial resolution of the micro-CT system. This result was further elaborated with the ultra-high contrast resolution bar chip phantom images shown in Fig.9. The projection image of the bar chip phantom acquired at $M=5.10$ reaches 14.9 lp/mm; whereas the z-direction (x-z) reconstructed image ranges between 12-13 lp/mm. Similarly from Fig.9, the bar lines corresponding to 5 lp/mm are completely resolved in the projection image acquired at $M=1.72$, while the reconstructed cross-plane (x-z) image cannot resolve those bar lines.

4. Discussion and Conclusion

The objective of this study was to investigate the spatial resolution characteristics of an in-vivo micro-CT (Quantum FX, Perkin Elmer) system through measurements in the in-plane (x-y), cross-plane (z) and the projection imaging modes using several imaging phantoms. The impact of the geometric magnification (M), field of views (FOVs) and interpolation (M_{Recon}) on the spatial resolution were studied. The in-plane spatial resolution was measured using a custom made tungsten wire phantom. The phantom was easy to fabricate and the EVA foam provided the flexibility to create necessary tension in the tungsten wires to remain straight without prone to breakage. The phantom can be used for routine evaluation of the performance of cone beam micro-CT systems. In the cone beam modality, in-plane (x-y) is the native reconstructed plane and much of the emphasis in this paper was given to this plane. The MTF curves indicated that the cutoff frequencies (10% MTF) ranged from 2.31 lp/mm ($M=1.72$, 2×2 data binning) to 12.56 lp/mm ($M=5.10$, 1×1 data binning). The MTF results for the in-plane image reconstructed with 918 projections were

better as compared to the image reconstructed with 511 projections. The improvement was seen prominently at the lower spatial frequencies on the MTF curves. For example, in the 60 mm FOV ($M=1.72$, 2×2 data binning), the spatial frequencies corresponding to a 0.7 MTF value for 918 and 511 projections were 0.80 lp/mm and 0.70 lp/mm. These frequencies correspond to detectable details of 512 μm and 571 μm , respectively. This improvement is likely due to the refinement in the angular sampling from 0.70° (511 projections) to 0.40° (918 projections) and reduction in the scanning speed which increases the number of views acquired and the result is in better signal to noise ratios (SNR). The micro-CT resolution bar pattern phantom validated the measured MTF results and the smallest visible lines on the bar pattern were in the 30 μm range. The influence of the reconstruction magnification (M_{Recon}) was also investigated, and our results indicated that $M_{\text{Recon}}=1.25$ has little or no impact on the in-plane spatial resolution, owing to a stable system. For example, for the 24mm FOV where no M_{Recon} was applied, the cutoff frequency was 7.25 lp/mm, while 7.20 lp/mm when $M_{\text{Recon}}=1.25$ was applied. Although, the application of M_{Recon} reduces slice the thickness, large values of M_{Recon} to fit a portion of the raw sinogram data into the 512×512 display/reconstruction matrix has some implications on the in-plane spatial resolution. This notion was evident when comparing the micro-CT resolution bar pattern images for the 5 mm and 10 mm FOVs as the latter yielded in a better contrast profile for the 30 μm lines. The z-direction MTF curves measured with the micro-CT SSP phantom revealed that the spatial resolution increases with a decrease in the slice thickness. The projection imaging MTF measured with the slit camera phantom yielded slightly higher cutoff frequencies as compared to the results of the in-plane and cross-plane. The cutoff frequency for projection imaging at $M=5.10$, 1×1 data binning was 13.75 lp/mm. As compared to the cross-plane images, more line pairs on the projection images were distinguishable on the ultra-high contrast resolution bar chip phantom, which validated the MTF curve calculations. At $M=5.10$, the cutoff resolution for projection and z-plane on the bar chip phantom images were 13.9-14.9 lp/mm and 12-12.5 lp/mm range respectively. There are several reasons that the projection spatial resolution is better than the in-plane. First, the reconstructed micro-CT image matrix is a group of discrete samples of the continuous signal produced from the back projection of the filtered projection images. Second, the application of different kinds of filters (low pass, ramp, cosine, etc) in the reconstruction process will discard additional samples to make the image matrix data smoother, sharper, etc. As compared to the MTF curves, the resolution on the bar pattern images is little deeper due to the fact that 10% MTF may not an absolute baseline limit for the cutoff frequency estimation. Several studies have used 5% baseline limit for the estimation of cutoff frequency, we chose 10% limit for consistency with majority of the studies. Curve fitting that was used for the generation of smooth ESF and LSF nominally impacts the cutoff frequency range. There are several reasons why the system's spatial resolution does not reach the limiting spatial resolution as defined by the Nyquist's criteria of a detector with an ideal point source: the finite focal spot of the x-ray tube, the detector blurring of the indirect flat panels, the reconstruction kernel. Data binning and geometric magnification had a direct impact on the spatial resolution, the spatial resolution increased as the geometric magnification (M) increased from 1.72 to 5.10. However, at large M , the micro CT performance is more affected by the finite focal spot blurring as compared to smaller M . For example, the sampling efficiency (10% MTF / Nyquist limit) in the in-plane mode at $M=1.72$, 1×1 data binning, 30 mm FOV was 60%,

whereas, it was 50% at $M=5.10$, 1×1 data binning, 10 mm FOV. Thus, the minimum resolution achieved by the system was in the 30–40 μm range. The geometric magnification employed in micro CTs provides a tradeoff between field of view and spatial resolution for a wide range of applications.

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Highlights

The spatial resolution characteristics of a cone beam micro CT system was investigated under different geometric magnifications.

By measuring the modulation transfer function (MTF) curves and presenting images of different spatial resolution phantoms, this work gives an in-depth understanding of spatial resolution of cone beam systems.

The in-plane MTFs were calculated as a function of number of projections, geometric magnification (M), data binning and reconstruction magnification.

The in-plane resolution was compared with the cross plane and projection resolutions

Figure. 1.

Phantoms used for the spatial resolution measurements. (a) Custom made phantom for the in-plane MTF measurement (b) Schematics of micro-CT SSP phantom (c) Schematics of micro-CT resolution bar pattern phantom.

Figure. 2.

Tungsten wire phantom used for the measurement of the in-plane spatial resolution. (a) Projection image of the phantom showing the 14 μm tungsten wire embedded in two needles on the EVA foam. The image was acquired at 90 kV, 160 μA , 33 msec, $M=5.10$, 2×2 binning; (b) Reconstructed image of the 14 μm tungsten wire showing an ROI for integration in the mentioned direction to produce the LSF needed for the determination of MTF.

Figure. 3.

MTF for different scan times to investigate the effect of frame combination on the reconstructed image; a) FOV 60mm, M=1.72; (b) FOV 20mm, M=5.10.

Figure. 4.

In-plane modulation transfer function (MTF) curves of the micro CT system measured with the tungsten wire at 90 kV, 160 μ A for the (a) FOV 75 mm, 148 μ m pixel size; (b) FOV 30mm, 59 μ m pixel size; (c) FOV 40 mm, 80 μ m pixel size; (d) FOV 20 mm, 40 μ m pixel size; (e) FOV 10 mm, 20 μ m pixel size.

Figure. 5.

The in-plane images of the micro-CT resolution bar pattern phantom acquired at 90 kV, 160 μA in (a) 5 mm FOV, 1x1 binning, $M=5.10$, $M_{\text{Recon}}=2.50$; (b) 10 mm FOV, 1x1 binning, $M=5.10$, $M_{\text{Recon}}=1.25$; (c) 30 mm FOV, 1x1 binning, $M=1.72$, $M_{\text{Recon}}=1.25$; (d) 40 mm FOV, 1x1 binning, $M=1.72$, $M_{\text{Recon}}=1.25$. The contrast profile corresponding to the dotted line is shown.

Figure. 6. Cross-plane (z-direction) MTF curves measured using the micro-CT SSP phantom at 90 kV, 160 μ A for the (a) FOV 5 mm, 10 μ m slice thickness; (b) FOV 20 mm, 40 μ m slice thickness

Figure. 7.

Z-direction (z-x) images of the ultra-high contrast resolution bar chip phantom acquired at 90 kV, 160 μ A for the (a) M=5.10, FOV 5 mm, 10 μ m slice thickness; (b) M=5.10, FOV 20 mm, 40 μ m slice thickness.

Figure. 8.

Projection imaging MTF curves measured using the slit camera phantom at 90 kV, 160 μ A, 3 fps with (a) M=5.10, 1×1 binning; (b) M=5.10, 2×2 binning.

Figure. 9.

Comparison of the spatial resolution between the projection and the CT reconstructed images with an Ultra-high contrast resolution bar chip phantom at 90 kV, 160 μ A (a) Projection image at M=5.10, 1 $\times$ 1 binning, 3 fps; (b) Z-direction (x-z) reconstructed image at M=5.10, 10 mm FOV; (c) Projection image at M=1.72, 1 $\times$ 1 binning, 3 fps; (d) Z-direction (x-z) reconstructed image at M=1.72, 30 mm FOV.

Table. 1

The configuration associated with each FOV of the micro CT system

FOV (mm)	M	Binning	M _{Recon}	x-y Pixel Size (μm)	z- Slice Thickness (μm)	Exposure Time	Number of Projections
75	1.72	2×2	1	148	148	17s, 122.4s	511, 918
60	1.72	2×2	1.25	118	118	17s, 122.4s	511, 918
30	1.72	1×1	1.25	59	59	26s, 180.4s	514, 902
40	2.54	2×2	1.25	80	80	17s, 122.4s	511, 918
24	5.10	2×2	1	50	50	17s, 122.4s	511, 918
20	5.10	2×2	1.25	40	40	17s, 122.4s	511, 918
10	5.10	1×1	1.25	20	20	26s, 180.4s	514, 902
5	5.10	1×1	2.50	10	10	26s, 180.4s	514, 902

Table. 2

The cutoff frequency (lp/mm) measured for different FOVs for configuration associated with each FOV of the quantum FX micro CT system

FOV (mm)	Pixel Size (μm)	M	Binning Mode	M_{Recon}	Cutoff freq (lp/mm)
75	148	1.72	2×2	1.00	2.31
60	118	1.72	2×2	1.25	2.32
30	59	1.72	1×1	1.25	4.85
40	80	2.54	2×2	1.25	4.25
24	50	5.10	2×2	1.00	7.25
20	40	5.10	2×2	1.25	7.20
10	20	5.10	1×1	1.25	12.56