PAPER

Machine characterization and central axis depth dose data of a superficial x-ray radiotherapy unit

Zhenyu Xiong, Yuncheng Zhong, Thomas I. Banks, Robert Reynolds, Tsuicheng Chiu, Jun Tan, You Zhang, David Parsons, Yulong Yan, Andrew Godley and Strahinja Stojadinovic

1 Department of Radiation Oncology, University of Texas Southwestern Medical Center, Dallas, TX, United States of America
2 Department of Radiation Oncology, Rutgers Cancer Institute of New Jersey, Rutgers Robert Wood Johnson Medical School, New Brunswick, NJ, United States of America

* Author to whom any correspondence should be addressed.

E-mail: Strahinja.Stojadinovic@UTsouthwestern.edu

Keywords: superficial radiotherapy, kV percent depth dose, commissioning of superficial radiotherapy unit

Abstract

Objectives. The purpose of this study is to present data from the clinical commissioning of an Xstrahl 150 x-ray unit used for superficial radiotherapy, Methods. Commissioning tasks included vendor acceptance tests, timer reproducibility, linearity and end-effect measurements, half-value layer (HVL) measurements, inverse square law verification, head-leakage measurements, and beam output calibration. In addition, percent depth dose (PDD) curves were determined for different combinations of filter/kV settings and applicators. Automated PDD water phantom scans were performed utilizing four contemporary detectors: a microDiamond detector, a microSilicon detector, an EDGE detector, and a PinPoint ionization chamber. The measured PDD data were compared to the published values in BJR Supplement 25, Results. The x-ray unit's mechanical, safety, and radiation characteristics were within vendor-stated specifications. Across sixty commissioned x-ray beams, the PDDs determined in water using solid state detectors were in excellent agreement with the BJR 25 data. For the lower (<100 kVp) and medium-energy (≥100 kVp) superficial beams the average agreement was within [−3.6,+0.4]% and [−3.7,+1.4]% range, respectively. For the high-energy superficial (low-energy orthovoltage) x-rays at 150 kVp, the average difference for the largest 20 × 20 cm² collimator was (−0.7 ± 1.0)%. Conclusions. This study presents machine characterization data collected for clinical use of a superficial x-ray unit. Special focus was placed on utilizing contemporary detectors and techniques for the relative PDD measurements using a motorized water phantom. The results in this study confirm that the aggregate values published in the BJR 25 report still serve as a valid benchmark when comparing data from site-specific measurements, or the reference data for clinical utilization without such measurements, Advances in knowledge. This paper presents comprehensive data from the acceptance and commissioning of a modern kilovoltage superficial x-ray radiotherapy machine. Comparisons between the PDD data measured in this study using different detectors and BJR 25 data are highlighted.

Introduction

Kilovoltage (kV) x-ray beams in the energy range 30-300 kVp are commonly used to treat skin cancers and benign skin conditions (Hill et al, 2014, Nahum, 1999, Poen, 1999, Williams and Thwaites, 2000). Superficial (50-150 kVp) and orthovoltage (150-500 kVp) radiotherapy is recommended for clinical use by the International Atomic Energy Agency (IAEA) (IAEA, 2008). Kilovoltage radiotherapy units are utilized in many radiation oncology departments due to their ease of use, quality of treatment, and relative cost-effectiveness. In many cases kV radiotherapy offers simple, yet very effective treatment options compared to megavoltage photon and electron alternatives. Beyond the radiotherapy horizon, kV x-ray beams are predominantly utilized for radiographic, fluoroscopic and tomographic diagnostic applications and research.
A characteristic feature of superficial kV beams is that dose is at a maximum at the skin surface which then falls off rapidly with depth due to beam attenuation and scattering. Therapeutic kV beams are therefore used predominantly for treatment of skin cancers and management of other dermatological conditions such as keloid scars, mycosis fungoides, and psoriasis. The doses prescribed for superficial radiotherapy can be relatively high and may utilize standard or hypofractionated regimens (Mcgregor et al, 2015). Treatments can cause unwanted acute reactions such as skin erythema, skin depigmentation, and hair loss, and minimization of these toxicities can be at odds with therapeutic goals. It is clear, then, that accurate assessment of dose to the patient is essential. A prerequisite for accurate treatment is the acquisition of treatment planning parameters. The PDD data tables for various filter/kV and applicator combinations and the corresponding dose rates are essential for determining the optimal prescription depth in superficial radiotherapy.

Methods and materials

The Xstrahl 150 x-ray system

The superficial treatment unit used in this study was an Xstrahl 150 x-ray system (Gulmay Medical Inc., Surrey, United Kingdom), shown in figure 1(a). The unit is equipped with an x-ray generator of 3 kW maximum power capable of producing x-rays from 10-150 kVp and tube currents up to 30 mA. The unit is available as either a floor- or ceiling-mounted system and offers a range of superficial treatment options for radiation oncology and dermatology departments. The tungsten (W) target angled at 30° has a focal spot size of 7.5 mm. The tube has inherent filtration of 0.8 ± 0.1 mm of beryllium (Be) and minimum and maximum HVLs of 0.2 mm Al and 0.5 mm Cu, respectively. Six of the nine available filter and tube potential settings were commissioned for clinical use. A built-in system interlock necessitates that one filter, which determines kVp/ma tube setting, is properly inserted in the gantry while the remaining filters are docked in a wall-mounted storage unit. The properties of each clinically used filter corresponding to fixed kVp/ma tube setting, added filtration and nominal HVL, are presented in table 1.

The unit has ten open-ended applicators made of steel, copper and Perspex: nine circular applicators ranging from 1.5 cm to 18 cm in diameter, and one 20 × 20 cm² square applicator. The focal spot distance (FSD) from the source to the applicator opening varies with applicator but is limited to 15 cm, 25 cm, 30 cm, or 50 cm; see table 2. All six clinically used filters, ID #s 3-7 and 9, and all ten applicators, IDs A-J, were used in this study. For all PDD measurements, the applicator openings were set at the water surface to mimic the standard clinical setup in which the applicator is flush against the patient’s skin.
Figure 1. (a) The Xstrahl 150 x-ray system. Figure 1(b) HVL measurement setup: ionization chamber at a focal spot distance of 75 cm from the x-ray source with gantry rotated 180° to facilitate insertion of additional filtration sheets. Figure 1(c) PDD measurement setup. The opening of each applicator was in contact with the water surface and the detector was placed in the center of the opening.

Table 1. The clinically used filter settings for the Xstrahl 150 kV x-ray system.

| Filter ID # | 3 | 4 | 5 | 6 | 7 | 9 |
|-------------|---|---|---|---|---|---|
| Tube Potential [kVp] | 50 | 80 | 100 | 120 | 120 | 150 |
| Tube Current [mA] | 10 | 10 | 10 | 5 | 10 | 10 |
| Added Filtration [mm] | 0.8 mm Al | 1.7 mm Al | 2.0 mm Al | 0.9 mm Al + 0.05 mm Cu | 0.5 mm Al + 0.1 mm Cu | 1.0 mm Al + 0.2 mm Cu |
| Half-Value Layer [mm] | 1.0 mm Al | 2.0 mm Al | 3.0 mm Al | 4.0 mm Al | 5.0 mm Al | 0.5 mm Cu |

Table 2. The clinically used applicators for the Xstrahl 150 kV x-ray system.

| Applicator ID | A | B | C | D | E | F | G | H | I | J* |
|---------------|---|---|---|---|---|---|---|---|---|---|
| Opening diameter d [cm] | 1.5 | 2.0 | 2.5 | 3.0 | 4.0 | 5.0 | 10 | 15 | 18 | 20 × 20* |
| Focal Spot Distance FSD [cm] | 15 | 15 | 15 | 15 | 15 | 25 | 25 | 30 | 50 |

*Square field
Mechanical and safety tests
The performed system’s checks included tube stand movements and brakes functionality, tube roll, pitch and yaw rotations and locking capability, smooth rotation and locking of applicators in the gantry, wall-mounted storage unit and gantry filter docking as well as generator water coolant circulation and cooling fans spinning. Furthermore, power on, x-ray on button, x-ray off button, emergency beam off switch, controlled area light illumination and audible x-ray on warning functionality was inspected. Lastly, the functionality of built-in system interlocks was tested, i.e., treatment room door interlock, filter-in-use and docked filters interlocks and timer interlock.

X-ray tube leakage measurements
Head leakage measurements were made using a Fluke 451P Survey Meter (Fluke Corporation, Everett, Washington). Circular applicator A, i.e., the smallest applicator with 1.5 cm diameter opening, and filter 9 corresponding to the highest 150 kVp and 10 mA tube setting, was used. The survey meter was placed at a fixed distance 1 m away and at 90° relative to the machine focal spot. The measurements were recorded for the gantry oriented at angles 0° and 180°, i.e., perpendicular to beam direction, and opposite to beam direction for the gantry angle at 270°.

Machine timer end effect
The end effect, Δt, is generally defined as the amount of beam delivery time not accounted for by a machine timing mechanism. It represents the time difference between the start of timing and the moment when the desired mA and kVp is achieved (Ma et al, 2001). Even a relatively small end effect (0.5-3 s) may lead to significant dosimetric errors due to the high dose rates and consequent short treatment times (1 min or less). The end effect for an x-ray unit can be determined using the graphical extrapolation method (Attix, 1986). The graphical solution of zero exposure, i.e., the intercept of the regression line on the time axis, yields the end effect. This method was used to determine the end effect for the entire range of clinical tube voltages for two different applicator sets using a PTW 31010 ionization chamber.

Timer reproducibility and linearity
The timer reproducibility and linearity tests were performed via in-air measurements using applicator F and a PTW 31010 ionization chamber placed at 15 cm distance from the focal spot. The reproducibility R was tested via a series of five consecutive 1-minute measurements at 50 kVp and 150 kVp tube potentials using equation

$$ R = \frac{\sigma}{M} \times 100\% $$

where σ was the standard deviation and M was the mean value of the charge collected in five measurements. The tolerance for R was set to ±0.1%. The reproducibility test was repeated by stopping and resuming the beam during 1-minute exposures at 150 kVp. The effect was quantified using

$$ R_i = \left( \frac{M_{int} - M}{M} \right) \times 100\% $$

where $R_i$ was the reproducibility with interruptions, $M_{int}$ was the mean value of the collected charge for five consecutive 1-minute measurements each with one interruption, and M was the mean value of the collected charge for five consecutive 1-minute measurements without any interruption.

The linearity test, performed for filters 3 and 9, included measurements of 0.5-, 3-, and 5-minute beam-on times. The linearity was calculated using the following expression:

$$ L = \left( \frac{M_i}{3 \text{min}} / \frac{M_i}{t_{\text{min}}} - 1 \right) \times 100\% $$

where $M_i$ was the mean charge collected for a 3-minute exposure time, $M_i$ was the mean charge collected for an exposure time of $t_{\text{min}}$. The tolerance for machine timer linearity L was set to ±1%.

The output constancy, for filters 3 and 9, was tested at gantry angles at 0°, 135°, and 225° for 1-minute beam-on times. The average measurements for gantry positions at 135° and 225° were compared to gantry 0° via equation

$$ OC = \left( \frac{M_G - M_0}{M_0} \right) \times 100\% $$

where $OC$ was the relative output difference, $M_G$ was the mean value of the charge collected during four 1-minute measurements at 135° and 225° gantry angles, and $M_0$ was the mean value of the charge collected during four consecutive 1-minute measurements at gantry angle 0°. The tolerance for the output constancy $OC$ was set to ±1%.

Inverse square law verification
For reference clinical dosimetry, the calibration point for each applicator was set at the center of the circular opening, i.e., at the applicator central axis at FSD listed in table 2. However, it was impossible to measure dose rate at this exact location for every setup due to the intrinsic dimensions of ionization chamber and the small applicator opening. In such situations the chamber was placed just downstream of the opening so that the chamber touched the applicator’s end, and an inverse-square correction was applied to determine the dose rate at midplane of the opening. Specifically, the measurement point and the intended calibration point differ by half the ionization chamber 6.9 mm outer diameter. The inverse square verification measurements were made using a PTW 31010 ionization chamber and a PTW Unidose E electrometer. Applicators A, C, and G and filters 3 and 9 were used in the measurements. The inverse square law was verified by taking two ionization chamber measurements. The first measurement position was at a point of contact with the applicator opening while the second one was at a point shifted 10 cm downstream along the applicator central axis. The measurements were
compared with the expected values calculated using the inverse square law.

**HVL measurements**

Central-axis HVL measurements were acquired using a narrow-beam geometry following the recommendations of the AAPM TG-61 protocol (Ma et al., 2001). The setup utilized an in-house ionization chamber mount which provided minimal scatter of the primary beam while the chamber was placed at a 75 cm FSD. As shown in figure 1(b), the gantry was rotated 180° to facilitate placement of aluminum (Al) and copper (Cu) sheets atop the applicator opening. Applicator E, with circular opening of 4 cm diameter, and an ADCL-calibrated PTW 31010 ionization chamber were used for the measurements. The measured HVLs were compared to the nominal vendor-provided values. In addition, the nominal and measured HVLs were compared to values obtained from the SpekCalc and SpekPy software (Poludniowski and Evans, 2007, Poludniowski, 2007, Poludniowski et al., 2021, Bujila et al., 2020, Healy and Hill, 2022).

**Output dose rate calibration measurements**

The in-air method of the AAPM TG-61 protocol (Ma et al., 2001) was used to determine the absorbed dose rate to water at the water surface, with the applicator-specific FSD listed in Table 2 chosen as the calibration point. A PTW UNIDOS E electrometer and a PTW 31010 ionization chamber, both ADCL calibrated by University of Wisconsin, were utilized for dose rate measurements. The absorbed dose rate to water at the water surface for a given FSD was determined according to:

$$D_{\text{air}, z=0} = (M_{\text{air}} P_{\text{ion}} P_{\text{pol}} P_{\text{dec}}) \times N \times B_{\text{air}} P_{\text{stem,air}} \times \left( \frac{\rho_{\text{air}}}{\rho_{\text{water}}} \right)^{\frac{w}{\text{ISF}}}$$

(5)

where $D_{\text{air}}$ is the dose rate to water at water surface; $M_{\text{air}}$ is the uncorrected in-air ionization chamber charge reading; $P_{\text{ion}}$, $P_{\text{pol}}$ and $P_{\text{dec}}$ are the temperature and pressure, ion recombination, polarity effect, and electrometer correction factors, respectively; $N$ is the air-kerma calibration coefficient for a given beam quality; $B_{\text{air}}$ is the backscatter factor; $P_{\text{stem,air}}$ is the chamber stem correction factor, which was taken as $1 - \left( \frac{\rho_{\text{air}}}{\rho_{\text{water}}} \right)^{w}$ is the water-to-air ratio of the mean mass energy absorption coefficients averaged over the incident photon spectrum, for an in-air measurement; and ISF is the inverse square factor needed if the measurement point is not at the calibration point, i.e., not at the applicator opening which indeed was the case here. Temperature and pressure were measured using a CNMC traceable digital thermometer and a Druck DPI 705 digital barometer. Output measurements were made for six filters and ten applicators, resulting in 60 clinical techniques. The dose rate for each filter and applicator combination was reported as dose rate in air $D_{\text{air}}$ in Gy min⁻¹, i.e., the backscatter factor $B_{\text{air}}$ was factored out from $D_{\text{air}}$. The reason for this is the requisite to calculate treatment beam-on time for clinical setups different than the reference calibration conditions.

In clinical practice, a beam-on time, $t$, needed to deliver the prescribed therapeutic dose, $D_{\text{Ref}}$, is calculated using the following formula:

$$t = D_{\text{Ref}} \times \left( \frac{D_{\text{air}} (\text{FSD}_{\text{ref}})}{D_{\text{air}} (\text{FSD}_{\text{off}}) \times \text{BSF} (\text{FSD}_{\text{off}}, d_{\text{SS}})} \right)^{2} \times \text{BSF} (\text{FSD}_{\text{off}}, d_{\text{SS}})$$

(6)

where $D_{\text{air}} (\text{FSD}_{\text{off}})$ is the calibrated dose rate in air for the reference FSD, $d_{\text{SS}}$ is the separation between applicator and the patient if the applicator is not flush with patient skin and $\left( \frac{\text{FSD}_{\text{off}}}{\text{FSD}_{\text{off}} + \text{gap}} \right)^{2}$ is the inverse square correction of output as needed when gap is nonzero, BSF (FSD, $d_{\text{SS}}$) is the backscatter factor for a given treatment FSD and field diameter $d_{\text{SS}}$. PDD (FSD$_{\text{off}}$, z) is the percent depth dose at the prescription depth $z$ for the reference FSD$_{\text{off}}$ measured during commissioning, and $F_{\text{ref}}$ is the Mayneord’s factor representing the ratio of the inverse square component of PDDs from the reference FSD$_{\text{off}}$ to treatment FSD also as needed when gap is nonzero.

**Percent depth dose measurements**

The PDD water tank measurements were carried out using an IBA Blue Phantom 2 (IBA Dosimetry, Schwarzenbruck, Germany) for combinations involving six filters and ten applicators, i.e., for 60 different clinical x-ray beams. The opening of each applicator was positioned at the water surface and the detectors were placed in the center of the applicator, see figure 1(c). The detectors used for PDD measurements were a synthetic diamond detector (microDiamond 60019, PTW, Freiburg, Germany), a diode detector (microSilicon diode 60023, PTW, Freiburg, Germany), another diode detector (EDGE, Sun Nuclear Corp., Melbourne, Florida) and an ultra-small-volume ionization chamber (PinPoint 3D 31016, PTW, Freiburg, Germany). The ionization chamber was positioned with its axis perpendicular to the beam central axis, while the diamond and diode detectors were positioned with their axes parallel to the beam central axis. Applicator-specific profile scans were used to determine the central axis of each field. Zero depth for the ionization chamber was established as the central chamber axis coincided with the water surface. Two assumptions were implicit here, first, the effective point of measurement coincided with the ionization chamber’s central axis, and second, the depth doses were directly proportional to the ionization readings without applying any corrections. A bias voltage of −300 V was used for the ionization
chamber scans. For the solid state detectors, zero depth was established as the line on the housing, indicating the reference point, was aligned with the water surface, and zero voltage bias was used for the scans. The detectors were pre-irradiated with vendor-recommended doses to stabilize the detector response. The PDDs were measured from 20 cm depth to water surface along the beams’ central axes, with the detectors moved upward at constant speed of 0.3 cm s\(^{-1}\) to minimize disturbances to the water surface. The detector readings were recorded in approximately 0.1 mm intervals, i.e., for each PDD scan the raw data had over 2000 data points. For efficiency and to minimize setup uncertainty, all PDDs for an applicator were measured in sequence before changing the setup for the next applicator.

The ISO methodology (ISO, 2011) provides a general basis for determining uncertainties associated with experimental measurements. A few exemplary uncertainty budget considerations relevant for kV x-rays can be found in the literature (Andreou et al., 2006; Ma et al., 2001, Hill et al., 2009, Gronberg et al., 2020). The contributing sources of uncertainty for PDD measurements in this study included setup repeatability, applicators FSD specifications and air gap (if necessary), detectors placement at the water surface, motors positional accuracy during scanning, detectors directional response in water, variation in the mass energy absorption coefficient ratios with beam size and depth in medium, machine output fluctuations during long exposure times necessary for data acquisition, and drift of the measuring equipment.

After obtaining the scanned data, the measured PDD points were fitted in Matlab (Mathworks, 2021) using the least-squares spline approximation by minimizing expression

\[
\sum_i w_i \left| y(i) - f(x(i)) \right|^2
\]

where \(w_i\) represents the weight factors with default values equal to 1, \(y_i\) represents the measured data points and \(f(x_i)\) is the spline function of polynomial order \(k\) with the knot sequence for which \(y(i) = f(x(i))\) for all \(i\). A 9th, 11th and 13th order polynomial were used for data fitting. The goodness of fit was judged by comparing the adjusted \(R^2\) values with a higher value closer to 1 signifying better fit. The spline technique was selected for fitting since it facilitated consistent data fitting for all detectors.

The measured PDD data were compared to the BJR 25 data by simply computing the differences, \(\Delta_{\text{PDD}}\), which were reported as percentages because PDD values are intrinsically relative numbers. The relative difference, or the ratio, of published and measured depth doses is not an adequate metric since the ratio of two small numbers with a small difference yields a large relative difference that creates a false perception of a large dose discrepancy.

**Results**

**Mechanical and safety tests**

The x-ray unit’s mechanical and safety features were all functional.

**X-ray tube leakage**

The survey meter readings of the x-ray treatment head leakage for gantry angles 0°, 180°, and 270° were 0.41, 0.03, and 0.19 mGy h\(^{-1}\), respectively. The measured values were much less than 1/1000 or 0.1% of the primary beam output, 262 Gy h\(^{-1}\), and thus in accord with the International Electrotechnical Commission IEC 60601-2-8 standard (IEC, 2010) for medical electrical equipment, which limit is 1 mGy h\(^{-1}\) at 1 m distance.

**Machine timer end effect**

The intrinsic operation of the Xstrahl 150 timer is such that for a given kVp and mA setting, the kV first rises to 90% of the required value before the mA begins ramping up to its required value. Once the mA ramp-up is 50% complete, the treatment timer starts. This approach enables the kV and mA to stabilize adequately and deliver a reproducible dose. The largest measured timer end effect, \(\Delta t\), was 0.19 s for filter 3, which has the lowest tube potential of 50 kVp, with a dosimetric impact of less than 1% for an exposure time of 20 s. For filters 4 and 5, \(\Delta t\) was 0.01 s, and for filters 6, 7, and 9, \(\Delta t\) was 0.06 s. In general, the measured end effects for all filters were determined to be negligible for clinically relevant beam-on times which are on the order of minutes.

**Timer reproducibility and linearity**

The measured timer reproducibility, \(R\), defined by equation (1), was 0.1% and 0.03% for filters 3 and 9, respectively. The reproducibility with intentional interruptions, \(R_i\), as defined by equation (2), was determined to be 0.04% for filter 9. This indicated that clinical treatment interruptions would have negligible dosimetric effect. The measured timer linearity, \(L\), defined by equation (3), was 0.2% and 0% for filters 3 and 9, respectively. The machine output constancy, OC, defined by equation (4), was less than 0.2% over the range of gantry angles tested, demonstrating that gantry orientation has negligible dosimetric impact on clinical treatments.

**Inverse square law verification**

In general, for two measurement points 10 cm apart on the beam central axis, the relative difference between measured and calculated inverse square factors, \(\Delta_{\text{ISF}}\), decreased with increasing applicator size. For instance, for applicator A, \(\Delta_{\text{ISF}}\) was 1.8% and 1.3% for filters 9 and 3, respectively. For applicator C, \(\Delta_{\text{ISF}}\) was 1.2% for both filters. For applicator G, \(\Delta_{\text{ISF}}\) decreased to 0.2% and 0.7% for filters 9 and 3,
respectively. Notably, a smaller applicator opening combined with a smaller FSD contributed to relatively larger in-scattering for applicator A compared to applicator G. Since the inverse square law for photons is universally valid, the main contribution to the observed variances was due to relative difference of scattered electrons from the applicator walls. Consequently, the ISF component for smaller collimators has an intrinsically larger relative systematic error relative to the larger collimators. This confirms the recommendation to use the smallest applicator for inverse square law validation, since a 4% difference was reported for a 2.5 cm diameter applicator at 10 cm FSD (Aukett et al., 2005).

**HVL measurements**

The measured HVLs were compared to the nominal vendor-stated values as well as to the HVLs calculated by SpekCalc and SpekPy software (Poludniowski, 2007, Poludniowski and Evans, 2007, Bujila et al., 2020, Healy and Hill, 2022, Poludniowski et al., 2021); see table 3. For the six clinically used filters, the SpekCalc and SpekPy HVLs were within 0.1 mm relative to the measured values. The agreement between all measured and nominal HVLs was within 10%. The observed HVL differences are common and acceptable and typically translate to less than 1.5% dosimetric change for tube potentials above 80 kVp. This is related to the choice of the beam quality points offered by the Accredited Dosimetry Calibration Laboratories (ADCL) for an ionization chamber calibration in terms of air kerma. Since the ADCL beam qualities are not only stratified based on the HVL values (mm Al or Cu) but are also based on the inherent filtration (mm Be), the additional filtration (mm Al or Cu), and the tube voltage (kVp), the choice of calibration beam quality may not be straightforward. In the event the ADCL calibration conditions do not match the user’s machine specifications a compromise of selecting the ‘nearest match’ point is one option, alternatively, an interpolation between two nearest neighbor calibration points can be utilized.

**Output dose rate calibration**

The in-air calibration was performed in accordance with the AAPM TG-61 protocol with the estimated combined standard uncertainty (1σ) of ±4.7%. Ten applicators and six filters yielded 60 clinical beam combinations whose dose rates are listed in table 4. As expected, the measured dose rates at a given FSD gradually increased with an increase of applicator size due to increased backscatter. At the same time, the measured dose rates were considerably reduced for applicators G to J owing to the increased FSDs and resulting inverse square law losses.

**Percent depth dose**

The four detectors used for PDD measurements were a PTW microDiamond 60019, a PTW microSilicon diode 60023, a Sun Nuclear EDGE diode and a spherical PTW ionization chamber PinPoint 3D 31016. Six filter and ten applicator combinations, i.e.,

### Table 3. The nominal, calculated, and measured HVLs for the Xstrahl 150 system.

| Filter ID | Nominal HVL [mm] | SpekCalc HVL [mm] | SpekPy HVL [mm] | Measured HVL [mm] |
|-----------|------------------|-------------------|-----------------|------------------|
| 3         | 1.0 mm Al        | 0.82 mm Al        | 0.83 mm Al      | 0.91 mm Al       |
| 4         | 2.0 mm Al        | 1.91 mm Al        | 1.98 mm Al      | 2.01 mm Al       |
| 5         | 3.0 mm Al        | 2.60 mm Al        | 2.70 mm Al      | 2.71 mm Al       |
| 6         | 4.0 mm Al        | 3.69 mm Al        | 3.83 mm Al      | 3.86 mm Al       |
| 7         | 5.0 mm Al        | 4.80 mm Al        | 4.91 mm Al      | 4.91 mm Al       |
| 8         | 0.5 mm Cu        | 0.46 mm Cu        | 0.46 mm Cu      | 0.47 mm Cu       |

### Table 4. Beam dose rate in air under reference conditions on the Xstrahl 150.

| Applicator ID | Diameter [cm] | FSD [cm] | Dose rate in air [Gy min⁻¹] |
|---------------|---------------|----------|----------------------------|
|               |               | Filter 3 | Filter 4 | Filter 5 | Filter 6 | Filter 7 | Filter 9 |
| A             | 1.5           | 15       | 3.265   | 3.602   | 4.539   | 2.604   | 4.116   | 4.366   |
| B             | 2             | 15       | 3.280   | 3.633   | 4.578   | 2.629   | 4.158   | 4.418   |
| C             | 2.5           | 15       | 3.312   | 3.667   | 4.628   | 2.652   | 4.188   | 4.466   |
| D             | 3             | 15       | 3.314   | 3.689   | 4.655   | 2.666   | 4.212   | 4.499   |
| E             | 4             | 15       | 3.324   | 3.704   | 4.682   | 2.676   | 4.230   | 4.523   |
| F             | 5             | 15       | 3.352   | 3.739   | 4.725   | 2.699   | 4.259   | 4.569   |
| G             | 10            | 25       | 1.179   | 1.329   | 1.677   | 0.952   | 1.519   | 1.620   |
| H             | 15            | 25       | 1.204   | 1.353   | 1.711   | 0.975   | 1.540   | 1.652   |
| I             | 18            | 30       | 0.822   | 0.925   | 1.170   | 0.665   | 1.052   | 1.133   |
| J             | 20 × 20"      | 50       | 0.289   | 0.328   | 0.413   | 0.239   | 0.377   | 0.405   |

*Square field
Table 5. Uncertainty budget associated with the experimentally measured kV depth dose data.

| Component of uncertainty                        | Uncertainty [%] |
|-------------------------------------------------|-----------------|
| Repeatability of setup                          | PinPoint detector | Solid state detectors |
| Detector placement at the water surface          | 0.5             | 0.5                 |
| Applicator FSD specification and air gap (if necessary) | 0.2             | 0.2                 |
| Motors positional accuracy during scanning       | 0.2             | 0.2                 |
| Variations of \( \frac{\Delta P}{P} \) with depth and field size | 1.0             | 0.5                 |
| Detector directional response in water           | 0.5             | 1.0                 |
| Drift of the measuring equipment                 | 0.5             | 0.5                 |
| X-ray unit output fluctuations                   | 0.5             | 0.5                 |
| Combined standard uncertainty (1σ)              | 1.5             | 1.5                 |

60 beams, were measured using four detectors in a motorized IBA Blue Phantom 2, generating 240 water-scanned PDD curves. The average PDD difference, \( \Delta_{PDD} \), between measured and published data was calculated for twelve discrete points at depths of 0, 0.5, 1, 2, 3, 4, 5, 6, 7, 8, 9 and 10 cm as tabulated in the BJR 25 report. A total of 101 discrete points, corresponding to 0-10 cm depth in 1 mm increments, were used for the cross comparison between four detectors utilized for the PDD measurements. In line with general textbook knowledge, it was observed that the measured PDDs increased with an increase in applicator size or beam energy.

The relative uncertainty of the performed percent depth dose measurements was estimated to be ±1.5% at the one standard deviation level; see table 5.

For the low-energy superficial x-rays (<100 kVp), i.e., 50 kVp and 80 kVp, corresponding to HVLs of 1.0 mm and 2.0 mm Al, respectively, all measured data for the solid state detectors were on average within [−3.6, +0.4]% range relative to the published BJR 25 data. The stated −3.6% mean was the outlier, corresponding to the largest applicator I (18 cm diameter) for which interpolated BJR 25 values could be obtained. As applicator size decreased, an improved average agreement relative to BJR 25 data was found for the microDiamond, microSilicon and Edge detectors. The PinPoint ionization chamber data did not follow any apparent trend and was within [−2.6, +3.0]% range relative to the published BJR 25 data. The difference, \( \Delta_{PDD} \), averaged over twelve BJR 25 PDD points and expressed as the mean within one standard deviation, is shown for applicators A and F in figure 2. The average \( \Delta_{PDD} \) values for applicator A and filter 3 in figure 2(a) were: (−1.9 ± 1.6)%, (−1.6 ± 1.6)% and (−1.1 ± 0.9)% and (−0.7 ± 0.7)% for microDiamond, microSilicon, Edge, and PinPoint 3D detectors, respectively. Similarly, for applicator F and filter 4 the PDD differences in figure 2(b) were: (−1.1 ± 1.2)% and (−0.6 ± 1.4)%, (−1.1 ± 0.9)% and (−0.6 ± 0.7)% for microDiamond, microSilicon, Edge and PinPoint 3D detectors, respectively.

For the medium-energy superficial x-rays, i.e., 100 kVp and 120 kVp, corresponding to HVLs of 3.0 mm, 4.0 mm and 5.0 mm Al in this study, on average all measured data for the solid state detectors were within [−3.7, +1.4]% range relative to the published BJR 25 data. The largest outlier −3.7% was again observed for applicator I (18 cm diameter) for which the comparison was based on interpolated BJR 25 values. For all applicators smaller than applicator I the average agreement with BJR 25 data improved across the board, for all PDDs measured with solid state detectors. The PinPoint ionization chamber data was within [−1.3, +3.7]% range relative to the published BJR 25 data. However, this only applies to applicators A to H, while for the two largest applicators I and J, the recorded data for PinPoint detector became very noisy rendering it unsuitable for obtaining reliable PDD curves.

The average difference, \( \Delta_{PDD} \), compared to twelve BJR 25 PDD points, is plotted for applicators E and D in figure 3. The average differences for applicator E and filter 9 in figure 3(a) were (−0.6 ± 1.2)% and (−0.9 ± 1.3)% for micro-Diamond, microSilicon, Edge and PinPoint 3D detectors, respectively. Likewise, for applicator D and filter 6 in figure 3(b) the results were: (0.7 ± 1.4)%, (0.1 ± 1.4)% and (−0.6 ± 0.7)% and (2.8 ± 1.7)% for micro-Diamond, microSilicon, Edge and PinPoint 3D detectors, respectively.

For the high-energy superficial x-rays, i.e., the low-energy orthovoltage x-rays at 150 kVp, corresponding to HVL of 0.5 mm Cu, there are no published BJR 25 data for circular applicators. The only comparison was possible for the largest 20 × 20 cm² square applicator J at 50 cm FSD. In absence of BJR 25 data, figure 4(a) shows the intercomparison of measured PDDs for applicator C and filter 9 relative to micro-Diamond detector. The average differences, \( \Delta_{PDD} \), were (−0.2 ± 0.4)% and (1.6 ± 0.7)% and (1.2 ± 0.6)% for microSilicon, Edge and PinPoint 3D detectors, respectively. Figure 4(b) shows a comparison of composite PDD curve i.e., an average PDD of all three solid state detectors (microDiamond, microSilicon and Edge) for applicator J and filter 9, relative to
twenty-one BRJ 25 data points available for this energy. The average difference, $\Delta_{PDD}$, for the composite data was $(-0.7 \pm 1.0)\%$. The PinPoint 3D data was not considered here as the scan was too noisy to obtain clinically acceptable PDD curve.

The composite PDD values measured by three solid state detectors are summarized in table 6. The data shown are for filter 7, HVL 5.0 mm, 120 kVp, for all ten clinically used applicators, A-J. The appendix includes tabulated composite PDD data for all combinations of available filters and applicators, i.e., for 60 different clinical x-ray beams. The composite PDD values do not contain the PinPoint measurements since the ionization chamber data was not reliable for the two largest applicators I and J and because the ionization chamber scans exhibited small but systemic overestimation of measured PDD curves compared to solid state detectors.

Figure 2. Comparison between measured and BRJ 25 PDD data for the low-energy superficial x-rays (<100 kVp). (a) PDD curves for filter 3 (HVL 1 mm Al, 50 kVp), applicator A (1.5 cm diameter, 15 cm FSD). (b) PDD curves for filter 4 (HVL 2 mm Al, 80 kVp), applicator F (5 cm diameter, 15 cm FSD).
Discussion

The aim of this study was to present commissioning data from an Xstrahl 150 radiotherapy system. Extra effort was expended to acquire PDDs in water for ten applicator and six filter combinations, i.e., sixty beams of unique quality, using four different detectors. Altogether 240 PDD curves were acquired using a high-accuracy motorized scanning water phantom. For x-ray energies in kV range, liquid water is the only medium recommended for reference dosimetry as well as for percent depth dose measurements (Ma et al., 2001, Aukett et al., 2005). This recommendation is notably supported by findings reported by Hill et al. (Hill et al., 2010a) which showed the PDD differences compared to liquid water were −21.7% for Plastic Water and +17.6% for polystyrene, for 50 kVp photons collimated by an 8 cm diameter applicator. In

![Figure 3](image-url)

Figure 3. The comparison between measured PDD and BJR 25 data for medium-energy superficial x-rays (100 & 120 kVp). (a) PDD curves for filter 5 (HVL 3 mm Al, 100 kVp), applicator E (4 cm diameter, 15 cm FSD). (b) PDD curves for filter 6 (HVL 4 mm Al, 120 kVp), applicator D (3 cm diameter, 15 cm FSD).
general, the water equivalency of plastic or solid water phantoms is energy dependent. As the energy of kV beams increase, the dosimetric variations between water and water-equivalent phantoms decrease. Most, but not all, commercially available solid phantoms are water-equivalent within ±2% in kV range (Hill et al., 2010a).

For absolute and relative measurements in the kV range, reference dosimetry protocols (Ma et al., 2001, Aukett et al., 2005, Klevenhagen et al., 1996) recommend air-filled cylindrical ionization chambers. Well-designed cylindrical ionization chambers are considered the gold standard due to their nearly constant energy response for tube potentials between 40-300 kV. For relative dosimetry measurements such as PDD curves and relative output factors, small-volume cylindrical or plane-parallel ionization chambers are widely used. For instance, Sheu et al. (Sheu et al., 2015)

**Figure 4.** The comparison between measured PDD and BJR 25 data for high-energy superficial x-rays (150 kVp), i.e., low-energy orthovoltage x-rays. (a) PDD curve comparison relative to microDiamond detector for filter 9 (HVL 0.5 mm Cu, 150 kVp), applicator C (2.5 cm diameter, 15 cm FSD). No BJR 25 data for this field size and energy. (b) Composite PDD curve of three solid state detectors (microDiamond, microSilicon and Edge) for filter 9 (HVL 0.5 mm Cu, 150 kVp), applicator J (20 × 20 cm² square, 50 cm FSD).
used a PTW plane-parallel chamber N23342 and a solid water phantom to generate the PDDs for a Sensus SRT-100 x-ray unit. Johnstone et al. (Johnstone et al., 2015) used IBA FC65-G and IBA CC13 ionization chambers for PDD measurements in water, and Aspradakis et al. (Aspradakis and Zucchetti, 2015) used a Wellhöfer/IBA IC-15 ionization chamber and a PTW Advanced Markus plane-parallel ionization chamber to determine PDDs both in water and water-equivalent plastic phantoms.

It is worth mentioning that ionization chambers are not ideal for measuring at depths close to the surface. Measurement depth is limited to no less than the outer radius of the chamber, due to the perturbation at the surface when the chamber is sitting partially out of the water and because of the difference in relative detector response in air and in water (Hill et al. 2009).

As a consequence, ionization chamber PDDs are flattened out near the surface, resulting in overestimation of relative dose by several percent. In this study, the PDD scans utilizing an ultra-small size PinPoint 3D ionization chamber with a nominal sensitive volume of 0.016 cm$^3$ and radius of 1.45 mm agreed with the BJR 25 data within $[-2.6, +3.7]$% range for all filters and applicators A to H. However, due to the ultra-small sensitive volume for two largest applicators I and J at 30 and 50 cm FSD, respectively, the signal was too noisy for a reliable fit compared to the solid state detectors.

The utilization and accuracy of solid state detectors for kV x-ray beam dosimetry has been the subject of a limited number of investigations to date. The advantages of diamond and silicon detectors include very small active volumes, small angular dependence, and minimal energy dependence. The PTW microDiamond detector was proven suitable for PDD measurements

Table 6. Composite PDD values for filter 7, HVL 5.0 mm, 120 kVp, for all ten clinically used applicators A-J, measured utilizing three solid state detectors microDiamond, microSilicon and Edge.

| Applicator | A | B | C | D | E | F | G | H | I | J |
|------------|---|---|---|---|---|---|---|---|---|---|
| FSD 15 cm  |   |   |   |   |   |   |   |   |   |   |
| 1.5 cm     | 100.0 | 100.0 | 100.0 | 100.0 | 100.0 | 100.0 | 100.0 | 100.0 | 100.0 | 100.0 |
| 2.0 cm     | 98.5 | 98.7 | 98.9 | 99.0 | 99.2 | 99.3 | 99.7 | 99.7 | 99.7 | 99.8 |
| 2.5 cm     | 96.1 | 96.7 | 97.1 | 97.3 | 97.6 | 97.8 | 98.9 | 99.0 | 99.1 | 99.4 |
| 3.0 cm     | 93.8 | 94.6 | 95.3 | 95.5 | 96.0 | 96.3 | 98.0 | 98.3 | 98.5 | 99.0 |
| 4.0 cm     | 91.1 | 92.2 | 93.1 | 93.5 | 94.1 | 94.5 | 96.9 | 97.4 | 97.7 | 98.4 |
| 5.0 cm     | 88.3 | 89.7 | 90.7 | 91.3 | 92.0 | 92.5 | 95.7 | 96.2 | 96.8 | 97.8 |
| 10.0 cm    | 85.9 | 87.1 | 88.3 | 88.9 | 89.9 | 90.5 | 94.3 | 95.0 | 95.8 | 97.0 |
| 15.0 cm    | 82.7 | 84.4 | 85.8 | 86.5 | 87.7 | 88.5 | 92.9 | 93.7 | 94.7 | 96.2 |
| 18.0 cm    | 79.9 | 81.8 | 83.4 | 84.1 | 85.3 | 86.4 | 91.4 | 92.4 | 93.6 | 95.4 |
| 20 × 20 cm | 77.2 | 79.2 | 80.9 | 81.8 | 83.3 | 84.3 | 90.0 | 91.1 | 92.4 | 94.5 |

As a consequence, ionization chamber PDDs are flattened out near the surface, resulting in overestimation of relative dose by several percent. In this study, the PDD scans utilizing an ultra-small size PinPoint 3D ionization chamber with a nominal sensitive volume of 0.016 cm$^3$ and radius of 1.45 mm agreed with the BJ 25 data within $[-2.6, +3.7]$% range for all filters and applicators A to H. However, due to the ultra-small sensitive volume for two largest applicators I and J at 30 and 50 cm FSD, respectively, the signal was too noisy for a reliable fit compared to the solid state detectors.

The utilization and accuracy of solid state detectors for kV x-ray beam dosimetry has been the subject of a limited number of investigations to date. The advantages of diamond and silicon detectors include very small active volumes, small angular dependence, and minimal energy dependence. The PTW microDiamond detector was proven suitable for PDD measurements...
across a wide range of kV x-ray energies (50-280 kVp) and field sizes (Damodor et al., 2018, Khan et al., 2020, Daniel et al., 2022, Butler et al., 2018). For all kV energies in this study, the intercomparison between microSilicon and Edge detectors relative to microDiamond was within $[-0.6, +2.4\%]$ and $[-3.3, +0.7\%]$ range, respectively. The composite PDD values, representing the average of measured data by all three solid state detectors are included in the appendix.

In clinical practice, if a suitable detector is not available for relative dosimetry measurements then the BJR supplement 25 data should be used (Ma et al., 2001). There have been several studies that compared PDD measurements against BJR supplement 25 data. Johnstone et al. (Johnstone et al., 2015) found differences of $-6\%$ to $6\%$ for an 80 kVp beam and $-14\%$ to $15.7\%$ for medium-beam energies ($>100$ kVp). However, these results are at odds with a study by Aspradakis et al. (Aspradakis and Zucchetti, 2015) which found PDD differences within $\pm5\%$ for medium-energy beams and differences of more than $10\%$ for beam energies between 50 kVp and 100 kVp. These results support the conclusions of Hill et al. (Hill et al., 2010b) stating that discrepancies between measured PDDs and BJR 25 data could be attributed to differences in machines, detectors, phantoms, and measurement methods across various institutions. The results in this study were better matched with the published BJR 25 data. For the lower (<100 kVp) and medium-energy ($\geq 100$ kVp) superficial beams the average agreement was within $[-3.6, +0.4\%]$ and $[-3.7, +1.4\%]$ range, respectively. For the high-energy superficial (low-energy orthovoltage) x-rays at 150 kVp, the average difference for the largest 20 $\times$ 20 cm$^2$ collimator was $(-0.7 \pm 1.0\%)$. In summary it is very difficult to make direct study-to-study comparisons of PDD measurements, especially considering the BJR 25 data are an aggregate of measurements contributed by eight UK radiotherapy centers. As the BJR 25 report points out, direct kV measurements of clinical PDDs must be carried out if higher accuracy is needed.

Conclusions

This study presents commissioning data from an Xstrahl 150 x-ray system with a special focus on PDD measurements in water utilizing different detectors. The mechanical, safety, and dosimetric machine performance agreed favorably with specifications. Overall, the measured PDD curves exhibited excellent agreement with the BJR 25 data. The appendix provides tabulated PDD values with more datapoints than the BJR 25 for clinical intercomparison. We conclude that the BJR 25 PDD data still represent a reliable benchmark for measurements, as well as a valid alternative for clinics without resources to perform machine-specific relative dosimetry measurements.

Data availability statement

All data that support the findings of this study are included within the article (and any supplementary files).

ORCID iDs

Zhenyu Xiong @ https://orcid.org/0000-0001-6905-5307
Yuncheng Zhong @ https://orcid.org/0000-0002-0794-3138
Thomas I. Banks @ https://orcid.org/0000-0003-4786-6891
Robert Reynolds @ https://orcid.org/0000-0003-1015-2123
Tsuicheng Chiu @ https://orcid.org/0000-0001-9280-3641
Jun Tan @ https://orcid.org/0000-0001-5827-4694
You Zhang @ https://orcid.org/0000-0002-8033-2755
David Parsons @ https://orcid.org/0000-0002-4466-0647
Yulong Yan @ https://orcid.org/0000-0002-2288-3671
Andrew Godley @ https://orcid.org/0000-0002-8896-5231
Strahinja Stojadinovic @ https://orcid.org/0000-0002-6840-5747

References

Andreo P, Burns D, Hohlfeld K, Huq MS, Kanai T, Laitano F, Smyth V and Vynckier S 2000 IAEA TRS-398–Absorbed dose determination in external beam radiotherapy: an international code of practice for dosimetry based on standards of absorbed dose to water International Atomic Energy Agency 18 35–6
Aspradakis MM and Zucchetti P 2015 Acceptance, commissioning and clinical use of the WOrMed T-200 kilovoltage x-ray therapy unit Br. J. Radiol. 88 20150001
Attix FH 1986 Introduction to radiological physics and radiation dosimetry (New York: Wiley)
Aukett RJ, Burns JE, Greener AG, Harrison RM, Moretti C, Nahum AE, Rosser KE and Party LW 2005 Addendum to the IPEMB code of practice for the determination of absorbed dose for x-rays below 300 kV generating potential (0.035 mm Al-4 mm Cu HVL) Phys. Med. Biol. 50 2739–48
Bujila R, Omar A and Poludniowski G 2020 A validation of SpekPy: a software toolkit for modelling x-ray tube spectra Phys. Med. 75 44–54
Butler DJ, Beveridge T, Lehmann J, Oliver CP, Stevenson AW and Livingstone J 2018 Spatial response of synthetic microdiamond and diode detectors measured with kilovoltage synchrotron radiation Med. Phys. 45 943–52
Damodor J, Odgers D, Pope D and Hill R 2018 A study on the suitability of the PTW microDiamond detector for kilovoltage x-ray beam dosimetry Appl. Radiat. Isot. 135 104–9
Daniel J, Yousif YA M, Zifodya J and Hill R 2022 An evaluation of solid state detectors for the relative dosimetry of kilovoltage x-ray beams Med. Phys. 49 4082–91
Desrosiers M, Dewerd L, Deye J, Lindsay P, Murphy M K, Mitch M, Macchiariini F, Stojadinovic S and Stone H 2013 The importance of dosimetry standardization in radiobiology J. Res. Natl. Inst. Stand. Technol. 118 403–18
Fontenot J D, Alkhathib H, Garrett J A, Jensen A R, McCullough S P, Olch A J, Parker B C, YANG C C, Fairobent L A and Staff A 2014 AAPM medical physics practice guideline 2.c commissioning and quality assurance of x-ray-based image-guided radiotherapy systems J. Appl. Clin. Med. Phys. 15 4528

Gronberg M P et al 2020 A mail audit independent peer review system for dosimetry verification of a small animal irradiator Radiat. Res. 193 341–50

Healy B J and Hill R F 2022 Use of calculations to validate beam quality and relative dose measurements for a kilovoltage x-ray therapy unit Phys. Eng. Sci. Med. 45 537–46

Hill R, Healy B, Holloway L, Kuncic Z, Thwaites D and Baldock C 2014 Advances in kilovoltage x-ray beam dosimetry Phys. Med. Biol. 59 R173–231

Hill R, Kuncic Z and Baldock C 2010a The water equivalence of solid phantoms for low energy photon beams Med. Phys. 37 4555–63

Hill R, Mo Z, Haque M and Baldock C 2009 An evaluation of ionization chambers for the relative dosimetry of kilovoltage x-ray beams Med. Phys. 36 3971–81

Hill R F, Tofts P S and Baldock C 2010b The bland–altman analysis: does it have a role in assessing radiation dosimeter performance relative to an established standard? Radiat. Meas. 45 810–5

IAEA 2008 Setting Up a Radiotherapy Programme (Vienna: International Atomic Energy Agency)

IEC 60601-2-8:2010 Medical electrical equipment - Part 2-8: Particular requirements for basic safety and essential performance of therapeutic x-ray equipment operating in the range 10 kV to 1 MV [https://webstore.iec.ch/publication/2684]

ISO 2011 ISO/IEC GUIDE 98-3:2008/SUPPL 2:2011 Uncertainty of measurement — Part 3: Guide to the expression of uncertainty in measurement (GUM:1995) — Supplement 2: Extension to any number of output quantities. International Organization for Standardization [https://iso.org/standard/50463.html]

Johnstone C D, Lafontaine R, Poirier Y and Tambasco M 2015 Modeling a superficial radiotherapy x-ray source for relative dose calculations J. Appl. Clin. Med. Phys. 16 5162

Khan A U, Culberson W S and Dewerd L A 2020 Characterizing a PTW micro diamond detector in kilovoltage radiation beams Med. Phys. 47 4553–62

Klevenhagen S C, Aukett R J, Harrison R M, Moretti C, Nahum A E and Rosser K E 1996 The IPEMB code of practice for the determination of absorbed dose for x-rays below 300 kV generating potential (0.035 mm Al-4 mm Cu HVL; 10–300 kV generating potential) Phys. Med. Biol. 41 2605–25

Ma C M, Coffey C W, Dewerd L A, Liu C, Nath R, Seltzer S M, Seuntjens J P and AMERICAN ASSOCIATION OF PHYSICISTS IN MEDICINE (AAPM) 2001 AAPM protocol for 40–300 kV x-ray beam dosimetry in radiotherapy and radiobiology Med. Phys. 28 868–93

MATLAB version 9.11.0.1769968 (R2021b) (2021) The Mathwork, Inc., Natick, Massachusetts, US [https://www.mathworks.com/products/matlab.html]

Mccullough S P, Alkhathib H, Antes K I, Castillo S, Fontenot J D, Jensen A R, Matney J E and Olch A J 2021 AAPM medical physics practice guideline 2.b: commissioning and quality assurance of x-ray-based image-guided radiotherapy systems J. Appl. Clin. Med. Phys. 22 73–81

Megregor S, Minni J and Herold D 2015 Superficial radiation therapy for the treatment of nonmelanoma skin cancers J. Clin. Aesthet. Dermatol. 8 12–4

Nahum A E 1999 kV x-ray dosimetry: current status and future challenges ed C. M. MA and J. P. Seuntjens Kilovoltage x-ray beam dosimetry for radiotherapy and radiobiology (Madison, WI: Medical Physics Publishing)

Palmer A L, Pearson M, Whittard P, McHugh K E and Eaton D J 2016 Current status of kilovoltage (kV) radiotherapy in the UK: installed equipment, clinical workload, physics quality control and radiation dosimetry Br. J. Radiol. 89 20160641

Pidliti R, Stojađinovic S, Speiser M, Song K H, Hager F, Saha D and Solberg T D 2011 Dosimetric characterization of an image-guided stereotactic small animal irradiator Phys. Med. Biol. 56 2585–99

Poen J C 1999 Clinical applications of orthovoltage radiotherapy: tumours of the skin, endorectal therapy and intraoperative radiation therapy ed C. M. MA and J. P. Seuntjens Kilovoltage x-ray beam dosimetry for radiotherapy and radiobiology. Madison (WI: Medical Physics Publishing)

Poludniowski G, Omar A, Bujila R and Andreo P 2021 Technical note: SpekPy v2.0-a software toolkit for modeling x-ray tube spectra Med. Phys. 48 3630–7

Poludniowski G G 2007 Calculation of x-ray spectra emerging from an x-ray tube J. Med. Imaging (Bellingham) 4 2585–93

Poludniowski G G and Evans P M 2007 Calculation of x-ray spectra emerging from an x-ray tube J. Med. Imaging (Bellingham) 4 2585–93

Sheu R D, Powers A and Lo Y C 2015 Commissioning a 50-100 kV x-ray unit for skin cancer treatment J. Appl. Clin. Med. Phys. 16 5162

Williams J R and Thwaites D I 2000 Radiotherapy Physics: in Practice (Oxford, UK: Oxford University Press)

Xiong Z, Vijayan S, Rudin S and Bednarek D R 2017 Assessment of organ and effective dose when using region-of-interest attenuators in cone-beam CT and interventional fluoroscopy J. Med. Imaging (Bellingham) 4 031210

Yin F-F et al 2009 The Role of In-Room kV X-Ray Imaging for Patient Setup and Target Localization Report No. 104 [https://doi.org/10.37206/104]

Zhong Y, Lai Y, Saha D, Story M D, Jia X and Stojađinovic S 2020 Dose rate determination for preclinical total body irradiation Phys. Med. Biol. 65 175018