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Monte Carlo model of the scanning beam digital x-ray (SBDX) source

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Abstract

The scanning-beam digital x-ray (SBDX) system has been developed for fluoroscopic imaging using an inverse x-ray imaging geometry. The SBDX system consists of a large-area x-ray source with a multihole collimator and a small detector. The goal of this study was to build a Monte Carlo (MC) model of the SBDX source as a useful tool for optimization of the SBDX imaging system in terms of its hardware components and imaging parameters. The MC model of the source was built in the EGSnrc/BEAMnrc code and validated using the DOSXYZnrc code and Gafchromic film measurements for 80, 100, and 120 kV x-ray source voltages. The MC simulated depth dose curves agreed with measurements to within 5%, and beam profiles at three selected depths generally agreed within 5%. Exposure rates and half-value layers for three voltages were also calculated from the MC simulations. Patient skin-dose per unit detector-dose was quantified as a function of patient size for all three x-ray source voltages. The skin-dose to detector-dose ratio ranged from 5–10 for a 20 cm thick patient to 1×10^3 – 1×10^5 for a 50 cm patient for the 120 and 80 kV beams, respectively. Simulations of imaging dose for a prostate patient using common imaging parameters revealed that skin-dose per frame was as low as 0.2 mGy.

1. Introduction

Medical x-ray imaging is an essential component of patient diagnosis and radiation therapy treatment planning. In general, conventional fluoroscopic x-ray imaging systems consist of a small x-ray source and a large-area detector. In such geometry, the dose to the skin limits the imaging time (Koenig *et al* 2001, Mahesh 2001). Serious radiation-induced skin damage can occur during fluoroscopically guided, complex interventional procedures (Shope 1996). It is therefore important to monitor the imaging dose and reduce it, if possible. To this end, a novel x-ray imaging device for interventional fluoroscopic imaging, the scanning-beam digital x-ray (SBDX) system, has been recently developed (Speidel *et al* 2006a, 2006b, Van Lysel *et al* 1997, Speidel *et al* 2001, Fahrig *et al* 2003). The SBDX imaging system offers a reduction in imaging skin dose. The SBDX is a unique imaging system capable of generating tomosynthetic x-ray images via its inverse geometry. X-ray tomosynthesis is a hybrid approach bridging two-dimensional (2D) fluoroscopic imaging and three-dimensional (3D) computed tomography imaging (Bushberg *et al* 2001). The 2D slices

within a small 3D volume are reconstructed by rotating the x-ray source over a small angle (Grant 1972). Moreover, the SBDX system operates in inverse geometry, in which a small-area detector detects x-rays generated by a large-area source (figure 1(a)). Inverse geometry fluoroscopic imaging possesses two advantages over conventional fluoroscopy, which result in safer and better image quality: (1) patient skin-dose is spread out over a large area, making it less of a limiting factor compared to conventional fluoroscopy; and (2) scatter is significantly reduced due to the small detector and the large distance between the patient and the detector (Speidel *et al* 2006a).

This work presents an accurate Monte Carlo (MC) model of the SBDX source. The model was validated against experimental measurements carried out with the SBDX system located at Triple Ring Technologies, Inc. (Newark, CA). The model was used to evaluate the dose to the patient as a function of patient size and to quantify the dose with 3D imaging of the prostate. The x-ray source parameters of the MC model can be easily changed making it a useful tool for optimization of the source design in terms of the target thickness and collimator and imaging geometry to achieve the best image quality for unit imaging dose. This MC model of the SBDX source has already been modified and investigated as a novel cost-effective radiotherapy source for treatment of deep-seated tumors (Bazalova *et al* 2011). This paper presents the first step toward modeling and design optimization of the entire SBDX system.

2. Materials and methods

2.1. Source design

The SBDX x-ray source consists of a large-area (23×23) cm² tungsten anode (figure 1(b)). An electron beam is magnetically steered across the anode and dwells for 1.04 μ s at 100×100 positions, generating 10 000 photon beams. Subsequently, the photon beams interact with 20 μ m of niobium (Nb) and 5 mm of beryllium, which encloses the x-ray source vacuum. The anode is cooled with a 3 mm thick layer of water and the beam is further filtered through 0.4 mm of aluminum. The photon beams are highly collimated with a 20.4 mm thick brass collimator with 100×100 divergent holes converging toward the detector. The entrance and exit collimator hole diameters are 1.04 and 1.50 mm, respectively. The collimator is designed so that each of the 10000 beams covers a (5.5×5.5) cm² detector positioned on the central axis 150 cm away from the x-ray source. The entire source is enclosed in a 3.1 mm thick polyetherimide window. Two scanning modes are employed in the SBDX source. In the first, all 100×100 focal spots corresponding to 16.1×16.1 cm² FOV are used, however, the commonly used 11.4×11.4 cm² FOV uses only 71×71 focal spots. Moreover, a number of focal spots at the edges of the field are not used and the scan pattern has a squircle shape created by the intersection of a square and a circle. The ratio of the diameter of the circle to the width of the square is 1.09. In this work, the 11.4×11.4 cm² FOV has been modeled and validated.

The source can be operated at voltages ranging from 80 to 120 kV. The peak power is 24.3 kW, corresponding to an x-ray source current of approximately 200 mA for the 120 kV beam. Scanning speeds up to 30 frames s⁻¹ can be achieved.

2.2. Dose measurements

Three 8" × 10" EBT Gafchromic films (ISP, Wayne, NJ) were sandwiched between four blocks of solid water, and the composite phantom was positioned on a table 16.5 cm from the polyetherimide window protecting the vacuum (figure 2). The thicknesses of solid water slabs were 1, 5, 5, and 1 cm, resulting in measurement depths of 1, 6, and 11 cm with additional 1 cm of solid water for backscatter.

The phantom was irradiated with 120, 100, and 80 kV beams using 16.8 kW for 10, 15, and 20 min, respectively. The irradiation times were chosen to approximately deliver 1 Gy to the films. The films were calibrated in a 100 kV beam generated by an RT250 x-ray tube (Royal Philips Electronics, Eindhoven, The Netherlands) using an ionization chamber. A Farmer type ionization chamber model TN30013 (PTW, Freiburg, Germany) was inserted in a solid water phantom at 1 cm depth with 9 cm for backscatter and the dose to the chamber was measured. EBT Gafchromic films were placed at 1 cm depth in the same phantom and irradiated to known doses using the 100 kV beam. According to our analysis of the effective energies of the studied SBDX beams (35–40 keV) and published data (Sutherland and Rogers 2010), the difference in dose response from the three tube voltages is within 3%. The films were read out in a flat-bed scanner (Epson Perfection V500, Long Beach, CA) with 508 dpi resolution 24 h post irradiation for both the calibration curve and SBDX measurements. As a result, SBDX source dose distributions perpendicular to the beam axis at 1, 6, and 11 cm depths for 80, 100, and 120 kV were experimentally measured.

2.3. MC simulations

The MC model of the SBDX source was built using the EGSnrc/BEAMnrc (Rogers *et al* 2006) code and validated against film measurements in the DOSYZnrc (Walters *et al* 2007) code. Beams with 120, 100, and 80 kV x-ray source voltages were modeled in BEAMnrc (see section 2.3.2) and the full phase-space files scored just below polyetherimide window were used for calculations of dose distributions in a solid water phantom to reproduce the experimental measurements. Additionally, patient dose as a function of patient size was calculated using an elliptical phantom for all three beams. The dose distribution for the 120 kV beam was calculated for imaging the prostate.

2.3.1. Electron beam—Due to the design of the BEAMnrc code, each electron dwell position was modeled in a separate simulation and 71 × 71 input files were automatically generated in MATLAB (The Mathworks, Natick, MA). Parallel circular electron beams (ISOURCE = 0) of 10 μm diameter were incident on a SLABS component module (CM) representing the target. The first layer of the SLABS was modeled as a 20 cm thick slab of vacuum to make it possible to simulate all electron beam dwell positions by modifying the incident angle of the beam (UINC, VINC), so that the variation in the electron incidence angle is taken into account. Uniform bremsstrahlung splitting with Russian roulette and a splitting number of 200 was used. Additionally, the bremsstrahlung cross-section in tungsten was enhanced by a factor of 200 (Ali and Rogers 2007).

2.3.2. Collimator design—The collimator was modeled using the BLOCK CM. The BLOCK CM models a number of openings in a slab angled with respect to the central axis

toward a single focal spot on the axis. As mentioned above, each collimator hole of the SBDX source is circular and divergent. The circular shape of the collimator holes was approximated by eight-sided polygonal openings. The divergence of each collimator hole was approximated by using five layers of BLOCK CM.

The default number of openings in the BLOCK CM is 20, however, 10 000 openings were needed to model the entire collimator (figure 3(a)). Extensive memory usage was required to run a simulation with all collimator holes, which resulted in long simulation times. Therefore, only a portion of the collimator was modeled for each electron beam dwell position (figure 3(b)). For each dwell position, 11×11 collimator holes uniformly distributed around the dwell position were modeled. This resulted in a significant acceleration of the simulations with less than 0.2% difference in the number of photons in the phase-space files.

In order to simulate the squircle beam shape, only the dwell positions actually used in the SBDX source scanning were identified and the individual phase-space files were added using the addphsp program (Rogers *et al* 2006). The scatter plot of 120 kV photons just below the collimator is plotted in figure 3(c).

All 71×71 input files were automatically created in MATLAB. For each electron beam dwell position, 5×10^6 histories were simulated, resulting in 400 to 500 phase-space photons. The number of histories per electron beam dwell position was chosen so that the final phase-space files contained at least 10^6 photons for each tube voltage. The photons from the phase-space file scored just below the collimator were then propagated through the polyetherimide window modeled by the SLABS CM. Simulations were run for three x-ray source voltages: 80, 100, and 120 kV.

All low energy physics processes, such as electron impact ionization, bound Compton scattering, and Rayleigh scattering, were included in the simulations. The energy cutoffs PCUT and ECUT, at which particle transport stops, were set to 0.01 and 0.521 MeV, respectively. Cross-section data from the XCOM database (Berger *et al* 2005) were used in all simulations.

2.3.3. Film phantom dose calculations—All dose calculations were performed in the DOSXYZnrc code. Dose was calculated in a solid water phantom defined following the experimental setup discussed in the next section. The absorbed dose to the $30 \times 30 \times 12 \text{ cm}^3$ solid water phantom was calculated using an isotropic voxel size of 0.4 cm. The phase-space was positioned 16.5 cm above the surface of the phantom and the beam was parallel to the 12 cm axis of the phantom. The dose from 80, 100, and 120 kV beams was simulated. The ‘HOWFARLESS’ algorithm (Walters and Kawrakow 2007) was used to increase the simulation efficiency. The statistical uncertainties in high-dose regions were below 1% and they were below 1.5% in regions of 50% of maximum dose.

2.3.4. Exposure rate and half-value layer (HVL) calculations—The exposure rates for 80, 100, and 120 kV beams were calculated and compared to presented data (Speidel *et al* 2006b) due to the unavailability of the system. Note that the x-ray target in the current

system is 17 μm thick as opposed to the older version with a 14 μm target. Similar to the experimental setup, dose to air was modeled by MC according to the specification of the PTW N233612 chamber (PTW, Freiburg, Germany) using the CHAMBER CM in BEAMnrc. The 8.5 cm in diameter and 5 mm thick chamber was placed in air 25 cm away from the exit window to simulate the experimental conditions. The dose to the air in the chamber was converted into exposure expressed in roentgens (R) using the following:

$$X(R) = \frac{100 \times D_{\text{air}}(\text{Gy})}{0.876 \left(\frac{\text{rad}}{R} \right)},$$

where D_{air} is the mean dose to the sensitive volume of the chamber. The exposure rate was calculated by normalizing the exposure by the irradiation time and source current.

HVLs, a commonly used characteristic for description of kilovoltage beams, were calculated based on the simulated spectra (Verhaegen *et al* 1999). The thickness of aluminum that reduced the air kerma of the incident x-ray beam to 50% of its original intensity was found for all three energies. Additionally, the mean energies and effective energies derived from the calculated HVL values were derived for all three beams based on XCOM attenuation coefficients (Hubbell and Seltzer 1996).

2.3.5. Effect on patient size calculations—The dependence of skin dose on patient size was studied. The skin dose was calculated for a number of elliptical phantoms. The skin dose was normalized to the detector dose assuming that similar detector doses produced comparable image quality in terms of image noise. Soft tissue elliptical phantoms mimicked patients with minor axes ranging from 20 to 50 cm and major axes ranging from 26 to 65 cm (depicted with the ellipse in figure 1(a)). They were designed with uniform voxel size of 5 mm. The center of the elliptical phantoms coincided with the imaging isocenter. A 2 mm thick $5.5 \times 5.5 \text{ cm}^2$ CdTe detector was modeled as a single crystal 150 cm away from the source. Particles were transported through the detector with all accurate physics, and the dose to the detector was calculated as the mean dose to the entire detector crystal. The dose to the skin was calculated as the mean dose to the first layer of phantom voxels over an area of $5.5 \times 5.5 \text{ cm}^2$. The skin-to-detector dose ratio (SDDR) was determined for each phantom size using the three phase-space files with 80, 100, and 120 kV beams.

2.3.6. Patient dose calculations—Imaging dose was also calculated for the prostate within a patient. The patient was modeled so that the center of the prostate was at the imaging isocenter. For accurate dose calculations with kilovoltage beams, the patient CT data set was segmented into 37 tissue types (Bazalova and Graves 2011), which were defined based on effective atomic numbers derived from CT numbers. The density of each $(4.3 \times 4.3 \times 5.0) \text{ mm}^3$ voxel was calculated using a calibration curve for our GE LightSpeed PET/CT (GE Healthcare, Milwaukee, WI). The imaging dose was calculated and displayed for the 120 kV beam.

3. Results

The geometry of the SBDX source was modeled based on the manufacturer's specifications. The phase-space files scored below the polyetherimide window contained at least 10^6 particles for all three energies. The energy spectra of the 120, 100, and 80 kV phase-space file photons averaged over the entire source are presented in figure 4. The $K\alpha_1$, $K\alpha_2$, $K\beta_1$, and $K\beta_2$ lines of tungsten can be seen in the 57–69 keV energy range. Surprisingly, the Nb $K\alpha$ and $K\beta$ lines at 16.6 and 18.7 keV are present in the spectrum and they are not filtered by the 0.4 mm of aluminum. Ideally, these low-energy x-rays should be removed as they do not contribute to image formation but do increase the patient dose. The mean energies calculated from the MC simulated spectra are 46, 53, and 58 keV for the 80, 100, and 120 kV beams, respectively.

The 2D imaging dose distributions measured by Gafchromic films and calculated by MC simulations at 1 cm depth for 120, 100, and 80 kV are presented in figure 5. All dose distributions are normalized to 200 mA and 15 min irradiation time. The measured dose shows higher intensity in the left lower quadrant for all energies. This can be due to an imperfect scanning pattern of the electron beam or due to the misalignment of the collimator with respect to the electron beam.

In-plane (along x-axis) and cross-plane (along y-axis) beam profiles at 1, 6, and 11 cm depths for 120, 100, and 80 kV beams are plotted in figure 6. The in-plane simulated profiles match the measured ones to within 5%, with the exception of the 120 kV profile at 1 and 6 cm depths. The right portion of the in-plane measured profile was by 11% greater than the simulated one, which is possibly due to the misalignment of the collimator as discussed previously. The discrepancy between measurements and simulations in the cross-plane profiles is more pronounced. The differences are up to 15% in the left side of the cross-plane profiles at 1 cm depth. As the depth increases, the differences between the measured and calculated profiles decrease.

Central-axis depth dose curves for 120, 100, and 80 kV beams calculated with MC for a 15 min irradiation with 200 mA are shown in figure 7. Only three points for central-axis depth dose could have been extracted from the film measurements for each source voltage. The points are also plotted in figure 7. The differences between MC simulations and measurements are within 5% for all nine points. Exponential fits with R^2 larger than 0.99 calculated for the MC data are included in the figure.

MC simulated and measured exposure rates and HVLs are summarized in table 1. Despite the fact that measurements were taken from a different system with a thinner x-ray target, the differences between simulated and measured exposure rates are within 6.5%. With a thicker x-ray target, it is expected that the MC simulated beams would be harder, which is reflected by larger HVL values compared to measurements. MC-simulated HVL values are 8.5%–12.3% larger than the measured ones. The effective energies calculated from the MC simulated HVL values are 35, 38, and 40 keV for the 80, 100, and 120 kV beams. The results for MC simulations of a 14 μ m target are also included in table 1. The simulated

HVL values match measurements within 3.8% when the target thickness of the previous version of the system is used.

The dose to skin per unit detector-dose (SDDR) for the three x-ray source voltages and a number of imaging object sizes is shown in figure 8. For pediatric patients with thickness of 20 cm, the dose ratio ranges from 5 to 10 for the 120 and 80 kV beams, respectively. As patient size increases, the dose ratio increases accordingly. For large patients with 50 cm anterior–posterior thickness, the SDDR increases significantly and corresponds to approximately 1×10^3 , 2×10^3 , and 5×10^3 for the 120, 100, and 80 kV beams.

Patient dose distributions for a single pelvic image acquired with 120 kVp, 23 kWp, and a frame rate of 15 frames s^{-1} in all three cardinal planes are presented in figure 9. The maximum skin dose is 200 μGy and the maximum dose to the pubic bone is 125 μGy . For lower source voltages, the bone dose would increase due to the larger contribution of photoelectric effect. The patient exit dose is approximately 5 μGy or 2.5% of the skin dose.

4. Discussion

A MC model of the SBDX source has been built. Modeling of the collimator is a challenging task, as it consists of a large number of divergent circular collimator holes that are focused at the detector. Such geometry is not available in the BEAMnrc code. The collimator was approximated by five layers of BLOCK CMs with openings in the shape of eight-sided polygons. This approximation was sufficient due to the small thickness of the collimator and small sizes of collimator holes. However, for larger dimensions, more BLOCK CM layers and more vertices approximating circular holes might be needed. For convenience, a flexible MATLAB routine generating BEAMnrc input files was coded. The number of BLOCK CMs, collimator thickness, number of collimator holes and their spatial distribution, number of vertices approximating the circular holes, source-to-detector distance and the detector size are entered as parameters in the routine. The routine is useful for optimizing the geometry of the SBDX source. As a result, various source designs and system geometries can now be easily investigated by means of MC simulations.

While simulating the x-ray source, we discovered that the electron beam should not be hitting the target at the center of collimator holes for the highest beam output. The target thickness needs to be taken into account due to the angular distribution of bremsstrahlung. To increase the output of the highly collimated SBDX source, the focal spot locations are calculated by projecting the collimator hole axes onto the front of the target (demonstrated in figure 10). This effect can also be conveniently studied using the MATLAB routine.

The MC simulated energy-spectra reveal two Nb peaks at 16 and 18 keV. These photons have a large linear attenuation coefficient for soft tissue, making them impractical for imaging. The Nb peaks should therefore be removed from the spectrum to decrease patient skin-dose. The peaks can be removed by hardening the beam using a 2.4 mm Al filter, which is a typical filter thickness for a 120 kV x-ray imaging source. With the additional filtration, the 18 keV peak will be attenuated to 5% of its original value.

We also investigated the shape of energy spectrum as a function of the position within the large-area source. We plotted spectra from the center of the source averaged over a circular area of 4 cm in radius (inside yellow line in figure 3(c)) and spectra averaged over an annular region between 4 and 8 cm (between yellow and green lines in figure 3(c)). The spectra did not differ significantly. It could be concluded that there was no distortion in spectral shape as a function of x-ray beam position.

HVL differences between MC simulations using the specification of the previous model of the SBDX source utilizing a thinner target and published measurements (Speidel *et al* 2006b) were within 3.8% for all studied x-ray source voltages. This result is used to validate the SBDX source model.

There is a shift of beam intensity toward the lower left quadrant in the measured dose distributions (figure 5, top row). According to the manufacturer, this could be due to alignment or to beam-shape effects. The electron beam becomes defocused and elliptical in cross-section as it is deflected away from the tube axis. The defocusing can be compensated, but that compensation may not be entirely accurate. The beam defocuses as it is deflected primarily because the total distance over which the beam travels increases approximately as the hypotenuse of the triangle formed by the deflected beam, the original axis of the beam, and the surface of the x-ray target. Therefore the image distance is increased and the lens needs to have a longer focal length. The source provides this increase by use of an additional focus coil winding that is dynamically changed as the beam is scanned across the surface of the target. The magnitude of this dynamic focus change is adjustable as a function of deflection and this adjustment is subject to errors in calibration. Magnets with quadrupolar fields compensate the elliptical beam cross-section. This compensation is also dynamic and the result of calibration and therefore subject also to error. There is another alignment-related effect seen in the measured dose distributions in figure 5. On one of the edges corresponding to the beginning of the scan, undulations in beam intensity are observed. This is caused by imperfect initiation of the fast scan of the electron beam at the end of a row, i.e. after the ‘fly-back’. The effect usually disappears approximately one quarter of the distance across the scan, and it is observed as a high intensity stripe across the film images in figure 5, top row.

The depth dose curves presented in figure 7 contain only nine measurement points due to the experimental setup, in which three films were placed perpendicularly to the beam axis so that 2D dose distributions were measured. Depth dose curves could be measured at high spatial resolution by sandwiching the film between slabs of solid water and orienting them parallel to the beam axis. However, phantom placement cannot be accurate, because no guidance lasers pinpointing the major imaging axes were available in the current system. Field corners can be used for alignment of films in the present experimental setup.

A number of beam energies were calculated for each studied x-ray source voltage. The mean and effective energies were extracted from the MC simulated spectra (figure 4) obtained just below the polyetherimide window enclosing the source. As expected, the mean energies are larger than the effective energies due to the energy dependence of the linear attenuation coefficient. Figure 7 demonstrates attenuation of the inverse-geometry SBDX source beam.

The depth dose curves reflect attenuation of each of the 71×71 sources as well as the convergence of the entire x-ray source beams toward the detector (figure 1). The dose fall off is not as fast as for a single beam with the energy spectrum presented in figure 4 as the edges of the imaging field approach the central axis at larger depths.

Figure 8 demonstrates the change of SDDR as a function of beam energy. It should be noted that in our simulation setup, the skin thickness is considered to be 5 mm, which is larger than the usual thickness of 2–3 mm. As such, the dose to the skin is underestimated by approximately 2.5%. The simulation data in figure 8 were fit with exponential curves and the R^2 values for all fits are larger than 0.98.

It is assumed that equal dose to the detector will result in similar image quality when patient dose as a function of patient size is investigated. This is not necessarily true, as higher image contrast can be expected for lower tube voltages. The results presented in figure 8 are however a good estimate of patient dose as a function of patient size and tube voltage. To fully understand imaging quality as a function of tube voltage for the current SBDX system, the presented MC model needs to be extended to incorporate the detector's impact. Ideally, dose deposition in each detector element should be simulated. As a result of the simulation, detector signal will be generated and processed. Finally, full tomographic SBDX images of any object will be reconstructed and their properties can be studied.

The patient imaging dose calculation demonstrates that it is possible to quantify 3D dose to a patient with our MC model. Though, imaging skin dose can be measured with dosimeters such as thermoluminescent dosimeters or even ionization chambers (Rybicki *et al* 2002), it is noteworthy that organ doses can only be calculated based on MC simulations. The presented model can be used to assess organ doses for imaging with the SBDX source. In addition, the results can be compared to conventional x-ray imaging systems, which have a small x-ray source and large-area detector.

5. Conclusions

A Monte Carlo model of the scanning beam digital x-ray source (SBDX) has been built and validated against Gafchromic film measurements. A general agreement of less than 5% between simulated doses and measured doses was found. The model was used to estimate patient skin dose as a function of patient size for three x-ray source voltages. Moreover, the three-dimensional dose distribution for imaging the prostate in a patient was quantified. The entire SBDX imaging system can be simulated by further expanding the current Monte Carlo model for the pixelated x-ray detector. It will thus be possible to study SBDX images in terms of contrast-to-noise ratio and spatial resolution. The extended model will be a useful tool for optimization of the SBDX imaging system, namely the collimator and the detector design.

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Figure 1.

A schematic drawing of the imaging geometry (a) with the SBDX source (b). The source is vacuum tight and enclosed with a 3.1 mm thick polyetherimide window.

Figure 2.
Experimental setup for measurements of imaging dose with Gafchromic EBT films at 80, 100, and 120 kV.

Figure 3.

Full collimator geometry with 100×100 collimator holes (a) and with only 121 holes for more efficient simulations (b). The beam is incident on the central collimator hole in (b). The distribution of 120 kV photons in the phase-space file scored below the collimator (c) with indicated circular areas for energy spectrum analysis.

Figure 4. MC-modeled energy spectra exiting the polyetherimide window for 120, 100, and 80 kV beams. The K-lines of Nb in the target at 16 and 18 keV are apparent.

Figure 5.

Film (upper row) and MC (lower row) dose distributions at 1 cm depth for the 120, 100, and 80 kV beams.

Figure 6.

In-plane (a)–(c) and cross-plane (d)–(f) beam profiles for 120 kV (a), (d), 100 kV (b), (e) and 80 kV (c), (f) at 1 cm (solid lines), 6 cm (dashed lines) and 11 cm (dotted lines) depth for film measurements (lines) and MC simulations (markers).

Figure 7.
Central axis depth dose for a 15 min irradiation time with 200 mA measured by film (markers) and calculated by MC (lines) for 120, 100, and 80 kV.

Figure 8.

Skin to detector dose ratio (SDDR—dose to skin D_{skin} per unit of detector dose $D_{detector}$) as a function of phantom size and x-ray source voltage. Exponential fits to data points are included.

Figure 9. Patient pelvic imaging dose distribution from an image acquired with 120 kVp, 23 kWp, and 15 frames s^{-1} overlaid on axial (left), coronal (upper right), and sagittal (lower right) CT images.

Figure 10.

Demonstration of electron beam positioning with respect to collimator holes for a higher output of the SBDX source. The electron beam should not be incident on the target just above the center of the collimator hole (black). Higher x-ray output is achieved when the electron hits the anode at the intersection of the collimator hole with the anode (red).

Table 1

Simulated and measured exposure rates and HVL values.

	Exposure rate (mR min ⁻¹ mAs ⁻¹)		HVL (mm Al)		
	MC simulated	Measured ^a	MC simulated	Measured ^a	
80 kV	24.5	23.0	3.29	2.92	3.03
100 kV	38.2	37.5	3.98	3.48	3.58
120 kV	53.0	56.5	4.56	3.98	4.06

^aMeasurements are from a previous version of the system with a 14 µm target, simulations are for a 17 µm target.

^bHVLs were also calculated by MC for a 14 µm target.