Monte Carlo calculation of dose to water of a $^{106}$Ru COB-type ophthalmic plaque

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Abstract. The concave eye applicators with $^{106}$Ru/$^{106}$Rh or $^{90}$Sr/$^{90}$Y beta-ray sources are world-wide used in brachytherapy for treating intraocular tumors. It raises the need to know the exact dose delivered by beta radiation to tumors but measurement of the dose to water (or tissue) is very difficult due to short range of electrons. The Monte Carlo technique provides a powerful tool for calculation of the dose and dose distributions which helps to predict and determine the doses from different shapes of various types of eye applicators more accurately. The Monte Carlo code MCNPX has been used to calculate dose distributions from a COB-type $^{106}$Ru/$^{106}$Rh ophthalmic applicator manufactured by Eckert & Ziegler BEBIG GmbH. This type of a concave eye applicator has a cut-out whose purpose is to protect the eye nerve which makes the dose distribution more complicated. Several calculations have been performed including depth dose along the applicator central axis and various dose distributions. The depth dose along the applicator central axis and the dose distribution on a spherical surface 1 mm above the plaque inner surface have been compared with measurement data provided by the manufacturer. For distances from 0.5 to 4 mm above the surface, the agreement was within 2.5 % and from 5 mm the difference increased from 6 % up to 25 % at 10 mm whereas the uncertainty on manufacturer data is 20 % ($2\sigma$). It is assumed that the difference is caused by nonuniformly distributed radioactivity over the applicator radioactive layer.

1. Introduction
For several decades, concave ophthalmic applicators, or plaques, are used in brachytherapy for the treatment of malignant melanoma, uvea, and other intra-ocular tumors [1]. The most suitable radionuclides are beta sources ruthenium-106/rhodium-106 ($^{106}$Ru/$^{106}$Rh) and strontium-90/ytrrium-90 ($^{90}$Sr/$^{90}$Y) used for treating surface tumors up to 5 mm depth due to the short range of beta particles [2,3], and low-energy gamma nuclides iodine-125 ($^{125}$I) and palladium-103 ($^{103}$Pd), used for treating deeper tumors up to 10 mm [4]. The brachytherapy techniques have been utilized in radiotherapy departments in Czech hospitals for many years, but so far there is no institution capable of calibration and verification of brachytherapy sources. Czech Metrology Institute is now preparing to establish a service of measurement of dose to water from brachytherapy sources, including concave eye applicators. For the task, a mathematical model of a COB-type $^{106}$Ru/$^{106}$Rh eye applicator manufactured by Eckert & Ziegler BEBIG GmbH has been created. The applicators of this type have a cut-out which allows irradiating tumors in immediate vicinity of the eye nerve [3]. The paper presents calculations of depth dose and dose distributions of dose to water for the above mentioned applicator using the Monte Carlo general-purpose computational code MCNPX v2.6c [5].
2. Materials and methods

2.1. Radionuclide source
The COB-type eye applicator contains radionuclide $^{106}$Ru which disintegrates via $^{106}$Rh to the stable nuclide $^{106}$Pd. $^{106}$Ru is a pure beta source with the half-life of 368.2 days. It has one beta transition with a maximum energy of 39.4 keV and mean energy of 10 keV. Its daughter $^{106}$Rh is a beta source with the half-life of 29.9 s but not pure, per one beta disintegration 0.34 photons are emitted. The mean and maximum energy of $^{106}$Rh beta radiation is 1.412 MeV and 3.54 MeV, respectively. The mean energy of gamma radiation is 0.5988 MeV, where the greatest yields belong to 0.512, 0.622, and 1.05 MeV transitions [6]. A 0.1 mm thin silver foil between the inner applicator surface (referred-to as “the surface” in the following text) and the radioactive layer absorbs electrons with the energy lower than 240 keV (derived from the MCNPX electron range tables) so only electrons from $^{106}$Rh contribute to the dose to a material outside the applicator. The nominal applicator activity 20 MBq gives the dose rate to water of 120 mGy/min on the surface [3]. The minimum dose delivered to any part of the treated region should not be lower than 100 Gy [7] and the recommended working life of the applicator is given by $^{106}$Ru half-life and it is 1.5 years [8,6].

2.2. Design of the applicator
Figure 1 shows the design of a COB-type eye applicator. It is a concave silver leaf with a radius of curvature of 12 mm [3] bounded by a cone with a vertex angle of 111°. The cut-out into the applicator body protects the eye nerve against the irradiation. Its depth in z-direction is ~4.8 mm and its vertex and edges are rounded with a radius of 3.5 mm [9]. The diameter of the applicator is 19.8 mm and height 5.2 mm [3]. The applicator thickness is 1 mm including 0.1 mm of the silver radiation window [3] and 0.2 mm thick silver foil with electrically deposited radioactive material [4,10]. The diameter of the radioactive layer is 17.1 mm [8] and hence the distance between the edge of the radioactive layer and the edge of the applicator should be ~1.2 mm. It was supposed that this distance was preserved in the cut-out as well.

![Figure 1: Schematic diagrams of a COB-type ruthenium-106 eye applicator. The origin of a coordinate system is placed at the intersection of the applicator central axis (equivalent to the x-axis) and the plaque inner surface.](image)

2.3. Simulation
Calculations were performed using the MCNPX 2.6c code. The Integrated Tiger Series style energy-indexing algorithm was used for the determination of energy losses instead of the default MCNP-style energy-indexing algorithm. The cut-off energy was set to 80 keV for electrons (corresponds with a range of ~0.01 cm according to [5]) and 10 keV for photons. The $^{106}$Ru/$^{106}$Rh activity was distributed uniformly over the radioactive layer. The applicator was surrounded with water (except front side). The depth dose in water along the applicator central axis (equivalent to x-axis shown in figure 1) was determined by scoring of energy absorbed inside the spheres placed successively one after another on
the applicator central axis. The radius of every sphere was 0.049 cm. This simulation was done separately for both beta (energy distribution taken from [4]) and gamma (energy distribution taken from [6]) of \( ^{106}\text{Rh} \) in order to estimate the contribution of gamma radiation to the total dose. Dose distributions for emitted beta only were calculated using the energy deposition mesh tally. To get the statistical uncertainties of the order of 1 % (2\( \sigma \)) in 0-10 mm above the surface on the central axis, \( 4.5 \times 10^7 \) primary particles were generated. Russian roulette/particle splitting variance reduction technique was used.

2.4. Experimental data
No experiments have been made so the calculated data were compared with the manufacturer information. The dose of beta radiation was measured by the manufacturer with a plastic scintillation detector in a water phantom [8,9]. The height of the detector was 0.5 mm and the diameter of a base 1 mm. This detector was calibrated at the National Institute of Standards and Technology, USA, and the dose rate values were determined with the 20 % uncertainty (2\( \sigma \)). The certificate includes protocol of measurements of the depth dose along the central axis from 0.6 to 10 mm above the surface and the dose distribution measured in 33 points on a spherical surface 1 mm above the surface. Coordinates of the measurement points were calculated from a drawing in the certificate [8] and using the interpolation method a map of the dose distribution was created in the Matlab® software.

3. Results and discussion
3.1. Depth dose
Figure 2 displays a simulated relative depth dose along the central axis compared to the manufacturer data. Figure 3 shows the difference between these values calculated as \( (D_{M}(1,x) - D_{E}(1,x))/D_{E}(1,x) \) where \( D_{M}(1,x) \) and \( D_{E}(1,x) \) are the relative doses at the distance of \( x \) mm normalized to the dose at the distance of 1 mm above the surface calculated by MCNPX and measured by the manufacturer, respectively. The differences were calculated for two versions - first beta radiation considered only and second both beta and gamma radiation considered. Thus it is demonstrated that the contribution of gamma radiation to the total dose is negligible at distances less than 10 mm above the surface. The influence of gamma radiation is deeply discussed below. Measured and calculated depth doses are within \( \pm 2.5 \% \) up to 4 mm above the surface. At greater distances the differences increase; at 10 mm from the surface reach \(~25 \% \). Converted to distances, calculated relative doses are displaced against the measured ones by 0.1; -0.04; -0.04; -0.26; -0.32; and -0.51 mm at distances of 0.5; 2; 4; 6; 8, and 10 mm above the surface, respectively.

In papers [1] and [11] the difference between simulations and measurements were observed as well. In [11] it was supposed to be due to the inaccurate source definition in simulations. In this study the difference could be caused by the source nonuniformity shown in detail below in figure 6. In the radioactive layer of the studied applicator, higher radioactivity (up to +20 %) is cumulated closer to the edge of the layer in comparison with the activity on the central axis. Due to this nonuniformity, more electrons contribute to the dose farther from the surface in comparison to the model, i.e. in relative units the measured depth dose curve is slightly above the calculated one. The second explanation could be that the area of radioactive layer is larger than stated in the certificate [8] though this issue is questionable. The additional calculations have shown that the extension of the radioactive layer closer to the applicator edge significantly improves the agreement of the two depth dose curves in farther distances. Nevertheless the real diameter of the radioactive layer would have to be found experimentally to support this hypothesis.

As all values presented in this paper were normalized, it is possible to convert calculated relative doses to absolute ones using the applicator activity of 24.0 MBq stated in the certificate [8]. Although the manufacturer does not state the activity uncertainty, it could be assumed to be \(~4 \% \) (2\( \sigma \)). And hence the simulation gave the dose rate of 108.6 mGy/min (4 %, 2\( \sigma \)) compared to 118 mGy/min (20 %, 2\( \sigma \)) measured by the manufacturer on the central axis 1 mm above the surface.
Figure 2: Measured (triangles) and calculated (circles) relative depth dose in water along the central axis. Uncertainties on the manufacturer data (20 %, 2 σ) are shown. Uncertainties on the simulated data (1 %, 2 σ) are smaller than the circle symbol.

Figure 3: Differences in depth dose in water along the central axis between simulation and manufacturer data. Triangles: gamma radiation is taken into account, circles: only beta radiation is considered.

3.2. Lateral dose profiles
The cut-out in COB-type applicators causes that the dose distribution is not symmetric around the central axis. Figure 4 shows two dose profiles along arcs of two concentric circles. Both arcs lie 1 mm above the surface inside planes XY and XZ. Short range of beta radiation causes steep dose gradients on the edge of the radioactive layer. The cut-out therefore significantly protects sensitive parts of the eye, e.g. eye nerve.

Figure 4: Calculated (solid line) and interleaved measured (dots) dose profiles along arcs of two concentric circles. Both arcs lie 1 mm above the surface inside planes XY (blue) and XZ (red; the arc passes the cut-out). The dose on the central axis has been normalized to 100 %. Points A, A’, B, and B’ represent in sequence: edge of the applicator, edge of the applicator in cut-out, edge of the radioactive layer, and edge of the radioactive layer in the cut-out (see figure 5).

Figure 5: Schematic diagram of the applicator and an arc of the 11 mm radius circle. Dose profiles along the arc are shown in Figure 4. The arc also lies on a spherical surface on which the dose distribution shown in Figure 6 was calculated.
3.3. Dose distributions
Several dose distributions were calculated: a distribution on a spherical surface 1 mm above the surface and three planar distributions on planes XY (horizontal plane), XZ (vertical plane passing the cut-out), and a plane just in front of the applicator parallel to the plane YZ. All values were normalized to the value on the central axis at the distance of 1 mm above the surface.

Projections of calculated and measured dose distributions on a spherical surface (11 mm radius, see figure 5) onto a plane parallel to the plane YZ are shown in figure 6. There is an apparent difference between these two distributions. Calculated distribution, supposing uniform distribution of the radioactivity, is symmetric with maximum doses slightly higher in the innermost region of the applicator in comparison with the dose on the central axis. Above the edge of the radioactive layer the doses are 50 % of the dose on the central axis. By contrast, doses measured by the manufacturer indicate that the radioactivity is distributed highly nonuniformly. Below the central axis the doses are even slightly lower than on the axis and, on the contrary, closer to the edge the doses are up to 20 % higher (see also figure 4). And more, above the edge of the radioactive layer the doses are as much as 80 % of the dose on the central axis compared to 50 % predicted by the simulation. This difference is most likely caused by the higher radioactivity over areas close to the edge of the radioactive layer in real applicator, as already stated. The nonuniformity of the radioactivity over the radioactive layer of concave $^{106}$Ru/$^{106}$Rh applicators was also observed in [12].

Figure 6: Projection of calculated (left) and measured (right) relative dose distributions on a spherical surface 1 mm above the surface onto a plane parallel to the plane YZ (see figure 5). The dose distribution in 33 points (black circles) measured by the manufacturer was interpolated in order to compare it with the simulation. The dose on the central axis has been normalized to 100 %. Grey circles mark the distance from the central axis in cm.

Figure 7 shows in sequence the dose distributions on the plane XY, XZ, and on a plane just in front of the applicator parallel to the plane YZ (4.6 mm far from the plane YZ). Shapes of the distributions are very similar to those presented in [10] which were measured using radiochromic films around a CIB-type applicator (this type of $^{106}$Ru/$^{106}$Rh plaque has a slightly larger cut-out than the COB-type).

3.4. Contribution of gamma radiation to the total dose
Relative contributions of gamma radiation of the daughter $^{106}$Rh and relative total dose (in parenthesis) both related to the total dose to water on the central axis in depths 1 mm, 10 mm, 15 mm, and 19 mm are as follows: 0.7 % (100 %), 4 % (2.5 %), 58 % (0.09 %), and 86 % (0.04 %). The maximum range of $^{106}$Ru/$^{106}$Rh electrons in water is roughly 16 mm thus the rest of the dose in greater distances is due to $^{106}$Rh photons and bremsstrahlung. In depths of tumor treatment (0-5 mm) it is not necessary to consider the influence of gamma radiation, but for radiation protection of healthy tissue, the dose from
bremssstrahlung and gamma radiation should be taken into account in greater distances from the applicator.

Figure 7: Dose distributions on the planes XY (left), XZ (center), and on a plane parallel to the plane YZ (4.6 mm far from the plane YZ; right). All dose normalizations have been done against the dose on the central axis at the distance of 1 mm above the surface. Grey circles mark the distance from the center of curvature (left and center) or the distance from the central axis (right) in cm.

4. Conclusion
The Monte Carlo general purpose code MCNPX has been used for the calculation of depth dose curves and dose distributions of $^{106}$Ru/$^{106}$Rh COB-type concave ophthalmic applicator manufactured by Eckert & Ziegler BEBIG GmbH. The difference is observed in depth dose between simulation results and data provided by the manufacturer for distances greater than 5 mm above the applicator inner surface. In depths from 0 to 4 mm the agreement was good – within 2.5 % in the dose rate (or 0.10 mm at the distance), but above 5 mm the difference increased from 6 % (0.16 mm) up to 25 % (0.51 mm) in 10 mm depth whereas the uncertainty on manufacturer data was ±20 % (2σ). It is assumed that the main reason of the difference is the nonuniform distribution of the radionuclide over the radioactive layer. This hypothesis has been supported by the comparison of calculated dose distribution on the spherical surface 1 mm above the applicator inner surface with the distribution measured by the manufacturer. The difference could be also caused by the different size of the radioactive layer than stated in the certificate although it is less likely. These not quite satisfactory results stressed the need to develop a suitable technique for measurement of dose distributions around the brachytherapy sources. Czech Metrology Institute will now focus on creation of the methodology for determination of 3D dose distributions of ophthalmic and other brachytherapy sources using radiocromic gel dosimeters and optical cone beam computed tomography.

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