Modelling of an Orthovoltage X-ray Therapy Unit with the EGSnrc Monte Carlo Package

Tommy Knöös
Per Munck af Rosenschöld
Elinore Wieslander

Radiation Physics, Lund University Hospital, S-221 85 Lund, Sweden
Tommy.knoos@med.lu.se

Abstract. Simulations with the EGSnrc code package of an orthovoltage x-ray machine have been performed. The BEAMnrc code was used to transport electrons, produce x-ray photons in the target and transport of these through the treatment machine down to the exit level of the applicator. Further transport in water or CT based phantoms was facilitated by the DOSXYZnrc code. Phase space files were scored with BEAMnrc and analysed regarding the energy spectra at the end of the applicator. Tuning of simulation parameters was based on the half-value layer quantity for the beams in either Al or Cu. Calculated depth dose and profile curves have been compared against measurements and show good agreement except at shallow depths. The MC model tested in this study can be used for various dosimetric studies as well as generating a library of typical treatment cases that can serve as both educational material and guidance in the clinical practice.

1. Introduction
The main purpose of this study was to investigate how well it is possible to reproduce measured data for an orthovoltage x-ray machine with simulations using the EGSnrc Monte Carlo codes. The BEAMnrc code was applied for the transport through the treatment unit and the DOSXYZnrc code for further transport in either water phantoms or patients based on x-ray CT data. These Monte Carlo codes have extensively been used in the mega voltage (MV) region but only a few studies have been performed in the kilo voltage (kV) region. For example, the cone beam kV unit on an Elekta accelerator have been modeled by Jarry et al 2006. Another application in the kV region has been published by Omrane et al 2003 regarding surface dose calculations for diagnostic x-rays. An earlier work studying the physics in this lower energy range with the EGSnrc code was published by Verhaegen 2002.
2. Material and Methods

1. The treatment unit
The orthovoltage machine in this study is a modern unit capable of delivering x-rays produced by accelerating voltages from 20 to 220 kV with tube currents up to 20 mA (D3225\footnote{ Latest name is Xstrahl 200 according to the company’s web page at \url{http://www.gulmaymedical.com}.} from Gulmay Medical Ltd, UK). The unit has a transmission chamber for monitoring the dose rate and built-in corrections in the computer control software for ambient conditions i.e. temperature and pressure. Filters with various thicknesses are available which in combination with tube voltage define the beam quality. The unit studied in this work uses fixed applicators with open ends to define the exposed field. The applicators come with different length (treatment distance) and openings (field sizes). For this study we have chosen the 120 and 200 kV with added filtration of 2.0 mm Al and 0.5 mm Cu respectively. All simulations were performed for a 10x10 cm\(^2\) field at 50 cm distance (Source to surface/skin distance – SSD).

2. Monte Carlo simulations
The Monte Carlo codes applied were the EGSnrc based BEAMnrc and DOSXYZnrc from the NRCC group (Rogers \textit{et al}, 1995). The EGSnrc was developed from the EGS4 code by Nelson \textit{et al} 1985. For this purpose the latest version of the EGSnrc engine is used which includes directional bremsstrahlung splitting (DBS) (Kawrakow \textit{et al}, 2003, Mainegra-Hing and Kawrakow, 2006) to increase the efficiency of energy transition from the electron current to x-ray photons. The electron impact ionization model is also implemented which significantly improves the shape of the x-ray spectra as described by Kawrakow 2002. The treatment head was simulated in the BEAMnrc code and all data was gratefully received from the vendor for making this feasible. In figure 1, a screen dump from BEAMnrc is shown describing the components used for the simulations.

Figure 1. Schematic description of the geometry used for the BEAMnrc simulations. To the left the used component modules (CMs) are listed together with the system/component in the x-ray machine they represents. The electron beam is impinging normally to the surface of the anode (target) as indicated with the red arrow.
Phase spaces i.e. the position, direction and energy of each particle when crossing a specific plane were scored at the exit level of the applicator. The energy spectrum for the central part (whole field except in the proximity of the applicator) was compiled from the phase space and the half-value layer (HVL) was determined for the radiation field. The following equation was used to determine the amount of material $t$ to attenuate the air KERMA, $K$ to half of its measure when no absorber was present:

$$K(t) / K(0) = \left( \sum_{i=1}^{N} \phi_i E_i \left[ \frac{\mu_{en}}{\rho} (E_i) \right]_{\text{Air}} \right) \cdot \exp \left( - \mu(E_i)_{\text{absorber}} t \right) \Delta E_i \cdot \left( \sum_{i=1}^{N} \phi_i E_i \left[ \frac{\mu_{en}}{\rho} (E_i) \right]_{\text{Air}} \right)^{-1} = 0.5$$

The summation was performed over $N$ energy bins with width $\Delta E_i$ and mid point energy $E_i$. The photon fluence per energy bin $\phi_i$ was extracted from the phase spaces with the utility program BEAMDP$^2$. The energy-dependent attenuation $\mu$ and mass absorption coefficients $\frac{\mu_{en}}{\rho}$ are from Hubbell and Seltzer, 1995. This summation was performed iterative until a solution was found giving the thickness of the absorber $t$ to half the $K$ in air.

The settings in EGSnrc were: Boundary crossing algorithm – EXACT, with skin depth 3 mean free path, electron steeping using PRESTA II, spin effects ON, simple bremsstrahlung angular sampling, and cross section for bremsstrahlung production according to NIST. Bound Compton scattering, electron impact ionization and atomic relaxations were set to ON while photo electron angular sampling was OFF.

\[2\] Included in the BEAMnrc package from NRCC.
Table 1. Performance statistics for different settings of electron transport parameters. The last row represents the data for the parameters used in the rest of the study. The first row represents the results presented during the workshop and the last row the present study.

| Parameters                  | Electrons per hour | x-ray photons per electron | Time to generate 50x10^9 photons in the phase space (days) |
|-----------------------------|--------------------|----------------------------|------------------------------------------------------------|
|                             | 120 kV             | 200 kV                     | 120 kV                                                     | 200 kV |
| Original (No DBS)           | 9.36x10^6          | 8.21 x10^6                 | 0.00469                                                   | 0.00603 |
| AE=512 keV                  | 1.40x10^7          | 3.45 x10^4                 | 2.38                                                      | 3.42    |
| ECUT=512 keV                | 2.28x10^7          | 1.23x10^7                  | 2.41                                                      | 3.43    |
| AE=512 keV                  | 2.40x10^7          | 1.35x10^7                  | 2.41                                                      | 3.44    |

Depth dose curves along the central axis and profiles at 2 cm depth along the major axis were scored in a 25x25x25 cm³ water phantom consisting of cubic voxels with 0.25 cm side. The code for this task was the DOSXYZnrc.

3. Patient simulations
Two common clinical situations where orthovoltage may be used were identified in our clinical database of patients. Their CT sets were unidentified and transferred to the DICOM toolbox (Spezi et al., 2002) where the studies were resampled to 0.5 cm cubic voxels with the average density based on the Hounsfield numbers. From these numbers the materials of the voxels were also mapped. The following default materials were used – air, lung, tissue and bone according to ICRU report 44 (1989). The first case selected was a thoracic wall treatment where the beam impinged normal to the surface. The other case was a head and neck treatment with a lateral beam at the level of the mandible.

4. Measurements
Depth dose curves as well as profiles were determined with three types of detectors in a scanning water phantom system. Two different ionisation chambers, a cylindrical Farmer type chamber and a plane parallel chamber (FC65-G and NACP-02 from Wellhofer-Scanditronix, Germany) were used. The IAEA TRS-398 protocol recommends that relative dosimetry in kV beams should be performed with cylindrical chambers (IAEA 2000). Measurements were made with the cylindrical chamber from 0.7 to 20 cm depth and complemented with plane-parallel chamber measurements in the shallow depths. This latter step was due to that the shaft of the cylindrical chamber would collide with the applicator. These measurements were complemented with results from a diamond detector (60003, from PTW, Germany). All detectors were connected to the built-in electrometer of the beam scanning system (RFA from Wellhofer-Scanditronix). The change in the ratio of water-to-air mass energy–absorption coefficient along the central depth dose has been estimated to be within 1.0% (Seuntjens and Verhaegen, 1996). This implies that the ionization charge directly can be assumed to be proportional to the absorbed dose for the two gaseous chambers. The signal from the diamond was corrected for the variation in dose rate (Björk et al. 2000). The diamond detector was also pre-irradiated to about 5 Gy before use.

3. Results and discussion
During the tuning of the Monte Carlo simulations it was found that a slight shift in the x-ray spectra towards higher energies resulted when Rayleigh scattering was turned on, thus all simulations are done with this photon interaction enabled, cf Figure 1. The influence from transport parameters on the x-ray spectra was investigated by using different AE (energy cut off for explicit electron interaction modelling) and ECUT (energy cut off for transport of electrons) for the electrons while the PCUT (energy cut off for transport of photons) always was equal to 1 keV. For electron energies lower than...
Figure 3. Depth dose curves for the 120 kV x-ray beam from the 10x10 cm applicator in water for Monte Carlo and measurements (left scale). Normalisation of each data set is performed at 2 cm depth. Differences (%) between Monte Carlo and measurements are shown for each point (right scale).

Figure 4. Depth dose curves for the 200 kV x-ray beam from the 10x10 cm applicator in water for Monte Carlo and measurements (left scale). Normalisation of each data set is performed at 2 cm depth. Differences (%) between Monte Carlo and measurements are shown for each point (right scale).
AE transport of electrons is based on the continuous slowing down approximation. Photons are terminated when their energy reach below PCUT. For all settings when Rayleigh scattering were included no significant difference was seen between the spectra.

For various combinations of values of these parameters the number of initial electrons per unit time that could be simulated was scored as well as the number of photons in the phase space resulting at average per initial electron.

From Table 1 it is clear that DBS improves the efficiency with a factor of 10 which have been shown earlier (Mainegra-Hing and Kawrakow, 2006). The choice of the electron transport parameters are of importance for the production of Bremsstrahlung photons in the target of the x-ray tube. Table 1 shows that raising the two parameters to 10 keV decreases the calculation times about a factor of two and four for the 120 and the 200 kV beams, respectively. Turning the Rayleigh scatter process “off” could further speed up the calculations a factor of two but since the inclusion resulted in a significant change of the x-ray spectra it was decided to include the process.

The initial simulation with all geometry parameters as the vendor had supplied gave HVL values that differed substantially (at least for 120 kV) from those determined experimentally. After reviewing all drawings, contacting the vendor, looking at data sheets from the x-ray tube maker, reviewing the measurements nothing were revealed that could explain the difference. It was then decided that additional material had to be added to increase the HVL. The results was that 0.52 mm Al (120 kV) and 5.2 mm Al (200 kV) had to be added for the two energies to match the measured HVL values.

Depth doses were scored in a homogeneous water phantom simulated with DOSXYZnrc. In figure 3 and 4 the simulations are compared to experimental measurements with an ionization chamber and a diamond detector. Except for the first centimetre or so a fairly good agreement between simulations and measurements were found for both energies. Deviations between parallel chambers as used at shallow depth in this work have been noticed by Hugtenberg et al 2001. They compared depth doses from parallel chambers and diamond detectors with Monte Carlo simulations and found a better agreement with the latter detector.

The dose profiles along the major axis were scored at 2 cm depth. These results are shown in figure 5 and 6. The profiles from the simulations were in principal similar to the measurements with the diamond detector. Agreement was rather well for 120 kV but little bit less good for the higher energy which partly maybe contributed to the uncertainty in set-up and alignment since this energy was studied last and prior to the scans one had to change the filter of the machine which may have lead to an un-noticed movement of the tube relative to the phantom. The so called heel effect is just about visible for the simulations for both energies. About 1 % difference in dose along the profile was found for the higher energy evaluated at plus/minus 2.5 cm indicating the presence of a heel effect. This was less pronounced for the lower energy.

The patient studies were done in two ways: First simulation set using the materials mapped from the Hounsfield numbers and the corresponding density. The second set assumed all materials present be equal to water with density given by the Hounsfield numbers. Dose normalisation was done according to the principle of output normalisation i.e. the dose distributions are relative to an absorbed dose of one gray or 100 % at 2 cm depth in homogenous water.

The first case, a single lateral beam impinging on the neck is presented in figure 7 for the 120 and 200 kV respectively. The simulations took about seven to nine hours to perform for 1.5x10^9 histories with an uncertainty of about 0.3 % in the high dose region when the two energy cases were running in parallel on a Pentium 4 3.2 GHz CPU with hyper threading enabled. The treatment of a thoracic wall is presented in figure 8. In the cases when the correct media are used one can note that absorbed doses within the voxels containing bone tissues are high, up to a factor of 3-4 compared to soft tissue. This effect is due to the fact that the energy absorption per unit energy fluence within bone is much higher than in soft tissue (or water). This ratio of energy absorption differences is also known as the f-ratio (c.f. Ma and Seuntjens 1999).
Figure 5. Dose profiles along the major axis at 2 cm depth (X – inplane i.e. along the x-ray tube, Y – crossplane). Measured data are performed with a diamond detector.

Figure 6. Dose profiles along the major axis at 2 cm depth (X – inplane i.e. along the x-ray tube, Y – crossplane). Measured data are performed with a diamond detector.
Figure 7. The head neck case with a beam impinging horizontally from the left side of the patient. Upper row shows the 120 kV while the lower row 200 kV. To the left are dose distributions calculated in media while to the right all voxels are filled with water. The isodose lines shown are 5, 10, 20, 30, 40, 50, 70, 80, 90, 95, 100, 105, 110, 150, 200 and 300 %.
Looking closer at the dose distribution when dose were calculated to the media and especially the thoracic case reveals that high dose areas are present within regions such as the heart etc where some voxels have been assigned bone due to high Hounsfield numbers. This just reminds us that the translation of Hounsfield numbers to media is of high importance when dose to tissue is calculated either with Monte Carlo principles as in this study or with other methods.

4. Conclusions

This study have shown that it is possible to Monte Carlo simulate an orthovoltage machine with rather high accuracy when compared to measured data. The time required for computer simulations have decreased substantially (a factor of 10-100) especially after the latest modifications regarding for example directional bremsstrahlung splitting (DBS) made by the NRCC group (Kawrakow et al 2004). Thus the earlier limitation with very low electron-to-x-ray conversion rates in the order of 1% resulting in extremely long simulation times have been removed. This will further be improved when the latest development becomes available to the public, i.e. the bremsstrahlung cross section enhancement, BCSE (personal communication Dave WO Rogers 2006). These improvements will together give the opportunity to more or less perform Monte Carlo treatment planning for orthovoltage treatment. Maybe that is not the goal but producing typical plans based on CT data which are available to the radiation oncologist during prescription would improve the treatments. The MC model tested in
this study can be used for various dosimetric studies as well as generating a library of typical treatment cases that can serve as both educational material and guidance in the clinical practice.

5. Acknowledgement
We acknowledge the fruitful discussions at the workshop (European Work Group for Monte Carlo Treatment Planning in Gent, Belgium 2006) especially with Dr Dave WO Rogers and Dr Iwan Kawrakow that led to much more efficient and accurate Monte Carlo simulations. The support from the Lund University Hospital Research funds especially for computer hardware is hereby also greatly appreciated making this study realized.

6. References
[1] Björk P, Knöös T and Nilsson P: Comparative dosimetry of diode and diamond detectors in electron beams for intraoperative radiation therapy, Med. Phys. 27, 2580-2588, 2000.
[2] Hubbell JM and Seltzer SM, Tables of X-Ray Mass Attenuation Coefficients and Mass Energy-Absorption Coefficients from 1 keV to 20 MeV for Elements Z = 1 to 92 and 48 Additional Substances of Dosimetric Interest Ionizing Radiation Division, Available from Physics Laboratory National Institute of Standards and Technology, Gaithersburg, MD 20899, 1995, See also http://www.nist.gov.
[3] Hugtenburg RP, Johnston K, Chalmers GJ, Beddoe AH, Application of diamond detectors to the dosimetry of 45 and 100 kVp therapy beams: comparison with a parallel-plate ionization chamber and Monte Carlo. Phys Med Biol 46 (2001) 2489–2501.
[4] IAEA, Absorbed Dose Determination in External Beam Radiotherapy: An International Code of Practice for Dosimetry based on Standards of Absorbed Dose to Water, IAEA TRS-398, 2000. (Available at http://www-naweb.iaea.org/nahu/dmrf/codeofpractice.shtml).
[5] Jarry G, Graham SA, Moseley DJ, Jaffray DJ, Siewerdsen JH, Verhaeghen F. Characterization of scattered radiation in kV CBCT images using Monte Carlo simulations. Med Phys. 2006 Nov;33(11):4320-9.
[6] Kawrakow I, Rogers DWO, Walters BR, Large efficiency improvements in BEAMnrc using directional bremsstrahlung splitting. Med Phys. 2004 Oct;31(10):2883-98.
[7] Kawrakow, I Electron impact ionization cross sections for EGSnrc, Med. Phys. 29, 1230, 2003.
[8] Kawrakow, I. Rogers DWO, The EGSnrc Code System: Monte Carlo Simulation of Electron and Photon Transport, NRCC Report PIRS-701, 2003
[9] Ma C-M, Seuntjens JP, Mass-energy absorption coefficient and backscatter factor ratios for kilovoltage x-ray beams, Phys Med Biol. 1999 Jan;44(1):131-43
[10] Mainegra-Hing E, Kawrakow I. Efficient x-ray tube simulations. Med Phys. 2006 Aug;33(8):2683-90.
[11] Nelson WR, Hirayama H, and Rogers DWO, The EGS4 Code System, Report SLAC–265, Stanford Linear Accelerator Center, Stanford, California, 1985.
[12] Omrane LB, Verhaeghen F, Chahed N, Mtimet S. An investigation of entrance surface dose calculations for diagnostic radiology using Monte Carlo simulations and radiotherapy dosimetry formalisms. Phys Med Biol. 2003 Jun 21;48(12):1809-24.
[13] Rogers DWO, Faddegon BA, Ding GX, Ma C-M, Wei J and Mackie TR, BEAM: A Monte Carlo code to simulate radiotherapy treatment units Medical Physics 22, 503 – 524, 1995.
[14] Seuntjens J, Verhaeghen F. Dependence of overall correction factor of a cylindrical ionization chamber on field size and depth in medium-energy x-ray beams Med Phys 23, 1789-1796, 1996.
[15] Spezi E, Lewis DG, Smith CW. A DICOM-RT-based toolbox for the evaluation and verification of radiotherapy plans, Phys Med Biol. 2002 Dec 7;47(23):4223-32.
[16] Verhaeghen F, Evaluation of the EGSnrc Monte Carlo code for interface dosimetry near high-Z media exposed to kilovolt and 60Co photons. Phys Med Biol. 2002 May 21;47(10):1691-705.