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Monte Carlo Simulation of an X-Ray Pixel Beam Microirradiation System

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Abstract

Monte Carlo simulations are used in the development of a nanotechnology-based multi-pixel beam array small animal microirradiation system. The microirradiation system uses carbon nanotube field emission technology to generate arrays of individually controllable X-ray pixel beams that electronically form irregular irradiation fields having intensity and temporal modulation without any mechanical motion. The microirradiation system, once developed, will be incorporated with the micro-CT system already developed that is based on the same nanotechnology to form an integrated image-guided and intensity-modulated microirradiation system for high-temporal-resolution small animal research. Prospective microirradiation designs were evaluated based on dosimetry calculated using EGSnrc-based Monte Carlo simulations. Design aspects studied included X-ray anode design, collimator design, and dosimetric considerations such as beam energy, dose rate, inhomogeneity correction, and the microirradiation treatment planning strategies. The dosimetric properties of beam energies between 80–400 kVp with varying filtration were studied, producing a pixel beam dose rate per current of 0.35–13 Gy per min per mA at the microirradiation isocenter. Using opposing multi-pixel-beam array pairs reduces the dose inhomogeneity between adjacent pixel beams to negligible levels near the isocenter and 20% near the mouse surface.

INTRODUCTION

State-of-the-art tools for small animal model research are in great demand for basic and translational cancer research. Impressive advancements in nanotechnology-based biomarkers (1, 2) promise major breakthroughs in cancer detection, treatment and treatment response detection. Concurrently, cancer modeling has entered a new era in which researchers genetically engineer the mice to mimic the formation, presentation and progression of disease in humans (3, 4). The advancements in both fields of cancer research underscore the need to develop new tools that can meet the unprecedented high spatial and temporal resolution requirements for clinically relevant investigations of new imaging and therapeutic agents using small animal models. This need is expected to grow rapidly in the near future as more biomarkers become available for preclinical study using the mouse models.

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To date, there are a number of commercially available small animal imaging tools, such as micro-CT (5), micro-MRI (6), and micro-CT-PET (7). However, the development of therapeutic tools for small animal models has lagged. For instance, most small animal irradiations performed in research laboratories today use either clinical radiation systems or conventional animal irradiators with limited spatial and temporal resolution. Using customized radiation collimation and immobilization devices, studies can achieve precise small animal irradiation using radiation field sizes of 5 mm or greater using conventional irradiators (8–11). The ability to produce smaller beams would allow conformal irradiation of smaller targets, such as tumors in an early stage of growth and interstitially implanted xenograft tumor models. Non-uniform (intensity-modulated) irradiation of larger targets, more accurately reproducing the types of treatments used on human patients, is also highly desirable (12, 13). Precise temporal resolution would be useful for studying specific radiobiological responses of tissue, such as blood flow. Realizing the need for better small animal irradiation technologies, Low's group pioneered the development of microirradiation systems (14). Their research group developed a microirradiation system that uses a commercially available radioactive ^{192}Ir source and physical collimator to generate a single small-diameter photon beam. Another microirradiation development group led by Wong developed the first image-guided small animal irradiator. It used a commercially available X-ray tube for both cone-beam CT imaging and radiation delivery (15). Other groups at Princess Margaret Hospital and Stanford University have used clinical simulators (16) and micro-CT devices (17), respectively, to produce radiation appropriate for small animal studies. All these microirradiation systems have demonstrated the capability of producing very small-diameter (mm or less) single-beam irradiation that can provide unprecedented spatial resolution for small animal irradiation. However, forming irregular-shaped and intensity-modulated radiation would pose challenges for these microirradiation systems, because high-precision mechanical fabrication and control would be needed. Furthermore, the single- and fixed-beam collimation systems are not ideal for generating intensity- and temporally modulated radiation, two basic components of state-of-the-art clinical radiotherapy.

We have proposed and begun the development of a novel microirradiation system that is based on carbon nanotube (CNT) field emission technology and an X-ray pixel-beam array system design (18). The key difference between this and other microirradiation system designs is that, in the CNT microirradiation system, the radiation field is formed by selectively turning on individually controllable X-ray pixel beams as opposed to using a broad beam that is mechanically collimated to a given shape. The proposed microirradiation system is capable of delivering arbitrarily shaped and intensity-modulated radiation with high temporal resolution for respiratory-gated radiotherapy for small animals. All of these features are important parts of the state-of-the-art clinical radiotherapy technology. Figure 1 illustrates the X-ray pixel-beam array capabilities for mouse irradiation, where the arbitrarily shaped irregular field is formed by six 2-mm X-ray pixel beams. The mouse CT image was obtained using a prototype CNT micro-CT already developed by our CNT field emission device fabrication group (13). A similar CNT micro-CT system will be incorporated into the microirradiation system to form an integrated micro-CT-irradiation system in the future.

Radiation dosimetry is a crucial consideration in the microirradiation system design. A variety of Monte Carlo packages have been used to simulate the design and resulting dosimetry of devices operating in the energy range relevant to small animal radiotherapy. Stepanek (19) used GEANT to evaluate the dosimetry of a synchrotron-based photon microbeam for energies between 50–200 keV. EGS-based codes have been used for the evaluation of ion chamber response to low-energy beams (20), the design of an ^{192}Ir -based small animal irradiator (14), and many other kV applications (21–23). Cellular-level Monte Carlo simulation for microbeams has been studied using PENELOPE (24), PITS (25) and EGS (26, 27). The breadth of application of the Monte Carlo technique to radiotherapy in the energy range relevant to small animal studies indicates that the technique is well suited for the task of dosimetry-based system design for this microirradiation system. This publication reports our initial results in the first phase of the CNT microirradiation system development research: Monte Carlo simulation-based system design and feasibility demonstration.

MATERIALS AND METHODS

Description of Prototype X-Ray Pixel-Beam Array Microirradiation System

The design of the proposed CNT microirradiation system is illustrated in Fig. 2. The entire system is shown in Fig. 2b and consists of several individual arrays arranged in a ring structure. Each of the arrays (Fig. 2a) can produce many parallel and narrow X-ray beams, referred to hereafter as X-ray pixel beams. The X-ray pixel beams are generated by a CNT field emission cathode array and collimated by a pixel-beam array-forming collimator. The control of each individual X-ray pixel beam is accomplished through the electronic control of the corresponding CNT field emission cathode. When the adjacent X-ray pixel beams are turned on, a continuous radiation field at the isocenter of the microirradiation system is formed. The prototype single-array CNT microirradiation system simulated in this publication is designed to operate at beam energies of 80–400 kVp. The source-to-axial (isocenter) distance is 8 cm and the tissue-equivalent target (mouse) is 3 cm in diameter. The single-array system under development is composed of 50 pixel beams in a 5×10 array that is capable of producing a continuous 1×2 -cm radiation field at the isocenter. Each pixel beam projects a square 2×2 -mm beam at the isocenter and adjacent pixel beams match (form a continuous radiation field) at the isocenter. The electron source, consisting of CNT field emission pixels and an electron-accelerating structure, was not modeled explicitly in the Monte Carlo simulations but was represented as a parallel electron distribution of varying diameter at the surface of the X-ray target (anode). The goal of this prototype system development, including Monte Carlo simulation, is to demonstrate the feasibility of the novel microirradiation system, which may or may not be the same as the prototype system under study here.

X-Ray Pixel-Beam Array Microirradiation Anode Design

We have investigated two different X-ray target or anode designs: a reflection target (Fig. 3a) and a transmission target (Fig. 3b). There are advantages and disadvantages for each of these anode designs. The reflection anode is better for cooling because of its large thermal mass; the transmission anode design is simpler to fabricate, which is an important consideration

for prototype device development. The reflection anode design might be susceptible to cross-talk between adjacent pixel beams, which breaks down the one-to-one correlation between electron and X-ray pixel beams, which is a core requirement for the X-ray pixel-beam array design. Another consideration in anode-type choice is X-ray production efficiency, because at this low-energy range the bremsstrahlung production efficiency is very low and anode heating can be a limiting constraint in prototype development. Both anode designs use a non-divergent collimator design. The prototype X-ray pixel-beam forming collimator is a copper block with 1-mm-square apertures and a center-to-center distance of 2 mm between adjacent apertures. In the reflection anode, the collimator begins immediately downstream of the aluminum filter (Fig. 3a), whereas the collimator for the transmission anode extends upstream all the way to the target. In both cases, the downstream end of the collimator is placed at the midpoint between the target and isocenter, producing adjacent 2×2 -mm-square pixels in the plane of the isocenter. While some small animal research requires a field size smaller than 2×2 mm, which can easily be achieved by existing microirradiation systems (14, 15), this prototype multi-pixel-beam array system design is capable of producing unique irregularly shaped, intensity- and temporally modulated radiation that may be important for other small animal research needs. Our focus in this early stage of development is to demonstrate feasibility of the X-ray pixel-beam array microirradiation system.

Monte Carlo Simulation

The Monte Carlo simulations in this work are focused on the prototype single-array X-ray pixel-beam device (Fig. 1a). The simulation process begins at the point where the electron pixel beam bombards the X-ray target (anode) and follows the resulting particles as they travel through the microirradiation system and deposit dose in the mouse phantom. A non-diverging electron beam is assumed. The simulated mouse dosimetry is used to optimize parameters of specific system components, such as the specifications of the X-ray pixel-beam array collimator and the CNT electron pixel-beam current. The Monte Carlo simulations of the microirradiation system were performed using EGSnrc-based Monte Carlo codes (28). The EGSnrc code system is the latest in a series of radiation transport algorithms used in high-energy, nuclear and medical physics and contains many improvements over the previous version (EGS4) that improve the accuracy of radiation transport in the keV energy region (28). The EGSnrc parameters used are shown in Table 1. The simulation procedure consisted of two phases: simulation of the microirradiation device, producing a phase space file or a rudimentary dose calculation, and the simulation of the interaction of the resulting beam with a water target or mouse model. The electron beam, anode designs, filter and collimator were simulated using BEAMnrc (29). BEAMnrc has numerous geometric modeling routines that can be used to easily represent the microirradiation components. The energy cutoffs in the anode were set to (PCUT) 1 keV for photons and (ECUT) 512 keV total energy for electrons. The calculation of the efficiency of photon production in the anodes was enhanced using directional bremsstrahlung splitting, in which a single bremsstrahlung event is repeated multiple times, with all photons produced being weight-adjusted and projected into a user-defined field (30). ECUT was set to 50 keV for electrons in the reflection target filter and collimator to reduce calculation time further by not tracking low-energy electrons as they scatter through the copper collimator. The

inclusion of this higher electron cutoff energy outside the anode, compared with the 1 keV kinetic energy cutoff employed in the anode, produced no discernable difference in dose or fluence results.

Dose-rate calculations were generated within BEAMnrc using the CHAMBER component module modeled as a water phantom. The phantom was configured as a cylinder having a diameter and thickness of 3 cm (the diameter of an average mouse), with the calculated dose rate averaged over 1×1 mm in the central region of the beam. Dose-rate results were calculated in terms of cGy/incident electron and were re-normalized to cGy per min per mA. All dose rate calculations assume d.c. emission current. With the exception of the above dose-rate calculations, the particles produced by the BEAMnrc microirradiation simulation were tallied as they exited the collimator and recorded in a phase space file containing each particle's type, energy, position and direction. This phase space file was used to calculate the 3D dose distributions in the water phantom and the mouse model. Phase space particles were recycled up to 20 times for each 3D dose calculation. These simulations were performed using DOSXYZnrc. Voxel dimensions varied depending on what property was being examined but were typically 0.25 mm and in all cases were less than 0.5 mm. The 3D mouse model was generated from a CNT micro-CT scanner (13) and was divided into $0.3 \times 0.3 \times 0.5$ -mm voxels.

RESULTS

Reflection Anode Design

The reflection anode (Fig. 3a) has an anode angle of 15° , and the parallel electron pixel beams are striking the surface of the anode at normal incidence. A low-energy filter was placed upstream of the collimator in the reflection anode design to eliminate low-energy X rays unlikely to penetrate sufficiently deep into the treatment target. The collimating device was modeled as a copper block having an array of parallel holes, with each hole aligned with an individual CNT cathode pixel. Each of the pixel-beam collimator apertures is 1×1 -mm square with a center-to-center distance of 2 mm. The downstream edge of the collimator was set to one half of the source-axis distance, so that the pixel beams would project to adjacent 2×2 -mm fields at the isocenter. The thickness of the collimator was set by geometric considerations and was considerably thicker than required for shielding considerations alone. The anode design assumes a source-axis distance of 8 cm. The extension of the collimator to 4 cm from the target leaves an 8-cm-diameter opening for the small animals. Figure 4 (inset) shows the simulation of the X-ray spectrum from 100 keV electrons striking the reflection anode. The raw X-ray spectrum contained a large number of low-energy photons that would be attenuated in the first few millimeters of tissue, resulting in an unacceptably high surface dose. Figure 4 shows percentage depth dose of a 100 keV microirradiation beam. These calculations show that X-ray photons of energies below 20 keV are responsible for the high skin dose and should be filtered from the beam. The Monte Carlo calculations indicate that a 2.5-mm aluminum filter was sufficient for this purpose.

For the multi-pixel-beam array microirradiation system, cross-talk refers to the bremsstrahlung X rays produced by one electron pixel beam going through, at least partially, not only the designated X-ray pixel beam opening but also the adjacent X-ray pixel-beam

openings in the collimator. The reflection anode design (Fig. 3a) can have a significant cross-talk effect, which will make the design unsuitable for the microirradiation application. This cross-talk effect can be largely mitigated by optimizing the design parameters including the collimator thickness and the position to reduce the solid angle available for cross-talk between pixel beams. Figure 5 shows the X-ray fluence profile generated by a single-electron pixel beam of the 100 kVp reflection anode design. The electron pixel beam is aligned with the center X-ray pixel-beam opening of the collimator. The resulting dosimetry through the “intrinsic” X-ray pixel-beam opening at the center and the “leakage” through the immediately adjacent and the next collimator openings are shown in Fig. 5 for different source-to-collimator distances. For the given design parameters, these results indicate that the clearance between the anode and upstream edge of the collimator must be 2 cm or less to reduce the cross-talk between adjacent pixels to less than 1%. The 200 and 400 kVp energy beams were not examined for the reflection anode configuration at this point because the chosen prototype system uses the transmission anode design.

Transmission Anode Design

This transmission anode featured a thin sheet of high-atomic-number material that is in good thermal contact with the upstream collimator array opening (Fig. 4b), allowing the copper to act as a heat sink for the anode. Direct contact between the target and collimator largely eliminates the cross-talk between adjacent pixel beams found in the reflection anode design. Tungsten and gold were studied as possible anode materials. Monte Carlo simulation is used to optimize the transmission anode thickness in terms of the efficiency of X-ray production and other design considerations. The dose rate for each anode material and thickness is shown in Table 2. Anode thickness was limited to 26 μm or thicker to provide sufficient filtering to reduce the surface dose from photons below 20 keV, as illustrated for the reflection target configuration in Fig. 4, and due to target heating and availability concerns. Dose rates for the tungsten targets were found to be slightly higher than those for the gold targets. These results, along with the higher melting point of tungsten, indicate that tungsten is the preferable transmission anode material. The highest dose rates resulted from the thinnest targets, indicating that the majority of useful photons are produced in the first few micrometers of the anode. Figure 6 shows the tungsten transmission anode X-ray energy spectrum simulation for the 100, 200 and 400 kVp electron energies. The result for 100 kVp indicates that the 26- μm -thick tungsten target provided sufficient self-filtering to remove photons below 20 keV, requiring no additional filter for this configuration. The inter-pixel cross-talk effect for all transmission anode configurations studied was calculated to be negligible (less than 0.1%). For all higher-energy configurations, the electron source is assumed to have the same dimensions as the collimator aperture ($1 \times 1\text{-mm}$ square). The raw spectra from the high-energy configurations contained a significant population of low-energy photons, requiring an additional layer of filtration downstream of the tungsten anode. The filtration material and thickness were chosen to match commonly used filters in therapeutic orthovoltage treatment machines. The 200 kVp configuration was evaluated for a 35- μm anode with a 2-mm aluminum filter and a 0.5-mm copper filter. The 400 kVp configuration was evaluated for a 50- μm anode with two double-layered filters, the first with 1 mm copper followed by 1 mm aluminum and the second with 1.5 mm copper and 1 mm aluminum. These filters were placed between the anode and the collimator. The data show

that the filters are needed to remove low-energy components of the bremsstrahlung X rays to reduce skin dose, especially for energies of 200 kVp and lower.

Microirradiation Dosimetry

Dosimetry for microirradiation is simulated using Monte Carlo simulations. Phase space files were generated based on the microirradiation parameters described in the previous section and were used as input for the 3D dose calculations. Simulation results for the reflection and transmission anode designs indicate that the X-ray fluence outputs for the two designs are very similar in terms of the shape and the sharpness of the dose distributions.

Single Pixel-Beam Dosimetry

The percentage depth dose (PDD) for transmission anode X-ray pixel beams at different energies and filtrations is shown in Fig. 7. The midpoint dose for a mouse 3 cm in diameter is 40–60% of the skin dose for an energy range of 100 to 400 kVp, which is consistent with human dosimetry under external-beam radiotherapy. The higher-energy configurations predictably exhibit more gradual falloffs with depth and differ significantly from the 100 kVp configuration. However, the PDDs for the 200 and 400 kVp configurations typically agree to within 5%. This suggests that the high-energy PDDs are dominated by the photon populations between 100 and 200 keV. Increasing the energy or changing the filtration was found to have only a minimal effect on the beam profile, as shown in Fig. 8 for a single X-ray pixel beam for a transmission target. Dose profiles for reflection anodes have similar divergence and penumbra characteristics but exhibit a heel effect (Fig. 5). The dose profiles are characterized by a small penumbra, resulting from the limited range of scattered electrons at these energies. Figure 9 shows the effect of the incident electron beam size on dose profiles in the phantom. Compared to the infinitesimal beam, the 1×1 -mm electron beam produced less steep beam penumbra as well as a lower dose rate. Our calculation shows that a reduction of the pixel-beam dose rate at isocenter to 35 cGy per min per mA from 50 cGy per min per mA when the electron pixel-beam size changes from infinitesimal to a 1×1 -mm square (Table 2). A sharp pixel-beam penumbra is highly desirable at both the field edge and the interior of each X-ray pixel to minimize inter-pixel dose inhomogeneities within the irradiation volume and to minimize dose to adjacent structures.

Dose Uniformity

The geometry of the microirradiation X-ray pixel-beam array is designed such that dose profiles from neighboring X-ray pixel beams are adjacent at the isocenter. Because individual pixel beams are unavoidably divergent, radiation from adjacent pixel beams of the same array will overlap at depths beyond the isocenter and will leave a gap at shallower depths. This effect leads to an intrinsic dose inhomogeneity within the irradiated volume. Simulations indicate that this dose inhomogeneity can be reduced by using opposing array pairs aligned such that the high-dose region from one array is superimposed over the low-dose region from the opposite array. Figure 10 shows that the use of opposed arrays reduces the degree of dose inhomogeneity to negligible levels in the vicinity of the isocenter and to 20% near the surface. The dose inhomogeneity can be reduced further in the multi-array microirradiation design (Fig. 2b), where multiple array pairs from different angles are used in radiation delivery. Figure 11 shows the central plane dose distribution from two

orthogonal sets of opposed beam arrays. The multi-array microirradiation system not only will improve the X-ray pixel-beam array dose uniformity but also will greatly improve the ability of the system to deliver highly dose-sculptured targeted irradiation for small animals.

Dosimetry in Inhomogeneous Tissue

For megavoltage X-ray therapy for human cancer patients, conventional dose calculation algorithms assume a patient is constituted entirely of one material (water) of varying density. This approach is appropriate in megavoltage therapy because the dominant X-ray-tissue interaction is Compton scattering that is insensitive to atomic number. However, the assumption is no longer valid for the lower-energy microirradiation dose calculation. The interaction between 100 kVp photons from the microirradiation system and the surrounding material is influenced by the photoelectric effect, which is highly dependent on the atomic number of the material in question. The photoelectric effect is proportional to the third power of atomic number of the material and is inversely proportional to the third power of the X-ray energy. Therefore, lower-energy X-ray beams are subject to large dose deposition inhomogeneities in tissue with a mixture of different elements (atomic numbers). To better understand the influence of photon energy on mouse irradiation dosimetry, we performed Monte Carlo simulations on two models: a tissue-bone phantom and a mouse CT image from a CNT micro-CT scanner. The bone-tissue phantom consists of two 1.5-mm-thick slabs of bone material ($\rho = 1.25 \text{ g/cm}^3$) in a water-equivalent ($\rho = 1.0 \text{ g/cm}^3$) phantom 3 cm thick. Figure 12 shows the Monte Carlo dosimetry results in the tissue-bone phantom for different photon energies and filtration from a parallel opposed beam pair. The figure shows that the 100 kVp photon beam without filtration deposits significantly more dose to bone (~four times higher) compared to tissue and that the relative bone dose reduces with increasing photon-beam energy. Calculations on the mouse model of a similar dimension produced qualitatively similar results. We found that the Monte Carlo results are strongly dependent on the assignment of materials, or atomic number, to a given range of Hounsfield numbers or electron density. In the above calculations we used the default conversion from CT value to material (atomic number) and density found in DO-SXYZnrc (31) and assumed specific material compositions for lung, tissue and bone as defined in ICRU Report 46 (32). Note the ICRU compositions are specified for human tissue and may not be precisely correct for mouse anatomy. While the approach of assuming human composition values has been used in recent studies of dose to mouse models from micro-CT (33) and radiopharmaceuticals (34), high-accuracy microirradiation mouse dose calculations may require a careful study of the applicability of the ICRU models to the anatomy of mice. These issues will be explored in greater detail in a future publication. Because of this uncertainty in material and density selection, the specific results shown in Fig. 12 are not quantitatively reliable. Despite this, the Monte Carlo study does demonstrate that special attention must be given to the unique tissue inhomogeneity effect seen in microirradiation.

DISCUSSION

Monte Carlo simulations have played an invaluable role in the development of the microirradiation system by providing dosimetry-based guidance to the system design and performance evaluation. The actual prototype system design took additional issues into

consideration, including anode cooling and device fabrication constraints. For example, the Monte Carlo studies indicate that increasing the X-ray energy would both increase the dose rate and minimize the bone dose. The technical challenges of operating at high voltage a confined space led to a lower-energy selection for the prototype device. Our aim for the initial phase of the carbon nanotube field emission technology-based microirradiation development is demonstration of feasibility. In the eventual multi-array microirradiation system (Fig. 2b), pixel beams from different arrays will be able to irradiate the same voxel in the mouse simultaneously. Therefore, the proposed microirradiation system is expected to be capable of delivering a dose at a rate as high as the clinical therapeutic dose rate of 5 Gy/min for an energy of 100 kVp, while the dose rate for the higher energy configurations will be as much as 20 times higher, as shown in Table 2. This high-dose-rate capability is crucial for high-precision small animal model radiation therapy in which organ motion is considered. The microirradiation is also capable of delivering extremely low dose rates by simply reducing the repetition rate of the CNT pixel-cathode beam field emission. We have shown that high bone dose due to high (average) bone atomic number is a major issue for a microirradiation system at energies below 200 kVp. We plan to fully explore the treatment planning flexibility of the unique X-ray pixel-beam array design for bone dose reduction, in addition to increasing the photon beam energy in our microirradiation system. Calculations of the overlap of adjacent diverging beam pixels demonstrate the critical importance of the opposed pair array design for achieving acceptable dose uniformity. Using opposing beam arrays reduces the cold spots from near 0% of the prescribed dose to 80% near the surface, with similar decreases of downstream hot spots. The use of additional array pairs increases the dose uniformity in the target area and decreases the relative importance of the remaining dose variations near the surface by reducing their absolute magnitude relative to the prescription dose at isocenter.

CONCLUSIONS

We have performed Monte Carlo simulation studies on the dosimetry of a nanotechnology-based X-ray pixel-beam array microirradiation system to guide the system design and performance estimation of the prototype system currently under development. Our studies show that the microirradiation system design and delivery techniques described here are capable of producing dosimetrically useful radiation fields with individually controlled pixels. Further refinement of the prototype design, especially to smaller field sizes and higher energies, should produce a useful tool for delivering conformal and intensity-modulated small animal irradiation.

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FIG. 1.

Illustration of the X-ray pixel-beam array for microirradiation of mice. An arbitrarily shaped irregular radiation field is formed by six individually controlled pixel beams. The CT image is from a prototype carbon nanotube field emission microirradiation developed earlier by our group. The micro-CT will be incorporated with the microirradiation under development to form an integrated micro-CT-irradiation system for small animal model research.

FIG. 2.
Schematic of a single-array (panel a) and multiple-array (panel b) X-ray pixel-beam array microirradiation system.

FIG. 3. Microirradiation system anode and beam collimation designs: reflection anode (panel a) and transmission anode (panel b) designs.

FIG. 4. Percentage depth dose for a reflective anode 100 kVp beam for various low-energy filters. The unfiltered spectrum is shown in the figure inset.

FIG. 5. Inter-pixel cross-talk effect for a reflection anode design with different source-collimator distances (SCD).

FIG. 6.

Transmission anode photon energy spectra. Panel A shows the 100 and 200 kVp configurations. Panel B shows the 400 kVp configurations. All spectra are calculated for tungsten targets. The 100 kVp configuration has no filtration other than the intrinsic filtration in the tungsten anode. The spectra are normalized for display purposes and do not reflect the dose-rate differences shown in Table 2.

FIG. 7.
Percentage depth-dose curves of the microirradiation mouse irradiation for different photon beam energies and filtration.

FIG. 8.
Beam profiles at 1.5 cm depth for transmission target X rays of different energies and filtration.

FIG. 9. Single pixel-beam profiles for a point electron source (solid line) and 1-mm-square extended electron source (dashed line). Profiles are shown for the 100 kVp 26- μ m tungsten transmission target configuration.

FIG. 10. X-ray pixel beam microirradiation dose profiles at different depths from two opposing arrays in a 3-cm water target. Each array has 3×3 pixel beams. The data show that the opposing array arrangement significantly improved dose uniformity.

FIG. 11.
Illustration of dosimetry from a multi-array microirradiation. Two orthogonal opposing array pairs generated a conformal dose distribution in the center.

FIG. 12.

Illustration of the effect of high-atomic-number bone on microirradiation dosimetry in a tissue-bone phantom. The phantom consists of two 1.5-mm-thick bone phantoms in an otherwise water-based phantom 3 cm in width. The dosimetry is computed under a parallel-opposed X-ray beam pair that is perpendicular to the bones. The high-dose regions correspond to the regions of simulated bone.

TABLE 1

Summary of EGSnrc Parameters Used in the Simulation of the Microirradiation Device and Phantom Simulations

Parameter	Value
Global ECUT	0.512
Global PCUT	0.001
Global SMAX	0.5
ESTEPE	0.25
XIMAX	0.5
Boundary crossing algorithm	PRESTA-I
Skin depth for BCA	0
Electron-step algorithm	PRESTA-II
Spin effects	On
Bremsstrahlung angular sampling	Koch and Motz
Bremsstrahlung cross sections	NIST
Bound Compton scattering	On
Pair angular sampling	Simple
Photoelectric angular sampling	On
Raleigh scattering	On
Atomic relaxations	On
Electron impact ionization	On

TABLE 2

Monte Carlo Simulation Results for Single X-Ray Pixel Dose Rate Relative to Transmission Anode Configuration

Anode material	Energy (kVp)	Thickness (μm)	Dose rate (cGy/min/mA)
W (pt. elec. source)	100	26	50
W	100	26	35
W	100	30	22
W	100	40	17
Au	100	26	29
Au	100	30	20
Au	100	40	17
W	200	35, 2 mm aluminum	241
W	200	35, 0.5 mm copper	153
W	400	50, 1 mm copper, 1 mm aluminum	1310
W	400	50, 1.5 mm copper, 1 mm aluminum	844

Notes. Tungsten (W) and gold (Au) transmission anodes were used. Except where specified, dose rates assume a parallel extended electron source having the same dimensions as the collimator opening (1 mm $\times$ 1 mm). The 100 kVp configurations have no filtration other than the intrinsic filtration of the tungsten target.