Impact of backscatter material thickness on the depth dose of orthovoltage irradiators for radiobiology research

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Abstract

The orthovoltage x-Ray energy frequently used in radiation research is prone to dosimetry errors due to insufficient backscatter conditions. In many radiobiology studies, especially for cell irradiations, precise dose calculation algorithms such as Convolution-Superposition or Monte Carlo are impractical and as such, less accurate hand calculation methods are used for dose estimation. These dose estimation methods typically assume full backscatter conditions. The purpose of this study is to demonstrate the magnitude of the dose error that results from insufficient backscatter, and to provide lookup tables to account this issue. The beam spectra of several widely used commercial systems (XRAD-225, XRAD-320, SARRP) were used in Monte Carlo (MC) simulations on a series of phantom setups to investigate the impact of varying backscatter conditions on dosimetry. The depth dose curves for different field sizes, water phantom thicknesses and beam qualities were generated. In addition, depth dependent backscatter factors for different field sizes and different beam qualities were calculated. It is demonstrated that as much as a 50% dose difference exists for different backscatter conditions at the beam qualities studied. The choice of cell dish size as well as other changes in the experiment setup can have more than 10% impact on the dose. The impact of backscatter is reduced with a decrease in field size. Further, the thickness needed to provide full backscatter can be approximated as being equal to the field size. It is imperative to ensure full backscatter conditions during system and dosimeter calibration, or to use the look-up table provided in this study.

Introduction

Many radiobiological effects are very sensitive to the radiation dose. For example, it has been demonstrated that a deviation of 3% in target dose can result in 10% change in the tumor control probability (Brahme, 1984). In addition, the responses of many organs at risk are sensitive to the change in the radiation dose, including spinal cord (Kirkpatrick *et al.*, 2010), lung (Marks *et al.*, 2010; Van Dyk *et al.*, 1981), gastro intestine (Kavanagh *et al.*, 2010), etc. Therefore, it is very important to ensure the accuracy of the dosimetry in radiation research. However, it has been shown that the dosimetry in many radiobiology studies were questionable (Desrosiers *et al.*, 2013). A survey of 4 radiobiology labs showed that dose errors on the order of 20%-40% existed in their experiments (Robbins, 2011). Another survey of the radiation biology irradiator at 12 laboratories showed that only one out of five facilities using x-ray irradiators delivered dose within 5% of the desired dose (Pedersen *et al.*, 2016). While it is recognized that the contributing reason is the lack of radiation physics training by many biologists, the complexity of the dosimetry at the energy levels used in radiobiology research poses significant challenges. Specifically, for radiobiology studies, due to the thickness of the cell plate and small animals, x-ray energies of 30-320 kVp are normally used (Yoshizumi *et al.*, 2011). Most dosimeters used clinically are unable to provide correct dose measurement due to the energy response, as shown by Desrosiers *et al.* (Desrosiers *et al.*, 2013). It has been recommended (Desrosiers *et al.*, 2013) that air-filled ionization chambers be used for calibration due to its flat energy dependence between 50 keV and 200 MeV and the Accredited Dosimetry Calibration Laboratory (ADCL) calibration, as well as many other advantages. The recommended calibration protocol is the AAPM TG-61 report (Ma *et al.*, 2001), which covers the x-ray energy of 40-300 kV.

AAPM TG-61 presented two calibration methods: “in-air” and “in-phantom”. The “in-phantom” method is only appropriate for energies >100 kV while the “in-air” method is good for all energies. Briefly, the

“in-air” method involves measuring air-kerma (“dose to air”) at a reference point in air (K_{air}^{in-air}) and converting it to water-kerma (K_w^{in-air}) by Eqn (1-2);

$$k_{air}^{in-air} = MN_k P_{stem,air} , \quad (1)$$

and

$$k_w^{in-air} = \left[\left(\frac{\mu_{en}}{\rho} \right)_{air}^{water} \right]_{air} k_{air}^{in-air} , \quad (2)$$

where M is the free-in-air ion chamber reading; N_k the air-kerma calibration factor for the beam quality; $P_{stem,air}$ the chamber’s stem correction factor, $\left[\left(\frac{\mu_{en}}{\rho} \right)_{air}^{water} \right]_{air}$ is the ratio of mass energy-absorption coefficients water to air, free in air. Finally, for dose to water at flat surface ($z=0$) at reference point ($D_{w,z=0}$) is established as:

$$D_{w,z=0} = B_w k_w^{in-air} , \quad (3)$$

where B_w is the backscatter factor which accounts for the phantom scatter effect. B_w is field size, beam quality and SSD dependent and TG-61 includes a table of B_w computed through Monte Carlo simulation. B_w can be over 1.4 for x-ray energies and SSDs commonly used in two popular commercially available radiobiology x-ray irradiator models (Xstrahl SARRP and Precision X-RAD) (Wong *et al.*, 2008; Deng *et al.*, 2007; Jaffray, 2007). However, the dose to water at the surface assumes there is enough material below the surface to fully build up the back-scatter. This is almost never the case for actual use. Often, the cell plate is placed on a plastic platform of a few millimeters thick. In the case of small animal irradiation, the small animal itself, plus the immobilization device, could provide several centimeters of backscattering material. However, it is likely far less than the B_w assumed in TG-61. There is no correction table or methodology to provide a better estimate of the dose at the surface under these conditions. As a result, researchers may either wrongfully assume that B_w given in TG-61 table applies to their setup, or they may use other dosimeters (such as film, OSLD) which add uncertainty and defeat the purpose of using the ADCL calibrated ion chamber.

The “in-phantom” method of TG-61 suffers similar issues in that the dose at the reference depth is calibrated for the backscatter condition of the calibration setup (usually has enough depth of water phantom to provide full backscatter). For the radiation experiments, using the calibrated dose at reference depth plus the percentage depth dose curve obtained in different setup will bring error in the estimation of dose administered.

Several sophisticated dose calculation engines have been developed, including convolution-superposition based(Jacques *et al.*, 2010; Cho *et al.*, 2018) and Monte Carlo based(Verhaegen *et al.*, 2011; van Hoof *et al.*, 2013; Bazalova *et al.*, 2009). While these dose calculation engines would be able to handle complex scatter conditions, they usually require the input of a volumetric representation of experiment setup. While high-end irradiators focusing on small animal use usually have integrated 3D imaging capabilities, many irradiator models designed for cell and tissue irradiation do not have these features. Even on irradiators with imaging capability, the imaging is usually not used for cell irradiation. Thus, in many occasions, dose has to be hand-calculated based on TG-61 calibration result. The correct interpretation of the scatter condition is critical in ensuring the dosimetry accuracy.

The dependence of backscatter build-up on the thickness of underlying material has been investigated previously with ion chamber(Klevenhagen, 1982), radiochromic film(Eaton and Doolan, 2013), and Monte Carlo simulations(Benmakhlouf *et al.*, 2012). While these studies provide a qualitative picture on the dependence, there are several deficiencies for it to be useful for the orthovoltage irradiators investigated in this study. First of all, the beam qualities in those studies do not match those used by the orthovoltage irradiators in this study. Both Benmakhlouf *et al.*(2012) and Klevenhagen(1982) only covers the diagnostic x-ray energy (50-150 kVp). While Eaton *et al.*(2013) measured a 220 kVp beam, it had a much harder beam (HVL: 2.2 mm Cu) than the 220/225 kVp irradiators for this study. Secondly, the backscatter factor determined through measurement could suffer from measurement uncertainty.

In Eaton et al.(2013), it was estimated to have 3.2% uncertainty on their result due to the EBT2 film. Klevenhagen(1982) used a specially designed parallel-plate ion chamber for surface dose measurement. While the ion chamber has much better reproducibility and consistency, the parallel chamber design usually has higher energy dependency varying by 20%-40% in kV x-rays(Ma *et al.*, 2001). Further correction may be required to account the spectrum change due to the increase of backscatter material thickness. Finally, those studies focused only on the influence to the surface dose. There were no results on whether the relative depth dose would change as the material thickness reduces. However, in cell irradiation experiments, the point of interest is usually at the bottom of cell dish with depth ranging from a few millimeters to centimeter. Therefore, it is desirable to have individual tabulated depth dose profiles for different material thickness.

The purposes of this study are to: (1) demonstrate the magnitude of the dosimetry error when insufficient backscatter exists for Precision X-RAD and Xstrahl SARRP irradiators; (2) provide a depth-dose look-up table as well as backscatter factor table at different thickness of water phantom for users of these two systems to use in their hand calculation. In addition, the impact of other materials and conditions in experimental setup, such as the use of cell-dish, stainless steel, low density polyurethane foam is also investigated.

Material and Methods

The PENELOPE Monte Carlo (MC) code (Salvat *et al.*, 2006) is used in this study. PENELOPE is a general-purpose Monte Carlo code system for the simulation of coupled electron-photon transport. It has been widely used and validated in various radiation therapy applications (Verhaegen and Seuntjens, 2003; Sterpin *et al.*, 2013; Sempau *et al.*, 2001), including low and medium energy x-ray (Ye *et al.*, 2004; Chica *et al.*, 2009). Since the present study focuses on the change in phantom backscatter at different phantom thickness, the detailed simulation of the x-ray tube and collimation was omitted. Instead, we

assume a point source with the photons' energy sampled from the photon spectra (Figure 1). The XRAD 225 and XRAD320 spectra and half value layer (HVL) values were provided by the manufacturer. The SARRP spectrum was provided by the original inventors at John Hopkins (Wong *et al.*, 2008) and the HVL value was provided by the manufacturer (Bazalova *et al.*, 2014). The detailed descriptions of the machines and beam qualities simulated in this study are listed in Table 1. For the rest of this paper, we will differentiate the two beam qualities of the XRAD 225 as X225(Cu) and X225(Al) to indicate the filter used.

Table 1 x-ray units and beam qualities studied.

Device	Energy	Added filter	HVL
Precision XRAD 225, Precision X-Ray Inc., North Branford, CT (X225)	225 kVp	0.3 mm Cu	0.92 mm Cu
		2.0 mm Al	0.37 mm Cu
SARRP, XStrahl Inc., Suwanee, GA (SARRP)	220 kVp	0.16 mm Cu	0.66 mm Cu
Precision XRAD 320, Precision X-Ray Inc., North Branford, CT (X320)	320 kVp	Thoraesus (0.8 mm Sn, 0.25 mm Cu, 1.5 mm Al)	3.86 mm Cu

Figure 1. Photon fluence spectra for the x-ray units and beam qualities studied.

As described in introduction, TG61 assumes there is enough material below the surface to provide full backscatter and eqn (1-3) were used to compute dose to surface. In the situation when there is not enough material to provide full backscatter, the backscatter factor will depend on the material thickness (T) as well. We define the thickness dependent backscatter factor as $B'_w(T)$ and call TG61's B_w as “full backscatter factor”. The eqn (3) is replaced with:

$$D_{w,z=0} = B'_w(T)k_w^{in-air}. \quad (4)$$

Note:
$$B_w = B'_w(\infty). \quad (5)$$

Equations 1,2 and 4 demonstrate how the dose to water at the phantom surface can be inferred from the ion chamber reading for the in-air measurement. On the other hand, the K_{air}^{in-air} , K_w^{in-air} , and $D_{w,z=0}$ can all be obtained directly through MC simulation in the geometry illustrated in Figure 2. In Figure 2a, the simulation is performed in air, with a voxel below the isocenter assigned a material of air but a physical density of 1.0 g/cm³ for better statistical precision of the MC dose. The dose tallied in the voxel is the air kerma K_{air}^{in-air} . Figure 2b is the same as 2a except the material of the tallying voxel become water. The dose tallied in this scenario is the water kerma K_w^{in-air} . In Figure 2c, a water phantom was placed with the top surface at isocenter. The water phantom is 51.2 cm by 51.2 cm with thickness (T) of [0.2, 0.4, 0.6, 0.8, 1.6, 2.4, 3.2, 6.4, 12.8, 41.2] cm simulated for our study. The maximum thickness simulated (41.2 cm) was assumed to be sufficient for the full backscatter buildup. The dose tallied at the surface is the surface dose D_w . From these simulation B_w and $[(\overline{\mu_{en}}/\rho)_{air}^{water}]_{air}$ can be computed following Equations 2-3 and compared with the TG-61 tables. The agreement can serve as a validation of our MC simulation.

Figure 2. Phantom setups in MC simulations. (a) A voxel with material of air right below the isocenter recorded the dose. (b) The same voxel with material of water recorded the dose. (c) A water phantom was placed with top surface at isocenter. Both the surface dose and the dose profile along the depth direction were recorded.

For Figure 2a and Figure 2b setup, $2e8$ photons per cm^2 were simulated. In addition, simulation was repeated 6 times with different random seed and dose in a single $2\text{ mm} \times 2\text{ mm} \times 2\text{ mm}$ voxel at isocenter was analyzed to estimate statistical uncertainty. For Figure 2c setup, simulation was not repeated since the statistical uncertainty can be reduced through spatial averaging. To compare with TG61 B_w table, square-field sizes of 1, 2, 3, 5, 10, 15, 20 cm were simulated. Note that the TG61 table assumes a circular aperture diameter for field size. To compare with square field size result we multiple the diameter by a factor of $\sqrt{\pi}/2$ to convert them to square field sizes (Podgorsak, 2005). For field sizes less than 5 cm, $2e8$ photons per cm^2 were simulated and depth dose at each 2mm depth increment was obtained by averaging a $6\text{ mm} \times 6\text{ mm}$ region around the beam axis. For field size greater than 5 cm, $2e7$ photons per cm^2 were simulated but spatial averaging was performed on $30\text{ mm} \times 30\text{ mm}$ region at each depth. For MC settings, electron cutoff energy of 1 keV and photon cutoff energy of 100 eV were used.

In addition to the standard TG61 calibration geometry, we simulated several scenarios that may be closer to the actual radiobiology experimental setup shown in Figure 3. The setups all involve a 30×30×1.2 cm³ square water block (simulating the supporting platform) with top surface at the isocenter. The difference is that: (a) another 30×30×1.2 cm³ square water block was placed on top as a reference; (b) a 1.2 cm thick water cylinder was placed on top to simulate cell dish, with diameter of the dish simulated as 3.5 cm, 6.0 cm and 10.0 cm; (c) same as (a) but with extra 4 mm stainless steel underneath; and (d) same as (a) but with extra 20 cm polyurethane foam with density of 0.03 g/cm³ underneath. Note that the parameters used in these simulations (material, size and thickness of supporting platform, size and thickness of cell dish, etc.) may not be the same as those used in actual experiment. The purpose of this simulation was to simply illustrate qualitatively the dosimetry impact with each setup. The top of the supporting platform is set at 50cm from the radiation source (dashed line). This put the SSD at 48.8 cm.

Figure 3. Different phantom setups to investigate the dosimetric impact of changes in phantom setup. (a) Same size of water blocks for reference. (b) Placing a cell dish on top of a supporting board. (c) Put a 4 mm thick stainless steel below the supporting board. (d) Putting a low density polyurethane foam below the supporting board.

Results

The air kerma, water kerma, and surface doses on a 41.2 cm-thick water block were simulated. The statistical uncertainty of air kerma and water kerma were 0.3% for each run and reduced to 0.12% after averaging. The statistical uncertainty for surface doses were <0.1% after spatial averaging. The $[(\overline{\mu_{en}}/\rho)_{air}^{water}]_{air}$ computed from our MC simulation were in good agreement with TG61 table lookup, as shown in Table 2. The X225(Al) result showed 1.29% deviation, most likely due to the difference between vendor-provided spectrum and the one used to generate the TG61 table.

Table 2. Comparison of the in-air mass energy absorption coefficient ratio from MC-simulation and from TG61 table look-up using the HVL of the beam.

Beam quality	$[(\overline{\mu_{en}}/\rho)_{air}^{water}]_{air}$		
	MC	TG61	Difference
X225(Cu) HVL: 0.92 mm Cu	1.074	1.072	0.14%
X225(Al) HVL: 0.37 mm Cu	1.053	1.040	1.29%
SARRP HVL: 0.66 mm Cu	1.066	1.060	0.54%
X320 HVL: 3.86 mm Cu	1.107	1.106	0.08%

Figure 4 shows the full backscatter factor simulated at different square field sizes and an SSD of 50 cm compared with the value from TG61 table. Note that the field diameter used in TG61 was already converted to square field size. There is excellent agreement between our MC and the TG61 table for X225(Cu). For X225(Al) there is around 1% difference near field size of 10cm but very good agreement elsewhere. For SARRP, the agreement is excellent at small field sizes but the difference grows to around 2% at field size of 20cm. The agreement for X320 is likely due to fact that the TG61 table around that HVL was created with filtered 280 kVp beam as well as mono-energetic 200 keV beam(Grosswendt,

1993, 1990), which are very different from the Thoraeus filtered 320 kVp beam. Overall, the agreements are very good, showing that our MC simulation settings and setups were reasonable.

The depth dose curves for Figure 2c setup is generated for X225(Cu) (Figure 5) and X225(Al) filter (Figure 6). The dose has been normalized with the water kerma at isocenter. The value at depth = 0 is the backscatter factor $B_w'(T)$ as defined in equation 5. For simplicity, only the field size of [20 cm, 10 cm, 3 cm, 1 cm], and the water phantom thickness of [4 mm, 8 mm, 24 mm, 64 mm, 412 mm] were plotted. For complete results, please refer to the PDD tables in Appendix A. Clearly, as the water thickness decreases, the dose also decreases due to the reduced contribution from phantom backscatter. This behavior is also affected by field size. For a 20 cm field size, the dose can vary by up to 50% as the water thickness changes. While at 1 cm field size, the impact of the water thickness is only a few percent.

Since the depth dose curves for the SARRP and X320 beams exhibit similar behavior as the X225 beams, we did not generate separate figures for them. The detailed depth dose relation of both beams is listed in Appendix A. Only the depth dose for the water thicknesses of 41.2 cm and 2.4 cm with 20 cm field size are shown in Figure 7 to demonstrate the effect of beam qualities on the depth dose curve. Under full backscatter condition (41.2 cm water thickness), the differences in beam quality are clearly reflected in the depth dose curve. SARRP, with HVL between the X225(Cu) and X225(Al), has the PDD curve between the X225 beams. In addition, the HVL difference between X225(Cu) and X225(Al) brings 10% difference in PDD curves at full scatter condition. However, the discrepancy reduced as the water thickness reduces. The PDD curve for X225(Cu) and X225(Al) and SARRP in 2.4cm-thick water block almost overlap each other. The X320, being on a much higher energy, exhibit a different depth dose curve than the other 3 beam, with lower surface dose and a slower dose fall-off.

Figure 4. Comparison of the full backscatter factor at different square field size and SSD of 50cm from our MC-simulation (line with circle symbols) and from TG61 table look-up using the HVL of the beam (asterisk symbols) with the field diameter converted to equivalent square field size.

Figure 5. Depth dose curves for different water thicknesses and field sizes for XRAD-225 with Cu filter. Dose is normalized to the water kerma at isocenter.

Figure 6. Depth dose curves for different water thicknesses and field sizes for XRAD-225 with Al filter. Dose is normalized to the water kerma at isocenter.

Figure 7. Comparison of Depth dose curves at different beam qualities and water phantom thickness. Solid line: X225 w Cu filter; Dashed line: SARRP; Dotted line: X225 w Al filter; Dash-dot line: X320. Symbols: water thickness of 2.4cm; No symbol: water thickness of 41.2cm.

The impacts of water phantom thickness (T) on the backscatter factor $B_w'(T)$ at different field sizes can be seen more clearly in Figure 8. Only the X225(Cu) and X225(Al) results are shown. The complete tables for all beam qualities are listed in Appendix B. As expected, the backscatter increases with the water phantom thickness and field size as shown in Figures 5 and 6. Of note, is that the water thickness has greater impact on the backscatter factor initially but gradually saturates as the thickness is increased. The saturated $B_w'(T)$ is the full backscatter factor B_w . For field size of 20 cm, the saturating water thickness is around 20 cm. As field size is smaller, the thickness required for full backscatter decreased accordingly. It seems that the field size is a good rule of thumb value when estimating how much material is needed to establish full backscatter condition. For example, the backscatter factor at 3cm field size for a 3.2cm-thick water block is only 1% off from the backscatter factor for 41.2cm-thick water block. The trending of the curve also showed that the maximum thickness used in the simulation (41.2 cm) provides enough backscatter for all field sizes.

Despite the difference in HVL, the shapes of the backscatter factor $B_w'(T)$ curves for two of the X225 beam qualities look similar, only the full backscatter factor at large water thickness differ. We

hypothesized that the $B_w'(T)$ of one beam quality $Q2$ can be converted to another beam quality $Q1$ through:

$$B_{w,Q1}'(T) = B_{w,Q2}'(T)^{B_{w,Q1}/B_{w,Q2}} \quad (6)$$

Figure 9 shows the backscatter factor $B_w'(T)$ for X225(Cu) beam quality as well as the same factor obtained from rescaling the $B_w'(T)$ curves for X225(Al) beam using Eqn. 7. The agreements are very good except for large field size at small water thickness where 3% errors are observed.

The depth dose curves for Figure 3 geometries using X225(Cu) beam are shown on Figure 10. As the figure shows, the variations in radiation setup can have significant impact to dose. At field size of 20 cm, a 4 mm stainless steel below the 2.4cm water phantom can increase the dose in the -12 mm to 0 mm region by 5%. The 0.03 g/cm³ polyurethane foam, while very low density, can still provide enough backscatter to boost the dose by 2% when it is 20 cm thick. On the other hand, when the 30×30×1.2 cm square water block was replaced with a 10cm diameter × 1.2 cm thick cell dish, the dose was decreased by 4%. And when the 10 cm diameter dish was replaced with 6 cm and then to 3.5 cm, the dose dropped further by 4% and 5% accordingly. Note the dose decrease is not limited to the cell dish (-12 mm to 0 mm region), it also extended to the supporting board (0 mm to 12 mm region) but with slightly reduced magnitude. The dependency on setup conditions is less pronounced when field size was reduced to 5 cm. The impact of 4 mm stainless steel became 1%-2.5% in the -12 mm-0 mm region. The polyurethane foam had negligible effect. The use of cell dish diameter of 10 cm and 6 cm also had negligible effect. Only when we switch to cell dish diameter of 3.5 cm, which is smaller than the radiation field size, showed a 4% decrease in dose.

The simulations for Figure 3 geometries were performed such that the distance from the radiation source to the surface of the supporting board was 50cm, resulting in an SSD of 48.8cm. The dose profiles obtained at 50cm SSD (Figure 5) were converted to 48.8cm SSD using an inverse square

correction and are shown as the dashed curve in Figure 10. Note the excellent agreement with the direct simulation at 48.8cm SSD.

Figure 8. The backscatter factor $B'_w(T)$ at field sizes of 1cm, 3cm, 5cm, 10cm and 20cm for X225(Cu) and X225(Al).

Figure 9. Comparison of the shape of backscatter factor $B'_w(T)$ under different beam qualities. The solid line and symbol shows the $B'_w(T)$ for X225(Cu), the dashed line shows the $B'_w(T)$ for X225(Al) rescaled with a $B_{w,Cu}/B_{w,Al}$ factor. Left shows the comparison at water thickness range of 0 – 120 mm, right is the zoomed in view showing the comparison in the 2-12 mm range.

Figure 10. Depth dose curves for Figure 3 geometries. The SSD is 48.8cm (SAD is 50.0cm). Depth doses are plotted from -1.2cm to 1.2cm around isocenter only. Left plot shows the result from field size of 20 cm, right plot shows the result from field size of 5 cm. Dashed curve is the depth dose curve for a 2.4cm thickness water phantom at 50.0cm SSD converted to 48.8cm SSD.

Discussion

Multiple publications have demonstrated close agreement between backscatter factors from MC calculation, measurements with radiochromic films and TLDs, and published tabulated values from IPEMB and AAPM protocols (Hewson *et al.*, 2018; Kim *et al.*, 2010; Nelson and Hill, 2011). Our MC simulation also confirmed this agreement in Figure 4. However, in our study, this comparison was served as a validation of our simulation design. The main focus of our study is to generate the tabulated value for the change of backscatter factor and depth dose profile when the full backscatter condition was not met, which was not covered in IPEMB and AAPM protocols.

The dose for orthovoltage x-ray units is strongly affected by the geometry and composition of the materials surrounding the sample. This behavior differs from that of megavoltage (MV) x-rays. While phantom scatter contributes significantly to dose at both energies, the scattering at MV is predominantly forward whereas there are significant backscatter contributions in the orthovoltage

range. Therefore, the dose in a phantom for MV beam depends primarily on the phantom material thickness before the point (depth) whereas for orthovoltage beam it depends on both the depth and the thickness of the material after the point (thickness – depth). This behavior can be clearly seen in the depth dose curves in Figure 5 and Figure 6, where the curve for 64 mm thickness phantom closely follows the curve for 412 mm thickness phantom until the last a few centimeter when there is insufficient material to provide phantom backscatter.

For the orthovoltage beam qualities studied, the forward scatter is also significant. This is indicated by the slight build-up section in the depth dose profile at field size of 20cm and phantom thickness of 41.2cm. In addition, the dose in a 3.5cm dish is smaller than the dose in reference conditions (illustrated in Figure 10) due to the same reason. If the dose primarily comes from backscatter, then dose in the 3.5cm dish should be higher because the backscatter from the region outside of the 3.5cm dish is higher than in the reference condition.

Understanding and quantifying the impact of the material thickness on the dose is very important in ensuring the dosimetric accuracy of the experiments performed using orthovoltage x-ray units.

Following TG-61's calibration procedure, the dose to water under full backscatter conditions can be determined. In contrast, typical cell or animal irradiations performed using orthovoltage x-ray irradiators are often conducted using geometries that do not create a full scatter condition, except when very small field sizes were used. Our simulations demonstrated up to 50% dose errors if this geometrical inconsistency remains uncorrected.

Radiochromic film, optically stimulated luminescence dosimeter (OSLD) and thermos-luminescent dosimeter(TLD), have also been used for irradiator dosimetry (Deng *et al.*, 2007; Rodriguez *et al.*, 2009). However, due to the energy dependent response of these dosimeters, a calibration factor has to be determined for the energy of interest. A common approach is to deliver a fixed amount of radiation to

the dosimeter and establish the relation between the responses to the known doses delivered, which is determined through TG-61 assuming full backscatter. It is important to ensure the full backscattering condition is met for the calibration irradiation. TG-61 calls for a water phantom with dimension of 30×30×30 cm³ or larger for in-phantom measurement. However, that is usually difficult to achieve given the limited space in many small animal irradiators. Various phantom thicknesses have been used for calibration, with the smallest being 3 cm (Rodriguez *et al.*, 2009). It is likely that the 3 cm thickness phantom only provided 80%-90% of the full backscatter for the beam quality (120kVp, HVL=5.6 mmAl) and field size (5 cm) used according to Klevenhagen (1982). It is recommended for X-RAD and SARRP users to refer to backscatter factor $B_w'(T)$ table in Appendix B1 or Figure 8 to ensure adequate phantom thickness is used for the given field size. In general, small field sizes (<4cm) can be used in generating film calibration curve to lessen the requirement of phantom thickness. However, the output factor of the smaller field needs to be considered if the TG-61 calibration is performed at a different field size.

Similar caution must be applied when calculating dose for an experiment setup using dose measured with TG-61 with an in-water measurement. Precision x-ray Inc. provides a 20cm diameter measuring cup-like container (Part No: XD1610-0100) with an ion-chamber at 2cm from the bottom for output and PDD measurement. This yields a backscatter thickness of 2cm, which is insufficient to provide full backscatter. Even the 6 cm phantom used by Azimi *et al.* (Azimi *et al.*, 2015) is likely to underestimate the dose under full scatter conditions by 4% at a 20cm field size, (refer to backscatter factor table in Appendix B1 for values at 6.4cm vs 41.2cm water phantom thickness). Therefore, it is strongly recommended that the tables in Appendix A and B be consulted to ensure the correct calibration and extensions to other backscatter conditions in experiments.

368 Although Appendix A only lists the depth dose table for a 50cm SSD, the dose at other SSD can be
369 computed through use of the inverse square law. Note that this assumes that the field sizes at each
370 depth for the two SSDs are equal. When the SSD change is large, such that this assumption does not
371 hold true, we can still perform the inverse square law correction and then estimating the residue error
372 from the backscatter factor $B_w'(T)$ table (Appendix B) by plugging in the backscatter thickness from the
373 point of interest, and the field size difference at the point of interest.

374 TG-61 created lookup tables for 36 beam qualities. In this study, we only have 4. However, the good
375 match in Figure 9 shows that for beam qualities slightly different from the four studied, Eqn. 7 can
376 provide a very good approximation for the backscatter factor at a given backscatter water phantom
377 thickness.

378 The limitation of this study is that most radiobiology experiments setups are more complex than simple
379 water blocks. The depth dose plots in Figure 10 illustrated that the depth dose curves for simple
380 geometries illustrated in Figure 2c can carry errors of up to 12.5% percent if it is used directly for
381 different experimental setups. Correction strategies similar to those used in second dose calculation in
382 human radiation therapy needs to be employed to reduce the discrepancy. Nevertheless, the tables
383 generated in this study still serves as a better starting point for the further correction. In addition,
384 Figure 10b illustrates a mitigation strategy of reducing the field size used so that the backscatter
385 condition has less impact. For the same 3.5 cm dish, reducing the field size from 20cm to 5cm, while still
386 maintaining the coverage of the entire dish, reduces the dose error from 12.5% to 5% before any
387 correction is applied. The dose error can be reduced further if a 3.5 cm field size is used, or if a
388 correction is estimated directly from Figure 10.

Conclusions

MC simulation was used to study the impact of the backscatter water phantom thickness on the depth dose profile and backscatter factor for two commercially available radiobiology x-ray irradiators on the market (Xstrahl SARRP and Precision X-RAD). Dose at the orthovoltage energies commonly used in radiobiology research is sensitive to the amount of backscatter material and other differences in setup (such as the size of cell dish used). Depth dose tables and backscatter factor tables for different thickness of backscatter material were created for Precision X-RAD and Xstrahl SARRP irradiator dosimetry. At the beam qualities studied, the material thickness for the establishment of full backscatter condition at different field size is determined. It is very important to either ensure full backscatter conditions during system and dosimeter calibration or to lookup the depth dose table and backscatter factor table in the Appendix for correction factors.

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