Implementation of a brachytherapy Ir-source in an in-house system and comparison of simulation results with EGSnrc, VMC++ and PIN

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Abstract: Today’s brachytherapy planning systems perform computation of energy deposition in patients by assuming homogeneous water medium and multiplicative transmission factors for shielding. Patient heterogeneities, shape and size are not fully taken into account. Aim of this study is the implementation of the microSelectron 192Ir high dose rate brachytherapy source in the Swiss Monte Carlo Plan, an in-house developed MC environment, where a defined geometry can be simulated by using one out of three different transport algorithms: EGSnrc, VMC++ or PIN. Additionally, the impact of different phantom shapes and clinical relevant inhomogeneities on dose distributions are studied. Radial dose functions, dose rate constants and anisotropy functions according to the AAPM TG-43 formalism have been determined. The implemented source has been validated by comparing dose distributions in a 30x30x30 cm³ water phantom derived from MC simulations using the three transport algorithms with literature data. Dose rate constants, radial dose functions and anisotropy functions agree within 3% with literature data. Placing the source toward the surface of a water phantom can result in local underdosage of up to 17% when compared with the dose distribution around the source at the centre of a 30x30x30 cm³ water phantom. Taking into account the presence of an air and cortical bone inclusion positioned at 1 cm from the source can lead to dose deviations in the region behind these inhomogeneities of up to +7% and -4%, respectively, if compared with the dose in a 30x30x30 cm³ water phantom. The same geometry can be used to compare different transport codes within one Monte Carlo system. Apart from the inverse square law, the impact of the size and the geometry of the phantom as well as heterogeneities on dose distributions have to be considered.

1. Introduction
The aim of radiotherapy is the local control of the disease with a minimal damage of the surrounding healthy tissue when applying ionizing radiation. Depending on tumour type, location and staging, different techniques are available. One of those is brachytherapy, which has become an increasingly popular treatment modality for localized tumours. The main advantage of this technique is the high conformal energy deposition in the target volume and the sparing of the organs at risk due to the influence of the inverse square law on the dose distribution around the source.

Due to the steep dose gradients determining the energy deposition near the source, MC has become an accepted dose calculation method in brachytherapy [1][2][3][4]. However, because of the very long computation time of a full MC simulation, many treatment planning systems estimate the dose in a
patient by superposition of the contributions of individual sources derived from pre-calculated dose matrices. These pre-calculations are performed assuming a 30x30x30 cm$^3$ water phantom. Since patient specific shape and inhomogeneities are neglected, this leads to inaccuracies of the calculated dose distribution [2][5][6]. In a study on oesophageal brachytherapy Anagnostopoulos et al (2004) [7] showed that, although the presence of patient inhomogeneities does not alter the delivery of planned dose distribution to the target volume, considering the patient geometry as a homogeneous water medium can lead to an overestimation of the dose to the spinal cord of up to 13% and to an underestimation of the dose to the sternum of up to 15%. For $^{192}$Ir HDR breast brachytherapy Pantelis et al (2005) [8] found that not taking into account patient shape and inhomogeneities causes overestimation of skin dose in the order of 5% in the central breast region and within 10% at all other points of the breast skin.

In this study we present the implementation of the microSelectron high dose rate (HDR) $^{192}$Ir source (new design) in our in-house developed Monte Carlo environment. Furthermore, simulations with the MC codes embedded in this environment were performed and all dosimetric constants and functions proposed by the Task Group 43 dose formalism [9][16] for the microSelectron-HDR $^{192}$Ir source have been derived and compared with literature data [10].

2. Methods and materials

2.1. Radioactive Source
The geometry and the materials of the microSelectron-HDR $^{192}$Ir source (Nucletron B.V., The Nederlands, part No. 105.002) were derived from [10] and are illustrated in Figure 1a). This source consists of a pure iridium core cylinder with rounded edges (length 3.6 mm and diameter 0.65 mm). The activity within the source is assumed to be uniformly distributed. The active core is covered by a stainless steel (AISI 316L) encapsulation with a total length of 4.5 mm and a diameter of 0.9 mm. The distal capsule tip has rounded borders with a curvature radius of 0.4 mm. This source is welded on a flexible woven stainless steel cable with a diameter of 0.7 mm.

![Figure 1. (a) Design of the microSelectron-HDR $^{192}$Ir source “new design” (part No. 105.002) [10]; (b) model of the microselectron-HDR $^{192}$Ir source utilized in our Monte Carlo simulations.](image)

The geometry of the modelled source is shown in Figure 1b). The main difference to the “real” geometry is the implementation of the source core tips. The design of the “real” source shows rounded edges, while the modelled core consists in an iridium cylinder with a density of 22.42 g/cm$^3$. The
length of this cylinder is 3.6 mm and the diameter is 0.65 mm. The encapsulation of the source consists of AISI 316L stainless steel (Mn 2 wt%, Si 1 wt%, Cr 17 wt%, Ni 12 wt% and Fe 68 wt%), with a density of 8.02 g/cm$^3$. For the cable also stainless steel is used, however, the effective density was 4.81 g/cm$^3$ (10).

2.2. Swiss Monte Carlo Plan

Within the Swiss Monte Carlo Plan (SMCP) project, a Monte Carlo environment with the aim of introducing Monte Carlo treatment planning in the clinical routine was developed (11).

SMCP is implemented in C++ using an object-oriented design. One of the main characteristics of this environment is that the objects responsible for geometry, sources, media and scoring are realised via specific methods of abstract classes, in particular they are separated from the transport codes (Figure 2). This represents an improved flexibility of the code allowing an easy update or add-on of single components. The design of SMCP allows the possibility of simulating exactly the same configuration with different transport codes. At the moment three transport codes are embedded: EGSnrc (12), VMC++ (13)-(14) and Photon Interactions (PIN). PIN is an in-house developed MC code. In this transport code the photon transport is very similar to the implementation of EGSnrc. Photoelectric effect, Compton photons, pair production, as well as Bremsstrahlung photons are simulated based on the cross sections taken from the NIST photon cross section database XCOM. In PIN a simplified electron transport was implemented. For example, Compton electrons do not have any angular distribution, hence are emitted in the same direction as the incident photons. Secondary and Compton electrons deposit their energy along their mean free path length.

The description of the operating mode of the interface between configuration and transport algorithms, is demonstrated for EGSnrc. For the HOWFAR and HOWNEAR definition in the EGSnrc transport code, the C-user interface provided by the EGSnrc distribution is used. These implementations call the methods of the according geometry class in the SMCP environment. With the return values of these methods, HOWFAR and HOWNEAR are determined and returned to EGSnrc. For the interfaces to PIN and VMC++ similar procedures have been implemented.

In this work, the SMCP environment has been extended to dose calculations for brachytherapy by implementing new geometries and source definitions.
2.3. Monte Carlo Calculations

For the low-energy photons emitted by $^{192}$Ir, secondary charged particle equilibrium can be assumed [8]. Hence, electrons resulting from photon interactions with a kinetic energy below 0.5 MeV were assumed to deposit their energy locally. Consequently, ECUT was chosen to be 1.011 MeV (0.5 MeV kinetic energy + 0.511 MeV mass) that means that electrons are not transported, but they deposit their energy locally.

In summary, the MC transport parameters for the EGSnrc code system were PCUT = AP = 0.010 MeV, ECUT = AE = 1.011 MeV, ESTEPE = 0.25. Rayleigh scattering was turned on. For the VMC++ code ESTEPE = 0.15, PCUT = 0.010 MeV, ECUT = 0.5 MeV (kinetic energy) and the kerma approximation KCUT = 0.5 MeV (kinetic energy) were used. Rayleigh scattering is not implemented in VMC++. However, comparisons of dose distributions using EGSnrc with and without Rayleigh scattering showed dose differences smaller than 1% for distances up to 5 cm. In PIN a PCUT of 0.01 MeV was used.

In this work only the photon spectrum of the $\gamma$ transitions and K- and L-shell x-ray from the iridium decays was used [15]. Since electrons are almost completely absorbed from the encapsulation material, they were not considered.

2.4. TG-43 dose calculation formalisms

In this work the calculation formalism proposed by the AAPM Task Group 43 [9][16] has been applied. The dose rate in medium at point $(r, \theta)$ is done by:

$$D(r, \theta) = S_k \cdot \Lambda \cdot \frac{G_L(r, \theta)}{G_L(r_0, \theta_0)} \cdot g_L(r) \cdot F(r, \theta),$$

where $r$ is the distance from the centre of the source to the measurement point and $\theta$ the polar angle relative to the longitudinal axis of the source.

$S_k$ is the source air-kerma strength in units of $U$ ($= \mu$Gy m$^2$ h$^{-1}$) and is defined as the product of the air-kerma rate due to photons of energy greater than $\delta$ at a calibration distance along the transverse plane of the source and the square of the distance,

$$S_k = K_\delta(b) \cdot b^2.$$

The dose rate constant $\Lambda$ [cGy h$^{-1}$ U$^{-1}$ = cm$^{-2}$] is the dose rate in water at a reference distance of 1 cm in the transverse plane of the source in unit of source strength,

$$\Lambda = \frac{D(r_0, \theta_0)}{S_k}.$$

The dimensionless radial dose function $g_L(r)$ accounts for photon attenuation and scattering in the medium and in the encapsulation material at any distance on the transverse plane of the source,

$$g_L(r) = \frac{D(r, \theta_o)}{D(r_o, \theta_o)} \cdot \frac{G_L(r, \theta_o)}{G_L(r_o, \theta_o)},$$

where: $r_o = 1$ cm and $\theta_o = \pi/2$.

The dimensionless anisotropy function $F(r, \theta)$ accounts for photon attenuation and scattering at any distance $r$ and any angle $\theta$ relative to the longitudinal axis of the source,

$$F(r, \theta) = \frac{D(r, \theta)}{D(r_o, \theta_o)} \cdot \frac{G_L(r, \theta_o)}{G_L(r_o, \theta)},$$

where: $r_o = 1$ cm and $\theta_o = \pi/2$.

The geometry function for the line source approximation $G_L(r, \theta)$ [m$^{-2}$] describes the variation of the relative dose due only to the geometry of the source and the spatial distribution of the activity in it. Photon attenuation and scattering are thereby ignored. $G_L(r, \theta)$ is given by:
\[
G_L(r, \theta) = \begin{cases} 
\frac{\beta}{L \cdot r \cdot \sin(\theta)}, & \text{if } \theta \neq 0 \\
\left( r^2 - \frac{L^2}{4} \right)^{-1}, & \text{if } \theta = 0
\end{cases}
\]

where \( L \) is the length of the active source and \( \beta \) is the angle subtended by the active source with respect to the point of calculation \((r, \theta)\). In this work the line approximation of the source was applied.

To derive these parameters the 2D dose distribution from the validation simulations were transformed to polar coordinates with a bilinear transformation. An additional error of 0.5% was assumed.

2.5. Validation

For the validation calculations the source was positioned at the centre of a 30 cm diameter liquid-water sphere. Hollow cylinders with the main axes corresponding to the main axis of the source, were used for the dose scoring. This scoring geometry is analogue to the one applied in the DOSRZnrc user code [17] and can partially account for the decrease of the photon fluence (and thus of statistics) at increasing distances from the source, since the scoring volume of the voxels increases proportional with the distance from the source. The water phantom was divided into approximately 10'000 hollow cylinders leading to a 2D dose matrix \((r-\text{and } z\text{-coordinates}).\) For the validation of the source geometry, the deposited energy was determined by using between \(10^8\) (VMC++) and \(10^9\) (EGSnrc and PIN) histories, yielding a statistical uncertainty of the dose distribution of 0.2% near the source and 0.5% for distances greater than 4 cm. With a bilinear interpolation the resulting dose distribution in 2D cartesian coordinates was transformed into a dose distribution in polar coordinates consisting of 44 angles and 17 radii. The additional dosimetric error of this transformation is estimated to be 0.5%. The resulting dose distributions in polar coordinates were used to express the dose distributions in the TG-43 formalism.

The air-kerma strength \( S_k \) needed for the calculation of the dose rate constant \( \Lambda \) was determined in a separate simulation of \(10^8\) histories. The source was positioned at the centre of an air sphere with a diameter of 5 m. The air-kerma was scored in a water hollow cylinder on the transverse plane of the source with an inner radius of 99.5 cm, an outer radius of 100.5 cm and a height of 1 cm. This configuration was chosen in order to simulate a real measurement with a therapy level detector calibrated in water as described in the Recommendations No. 13 of the Swiss Society of Radiobiology and Medical Physics (SSRMP) [18]. The scored energy deposition in water was transformed to air-kerma by multiplication with the ratio of the mass attenuation coefficients of air and water from the NIST database:

\[
\left( \frac{\mu_{\text{en}}}{\rho} \right)_{\text{Air}}^{\text{Water}} = 0.899
\]

Air attenuation and scattering were corrected with the factor \( f_{\text{asc}} = 1.012 \) as proposed by [19][20].

To ensure that the simulation of the air-kerma strength \( S_k \) as described above delivers reliable results a simulation with the EGSnrc transport code was carried out. In this simulation the air-kerma strength was determined by using the energy fluence to estimate the air-kerma as presented in [22]. Values from the two methods agreed within 2%. Furthermore, the comparison with literature data as shown in Table 1 confirmed the reliability of the simulation configuration presented in this work.
2.6. Inhomogeneities and phantom shape

The effects of inhomogeneities were studied by placing the source at the centre of a 30x30x30 cm³ water phantom together with a 1x2x2 cm³ cuboidal inhomogeneity at a distance of 1 cm from the source in its transverse plane. Dimensions and the location are shown in Figure 3. The inhomogeneities are made of ICRP cortical bone and dry air to simulate patient heterogeneities.

Figure 3. Effects on dose distributions due to inhomogeneities were simulated by placing a 1x2x2 cm³ cuboidal inhomogeneity at a distance of 1 cm from the source.

Dose distributions of each inhomogeneity have been simulated using all three MC transport algorithms. For the comparison of the results from the three transport algorithms, the gamma analysis [21] (with distance to agreement and dose criteria set to 3 mm and 3%, respectively) has been carried out. In this analysis the dose distributions are divided by the geometry function and only voxels with dose values greater than 3% of the maximum dose are included in the gamma evaluation.

For the evaluation of the effects of the phantom dimensions on the dose distribution, the source was located at 1 cm from the surface of the phantom and the deposited energy was scored at distances of 1, 2, 4 and 8 cm from the midplane through the source (Figure 4). The dose distributions were then compared with the calculated dose distribution when the source is located at the centre of a 30x30x30 cm³. By placing the source close to the surface a lack of scattering photons resulting in a significant drop in dose is expects. With this arrangement conditions are reproduced, where the source is located close to the skin of the patient. A typical example is the breast HDR brachytherapy.

Figure 4. The effect of different phantom shapes on the dose distribution was simulated by placing the source at 1 cm from the water phantom surface. The dose was scored at 1, 2, 4 and 8 cm from the centre of the source.
Since the rotation symmetry around the source axis is broken by both inclusion of inhomogeneities and variation of the distances of the source from the phantom surface, the energy deposition was scored within cuboidal 2x2x2 mm$^3$ scoring voxels like in the DOSXYZnrc user code ([23]).

Simulations evaluating the effects of inhomogeneities and phantom shape on the dose distribution were performed by tracing 10$^8$ histories.

3. Results

3.1. Dose rate constant

The dose rate constants $\Lambda$ derived from simulations with the three transport algorithms are shown in Table 1. The three values agree within 0.4% and are close to the values proposed by other authors for the same source [10][24][25]. From the simulations using the three MC transport codes, an average dose rate constant of 1.119 ± 0.002 cGy h$^{-1}$ U$^{-1}$ is proposed.

Dose rate constants from this work also agree within their uncertainties with values from simulations and measurements on the classic design of the microSelectron-HDR $^{192}$Ir source proposed by Papagiannis et al (2002) [24] and Russel et al (2005) [25].

Table 1. Dose rate constants according to the TG-43 formalism derived from simulations with the three transport algorithms EGSnrc, PIN and VMC++.

| Source | $\Lambda$ [cGy h$^{-1}$ U$^{-1}$] |
|--------|---------------------------------|
| this work, (EGSnrc) | 1.119 ±1.6% (2σ) |
| this work, (VMC++) | 1.118 ±1.4% (2σ) |
| this work, (PIN) | 1.121 ±1.7% (2σ) |
| this work, (EGSnrc)$^1$ | 1.100 ±1.6% (2σ) |
| Daskalov et al (1998) [10] | 1.108 ±0.13% |
| Papagiannis et al (2002) [24] | 1.109 ±0.45% |
| Russel et al (2005) [25] | 1.118 ±0.8% |

$^1$: Air-kerma determined by using the energy fluence to estimate the air-kerma as presented in [22].

3.2. Anisotropy functions

Anisotropy functions at different distances from the source are shown in Figure 5. At angles between 15° and 165° the simulations using the three algorithms and literature data [10] agree within 2%. At angles <15° and >165° deviations up to 7% are seen. However, the uncertainty of the dose values in these regions reaches also 7%. In fact, in these two regions very small voxel sizes were chosen to have a high resolution of the anisotropy which in turn increases the statistical uncertainty.

Because the modelled source core is not a line, but a cylinder, the line source approximation of the geometry function, $G_L(r, \theta)$, cannot correctly reproduce pure geometrical variations of the relative dose distribution close to the source. For this reason, comparisons of the anisotropy functions calculated using the line approximation with literature data could not be carried out for distances smaller than 0.5 cm. However, an extrapolation of the anisotropy function at 0.25 cm from values between 0.5 cm and 1.0 cm reproduces the data of Daskalov et al (1998) [10] at this distance. The same observations were given by Karaiskos et al (2000) [26]. The authors pointed out that errors greater than 3% are observed at radial distances very close to the source ($r \leq L/2$), at polar angles far away from their transverse bisector.
3.3. Radial dose function

The radial dose functions of the microSelectron-HDR $^{192}$Ir source obtained from simulations with the three transport algorithms are shown in Figure 6. The radial dose function from simulations using EGSnrc agrees within the statistical uncertainty of 0.8% with literature data ([10]) at any distances. Simulations using PIN show very good agreement to literature data (<1%), except at distances between 4 and 5 cm where differences of up to 1.6% are present. For distances larger than 3 cm VMC++ seems to systematically underestimate the deposited dose by up to 2%.

Further investigations need to be carried out in order to understand the cause of these deviations.
Figure 6. Dose rate function from simulations using EGSnrc, VMC++, and PIN as MC transport code compared with literature data [10].

Figure 7. Monte Carlo calculated isodose lines for the microSelectron-HDR $^{192}$Ir source positioned at the centre of a 30x30x30 cm$^3$ water phantom in the presence of two different inhomogeneities at 1 cm distance: a) air and b) cortical bone. Isodose values are in an arbitrary scale: 10’000, 1’000, 300, 100 and 40. Dashed lines represent isodoses in the case of energy deposition in a 30x30x30 cm$^3$ water phantom without inhomogeneities. The according relative dose difference profiles along the red dashed line for air and cortical bone inhomogeneities are shown in c) and d) respectively.
Table 2. Gamma analysis of the dose distributions from the Monte Carlo simulations normalised with the geometry function, $G_L(r, \theta)$. Simulations with EGSnrc were used as reference data. Reported values represent voxels or volume (in %), for which the gamma values are less or equal than 1.0 (3 mm and 3% criteria). Only values greater than 3% of the maximum value of the reference data were included in the evaluation.

|            | air      | cortical bone |
|------------|----------|---------------|
| VMC++      | 99.9     | 99.9          |
| PIN        | 98.0     | 98.0          |

3.4. Inhomogeneities

In Figure 7a) isodose contours in the case of an air inclusion in a water phantom are presented. The decreased attenuation in air results in an increase of the dose behind the inclusion of about 5% to 7% (Figure 7c). The increased attenuations in bone (Figure 7b) cause a dose decrease behind the inhomogeneities of up to 4% (Figure 7d).

Gamma analysis was performed for the resulting dose distributions from EGSnrc, VMC++ and PIN, divided with $G_L(r, \theta)$ to remove the strong influence of the geometry function. Results are shown in Table 2. Simulations using EGSnrc were used as reference data. For all the simulations regarding the different inhomogeneities, comparison of the dose distributions using VMC++ and EGSnrc show a gamma-index smaller than 1 in more than 99% of the voxels. With the same criteria, distributions using PIN and EGSnrc agree in at least 98% of the voxels.

Figure 8. Ratio of dose from a source positioned at 1 cm from the surface of a water phantom (see Figure 4) and dose from the same source at the centre of a 30x30x30 cm$^3$ water phantom.
3.5. Varying phantom shapes
The effect of missing backscattering material around the source on dose distribution has been simulated by reducing the distance of the source from the phantom surface. Results can be seen in Figure 8. Data points represent the ratio of the energy deposition in the case where the source is located at 1 cm from the surface of a water phantom and the case where the source is located at the centre of a 30x30x30 cm³ water phantom. This ratio was built for points at 1, 2, 4 and 8 cm from the source (see Figure 4).

The smaller the distance to the surface, the lower the dose at the sampling points. This effect becomes more pronounced at larger distances from the source. Since current planning systems do not account for this effect, they overestimate dose values around the source when it is placed close to the skin, low density tissues or air inclusions in the patient. This overestimation of the energy deposition of the planning systems is more pronounced for points toward the surface and at increased distances from the source, since at small distances the missing scatter component is overlapped by the primary dose component. Points at a distance of 8 cm from the source show an underdose of about 17%. For the same position of the source, the underdose at a distance of 2 cm is on the order of 2 to 3%.

4. Discussion and conclusions
In this study the “new design” of the microSelectron-HDR ¹⁹²Ir source was implemented into the SMCP environment and simulations with the three MC transport algorithms EGSnrc, PIN and VMC++ were performed. Since no confirmatory experimental measurements have been performed, results were validated with published data. Radial dose functions and anisotropy functions calculated using the three transport algorithms show very good agreement with literature data for radii greater than 0.25 mm. Dose rate constants agree within their uncertainties with literature data.

The radial dose functions agree with the available published data within 2.1% for distances up to 9 cm. Data calculated with VMC++ show systematic underestimation of about 2% for distances greater than about 3 cm. In this regard, further investigations will be carried out. The implementation of Rayleigh scattering may reduce this difference, however comparison of the dose distributions using EGSnrc with Rayleigh scattering turned on and off showed dose differences smaller than 1% for distances of up to 5 cm.

Anisotropy functions calculated with the three transport algorithms agree for polar angles 15°<θ<165° within 2% with literature data. At angles 15°<θ and θ>165° deviations are in the order of 5-10%. However, the uncertainty of these values is also in the same order of magnitude.

Dose rate constants from the three MC transport algorithms agree within 0.4%. An average dose rate constant of 1.119 ± 0.002 cGy h⁻¹U⁻¹ is proposed.

The influence of inhomogeneities and phantom shape on dose distribution has also been investigated. The same geometrical configurations were simulated with the three MC transport algorithms and the resulting dose distributions were compared. Gamma analysis shows that for all considered inhomogeneities the resulting dose distributions simulated with the three MC algorithms agree in more than 98% of the volume.

Decreasing the distance between the source and the surface of the water phantom causes a decrease of the scored dose near to the phantom surface. This effect gets more significant at higher distances from the source where the scatter dose component gets more important.

In this work we show that in our SMCP environment the same geometry can be easily used to validate transport codes by comparing their results with results from the already benchmarked EGSnrc code.

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