Characterization of an x-ray source with a partitioned diamond-tungsten target for electronic brachytherapy with 3D beam directionality

Daniel S Badali¹, Guillaume R Plateau¹, Christopher W Ellenor¹, Chwen-Yuan Ku¹, Petre Vatahov¹, Jane Esterline¹, Brian P Wilfley¹, Christopher R Mitchell¹, Kalman Fishman¹,² and Tobias Funk¹,³

¹ Triple Ring Technologies, Newark, CA United States of America
² Sensus Healthcare, Inc., Boca Raton, FL, United States of America
³ Author to whom any correspondence should be addressed.

E-mail: kal@sensushealthcare.com and tfunk@tripleringtech.com

Abstract

The Sculptura™ is a new high-dose-rate electronic brachytherapy system developed by Sensus Healthcare. By combining a steerable electron beam with a partitioned diamond-tungsten x-ray target, the x-ray source of the Sculptura™ is capable of producing highly anisotropic dose distributions, thus achieving true 3D beam directionality. This article reports the spectral and dosimetric characterization of the Sculptura™ x-ray source through a combination of measurements and Monte Carlo simulations for operating points between 50–100 kV. Excellent agreement (~5% discrepancy) between the simulations and measurements was obtained for in-air dose rate characterization. The validated simulations were then used to calculate the dose distribution in water. Dose rates of >2 cGy/min/μA can be produced at 100 kV, thus delivering 10 Gy in 1 min for typical operating conditions. The dose distributions are sharply peaked, with a full-width at half-maximum azimuth of about 100°.

1. Introduction

Electronic brachytherapy is a form of high-dose-rate brachytherapy in which a miniature x-ray tube is inserted into the surgical cavity to perform intracavitary or intraoperative radiation therapy (IORT) (Eaton 2015, Ramachandran 2017). Conventional electronic brachytherapy, much like radionuclide brachytherapy, produces an isotropic dose distribution (Eaton 2015, Ramachandran 2017), which limits the ability to design treatment plans which spare healthy tissue while reaching the clinical dose targets. Although intensity-modulation (Shi et al 2010) is currently used in clinical practices to produce conformal dose distributions by tuning the dwell positions, dose rates, and penetration depths, the shape of the dose distributions remains axially symmetric.

In recent years, rotating-shield brachytherapy (Yang et al 2013) has been demonstrated to produce anisotropic dose distributions by partially blocking the radiation field in a controlled fashion. By varying the directionality of the dose and the dwell time at each shielding position, rotating-shield brachytherapy has been shown to be able to produce arbitrary conformal dose distributions (Yang et al 2013).

As an alternative to rotating-shield brachytherapy, this manuscript introduces the novel design of the Sensus Healthcare Sculptura™, a device which can produce anisotropic dose distributions for conformal radiation therapy. To achieve this, the device contains a scanning electron beam source which impinges on a partitioned diamond-tungsten target which can produce 18 unique anisotropic dose distributions for a range of acceleration potentials.

To characterize this novel electronic brachytherapy source, the absorbed dose-to-tissue produced by the device at all operating points must be established. Because human tissue is largely composed of water, absorbed dose-to-water is considered to be an appropriate proxy for absorbed dose-to-tissue (Gurjar et al 2014). However, we note that inhomogeneities in tissue composition can cause deviations from homogeneous water models (Spiers 1946, Sethi et al 2015).
At the energies considered here (50–100 kV), ionization chambers are typically the most straightforward and accurate technology to provide calibrated measurements of radiation dose. Ionization chambers measure air kerma, which is numerically equivalent to the absorbed dose-to-air for the energies considered here (International Atomic Energy Agency 2007).

The National Institute of Standards and Technology (NIST) has not, however, defined a primary standard for dose-to-water in the energy range in use (Fulkerson et al 2014), and determination of patient dose strictly through measurement relies substantially on correction factors that interpret the application of primary air kerma standards (see International Atomic Energy Agency (2000), for example). To address the challenges of measuring absorbed dose-to-water, we use a combined approach of Monte Carlo (MC) simulations and experimental measurements to characterize the spatial dose profiles produced by the device presented here. This approach consists of two steps: first, establish the ability of MC simulations to accurately reproduce experimental dose-to-air measurements through a rigorous validation procedure. Once validated, the second step is to use the MC simulations as a predictive tool and simulate the spatial profiles of the absorbed dose-to-water.

Paramount to this approach is establishing the ability of MC simulations to accurately reproduce the dose measured by validated laboratory methods. Although there is a body of literature that establishes the validity of MC simulations for dose simulations in the field of radiation therapy (Carrier et al 2004, Rodrigues et al 2004, Poon and Verhaegen 2005, Amako 2005, Guimaraes et al 2008, Liu et al 2008, Carver et al 2015), we have striven to establish this equivalence in the context of the Sculptura™ device under consideration here.

The validation procedure consists of a series of logical steps to validate the MC simulations. First, half-value layer measurements are replicated in simulation to ensure that the MC simulations are using a correct representation of the spectrum of the x-rays produced by the device. Second, it is established that MC simulations can produce dose that is equivalent to the measured dose on an absolute scale. Finally, measurements of the spatial dose profiles in air are compared to their simulated counterparts. These steps will serve to establish that the MC simulations can faithfully reproduce the geometric dose distributions in air.

Finally, we note that the Sensus Healthcare Sculptura™ device has received 510(k) clearance from the FDA (K number K182641), and we recommend all industry guidelines (e.g. Nath et al (2016) and Tom et al (2019)) be followed when using the device.

2. Materials and methods

2.1. Sensus Healthcare Sculptura™ x-ray source

The x-ray source of the Sensus Healthcare Sculptura™ system consists of a miniature electron gun, deflection and focusing electron optics, and a partitioned diamond-tungsten target for x-ray production. The electrons are thermionically emitted from an yttria-coated iridium disc cathode and accelerated in a single stage. For this work, results are reported for potentials of 50 kV, 85 kV, and 100 kV. The high-voltage power supply includes an emission current control (ECC) circuit to maintain a constant emission current (and thus a constant x-ray output). The ECC achieves an emission current stability of <0.2%.

A solenoid magnetic lens is used to focus the electron beam, and a second magnet, consisting of windings around a square steel yoke, is used to deflect the electron beam perpendicularly. The electrons propagate down the drift tube and impinge on the partitioned x-ray diamond-tungsten target. The distal 2 cm of the drift tube is constructed of a silicon carbide (SiC) tube mounted to the stainless-steel drift tube. The purpose of the SiC component is to allow radiation to emerge from the drift tube from behind the target, thus acting as an x-ray window for the x-ray backflow from the target. The entire drift tube is encased in an external sheath which provides a channel for coolant flow over the external surface of the diamond window. In the current design, the cooling jacket is 221.3 mm long and 27 mm in diameter, although a new version that is 246.15 mm long and 17 mm in diameter is currently being produced.

The target consists of a thin tungsten layer deposited on the interior surface of the diamond window. The window forms a vacuum seal with the drift tube. A molybdenum partition structure is mounted on both sides of the diamond window. As shown in figure 1, the partition structure consists of 6 symmetric fins which divide the target into sextants. Within each sextant, the deflection coil can position the electron beam at three radial positions: 3 mm (innermost—IM), 5 mm (medial—M), and 7 mm (outermost—OM) from the center of the target. The molybdenum partition structure acts as a collimating septa, such that by aiming the electron beam at different target positions relative to the partition structure, the resulting x-ray beam can be collimated in a spatially variable way. Figure 1(b) illustrates this basic concept. Focusing the electron beam near the collimating septa (e.g. the 3 mm (innermost) radial position) will form a beam with a narrow angular range, whereas focusing the electron beam away from the center (e.g. the 7 mm (outermost) radial position) will irradiate a broader swath of tissue.

2.2. Experimental measurements

The Sculptura™ x-ray source was mounted horizontally for characterization of its radiation output. The tube’s focal spot was imaged through a 100 μm pinhole onto a CMOS detector (Xineos with a 99 μm pixel size, Teledyne DALSA, Waterloo, Ontario).
Dose rate measurements were performed with an Exradin™ A20 ionization chamber with a MAX-4000™ Plus Electrometer (Standard Imaging, Middleton, WI). The Exradin™ A20 is a low-energy parallel plate chamber designed for soft x-rays and was outfitted with a 120 kV buildup cap for absolute measurements. The ionization chamber was calibrated at three different American Association of Physicists in Medicine (AAPM) Accredited NIST-Traceable beam qualities by the University of Wisconsin Accredited Dosimetry Calibration Laboratory (UWADCL) to cover the expected range of operation of the Sculptura™ device (Beam Codes UW50-L, UW80-L, and UW100-L). Measurements with the ionization chamber were always performed 5–30 s after initiating the x-ray exposure to allow the ECC circuit to stabilize the radiation output. During the stabilization time we observed deviations up to 25% in the radiation output, although future designs will have an improved stabilization time on the order of hundreds of milliseconds.

At all accelerating potentials the emission current setpoint was chosen to give an electron beam power of 50 W.

2.2.1. Half-value layer measurements

The aluminum half-value layer (HVL1), in combination with the x-ray tube potential, is typically reported to characterize the beam quality of an x-ray source (Ma et al 2001). The first half-value layer (HVL 1) is defined as the thickness of aluminum at which the air kerma is attenuated to half the value of that in the unfiltered beam (Ma et al 2001). Similarly, the second half-value layer (HVL 2) is defined as the additional thickness of aluminum which must be added to decrease air kerma to a quarter of that in the unfiltered beam. Because air kerma and dose rate in air are equivalent for the energies considered here (International Atomic Energy Agency 2007), the HVLs are instead interpreted in terms of the attenuation of the measured dose rate.

The experimental setup used to measure the half-value layers (figure 2(a)), was guided by AAPM Report No. 25 (Lin et al 1988). Owing to spatial constraints in the laboratory, the dimensions were slightly smaller than those recommended in the report. The thickness of aluminum between the lead aperture and the ionization chamber was varied, and the dose rate measured by the ion chamber was recorded. The electron beam was positioned at the 5 mm (medial) radial position in one of the sextants, and >99% purity aluminum (alloy 1100, McMaster-Carr) was used to attenuate the radiation. Measurements were repeated three times for each thickness of aluminum.

The experiment (or replicated simulation) resulted in an attenuation curve (dose rate as a function of aluminum thickness), which was fitted to the following empirical equation:

\[
\hat{D}(x) = A_1 \exp (-B_1 x) + A_2 \exp (-B_2 x)
\]

Figure 1. (a) Simplified drawing of the Sensus Healthcare Sculptura™ x-ray source, with an insert showing the structure of the partitioned diamond-tungsten target. (b) Top-down view of the partitioned diamond-tungsten target, illustrating the three radial positions of the steered electron beam within each sextant. The collimating effect of the partition is illustrated.
2.2.2. Dose mapping measurements

To measure the spatial dose profiles, the ionization chamber was mounted to an acrylic fixture attached to the end effector of a 7-axis KUKA LBR iiwa 14 R820 robot (figure 2(b)). The robot was used to position the ionization chamber at a fixed radial distance of 12.5 cm from the center of the diamond-tungsten target and to orient the chamber so that it was always facing the center of the target. Measurements were spaced by approximately 10 degrees in the polar direction, and 15 degrees in the azimuthal direction. The offset between the robot’s coordinate system and the x-ray source’s coordinate system was measured with a caliper. The result of this measurement was a series of circular trajectories at different $z$ distances from the target plane, which was designated $z = 0$. Positive $z$ values indicate planes in front of the x-ray target, and negative $z$ values indicate planes behind the target. Over 520 measurement positions were used.

2.3. Monte Carlo simulations

In this study, Monte Carlo simulations were performed using the Geant4 toolkit, version 10.4p2 (Agostinelli et al 2003, Allison et al 2006, 2016). Geant4 has been extensively validated in the literature for the energy ranges considered here (see, for example Carrier et al (2004), Poon and Verhaegen (2005) and Amako (2005)). In particular, it has been found that the dose simulated by Geant4 agrees with measured dose using ionization chambers within a few percent (Rodrigues et al 2004, Guimaraes et al 2008, Liu et al 2008, Carver et al 2015).

In designing the Monte Carlo simulations, we strove to replicate the conditions of the laboratory measurements as faithfully as possible. In order to capture the complicated interactions that arise during x-ray production, the simulations started with the electrons in the x-ray source. The electrons were modelled as a parallel, monoenergetic beam propagating down the drift tube and impinging on the x-ray diamond-tungsten target. The transverse spatial distribution of the electrons was sampled from the measured focal spot distribution from the pinhole imaging experiment described above.

All components of the x-ray source were modelled with the G4TessellatedSolid class, with the tessellated geometry coming from STL files (3D Systems, Inc. 1989) produced from the CAD models of the device. The material of each component was derived from the CAD model.

The physics of the simulation were modelled using the Geant4 EM standard physics list option 4, which was specially designed for highly accurate electron tracking at the energies considered here (Geant4 Collaboration 2017). The following types of interactions were included: photoelectric absorption, Compton scattering,
Rayleigh scattering, gamma conversion, pair production, multiple scattering, ionization, and bremsstrahlung. Note that, although they were included in the simulation, gamma conversion and pair production, as well as all positron processes never occur because all energies remain below the pair production threshold of 1.022 MeV.

In sections 3.3 and 3.4 the simulations were used to calculate the 3D distribution of dose, in which case 1 mm $\times$ 1 mm $\times$ 1 mm voxels were used.

### 2.3.1. Variance reduction—Bremsstrahlung splitting

Because of the inefficiency of x-ray production from the bremsstrahlung interactions between the electrons and the target, uniform bremsstrahlung splitting was employed as a variance reduction technique. Bremsstrahlung splitting is a procedure by which $N$ photons are generated from a single bremsstrahlung event, each with a statistical weight of $1/N$ (Kawrakow et al. 2004). This technique is natively supported in Geant4, and a splitting factor of $N = 1000$ was used in these simulations.

### 2.3.2. Variance reduction—phase-space scoring

When replicating the geometry of the HVL experiment shown in figure 2(a), the statistics of the simulation were very low, even with the Bremsstrahlung splitting discussed above. This is likely due to the small collecting volume ($<0.1$ cm$^3$) of the ionization chamber, and the divergence of the radiation output. To compensate for this, we employed a variance reduction approach in which the phase-space data (position, direction, and energy) of each photon exiting the x-ray source’s cooling jacket was written to a file. Once the photon’s data was saved, the photon was ‘annihilated’ in the simulation. We note that the storing of phase-space data has been used by other authors (Poon and Verhaegen 2005, Liu et al. 2008) to improve simulation times.

To replicate the HVL experiments, the spectrum of photons headed towards the ionization chamber was calculated from the phase-space data. Although the ionization chamber subtends a cone with a half-angle of $\sim 0.3^\circ$, a half-angle of $5^\circ$ was used to calculate the spectrum to improve the statistics. This approximation was justified by examining the phase-space data and noting the relative insensitivity of the spectrum to the emission angle (for the small range of angles subtended by the ion chamber). The spectrum was then smoothed using a 5th degree Savitzky–Golay filter (Savitzky and Golay 1964). This approach dramatically reduced the variance in the simulation because the spectrum is broad and featureless (except for the fluorescence lines), and so does not require as many photons to sufficiently describe its shape.

The smoothed spectrum was then attenuated through the amount of air and aluminum used in the HVL experiment. The attenuation coefficients were taken from NIST Report 5632 (Hubbell and Seltzer 1995) and the attenuation calculations were performed in MATLAB$^\text{®}$ Release 2017a (The MathWorks, Inc., Natick, Massachusetts).

A second Monte Carlo simulation was performed to calculate the amount of dose deposited into the ionization chamber. This simulation consisted of propagating x-ray photons into a cylinder of air (which represents the ionization chamber), and calculating the total dose deposited into the air volume. The energy of each photon was sampled from the attenuated spectrum, calculated in the previous step. The x-ray photons were represented as a parallel beam uniformly distributed over the circular entrance to the air cylinder. This simulation was extremely fast, taking only a few minutes for each aluminum thickness to get sufficient statistics ($<1\%$ statistical uncertainty) on a standard PC machine.

The output of the final step in the simulation procedure described above is the dose (in Gy) per x-ray photon for each kV and each thickness of aluminum filtration. To convert this to the same units as the measured dose rate (Gy s$^{-1}$), the following conversion was applied:

$$\text{dose rate (Gy s}^{-1}\text{)} = \frac{\text{dose (Gy photon}^{-1}) \times t_{\text{Al}} \times \varepsilon (\text{kV}) \times \text{emission current used for measurement}}{1.602 \times 10^{-19} \text{coloumbs/electron}} \quad (2)$$

where $t_{\text{Al}}$ is the transmittance through the air and aluminum, and $\varepsilon$ (kV) is the efficiency of the ion chamber collection, defined as the number of photons from the phase space which are projected to hit the ionization chamber per electron.

Finally, saving the phase-space at the boundary of the x-ray source’s cooling jacket has the benefit that the surrounding medium can easily be changed between air and water without needing to rerun the computationally expensive simulations starting from electrons. As such, the dose distribution in water was simulated starting from the photon phase-space data into a homogenous volume of water. The dose was scored in 1 mm $\times$ 1 mm $\times$ 1 mm voxels, and more than 5 cm of backscattering material was included beyond the scoring volume as per AAPM recommendations (Rivard et al. 2004). The simulated data was then smoothed with a three-dimensional $7 \times 7 \times 7$ voxel median filter to reduce statistical noise.

### 2.4. Uncertainty analysis

An uncertainty analysis is presented for all the measurements and calculations performed in air (and the simulations performed in water), as recommended by the 2004 AAPM TG-43U1 report (Rivard et al. 2004) and
expanded upon in a subsequent report (DeWerd et al 2011). This analysis accounts for all Type A (statistical) and Type B (systematic) uncertainties and follows the laws of propagation of uncertainty (Taylor and Kuyatt 1994) by adding the relative uncertainties (%) in quadrature. The resulting combined uncertainty $\sigma_{\text{c}}(\%)$ is reported as expanded uncertainty $k\sigma_{\text{c}}$, where $k = 1$ defines a confidence interval of 68% and $k = 2$ defines a confidence interval of 95%. The AAPM recommends reporting $k = 2$ uncertainties (DeWerd et al 2011), and this convention will be followed for all measured and simulated data presented in the following sections.

The total percent uncertainty $\%\sigma_{D}(k = 1)$ in the measured or simulated dose rate $D$ is given by Rivard et al (2004):

$$\%\sigma_{D}^2 = \sum_{i=1}^{N} \left( f_i \times \%\sigma_{x_i} \right)^2$$

(3)

where the sum runs over each of the $N$ quantities $x_i$ that influence the dose rate, $\%\sigma_{x_i}$ is the percent uncertainty ($k = 1$) in the quantity $x_i$, and $f_i$ is the relative uncertainty propagation factor for $x_i$. It is expressed as Rivard et al (2004):

$$f_i = \frac{x_i}{D} \times \frac{\partial D}{\partial x_i}$$

(4)

2.4.1. Measurement uncertainties

A number of factors contribute to the uncertainty in the measured dose rate. The uncertainties in the calibration of the ionization chamber and electrometer were 0.5%, according to the UWADCL’s calibration report. Precision in the source-detector position and transverse alignment of the ionization chamber were ±0.5 mm. The influence of these geometrical misalignments on the dosimetric uncertainty was established by performing parametric Monte Carlo simulations with the ionization chamber offset by ±1 mm in all dimensions. The emission current of the x-ray source (which the dose rate is proportional to) was measured to have an uncertainty of 0.2%. The output of the high voltage power supply was stable to 0.05%, and so was neglected. According to the manufacturer, the tolerance in the thickness of the aluminum plates was ±10%. By examining the slope of the measured dose rates as a function of aluminum thickness (figure 3 below), the corresponding uncertainty propagation factor ranged from about −0.14 to −1.2, depending on the thickness and the tube potential. A mean value was −0.78 was used for propagating the uncertainty. When all sources of uncertainty were appropriately added in quadrature, the resulting total uncertainty was 2.7% ($k = 1$), with an expanded uncertainty of 5.4% ($k = 2$). A summary of all these values can be found in table 1. These uncertainties are comparable to those reported for other electronic brachytherapy devices with similar energies (Rivard et al 2004).

2.4.2. Calculation uncertainties

The uncertainties budgeted for the Monte Carlo simulations include the statistical, geometrical, and cross-section uncertainties. Other factors, such as the composition of the partition structure and the alloy used for the attenuation measurements were investigated and found to have negligible impact on the dose rate. The estimated dosimetric impact of source geometry uncertainties due to variation in cooling water and drift tube
2.4.3. Uncertainties in the half-value layers

Because the HVLs were calculated by fitting the dose rate to (1), care must be taken to relate the uncertainty in the dose rate to the uncertainty in the HVL. To do so, the dose rate of each point in the attenuation curve was randomly varied, with the perturbation chosen from a normal distribution with a specified percent standard deviation. For each chosen standard deviation in the dose rate, 1000 realizations were calculated, resulting in a family of attenuation curves. The HVLs were calculated for each curve, and standard deviation of the HVLs over the family of curves was calculated. The procedure was repeated for a range of percent dose rate uncertainties, and it was found that the uncertainty in the HVL was linearly proportional to the uncertainty in the dose rate. The proportionality constants are listed in table 3. These values were multiplied by the expanded uncertainties in the dose rate (listed in table 1 for the measurements and table 2 for the simulations) to obtain the expanded uncertainties in the HVLs.

3. Results

3.1. Half-value layers

A comparison between the measured and simulated dose rates for various thicknesses of aluminum attenuation is shown in figure 3. The absolute magnitude of the simulated data has been scaled to the measured dose rates,
so that the relative agreement can be compared. It can be seen that the agreement between the measured and simulated values is quite good, with the simulations reproducing the shape of the attenuation correctly.

The resulting HVLs are summarized in Table 4. The reported uncertainties are the $k = 2$ values described in section 2.4.3. According to the criterion defined in ISO 4037-1, two x-ray beams are considered to be the same quality if the first and second HVLs agree within $\pm 5\%$ (ISO 4037-1:1996). For all tube potentials, the differences listed in Table 4 are less than 5%, except for the 2nd HVL at 100 kV, which is $\sim 10\%$. However, because the uncertainties are relatively large, it can be concluded that the ISO 4037-1 criteria is met, and that the simulations adequately replicate the experimental beam qualities. Similar agreement has been found between ion chamber measurements and Geant4 in characterizing the HVLs of an electronic brachytherapy source (Liu et al. 2008).

In light of this conclusion, the simulations were used to analyze the spectra produced by the x-ray source. The unique design of the partitioned diamond-tungsten target (see figure 1) results in x-rays being produced in the ‘forward’ and ‘backward’ directions relative to the x-ray target plane (as defined in figure 2(b)). The spectra in these directions are shown in figure 4(a). The forward direction has more low energy content in the spectrum, whereas the backward direction resembles a filtered version of the forward spectrum. This can be attributed to the additional layer of silicon carbide that the ‘backward’ photons must penetrate through, which acts as a filter. The additional filtration of the silicon carbide layer is only present in the ‘backward’ direction, which explains why the ‘backward’ spectrum is harder than the ‘forward’ spectrum. Additionally, there is an intrinsic difference between ‘forward’ and ‘backward’ radiation due to the angular dependence of bremsstrahlung production. Although only the data for 85 kV is shown, similar trends were observed at 50 kV and 100 kV.

Figure 4(b) compares the spectra produced at the three different radial positions. There is a significant increase in the molybdenum K-characteristic radiation at 3 mm, which is expected since the x-ray focal spot is much closer to the molybdenum partitions. The shape of the remainder of the spectra is the same for all radial positions. Again, similar trends were observed at all tube potentials and also in the backward direction.

### Table 4. Measured and simulated first and second half-value layers. The reported uncertainties are the expanded values ($k = 2$) described in section 2.4.3.

| Tube potential (kV) | Measured       | Simulated       | Difference (%) |
|--------------------|----------------|----------------|----------------|
| 50                 | 1.15 ± 0.14    | 1.15 ± 0.21    | −0.05          |
| 85                 | 1.52 ± 0.31    | 1.51 ± 0.45    | −1.0           |
| 100                | 1.68 ± 0.36    | 1.60 ± 0.54    | −4.7           |

| Tube potential (kV) | Measured       | Simulated       | Difference (%) |
|--------------------|----------------|----------------|----------------|
| 50                 | 2.82 ± 0.21    | 2.72 ± 0.31    | −3.3           |
| 85                 | 4.40 ± 0.34    | 4.26 ± 0.50    | −3.2           |
| 100                | 5.10 ± 0.37    | 4.59 ± 0.55    | −10            |
3.2. Absolute dose

As discussed in section 3.1, the simulated attenuation curves in figure 3 were scaled to the measured dose rates, with a single scaling constant used for each source potential. In all cases, the absolute value of the simulated dose rate was lower than the corresponding measured dose rate. This discrepancy is attributed to a systematic error that results from using the emission current reported by the high voltage power supply when performing the unit conversion using (2). In (2), the numerator in the final term should in fact be the beam current used for the measurement; that is, the current of the electrons reaching the target. Instead, the power supply measures the emission current, which is the current of electrons leaving the cathode. The two are related by the following equation:

$$\text{beam current} = \beta (\text{kV}) \times \text{emission current}$$

(5)

where $\beta (\text{kV}) \leq 1$ represents the transmission of the electrons from the cathode to the target, and accounts for losses due to, for example, the electron optics.

The x-ray source potential-dependent transmission factor $\beta (\text{kV})$ was calculated by taking the ratio of the average simulated and experimental attenuation data for each kV, and is exactly the scale factor that was used in plotting figure 3. These values are summarized in table 5, which shows that the transmission factor decreases with the x-ray source potential, a trend that has been found in other electron guns with similar designs (Kimball Physics, Inc. 2017).

| Tube potential (kV) | Beam current/emission current, $\beta$ (kV) |
|---------------------|---------------------------------------------|
| 50                  | 0.93                                        |
| 85                  | 0.84                                        |
| 100                 | 0.81                                        |

3.3. Spatial dose distributions in air

Figure 5 shows the measured and simulated circular dose profiles at 100 kV in air, measured along different lines of latitude around the source (with $z = 0$ defined in the target plane, as illustrated in figure 1). The offset between the robot’s and x-ray source’s coordinate systems was optimized by minimizing the squared difference between the measured and simulated circular profiles, with the measured offset used as a starting guess. The measured and optimized offsets differed by about 4 mm, 0.6 mm, and 1 mm for the X, Y, and Z axes, respectively. It was found that improved agreement between the measurements and simulations could be obtained by increasing the electron transmission factors for 85 kV and 100 kV (defined in table 5) by about 20%. This additional scaling also improved the gamma pass rate discussed below.
The dose distribution becomes more peaked as the electron beam is positioned closer to the center of the target. With the electron beam positioned at a radial position of 3 mm, the full-width at half-maximum of the peak is approximately 90°, and it increases to ~195° at 7 mm. This is as expected due to the decreased collimation of the radiation by the partitions as the electron beam is positioned at larger radial positions.

Furthermore, the dose rate is highest in front of the x-ray tube, decreases near the target plane, and then increases again in planes behind the target. The low values near \( z = 0 \) reflect the fact that x-rays tend to be absorbed when exiting nearly parallel to the plane of the tungsten target. The lower intensity in the ‘backward’ direction compared to the ‘forward’ direction can be attributed to the additional absorption of the silicon carbide tube. The width of the peak is fairly constant with the different \( z \) positions, varying <1° at 3 mm and ~6° at 7 mm.

The azimuthal anisotropy is well captured by the simulations, indicating that the geometric modulations of the radiation field due to the partitioned diamond-tungsten target are well reproduced in simulations. The standard deviation of the residual between the simulated and measured dose distribution is less than

| 50 kV  | 85 kV  | 100 kV |
|-------|-------|--------|
| 3 mm  | 5%/3 mm | 5%/5 mm | 93.8% | 100.0% | 81.0% | 99.3% |
| 5 mm  | 85.4% | 100.0% | 82.8% | 100.0% | 93.1% | 100.0% |
| 7 mm  | 75.1% | 99.5% | 83.6% | 99.5% | 82.2% | 99.6% |

**Figure 6.** Spatial distribution of measurement points showing those that passed (green) and those that failed (red) the 5%/local)/3 mm acceptance criterion for a 3D gamma index analysis. The measured dose rates were used as the reference. Measured dose rates that were below 10% of the maximum measured dose rate were not considered and are shown in gray. The end of the x-ray tube’s cooling jacket is shown for reference.
0.5 mGy s$^{-1}$ for all source potentials and radial positions. These discrepancies are between 5%–20% at the peak dose rate (at an azimuthal angle of 0 degrees), with the agreement being considerably better at all other azimuths. Furthermore, the absolute agreement between the simulations and the measurements is quite good, as expected since the scaling of the beam current to the emission current (as discussed in section 3.2) was taken into account.

To further quantify the agreement between the measurements and the simulations we used the gamma index analysis introduced by Low et al (1998). Using the measured dose rates as a reference, the 3D voxelized simulated dose distribution was analysed against a specified dose-difference and distance-to-agreement criterion. Local normalization was used for the dose-difference, and a low-dose threshold of 10% was used, as recommended by AAPM Task Group 119 (Ezzell et al 2009). The gamma pass rates with acceptance criteria of 5%/3 mm and 5%/5 mm for various operating points are shown in table 6, and the average pass rates were 84.0% at 5%/3 mm and 99.6% at 5%/5 mm. To assess whether there was any systematic discrepancy, the distribution of which measurement positions passed/failed the test were plotted in figure 6 for the more stringent criterion of 5%/3 mm. As can be seen in this figure, the measurement positions that failed do not appear to have any apparent correlation. The gamma analysis did not account for the uncertainties in the dose rates, which could further improve the agreement between the simulations and measurements.

### 3.4. Source characterization in water

Although the dose distributions would ideally be measured directly in water, the faithful reproduction of the HVLs and dose distributions in air validates the simulations’ ability to accurately reproduce the radiation output of the x-ray source. In addition, even if the experiments are performed in water, ionization chambers inherently measure dose-to-air and an empirical conversion to dose-to-water has to be performed (Ma et al 2001).
Figure 8. Simulated dose distributions in water for different tube potentials at a radial position of 5 mm. A representative cannula and 2.5 cm radius balloon have been masked out. The dose rate is in units of cGy/min/μA. Isodose contours at 0.02, 0.05, 0.1, 0.2, 0.5, 1, and 2 cGy/min/μA are shown.

Figure 9. (a) Simulated percent dose depth (PDD) curves in water at three different source potentials for the forward and backward directions with a 7 mm radial electron beam position, with the depths defined relative to a 2.5 cm radius balloon. (b) The dependence of the PDD curves on the balloon radius. The cooling jacket has a radius of about 20 mm relative to the center of the target. This data is tabulated in table A1 in Appendix A.
Figure 7 shows the simulated dose distributions at 100 kV for the three different radial positions in one of the sextants. The slices represent orthogonal planes passing through the origin, which is located at the center of the x-ray target. The dose rates are in units of cGy/min/μA. For typical operating conditions of 100 kV at 50 W the beam current is 500 μA, which means that a dose rate of 1 cGy/min/μA can produce 10 Gy in 2 min.

The anisotropy of the dose distributions produced by the x-ray source is evident in figure 7. As was observed in the measurements in air shown in figure 5, the distribution is more peaked when the electron beam is positioned closer to the center of the target (e.g. 3 mm radial position as opposed to the 7 mm radial position). This is again due to the increased collimation of the radiation by the septa at smaller radial positions. Additionally, it can be observed that the dose rate is higher at a radial position of 7 mm than at 3 mm. This can be explained by the inverse squared decrease in the intensity of the x-ray radiation, since photons generated at a 3 mm radial position need to travel further to the surface of the balloon than those generated at a 7 mm radial position. Similar trends were observed at 50 kV and 85 kV.

The effect of the source potential on the dose distribution is shown in figure 8. As expected, the higher the kV, the larger the dose rate produced in water. Apart from the difference in the absolute dose rate, the shape of the distribution produced at the different source potentials is essentially the same. Similar observations can be made at the 3 mm and 7 mm radial positions (data not shown for brevity).

Figure 9 shows the simulated percent dose depth (PDD) curves in water at the different source potentials, and the data is tabulated in table A1 in Appendix A. The PDD curves were calculated separately for the ‘forward’ and ‘backward’ spectra, defined relative to the plane containing the x-ray target. In figure 9(a) the data has been normalized to the dose rate at 2.5 cm from the x-ray source drift tube, which is the expected distance-to-tissue. As expected, the curve at 50 kV decays much faster than the higher energy curves, since the low-energy x-rays are absorbed more quickly in the water. Furthermore, the ‘forward’ spectra decay more quickly that the ‘backward’ spectra. This can be explained by referring to the spectra shown in figure 4(a), which shows that the backward spectra are harder than the forward spectra, and so penetrate deeper into the water.

Because of the $1/r^2$ falloff that results from the beam divergence, the PDD curves depend strongly on the balloon radius. In particular, smaller balloons have PDDs that falloff more quickly, since the $1/r^2$ divergence is the strongest. Alternatively, for larger balloons the divergence is less significant, and so the PDDs penetrate more deeply into the tissue. This is illustrated in figure 9(b), which show the PDDs for different balloon radii.

It was additionally found that there was negligible dependence of the PDD curves on the electron beam’s radial position (data not shown).

4. Conclusions

A novel x-ray source for high-dose-rate electronic brachytherapy, the Sculptura™ by Sensus Healthcare, has been characterized. Excellent agreement between Monte Carlo simulations and dose rate measurements in air were obtained. The validated Monte Carlo simulations were then used to calculate dose distributions in water.

The unique design of the Sculptura™ x-ray source produces highly anisotropic dose distributions with true 3D beam directionality, which can improve target dose conformity relative to delivery systems that emit radially-symmetric dose distributions. Further work developing a treatment planning system to make use of such unique directed dose distributions needs to be done, and clinical outcomes will further elucidate the significance of such a source for intracavitary and intraoperative radiation therapy.

Conflict of interest disclosure

Triple Ring Technologies was contracted by Sensus Healthcare to design and develop the Sculpura™ x-ray source. Kalman Fishman is the Chief Technology Officer of Sensus Healthcare.

Appendix. Tabulated simulated PDD curves in water

Because the Sculpura™ x-ray source has the ability to produce highly anisotropic dose profiles, the full 3D dose distributions are required to completely characterize the output capabilities of the source. However, there is some desire and benefit to using familiar metrics, such as the PDD curve, that are used to characterize conventional isotropic x-ray sources. While this is possible, it should be noted that the PDD curves also exhibit some spatial dependence due to the anisotropic nature of the source. Because of this, the PDDs tabulated in table A1 offer only a partial characterization of the source output.
Table A1. Simulated percent dose depth curves in water for different kVs in both the forward and backward directions. This data is plotted in figure 9. The PDDs are normalized to the dose rate at the cooling jacket surface. The PDDs for various balloon radii can be calculated by normalizing the tabulated data by the PDD value at the appropriate distance from the cooling jacket.

| Distance from cooling jacket (mm) | Percent dose depth (%) | 50 kV | 85 kV | 100 kV | 50 kV | 85 kV | 100 kV |
|----------------------------------|------------------------|-------|-------|--------|-------|-------|--------|
| 0.00                             | 100.00                 | 100.00| 100.00| 100.00 | 100.00| 100.00|
| 1.50                             | 84.75                  | 84.80 | 87.08 | 88.65  | 87.72 |
| 3.00                             | 72.43                  | 72.67 | 75.77 | 78.75  | 77.75 |
| 4.50                             | 62.60                  | 62.79 | 66.57 | 69.69  | 69.86 |
| 6.00                             | 53.92                  | 53.74 | 58.20 | 61.63  | 61.90 |
| 7.50                             | 45.93                  | 45.63 | 50.16 | 54.18  | 54.02 |
| 9.00                             | 39.89                  | 39.31 | 44.01 | 48.11  | 48.02 |
| 10.50                            | 35.35                  | 34.78 | 39.48 | 43.51  | 43.54 |
| 12.00                            | 31.77                  | 31.77 | 35.36 | 39.39  | 39.82 |
| 13.50                            | 28.49                  | 27.81 | 31.87 | 36.07  | 36.36 |
| 15.00                            | 25.72                  | 25.16 | 28.87 | 32.91  | 33.27 |
| 16.50                            | 23.10                  | 22.53 | 25.76 | 30.02  | 30.45 |
| 18.00                            | 20.52                  | 20.08 | 22.68 | 27.09  | 27.21 |
| 19.50                            | 18.61                  | 18.07 | 20.56 | 24.59  | 25.03 |
| 21.00                            | 17.02                  | 16.34 | 18.74 | 22.74  | 22.88 |
| 22.50                            | 15.44                  | 14.97 | 17.02 | 20.91  | 21.42 |
| 24.00                            | 14.19                  | 13.70 | 15.48 | 19.04  | 19.74 |
| 25.50                            | 13.10                  | 12.55 | 14.28 | 17.85  | 18.47 |
| 27.00                            | 12.16                  | 11.57 | 12.96 | 16.71  | 17.13 |
| 28.50                            | 11.09                  | 10.57 | 11.71 | 15.32  | 15.76 |
| 30.00                            | 10.16                  | 9.54  | 10.75 | 14.19  | 14.71 |
| 31.50                            | 9.41                   | 8.82  | 9.74  | 13.07  | 13.78 |
| 33.00                            | 8.79                   | 8.20  | 9.07  | 12.10  | 12.78 |
| 34.50                            | 8.15                   | 7.66  | 8.41  | 11.45  | 11.98 |
| 36.00                            | 7.64                   | 7.07  | 7.82  | 10.77  | 11.29 |
| 37.50                            | 7.07                   | 6.66  | 7.21  | 10.10  | 10.62 |
| 39.00                            | 6.57                   | 6.18  | 6.60  | 9.32   | 9.93  |
| 40.50                            | 6.16                   | 5.70  | 6.61  | 8.73   | 9.28  |
| 42.00                            | 5.74                   | 5.33  | 5.56  | 8.11   | 8.68  |
| 43.50                            | 5.34                   | 4.95  | 5.15  | 7.76   | 8.23  |
| 45.00                            | 5.02                   | 4.72  | 4.84  | 7.20   | 7.73  |
| 46.50                            | 4.78                   | 4.40  | 4.46  | 6.73   | 7.32  |
| 48.00                            | 4.52                   | 4.16  | 4.20  | 6.42   | 6.93  |
| 49.50                            | 4.25                   | 3.91  | 3.92  | 6.10   | 6.58  |
| 51.00                            | 3.96                   | 3.65  | 3.56  | 5.68   | 6.25  |
| 52.50                            | 3.73                   | 3.41  | 3.32  | 5.34   | 5.89  |
| 54.00                            | 3.54                   | 3.18  | 3.12  | 5.10   | 5.54  |
| 55.50                            | 3.36                   | 3.03  | 2.93  | 4.81   | 5.17  |
| 57.00                            | 3.15                   | 2.89  | 2.70  | 4.57   | 5.04  |
| 58.50                            | 3.03                   | 2.75  | 2.58  | 4.34   | 4.77  |
| 60.00                            | 2.85                   | 2.59  | 2.42  | 4.14   | 4.51  |
| 61.50                            | 2.69                   | 2.45  | 2.24  | 3.92   | 4.27  |
| 63.00                            | 2.55                   | 2.29  | 2.07  | 3.71   | 4.03  |
| 64.50                            | 2.43                   | 2.17  | 1.98  | 3.44   | 3.83  |
| 66.00                            | 2.30                   | 2.02  | 1.83  | 3.28   | 3.65  |
| 67.50                            | 2.19                   | 1.96  | 1.74  | 3.15   | 3.55  |
| 69.00                            | 2.10                   | 1.83  | 1.63  | 3.03   | 3.37  |
| 70.50                            | 1.99                   | 1.77  | 1.53  | 2.86   | 3.21  |
| 72.00                            | 1.89                   | 1.65  | 1.43  | 2.71   | 3.08  |
| 73.50                            | 1.81                   | 1.57  | 1.36  | 2.61   | 2.88  |
| 75.00                            | 1.72                   | 1.51  | 1.24  | 2.48   | 2.77  |
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