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ORIGINAL PAPER

Validation of XiO Electron Monte Carlo-based calculations by measurements in a homogeneous phantom and by EGSnrc calculations in a heterogeneous phantom

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Received 25 February 2012; received in revised form 3 August 2012; accepted 11 August 2012

KEYWORDSTreatment planning;
Electron beams
dosimetry;
Heterogeneous
phantom;
XiO Electron Monte
Carlo

Abstract The purpose of the present study is to perform a clinical validation of a new commercial Monte Carlo (MC) based treatment planning system (TPS) for electron beams, i.e. the XiO 4.60 electron MC (XiO eMC). Firstly, MC models for electron beams (4, 8, 12 and 18 MeV) have been simulated using BEAMnrc user code and validated by measurements in a homogeneous water phantom. Secondly, these BEAMnrc models have been set as the reference tool to evaluate the ability of XiO eMC to reproduce dose perturbations in the heterogeneous phantom. In the homogeneous phantom calculations, differences between MC computations (BEAMnrc, XiO eMC) and measurements are less than 2% in the homogeneous dose regions and less than 1 mm shifting in the high dose gradient regions. As for the heterogeneous phantom, the accuracy of XiO eMC has been benchmarked with predicted BEAMnrc models. In the lung tissue, the overall agreement between the two schemes lies under 2.5% for the most tested dose distributions at 8, 12 and 18 MeV and is better than the 4 MeV one. In the non-lung tissue, a good agreement has been found between BEAMnrc simulation and XiO eMC computation for 8, 12 and 18 MeV. Results are worse in the case of 4 MeV calculations (discrepancies $\approx$ 4%). XiO eMC can predict dose perturbation induced by high-density heterogeneities for 8, 12 and 18 MeV. However, significant deviations found in the case of 4 MeV demonstrate that caution is necessary in using XiO eMC at lower electron energies.

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Introduction

In the patient radiotherapy treatment process, accurate dose calculation is one of the important steps between the dose prescription to the clinical target volume and the actual dose delivery. Ensuring treatment quality in radiation therapy requires that the best equipment and techniques should be available for treatment planning. This includes the actual purchase and clinical implementation of a treatment planning system (TPS) which is able to compute absorbed dose distributions with an acceptable accuracy, especially if tissue heterogeneities are present. Monte Carlo (MC) methods have proven to be among the best tools for computing radiotherapy dose [1–4]. One of the important developments in the MC technique was the release of the BEAMnrc [5], an EGSnrc-based [6] radiotherapy simulation package. As the MC method is by its nature very time consuming and impractical for radiotherapy applications, a number of approximations and simplifications to speed up calculations have been included in the commercial MC dose calculation engines, leading to the development of fast MC algorithms [7–12] implemented in some TPS for patients dose computation. Traditionally, the clinical implementation of every MC based TPS involves design compromises with possible errors. Therefore, it is important to perform independent validation under conditions similar to those found in clinic [13–15]. Recently, our radiotherapy department of the University Hospital Saint Luc, bought a license of a new MC based TPS for four electron beam energies 4, 8, 12 and 18 MeV. The above TPS, XiO 4.60 electron MC (XiO eMC) is developed and commercialized by ELEKTA CMS SOFTWARE group (ECMSSG). In the present study, we followed the commissioning approach already presented in our previous work [15]. As this TPS was not yet studied for ELEKTA SL25 linear accelerators, we have been careful in configuring XiO eMC. In addition to dosimetric data for beam characterization (percentage depth dose, off-axis profiles and output factors), as specified in the user manual, the accuracy of XiO eMC model for a given linear accelerator type depends on other manufacturer specifications, e.g. the opening jaws for each field size, the applicator designs, applicator material and serial numbers.

Materials and methods

Measurements

Homogeneous water phantom measurements

Dose distributions in the water tank ($40 \times 40 \times 40 \text{ cm}^3$) for electron beam of 4, 8, 12 and 18 MeV delivered by an ELEKTA SL25 medical linear accelerator have been measured within an IBA computerized water phantom system and analyzed with OmniPro-Accept, version 6.6B software. Measurements have been carried out for open field sizes of $6 \times 6 \text{ cm}^2$, $10 \times 10 \text{ cm}^2$, $14 \times 14 \text{ cm}^2$, $20 \times 20 \text{ cm}^2$ and the $5 \times 5 \text{ cm}^2$ insert within the $14 \times 14 \text{ cm}^2$ open applicators, at two SSDs of 100 and 115 cm. Percentage depth dose and off-axis profiles have been measured with a Wellhöfer type CC01 thimble ionization

chamber having 0.01 cm^3 collecting volume. Depth ionization curves have been converted to the percentage depth dose by multiplying the ionization charge at each measurement depth by the corresponding water/air stopping-power ratios derived from the IAEA, TRS 398 protocol [16]. Output factors have been measured for open field size of $6 \times 6 \text{ cm}^2$, $10 \times 10 \text{ cm}^2$, $14 \times 14 \text{ cm}^2$ and $20 \times 20 \text{ cm}^2$ open applicators and several highly asymmetric insert cutouts. Measurements have been performed at the individual depth, R_{100} , of each field based on its PDD, with two ionization chambers: a plane-parallel ionization chamber type NACP17-02, active volume 0.16 cm^3 for large field sizes measurements and a Scanditronix-Wellhöfer IC13 thimble ionization chamber, having a cavity volume of 0.13 cm^3 , for measurements in small field sizes. All cross-beam profiles have been measured at several depths, namely, R_{100} , R_{90} , R_{80} , R_{50} , R_{20} and at 2 cm away from the practical range ($R_p + 2 \text{ cm}$), where R_j is the depth, beyond the depth of maximum dose R_{100} , at which the dose at the central axis falls off by $J\%$ of the maximum dose. In this work, output factors (OF) are defined as:

$$\text{OF} = \frac{D_{\max}(E, S, \text{SSD})}{D_{\max}(E, 10 \times 10 \text{ cm}^2, \text{SSD})} \quad (1)$$

where $D_{\max}(E, S, \text{SSD})$ is the maximum dose along the central axis of the field of interest for a given beam energy E , field size S and SSD. The factor in the denominator has the same meaning and the $10 \times 10 \text{ cm}^2$ open applicator has been set for the normalization. The evaluation of the accuracy of XiO eMC in the water as well as in the heterogeneous phantom has been performed at a single SSD of 100 cm.

Heterogeneous phantom description

We have used the thorax phantom constructed by Seuntjens et al. [17], which has densities close to realistic human tissues and appropriate dimensions (see Fig. 1). This phantom is composed of three layers of tissue-equivalent materials with thicknesses obtained from averaged measurements on CT-scan of a group of 18 patients performed by the authors. A detailed description (composition and density) of the above phantom can be found elsewhere [15,17]. This phantom has been scanned and CT-scan data have been introduced in DOSXYZnrc [18] using CTCREATE. Distances have been referred from the lung surface for all calculations, i.e. 1 cm depth is equivalent to 3.3 cm for 8, 12 and 18 MeV (because of 2.3 cm of adipose and polyethylene tissue-equivalent materials on top of that surface). Due to the shortest range of lower electron beam energy, calculations have been performed in the thorax phantom without the first 1.1 cm layer of polyethylene material in the case of the 4 MeV electron beam irradiation. Thus, 1 cm from the lung surface was equivalent to 2.2 cm from the top of the phantom surface. The 4 MeV electron beam is obviously not of clinical interest for chest wall irradiation, but this energy was chosen to evaluate the accuracy of XiO eMC for low electron beams irradiation. The accuracy of XiO eMC in the presence of heterogeneities has been investigated from off-axis profile and PDD calculations in the thorax phantom described above and results were compared with the reference BEAMnrc data at one SSD of 100 cm. In the lung tissue, calculations have been carried out at several

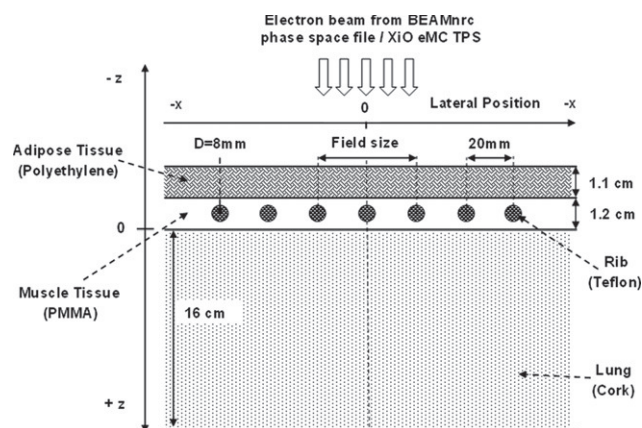


Figure 1 Graphic representation of the thorax phantom. It consists of three layers of different tissue-equivalent materials: adipose tissue (polyethylene), striated muscle (PMMA) in which are embedded a series of seven equidistant solid Teflon (ribs), and lung tissue (cork). The radiation beam is entering on the top surface at 100 cm SSD and the beam axis is perpendicular to the central rib axis. 200 MU have been used for all computations and the statistical uncertainty was less than 1% for each calculation.

depths from the lung surface, spaced by 0.5 cm for 4 MeV, 0.6 cm for 8 and 12 MeV, and 1 cm for 18 MeV beam energies. Off-axis profiles have also been calculated in the non-lung tissue (1.2 cm-thick of PMMA) at 0.6 cm from the lung surface. In the above selected depths, hot and cold spots are more pronounced. So that, the evaluation of the ability of XiO eMC to reproduce dose perturbation due to ribs inhomogeneities, which are narrow, compared to the field size, can be considered as a severe test at these depths. The field sizes have been $6 \times 6 \text{ cm}^2$, $10 \times 10 \text{ cm}^2$ and $14 \times 14 \text{ cm}^2$ open applicators. This has enabled the quantification of both the effect of heterogeneities in dose distributions in the lung and non-lung tissues, and the accuracy of XiO eMC in reproducing dose perturbations in the presence of heterogeneities of high density such as bone structures irradiated with electron beam. The normalization has been performed at the average dose, D_{av} , as follows: For each field size at the depth of interest, let D_{max} be the average maximum dose between two ribs and D_{min} the average minimum dose beneath the ribs. D_{max} and D_{min} are calculated from equations (2) and (3) given below:

$$D_{max} = \left(\frac{1}{m} \right) \cdot \sum_{j=1}^m D_{max}(j) \quad (2)$$

$$D_{min} = \left(\frac{1}{m-1} \right) \cdot \sum_{j=1}^{m-1} D_{min}(j) \quad (3)$$

where “ m ” is the number of maximum peaks dose involved in each field size at the depth of interest. The normalization has been performed at the average dose $D_{av} = (D_{max} + D_{min})/2$.

Monte Carlo simulations

Monte Carlo simulation model of the ELEKTA SL25 linear accelerator

A MC model of an ELEKTA SL25 linear accelerator has been built for 4, 8, 12 and 18 MeV electron beams, using the popular MC code BEAMnrc, an EGSnrc-based radiotherapy simulation package. A detailed description of the simulation process can be found in our early work [15] aimed on the evaluation of VMC++ MC based TPS using experimental measurements and the BEAMnrc MC simulation. Thus, this part will be briefly presented, with the emphasis being on the differences between the two works, namely, the number of history and the electron beam spectra parameters. In this work, new models have been built in order to “match” the measured and calculated depth dose in the water phantom. To this end, the mean energy and full width at half maximum (FWHM) have been slightly adjusted. The above parameters are summarized in Table 1. The code has been run repeatedly by varying these two parameters until the difference between measured and BEAMnrc calculated depth dose parameter R_{50} , representing the beam quality, was less than 0.5 mm. About $(0.6-2) \times 10^9$ particle histories have been simulated in BEAMnrc and DOSXYZnrc, depending on the energy and field size in order to keep a statistical uncertainty under 0.5% for all calculations (depth dose as well as off-axis profile). $(3-400) \times 10^6$ particles (photons and electrons) were stored in the constructed phase space files. For both BEAMnrc and DOSXYZnrc codes, the CPU time per history depends on energy, field size as well as parameters of simulation.

XiO Electron Monte Carlo dose calculation

The XiO is a new commercial TPS developed by ECMSSG. An Electron Monte Carlo (eMC) algorithm has been incorporated into the XiO TPS as a powerful choice for dose calculation with electron beams. The XiO eMC algorithm is based on the original X-ray Voxel MC (XVMC) initially developed by Universitätsklinikum Tübingen [11] to provide raw calculations that involved electron and photon transport for research in the clinical radiotherapy oncology. To be used as eMC dose engine in the XiO TPS, the base code was modified by ECMSSG in order to greatly extend its modeling capabilities. For instance, a flexible and expandable set of primary source modeling routine was developed to provide accurate reproduction of broad

Table 1 Energy spectrum parameters of electron beams used for the BEAMnrc simulation. FWHM = full width at half maximum. Parameters of smaller peaks sometimes introduced in the spectrum are omitted here.

Nominal energy (MeV)	Peak energy (MeV)	FWHM (MeV)
4	4.85	1.45
8	8.59	2.0
12	13.15	2.62
18	18.71	2.90

variety of incident electron beams' characteristics. An electron fluence function was included in order to account for electron scattering in air. The use of XiO eMC starts with the characterization of the treatment unit from conventional measurements. This characterization procedure can be summarized below in three main steps.

Collimation geometry

The user has to provide the X- and Y-jaws setting for every combination of energy and applicator. For patient specific applicator (cutout): the material composition, thickness, and distance from the nominal source position of the accelerator to lower surface of the cutout must be provided. The accuracy of XiO eMC depends also slightly on the serial number for a given type of applicator.

Scanning measurements in the water

Scanning measurements with applicator in place

For each energy and applicator at 100 and 115 cm SSD, depth dose and off-axis profiles at several depths (R_{100} , R_{90} , R_{80} , R_{50} , R_{20} and $R_p + 2$ cm) are required by the manufacturer.

Scanning measurements with no applicator in place

Further measurements (depth dose and lateral dose profiles) with no applicator in position and the jaws opening to 40×40 cm², 15×15 cm², 10×10 cm² and 5×5 cm² are needed for each energy. Depth dose and lateral dose profiles have been used to determine the energy spectrum and off-axis fluence variations of the beam, respectively.

Non-scanning measurements in the water

Before clinical use, the electron output produced by external beam radiotherapy machines must be calibrated. This is one of the important steps constituting the chain representing an accurate dose delivery to the patient. The calibration has been performed at reference depth, Z_{ref} ($Z_{ref} = 0.6R_{50} - 0.1$ cm), according to the NCS Report 18 protocol [19]. As XiO eMC calculates the absolute dose to the medium, a direct representation of the dose in terms of monitor unit (MU) for a given applicator in a specific electron beam requires an absolute dose measurement for a fixed number of MU. Measurements have been performed at 100 and 115 cm SSD for the four open applicators 6×6 cm², 10×10 cm², 14×14 cm², 20×20 cm² and the 5×5 cm² insert cutout within the 14×14 cm² applicator. These data have been used to model the amount of scatter in the beam.

All the above data have been collected by ECMSSG which performs the characterization of the radiation treatment unit. In this process, beams are "tuned" by the manufacturer and the user is provided with a ready-to-use virtual treatment unit.

User control parameters affecting the accuracy of XiO eMC dose calculation

There are three parameters that XiO eMC user can modify to qualitatively improve the final result during the patient dose calculation, namely:

- (1) The number of history that can be manually fixed and changed, with a maximum value of $1.0E + 12$, until the isodose lines are smooth;
- (2) The mean relative statistical uncertainty (MRSU). As specified in the user manual, the MRSU fixes the largest average uncertainty the user is willing to accept for the final dose calculation. XiO eMC will generate as many histories as necessary to achieve the MRSU value set by the user (as long as it is not more than maximum number of history allowed). The manufacturer advises typical values in the range between 1% and 2%, although, smaller values which require a long computation time can be set. In the present study, we fixed the MRSU at 0.5% and checked periodically when isodose lines were smoothed before stopping calculation;
- (3) The voxel size. This parameter can be independently set for each one of the three directions.

Results

Homogeneous water phantom calculations

Central axis depth dose and off-axis profiles

After the fine-tuning of the primary electron spectrum energy, simulations have been performed to verify the BEAMnrc MC models with respect to measured dose distributions in the water. The verification has been done by comparing depth dose and off-axis profiles obtained from BEAMnrc models and measurements in a homogeneous water phantom. The accuracy of XiO eMC calculation in the water phantom has been investigated by similar comparisons. The results for depth dose profiles corresponding to 4, 8, 12 and 18 MeV for open field sizes of 10×10 cm² and 14×14 cm² are displayed in Fig. 2(a, b), respectively. Agreement of R_{50} representing the beam quality (see Table 2), i.e. the difference between calculations and measurements was better than 0.5 mm for both BEAMnrc and XiO eMC. Similar results have been obtained for the open 6×6 cm², 20×20 cm² applicators and other insert cutouts. Off-axis profiles in the inplane and crossplane directions at the depth of maximum dose [see Fig. 3(a, b)] and other depths R_{90} , R_{80} , R_{50} , R_{20} have also been calculated with both codes and compared to measurements. Their agreement with measurements is better than 0.5 mm in the penumbra region and less than 1% in the homogeneous region. However, in the case of 4 MeV, some limitations of XiO eMC values to match lateral dose profiles near the field edge are evident for the open applicators 14×14 cm² (Fig. 3a) in both directions, and 20×20 cm² in the inplane direction, respectively. Table 2 summarizes the above results for a depth dose profile comparison of the R_{50} parameter. We also found good agreement (discrepancies of 0.5 mm or less) for the depths of maximum dose computation (BEAMnrc, XiO eMC) relative to the measurements.

Output factors comparison

As XiO eMC computes absolute dose for a fixed number of MU, we have compared experimental and predicted XiO

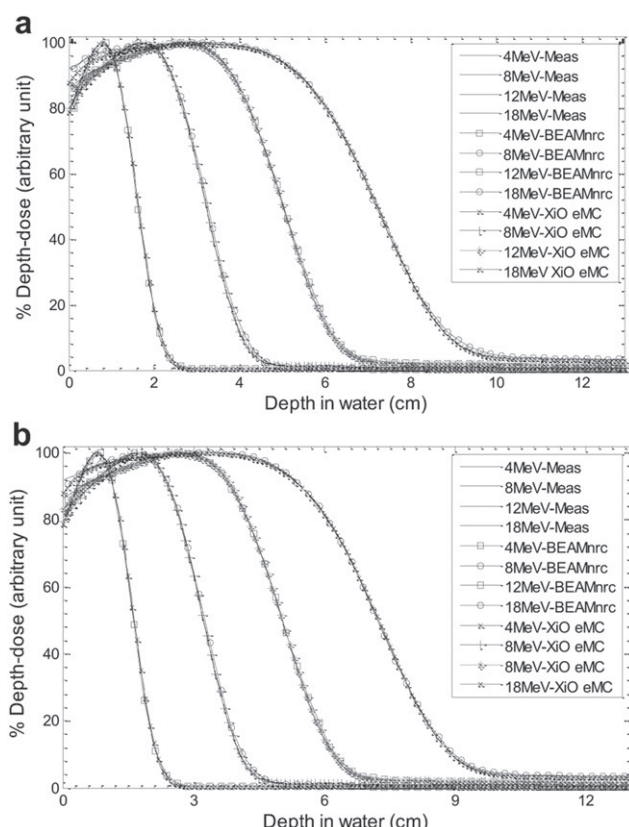


Figure 2 Central axis depth dose curves for $10 \times 10 \text{ cm}^2$ (a) and $14 \times 14 \text{ cm}^2$ (b) field size for 4, 8, 12 and 18 MeV at 100 cm SSD. Measurements (Meas) are compared to XiO TPS (XiO eMC) and BEAMnrc values. The normalization has been performed at the depth of maximum dose along the central axis.

eMC output factors, for open applicators of $6 \times 6 \text{ cm}^2$, $10 \times 10 \text{ cm}^2$, $14 \times 14 \text{ cm}^2$ and $20 \times 20 \text{ cm}^2$ and several highly asymmetric insert cutouts. The accelerator is calibrated to deliver 1 cGy/MU at R_{100} and output factors have been measured according to the NCS Report 18 calibration protocol [19]. The $10 \times 10 \text{ cm}^2$ open applicator has been set as the reference field for normalization. Before performing calculations, the absolute dose predicted from XiO eMC has been firstly compared to the calibrated one in the reference conditions. As presented in Table 3, agreement is under 0.8% with a maximum discrepancy at 4 MeV.

Moreover, output factors calculated with XiO eMC for opened field size of $6 \times 6 \text{ cm}^2$, $14 \times 14 \text{ cm}^2$ and $20 \times 20 \text{ cm}^2$

have been compared with measurements for all energies. As expected, we have found good coincidence between XiO eMC and measurements for all open applicators (differences are below 1%). A comparison between measured cutout factors and those predicted from BEAMnrc simulation as well XiO eMC calculation is shown in Table 4. XiO eMC predicts worse results at lower electron beam energy in most cases.

Heterogeneous thorax phantom calculations

Percentage depth dose comparison

Percentage depth dose has been calculated with XiO eMC and compared to those predicted from BEAMnrc simulation for three open applicators of $6 \times 6 \text{ cm}^2$, $10 \times 10 \text{ cm}^2$ and $14 \times 14 \text{ cm}^2$. The comparison has been made for the four energies of 4, 8, 12 and 18 MeV at 100 cm SSD. Results presented in the Fig. 4(a, b) show a good match between both calculations in the case of 4 MeV (a) and 18 MeV (b). Similar agreement has been observed for 8 and 12 MeV electron beam energies.

Off-axis profile in the non-lung tissue

The evaluation of XiO eMC to accurately reproduce dose distribution in the non-lung tissue has been investigated from off-axis profile computation in the muscle tissue (1.2 cm-thick of PMMA) at 0.6 cm from the lung surface. Calculations were carried out for the four energies involved 4, 8, 12 and 18 MeV at 100 cm SSD. The field sizes were $6 \times 6 \text{ cm}^2$, $10 \times 10 \text{ cm}^2$ and $14 \times 14 \text{ cm}^2$ open applicators. We have found very good coincidence between XiO eMC and BEAMnrc set as reference, for calculation with 8, 12 and 18 MeV (differences are less than 1%). Figure 5(a, b) shows a comparison between both computations for 8 MeV (a) and 12 MeV (b). At 4 MeV XiO eMC results are somewhat less accurate (differences up to 4% were observed).

Off-axis profile in the lung tissue

The ability of XiO eMC to accurately predict dose perturbation in the presence of heterogeneities has been investigated in the lung tissue. To this end, off-axis profiles in the direction perpendicular to the ribs in the cork material (thorax phantom depicted in Fig. 1) simulating lung tissue have been evaluated using XiO eMC and compared with the predicted BEAMnrc models, set as the reference. Computations have been done at several depths from the lung surface located approximately at 0.2 cm beneath the ribs. For each energy and field size, computations have been carried out in depth

Table 2 Comparison of R_{50} between BEAMnrc/XiO eMC values and measurements (Meas) at 100 cm SSD. Diff.1 = (Meas – BEAMnrc) and Diff.2 = (Meas – eMC). Good coincidence was achieved between BEAMnrc computations and measurements as well as between XiO eMC calculations and measurements.

Field size (cm^2)	4 MeV/ R_{50} (cm)			8 MeV/ R_{50} (cm)			12 MeV/ R_{50} (cm)			18 MeV/ R_{50} (cm)		
	Meas	Diff.1	Diff.2	Meas	Diff.1	Diff.2	Meas	Diff.1	Diff.2	Meas	Diff.1	Diff.2
6×6	1.647	0.047	-0.02	3.23	0.00	0.00	4.99	0.00	0.01	7.05	-0.03	0.00
10×10	1.640	0.024	0.00	3.22	0.00	0.00	5.02	0.02	-0.05	7.24	0.04	0.03
14×14	1.643	0.032	0.01	3.24	0.03	0.01	5.02	0.02	0.01	7.22	-0.03	-0.01
20×20	1.640	0.022	0.01	3.23	0.00	0.00	5.05	0.02	0.02	7.23	0.04	0.04

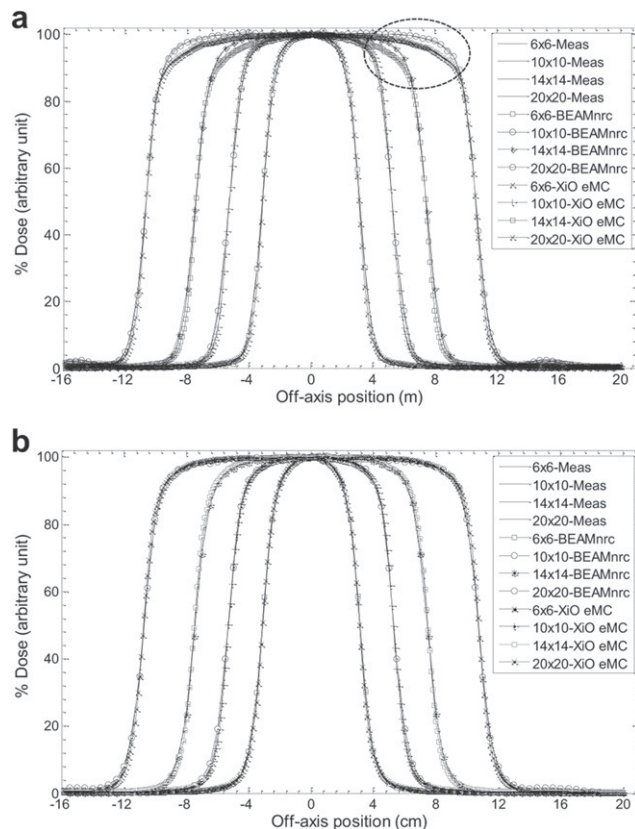


Figure 3 Off-axis profiles (crossplane direction) for four open field sizes of $6 \times 6 \text{ cm}^2$, $10 \times 10 \text{ cm}^2$, $14 \times 14 \text{ cm}^2$ and $20 \times 20 \text{ cm}^2$ at 100 cm SSD. Comparison between measurements (Meas), BEAMnrc simulation and XiO eMC for 4 MeV (a) and 8 MeV (b) electron beam energies. The depth of comparison in the water phantom is the depth of maximum dose along the central axis. Data have been normalized at this depth.

until where the effect of heterogeneity is less pronounced, namely, 2.0 cm (4 MeV), 1.8 cm (8 MeV), 2.4 cm (12 MeV) and 4 cm (18 MeV). Both in the XiO eMC and BEAMnrc/DOSXYZnc, voxel size has been set in the same way to $0.25 \times 0.25 \times 0.25 \text{ cm}^3$. We have found similar agreement for 8, 12 and 18 MeV calculations for all field sizes. Figure 6(a, b) shows a comparison between BEAMnrc and XiO eMC in the case of $10 \times 10 \text{ cm}^2$ open applicator at 8 MeV energy. The latter slightly underestimated the dose between two ribs and overestimated it below the ribs. For the above three energies (8, 12 and 18 MeV), discrepancies between BEAMnrc models and XiO eMC are around 2.5% (2.0–2.6% between two ribs and

2.0–2.5% below the ribs). The depths of comparison from the lung surface were 0.6 cm (a) and 1.2 cm (b). As expected, the agreement is quite good (differences are less than 1.0%) beyond these depths for the three energies. Results are less accurate in the case of 4 MeV, which probably stems from the fact that XiO eMC dose algorithm has been developed for electron beam energies in the range 6–25 MeV. Deviations between BEAMnrc and XiO eMC lie within 2.0–3.5% between two ribs as well as below the ribs. Large deviations have been found at the large field size of $14 \times 14 \text{ cm}^2$. The depth of comparison is 0.5 cm from the lung surface. At 1 cm to this surface, the above deviations decrease slightly within 2.0–2.4%. As expected, the accuracy of XiO eMC can be improved with the number of histories generated. In the present study, the number of histories needed to obtain smooth isodose lines (statistical uncertainty less than 1% in all cases) depends on energy, field and voxel sizes. Typical values in the range of $(3\text{--}20) \times 10^7$ histories were used for all energies and field sizes investigated. No much qualitative improvement to the dose distribution has been found beyond these numbers of histories. We have studied the accuracy of XiO eMC with additional voxel size variation, i.e. $0.55 \times 0.55 \times 0.55 \text{ cm}^3$, $0.45 \times 0.45 \times 0.45 \text{ cm}^3$, $0.35 \times 0.35 \times 0.35 \text{ cm}^3$ and $0.15 \times 0.15 \times 0.15 \text{ cm}^3$ for all previous electron beam energies. The general trend is that the accuracy of XiO eMC increases with the decrease of the voxel size. In the case of 8 MeV (field size $10 \times 10 \text{ cm}^2$) at 0.6 cm depth, deviations up to 4.5% have been found for calculations with the sizes of $0.55 \times 0.55 \times 0.55 \text{ cm}^3$ and $0.45 \times 0.45 \times 0.45 \text{ cm}^3$. Similar behavior has been found for calculations performed with 4, 12 and 18 MeV electron beam energies. Changing the voxel size from $0.25 \times 0.25 \times 0.25 \text{ cm}^3$ to $0.15 \times 0.15 \times 0.15 \text{ cm}^3$ does not significantly affect the accuracy of XiO eMC. Deviations between the two voxel sizes calculation are less than 0.8% for $10 \times 10 \text{ cm}^2$ open applicator at the depths of comparison of 0.5 cm (4 MeV), 0.6 cm (8 and 12 MeV) and 1 cm (18 MeV). As XiO eMC uses several simplifications to speed up calculations, we compared the CPU times between BEAMnrc simulation and XiO eMC in the heterogeneous phantom calculation, for three field sizes of $6 \times 6 \text{ cm}^2$, $10 \times 10 \text{ cm}^2$ and $14 \times 14 \text{ cm}^2$. However, this is a rough estimation since calculations have been performed on personal computers (PCs) with different characteristics. BEAMnrc models have been built under a Mandriva 2010.2 operating system with g77 compiler installed on a cluster composed with a set of five connected computers. Each computer was equipped with a double processor Xeon 3.0 GHz and 2.0 GB RAM, whereas, XiO eMC engine has been installed on a virtual machine (VMware Player 4.0.3) installed on a PC with a processor Intel(R) Core(TM) i5, 3.47 GHz and 4.0 GB RAM. About 80% of CPU times indicated for BEAMnrc simulation have been devoted to the generation of the phase space file. Results summarized in the Table 5 show clearly that XiO eMC is faster than BEAMnrc user code, while maintaining a level of accuracy which is clinically acceptable.

Discussion and conclusion

All MC calculations in the present study are in very good agreement with measurements data performed in the

Table 3 Absolute dose per monitor units (MU) calculated with XiO eMC in the homogeneous water phantom in the reference conditions. Computations have been performed for a standard field size of $10 \times 10 \text{ cm}^2$ at 100 cm SSD.

Nominal energy (MeV)	Dose (cGy/MU)
4	0.992
8	1.001
12	1.000
18	1.006

Table 4 Comparison of calculated and measured cutout factors at 100 cm SSD for electron beams of 4, 8, 12 and 18 MeV. Diff.1 = $[(\text{Meas} - \text{BEAMnrc})/\text{Meas}]$ and Diff.2 = $[(\text{Meas} - \text{eMC})/\text{Meas}]$.

Field size (cm ²)	Cutout (cm ²)	4 MeV			8 MeV			12 MeV			18 MeV		
		Meas	Diff.1	Diff.2	Meas	Diff.1	Diff.2	Meas	Diff.1	Diff.2	Meas	Diff.1	Diff.2
6 × 6	5 × 2.5	0.762	3.24	4.19	0.892	1.90	1.34	0.914	1.28	-1.31	0.941	1.83	-1.53
10 × 10	2.5 × 5	0.929	1.45	0.55	0.922	0.32	2.17	0.908	0.67	-0.55	0.973	-1.78	1.68
14 × 14	5 × 5	1.036	-1.93	-1.86	0.979	-1.63	0.92	0.964	-1.03	-0.73	0.980	1.36	1.33
	3 × 8	1.042	1.06	4.22	0.943	-1.25	0.34	0.939	-0.73	1.06	0.979	1.92	2.35
	5 × 10	1.035	0.97	-0.77	1.050	-1.30	1.25	0.967	0.41	-1.34	0.999	-0.70	-2.43
20 × 20	6 × 16	1.020	1.08	-3.71	0.991	0.20	-1.92	0.956	-0.26	-1.78	0.966	0.38	-0.62
	16 × 10	1.024	1.27	-3.42	1.003	1.29	-0.26	0.961	-0.21	-1.98	0.974	-0.90	0.31

water phantom, providing a proof of the accuracy of our BEAMnrc models. Discrepancies between BEAMnrc and measurements are almost always under 2% in the homogeneous dose regions, and 0.5 mm shifting in the high dose gradient regions. These have been our acceptance criteria

based on the NCS Report 15 [20] recommendations (2%/2 mm for small dose/large dose gradient in the homogeneous phantom), for MC modeling as well as a prerequisite before considering BEAMnrc models as reference tool in the heterogeneous phantom calculations. Accuracy of XiO eMC is benchmarked for a wide range of electron beam energies of 4, 8, 12 and 18 MeV both in the homogeneous water phantom and heterogeneous phantom. We have compared experimental dose distributions with predicted values from XiO eMC and full MC simulation. For all field sizes that have been tested in the water phantom

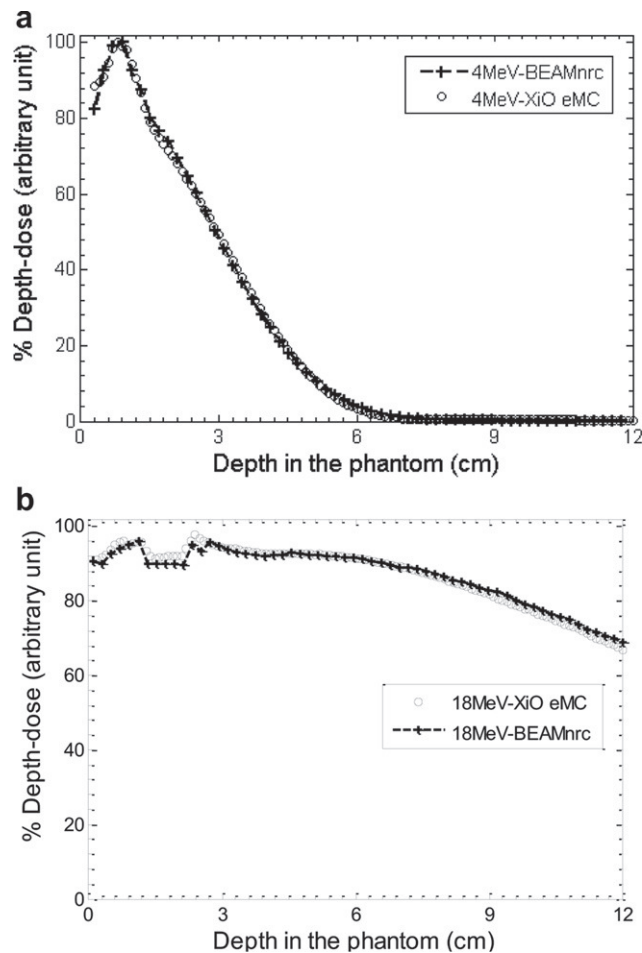


Figure 4 Comparison of central axis depth dose curves in the heterogeneous phantom for 4 MeV (a) and 18 MeV (b) field size $10 \times 10 \text{ cm}^2$. Data have been normalized at the depth of maximum dose along the central axis for 4 MeV. For 18 MeV, data have been normalized relative to the maximum dose along the central axis in the homogeneous water phantom of the corresponding field size.

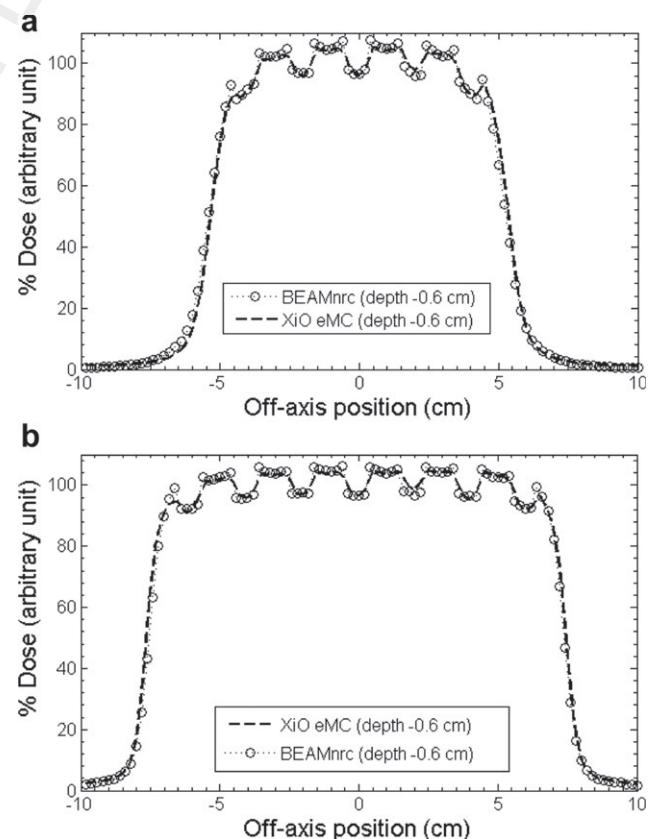


Figure 5 Off-axis profiles in the muscle tissue (1.2 cm-thick of PMMA), in the crossplane direction at 100 cm SSD. Comparison between BEAMnrc simulation and XiO eMC for 8 MeV and $10 \times 10 \text{ cm}^2$ field size (a), 12 MeV and $14 \times 14 \text{ cm}^2$ field size (b).

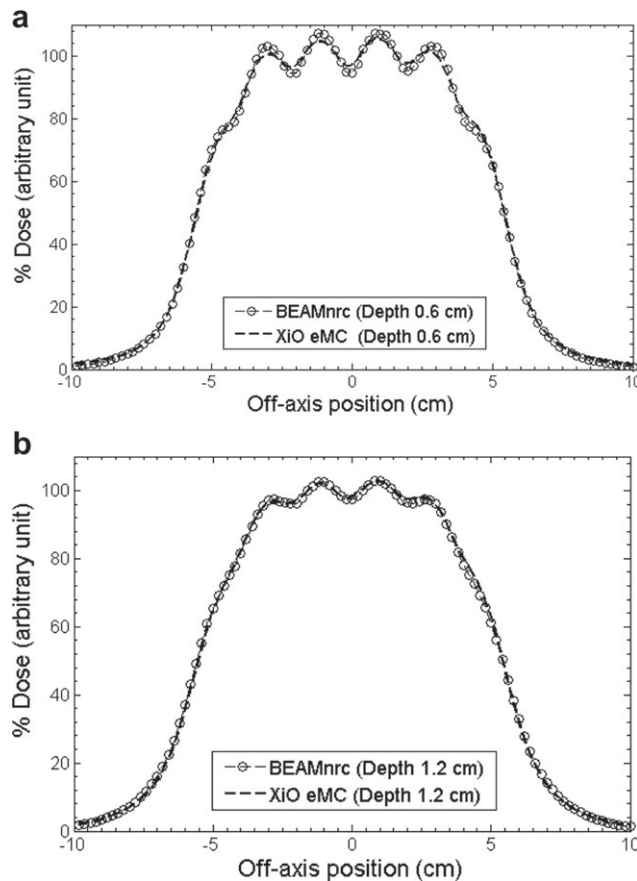


Figure 6 Off-axis profiles in the lung tissue, in the crossplane direction for an open field size of $10 \times 10 \text{ cm}^2$ at 100 cm SSD. XiO eMC and BEAMnrc are compared for 8 MeV electron beam energy. Depths from the lung surface are 0.6 cm (a) and 1.2 cm (b).

calculations, a good agreement is found between BEAMnrc/XiO eMC results and measurements for 8, 12 and 18 MeV. Similar accuracy is obtained in the case of 4 MeV, except the large field sizes of $14 \times 14 \text{ cm}^2$ and $20 \times 20 \text{ cm}^2$ for which off-axis profiles do not match measurements near the edge of the field. Note that BEAMnrc models agree well with measurements in these cases. We have also found that XiO eMC can predict MU with good accuracy for open field size. Cutout factors computed with XiO eMC are quite similar to those predicted from BEAMnrc simulation, except at low energy for which the results are worse in the most cases. In the heterogeneous phantom, we have found good coincidence of dose distribution (PDD and lateral dose

profile) between BEAMnrc simulation and XiO eMC in the non-lung tissue for 8, 12 and 18 MeV (differences less than 1%). PDD obtained from XiO eMC is in good agreement with BEAMnrc simulation at 4 MeV, while deviations up to 4% were observed for lateral dose profile at the same energy. In the lung tissue, discrepancies between BEAMnrc and XiO eMC are around 2.5% for 8, 12 and 18 MeV and 3.5% for 4 MeV. Results from 4 MeV calculations are not surprising because XiO eMC dose algorithm was developed for electron beam energies range 6–25 MeV. Discrepancies between BEAMnrc and XiO eMC for the other energies may be partially explained from some aspects. Firstly, the depth of comparison is closer to the bone structure where the effects of heterogeneities are more pronounced in that region. A $\pm 0.1 \text{ cm}$ positional uncertainty in the depth analysis may lead up to 1% difference in dose. Moving away from the heterogeneity rapidly reduces the disagreement and a good consistency is achieved. Secondly, at this region, scattered electrons from two ribs contribute to the increase in dose (i.e. increase in electron fluence) laterally to both sides of the ribs and the decrease in dose beneath the ribs (i.e. decrease in primary electron fluence). This phenomenon seems not to be well modeled, probably due to simplifying approximations used in the implementation of electrons transport and secondary particles creation in the XiO eMC algorithm. Accuracy of XiO eMC can be improved by decreasing the voxel size. However, reducing it from $0.25 \times 0.25 \times 0.25 \text{ cm}^3$ to $0.15 \times 0.15 \times 0.15 \text{ cm}^3$ does not lead to significant change in accuracy. Therefore, the reduction of the voxel size does not indefinitely improve XiO eMC dose calculation. We have found that the size $0.55 \times 0.55 \times 0.55 \text{ cm}^3$ can be considered as large and leads to wrong results. A typical voxel size in the range $0.25 \times 0.25 \times 0.25 \text{ cm}^3$ – $0.35 \times 0.35 \times 0.35 \text{ cm}^3$ can be considered as acceptable for dose calculation. However, smaller voxel sizes should be selected for the calculations of dose distribution in some critical situations where three-dimensional heterogeneities are always encountered (e.g. in head and neck area).

A similar study has been reported by Edimo et al. [15] on the commissioning of Oncentra MasterPlan TPS (OMTPS) which implements a 'VMC++ Kawrakov' algorithm [7,8] for two electron beam energies, 4 and 12 MeV. The similarities between the two studies are:

- the use of the same heterogeneous phantom and commissioning approach;
- the calculations depths are identical for 4 MeV and comparable for 12 MeV.

Table 5 Comparison of CPU times in hours (h) between BEAMnrc and XiO eMC calculations in the heterogeneous phantom for the entire dose distribution (PDD and lateral dose profile) with a voxel size of $0.25 \times 0.25 \times 0.25 \text{ cm}^3$ and a treatment distance of 100 cm SSD.

Field size (cm^2)	4 MeV/CPU times (h)		8 MeV/CPU times (h)		12 MeV/CPU times (h)		18 MeV/CPU times (h)	
	BEAMnrc	XiO eMC	BEAMnrc	XiO eMC	BEAMnrc	XiO eMC	BEAMnrc	XiO eMC
6×6	158	0.28	386	0.47	593	1.05	721	0.92
10×10	446	0.35	540	0.65	634	1.15	761	1.28
14×14	460	0.48	560	0.67	792	1.35	787	1.58

The ability of both TPSs to predict dose perturbation is compared here for the above two energies. For 4 MeV, at the same depth, both TPSs predict dose perturbation caused by ribs heterogeneities with similar accuracy. However, XiO eMC is more robust in the case of 12 MeV. Deviations range from 2.0 to 4.0% for OMTPS and they are less than 1% for XiO eMC at comparable depths. Moreover, the capability for the user control over the voxel size is a good feature compared to OMTPS in which it is automatically assigned by the system. However, as mentioned in the MC simulation paragraph, the BEAMnrc peak energy and FWHM are different between the two works, which could limit our findings in the above comparison. The classic paper of Van Dyk et al. [21] suggests that an electron dose calculation engine should achieve 2% accuracy along the central axis ray (except around the dose maximum region) and 4%/4 mm accuracy throughout the dose distribution. The goal of ECMSSG was to exceed those criteria and to achieve 2%/2 mm accuracy throughout most of the dose distributions. In view of our results, the above criteria have been achieved in the homogeneous phantom calculations. Deviations observed between BEAMnrc and XiO eMC in the heterogeneous phantom are close to these criteria of acceptability for the most tested dose distribution, demonstrating the reliability of using XiO eMC 4.60 for planning treatment of cancer patients with electron beams. However, significant deviations found in the case of 4 MeV electron beam calculations with XiO TPS demonstrate the need to be careful when using XiO at lower electron beam energies.

Acknowledgments

This work was partly supported by the ICTP through CEPAMOQ (OEA-AC-71 project) at the University of Douala (Cameroon) and the Fellowships Program of the "Agence Universitaire de la Francophonie (AUF)". The authors are greatly indebted to the radiation oncology physicists J. M. Denis and A. Delor of the Radiotherapy Department of Cliniques Universitaires Saint-Luc (Brussels, Belgium) for their help in measurements performed in the water phantom and wish to thank M. Coevoet for his technical support during the entire work. Finally, we are also grateful to the referees for their pertinent comments and helpful suggestions which have been used in this work.

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