Monte Carlo Model for Estimation of Dose Delivered to Small Animals During 3D High Resolution X-ray Computed Tomography

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Abstract-- **Biological research in recent years has generated significant interest for in vivo small animal imaging technologies. 3D small animal x-ray Computed Tomography (CT) provides anatomical images with high spatial resolution and good bone-to-soft tissue contrast. Radiation doses to the subject can be significant when soft tissue contrast and high-resolution images are desired.**

We have used the MCNP Monte Carlo simulation, and calibrated thermoluminescent dosimeters (TLD's), in combination with high resolution x-ray spectra obtained with a Cadmium Zinc Telluride (CZT) detector, to calculate the depth dependent dose in 3D high resolution x-ray CT. Three spectra (30kVp with 0.25mm of aluminum filtration, 40kVp with 0.50mm Al, and 50kVp with 1.00mm Al) were chosen as representative of soft, medium, and hard beams. MCNP was used to simulate the dose from these x-ray spectra incident upon a cylindrical mouse-sized phantom (2.54×6.1cm).

The same phantom was also constructed from solid lucite material with thermo-luminescent dosimeters (TLD's) placed at the positions where we sampled the dose with MCNP. The maximum and minimum dose observed in this study is 19.7 ± 1.7 cGy from the soft beam measured nearest the surface, and 5.2 ± 0.1 cGy from the hard beam measured furthest from the source, for a typical data acquisition with 196 angles.

I. INTRODUCTION

Recent advances in biological research have generated significant interest for in vivo imaging technologies dedicated to small animals, predominantly mice. Small animal 3D x-ray Computed Tomography (CT) provides excellent bone versus soft tissue contrast, with good images of the anatomy that can be used in conjunction with molecular imaging technologies like Positron Emission Tomography. In order to realize the high spatial resolution capabilities of the x-ray imaging systems (~50 microns), significant radiation doses can be delivered to the studied subject. These radiation doses can interfere with the biological model, especially during longitudinal studies. It is important to understand the magnitude of these doses and build on the capability to simulate the multiple x-ray CT imaging parameters in 3D (x-ray tube potential, filtration amount and material type, exposure time, etc.), in order to optimize the imaging technique and minimize the radiation dose. Similar work has been previously performed for human CT scanners. These methods though cannot be directly ported over to small animal imaging work, because they are predominantly 2D in nature and they involve a much higher x-ray photon energy [1,2]. In this work, we have used the MCNP Monte Carlo simulation code developed at the Los Alamos National Laboratory, in combination with high resolution x-ray spectra acquired with the x-ray source and a Cadmium Zinc Telluride (CZT) detector. We have simulated the dose distribution on a geometry-based simple mouse phantom, with representative spectra from the x-ray source operating at variable kVp's and containing various amounts of aluminum filtration.

II. MATERIALS AND METHODS

A. X-ray Tomograph

We have used a 3D high-resolution x-ray CT system (MicroCAT, ImTek Inc., Knoxville TN) [3], dedicated for imaging small animals. The tomograph is illustrated in Figure 1. The system has a microfocus x-ray source with a maximum voltage of 50 kVp, a maximum anode current of 1.5 mA, a 22° cone angle, a fixed tungsten anode and a thin beryllium exit window. The x-radiation detector is a 7.5×7.5cm gadolinium-based phosphor screen, coupled via a tapered optical fiber bundle to a 1024×1024 low noise CCD camera, with 24-micron pixels. The imaging field of view is 5×5cm, while the best imaging spatial resolution is ~50 microns, achieved with the magnification factor inherent in the cone beam imaging geometry. The projection images are formed into 3D sinograms, which are subsequently reconstructed with a 3D cone beam reconstruction algorithm [4].

B. X-ray Spectra

High resolution x-ray spectra (1.5% FWHM) were measured with a 3x3x2mm CZT detector (XR-100T-CZT,

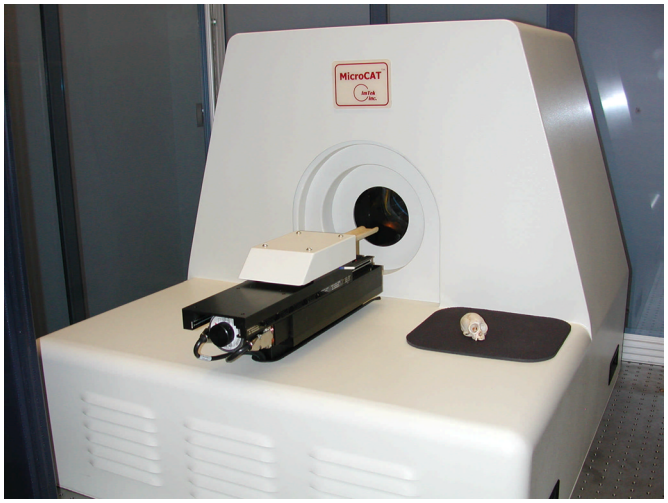

Figure 1. MicroCAT, the small animal x-ray CT scanner from ImTek Inc.

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Amptek Inc., Bedford MA) and an identical x-ray source in a lab bench-top setting, Figure 2. The source to detector distance was 25 cm. The x-ray beam was collimated with two pinhole tungsten collimators (200 micron and 400 micron), spaced 3.8cm apart between the source and the detector. Three x-ray beam qualities were selected: *soft* - 30kVp with 0.25mm Aluminum filtration (having an average photon energy, E_{avg}=17keV), *medium* – 40kVp with 0.50mm Al (E_{avg}=22keV), and *hard* – 50kVp with 1.00mm Al (E_{avg}=27keV). The spectra were acquired until 10^6 photons were collected and are shown in Figure 3. Currently, our default technique for acquisitions is 40kVp with 0.50mm Al.

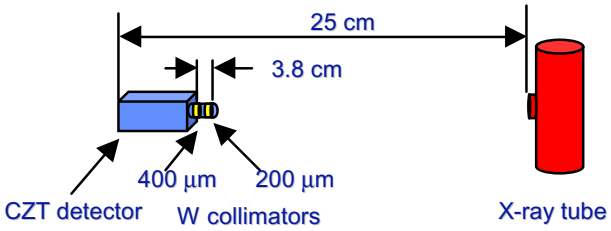

Figure 2. Setup for acquiring X-ray spectra.

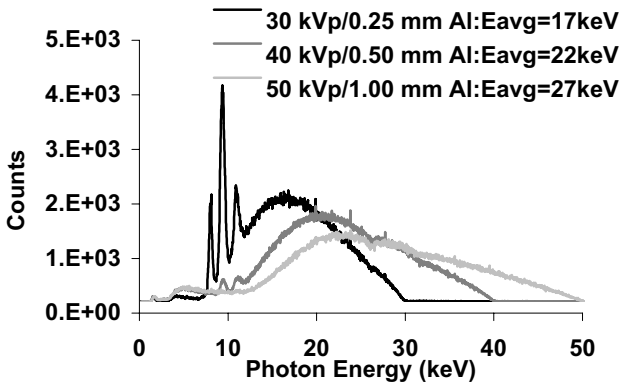

Figure 3. X-ray spectra representing the soft, medium, and hard beams in the experiment acquired using the CZT detector.

C. Thermoluminescent Dosimetry

We constructed a 31cc cylindrical, mouse-sized phantom (2.54cm length × 6.1cm diameter) made of solid lucite material. An illustration of the phantom is in Figure 4. Slots were created in the phantom to allow placement of TLD's at 5mm sampling intervals at three radial locations: surface (r=1.006cm), center (r=0cm), and off-center (r=0.635cm). Lithium fluoride (TLD-100, Harshaw Chemical Co., Solon OH) chips (3.2×3.2×0.9mm) were used to measure the absolute dose at each of the 33 locations within the phantom. The phantom was placed on the bed of the MicroCAT scanner and centered in the field of view by acquiring a scout image. Measurements of dose from one screening scan (196 angles) were repeated three times for each of the three x-ray beams. The exposures for each beam quality were not fully optimized for image quality, but were representative of the technique used. The tube current was held constant at 400 µA, while the exposure time was changed to get the same exposure in the background (800ms for the soft beam, 450ms for the medium beam, and 290ms for the hard beam). The TLDs were read out with a standard TLD reader (Harshaw model 5500, Harshaw Chemical Co., Solon, OH) interfaced through a Windows NT PC. Using the ratio of mass attenuation coefficients, the maximum over-response of LiF at 20-30 keV as compared to ^{60}Co energies is 1.3 as illustrated in Figure 5 [5]. The calibration curve from ^{60}Co can therefore be applied to the data to obtain results in cGy.

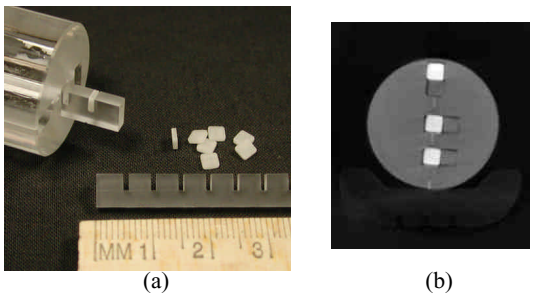

Figure 4. (a) The mouse-sized phantom showing the inserts that hold the TLD chips in place. (b) Transverse slice showing the 3 radial locations of the TLD chips.

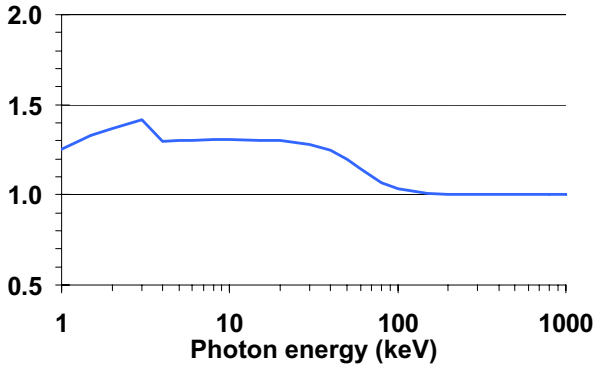

Figure 5. Plot of the calculated energy dependence of LiF relative to that of ^{60}Co (1.25 MeV).

D. Monte Carlo Simulation

The high resolution x-ray spectra, the system geometry, and the exact phantom geometry were used as inputs to the MCNP version 4C Monte Carlo code [6] for a simulation of the TLD measurement as described above. In order to achieve a 1% statistical uncertainty in the Monte Carlo exposure calculations, 10^9 photons were simulated for a full 195^0 scan of the phantom, requiring ~11 hours of computation time on a 1.7 GHz Pentium Xeon computer system. Output from the Monte Carlo code is in the form of dose per photon simulated. The exact number of photons produced for each imaging technique depends on tube voltage, filtration, exposure time, and tube current. In order to cross calibrate the Monte Carlo results with the TLD measurements, we normalized the MCNP results by the TLD measurement at the center of the phantom.

III. Results & Discussion

The TLD measurements and Monte Carlo simulation results are plotted in Figure 6. A summary of the measured

dose at the central axial position for each of the three beams and three radial sampling locations appear in Table 1. To calculate the degree of agreement between the measured and simulated values, each set of depth dependent doses is normalized to the average value at the surface. The accuracy is a simple percent deviation from the measured TLD value. This is tabulated for the central axial position in Table 2. The depth-dependent behavior of radiation dose is well modeled in the Monte Carlo simulation. The slope observed in the measured data is most likely due to a mechanical shutter latency effect. The slope varies from 3-9% for the soft beam, 5-12% for the medium beam, and 8-13% for the hard beam. This increase is expected since the exposure rate is higher at higher energies.

The low energy Tungsten L lines of the x-rays are essentially eliminated by 0.50mm Al, as shown in Figure 3, and contribute to surface dose for the soft beam as expected. Our Monte Carlo calculations indicate that the soft beam dose at the center location on the surface can be 1.3 times higher than the hard beam dose. However, we measured this ratio to be 1.8 in reality. This difference is most likely exaggerated by the shutter latency effect not modeled in the Monte Carlo simulation. The maximum and minimum dose observed in this study is 19.7 ± 1.7 cGy from the soft beam measured at the surface, and 5.2 ± 0.1 cGy from the hard beam measured at the off-center location.

From these results, we can estimate the dose in the bladder of a mouse during the high resolution acquisition shown in Figure 7 (390 angles, 40kVp/0.50mmAl/250ms/400μA) is 10.4 ± 0.2 cGy.

IV. CONCLUSION

We have used the MCNP Monte Carlo simulation and the measured spectrum of the x-ray source, to estimate the depth dependent dose for a geometry-based, mouse-like phantom. Although these results are not optimized for image quality, they provide an initial estimate of the radiation exposure to small animals during high-resolution x-ray imaging. Although the measured dose (~10cGy) are much smaller than the LD50 for mice is 6-8 Gy, longitudinal high resolution serial studies can deliver significant accumulated dose. Simulation results are only as good as the parameters accounted for. Our measurements show that our simulation does not adequately describe the CT system as evident in the axial slope of the dose measurements. Even minimal filtration of the x-ray beam changes dramatically the spectrum and the depth profile of the radiation dose, in favor of sparing the skin and surface dose. The increase of Al filtration will cause a reduction of soft tissue contrast, but the tradeoff can be optimized as a function of the imaging task. We have shown that Monte Carlo can be used to determine the depth dependent dose to within 10% for three techniques. In the near future, we plan on verifying the TLD measurements with calibrated ion chamber measurements. With the ability to estimate the dose from a Monte Carlo simulation system, we can then optimize the dose delivered versus imaging procedure. This future work will require the use of a more accurate voxel based input geometry,

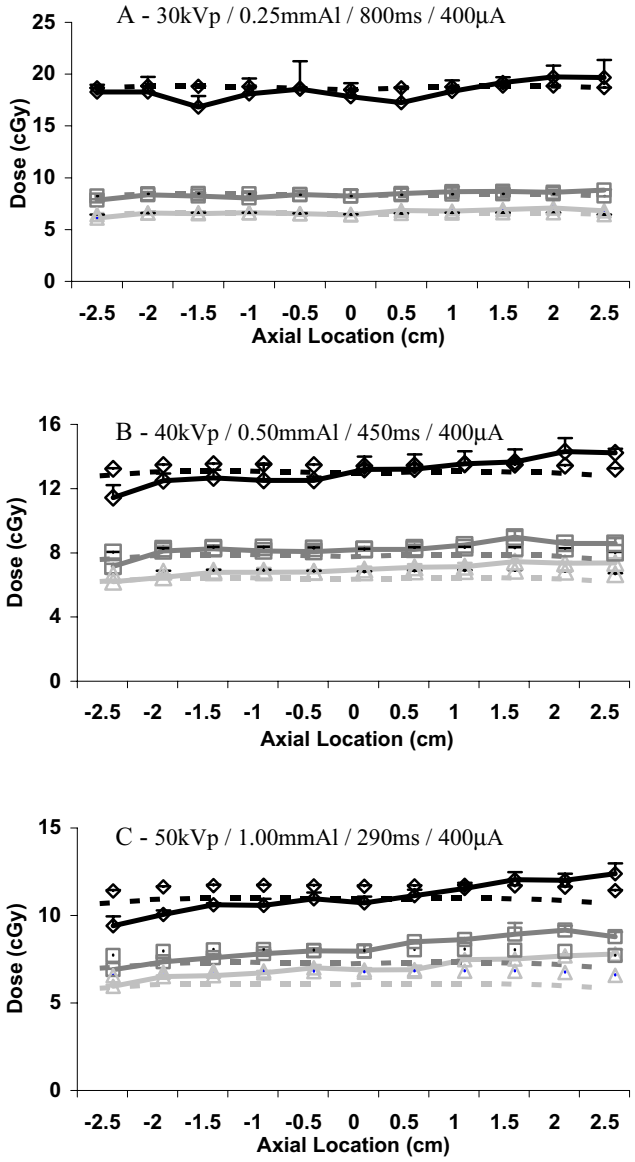

Figure 6. Plot of the measured doses (solid lines) and the simulated (dashed) results from Monte Carlo for the a) soft, b) medium, and c) hard beams for one 196-angle scan. The colors indicate the radial positions of the measurements: black–surface, grey–center, and light grey–off-center.

Table 1. Average TLD dose (cGy) at the central axial position for each beam.

N=3	Soft	Medium	Hard
Surface	17.8 ± 1.3	12.8 ± 0.8	10.0 ± 0.3
Center	8.2 ± 0.6	7.8 ± 0.2	7.2 ± 0.3
Off-Center	6.4 ± 0.2	6.5 ± 0.2	6.2 ± 0.1

Table 2. Accuracy of simulated results at the central axial position.

	Soft	Medium	Hard
Surface	-1.62%	1.23%	-3.55%
Center	1.89%	2.86%	5.43%
Off-Center	1.31%	5.15%	7.25%

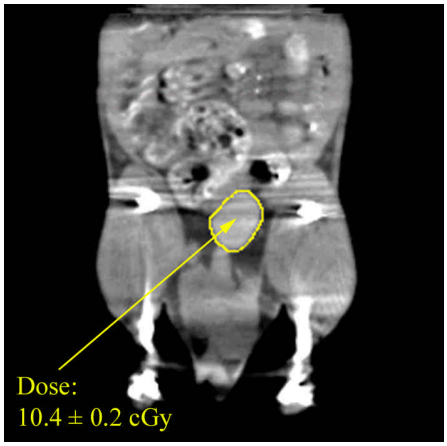

Figure 7. Coronal image of mouse and dose estimation using MCNP assuming bladder is at the center of our phantom. Acquisition parameters: 40kVp / 0.50mmAl / 250ms / 400mA / 390 angles. Reconstructed in 3D with 256^3 matrix.

that will be based on animal data acquired with the x-ray CT scanner.

V. ACKNOWLEDGMENTS

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