Multidimensional dosimetry of $^{106}$Ru eye plaques using EBT3 films and its impact on treatment planning

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Purpose: The purpose of this study was to establish a method to perform multidimensional radiochromic film measurements of $^{106}$Ru plaques and to benchmark the resulting dose distributions against Monte Carlo simulations (MC), microdiamond, and diode measurements.

Methods: Absolute dose rates and relative dose distributions in multiple planes were determined for three different plaque models (CCB, CCA, and COB), and three different plaques per model, using EBT3 films in an in-house developed polystyrene phantom and the \textsc{mcr6} MC code. Dose difference maps were generated to analyze interplaque variations for a specific type, and for comparing measurements against MC simulations. Furthermore, dose distributions were validated against values specified by the manufacturer (BEBIG) and microdiamond and diode measurements in a water scanning phantom. Radial profiles were assessed and used to estimate dosimetric margins for a given combination of representative tumor geometry and plaque size.

Results: Absolute dose rates at a reference depth of 2 mm on the central axis of the plaque show an agreement better than 5% (10%) when comparing film measurements (\textsc{mcr6}) to the manufacturer’s data. The reproducibility of depth-dose profile measurements was <7% (2 SD) for all investigated detectors and plaque types. Dose difference maps revealed minor interplaque deviations for a specific plaque type due to inhomogeneities of the active layer. The evaluation of dosimetric margins showed that for a majority of the investigated cases, the tumor was not completely covered by the 100% isodose prescribed to the tumor apex if the difference between geometrical plaque size and tumor base $\leq$ 4 mm.

Conclusions: EBT3 film dosimetry in an in-house developed phantom was successfully used to characterize the dosimetric properties of different $^{106}$Ru plaque models. The film measurements were validated against MC calculations and other experimental methods and showed a good agreement with data from BEBIG well within published tolerances. The dosimetric information as well as interplaque comparison can be used for comprehensive quality assurance and for considerations in the treatment planning of ophthalmic brachytherapy. © 2015 Author(s). All article content, except where otherwise noted, is licensed under a Creative Commons Attribution 3.0 Unported License. [http://dx.doi.org/10.1118/1.4929564]

Key words: brachytherapy, eye plaque, radiochromic film, dosimetry, quality assurance

1. INTRODUCTION

In the treatment of uveal melanoma different external beam therapy techniques, i.e., stereotactic radiotherapy using protons,$^{1-3}$ particle beam therapy with protons$^{4,5}$ and helium ion$^{6,7}$ beams have been employed as well as brachytherapy utilizing different radionuclides, e.g., $^{125}$I, $^{90}$Sr, and $^{106}$Ru. Whereas $^{125}$I eye plaques are most common in the United States, $^{106}$Ru
plaques are preferred across Europe. Due to lower penetration depths of their beta particles, $^{106}$Ru plaques are likely to cause fewer severe side effects in small-apex tumors adjacent to the optical nerve and macula. The sharper penumbra, however, requires an accurate positioning of the plaque by the surgeon to achieve a high local tumor control.

The steep dose gradients and the generally confined geometry of the different $^{106}$Ru plaque models imply dosimetric challenges. Hence, sufficiently small sensitive volumes are a prerequisite for accurate dosimetry of ophthalmologic brachytherapy plaques, as demonstrated in several studies. A comprehensive comparison was presented by Soares et al., who compared most commonly used detectors in beta-ray dosimetry for one concave plaque model (CCB). They were able to reduce the uncertainty down to $\pm 15\%$ as compared to the 95% confidence interval of $\pm 25\%$ specified by the manufacturer. Other studies investigated the use of radiochromic films, $^{10,12-17}$ TLDs, $^{15,16,18-23}$ extrapolation ionization chambers, $^{24}$ and diodes $^{25}$ which are considered to be a standard for $^{125}$I plaque dosimetry. Other dosimetric techniques which are less widely used, such as plastic scintillators, $^{11,26,27}$ three-dimensional liquid scintillators, $^{28}$ and polymer gel dosimetry $^{29}$ were explored as well. The manufacturer of BEBIG eye plaques also uses plastic scintillators to provide the reference data for the customer.

The purpose of this study is to present a method to determine two dimensional (2D) relative dose distributions and also absolute dose rates at reference depths for three different models of $^{106}$Ru eye plaques (CCA, CCB, and COB) in a specially designed phantom with radiochromic films. In this context, 2D measurements were compared with Monte Carlo (MC) simulations and 1D relative depth profiles and off-axis profiles were benchmarked against a diamond detector and a Si-diode in a standard water scanning phantom, to validate the results from radiochromic film measurements and MC calculations. Inhomogeneities in the active layer among different plaques of the same type and their impact on treatment planning were investigated as well. Finally, results of 2D measurements were used to determine margins for typical uveal melanoma treatments. According to the authors’ knowledge this has not been addressed by others previously.

2. MATERIALS AND METHODS

2.A. $^{106}$Ru eye plaques

In this paper, three different types of BEBIG plaques were used that are typically applied for different tumor sizes, i.e., CCA, CCB, and COB. For each plaque model, three different plaques were evaluated (CCA model numbers 1263, 1382, and S1505, CCB model numbers 1780, 1855, and S2042, and COB model numbers 935, 980, and S1027, Eckert & Ziegler, Germany). The COB type has a small indentation for the optical nerve that allows for the irradiation of juxtapapillary melanomas [see Fig. 2(b) below].

All $^{106}$Ru sources are applied on a 0.2 mm silver shell which is coated with a 0.7 mm silver backing to shield the beta-rays on the convex side of the plaque. The low-energy beta-particles ($E_{\text{max}} = 39 \text{ keV}$) produced by the embedded layer of $^{106}$Ru are absorbed by a 0.1 mm silver radiation exit window. Having a half-life of 368 days, Ruthenium-106 decays to Rhodium-106 ($^{106}$Rh), having a maximum energy of 3.54 MeV and a mean energy of 1.42 MeV. $^{106}$Rh then decays into stable palladium ($^{106}$Pd) with a half-life of 2.2 h. The spherical plaques have an inner radius of 12 mm to conform to the eye, with an area of 15.3–20.2 mm in diameter. These radii include an inactive rim of 0.7–1 mm (according to the manufacturer), depending on the type of plaque.

2.B. Radiochromic film calibration and measurements

Gafchromic EBT3 (ISP Technologies, Inc., Wayne, NJ) films were used according to published guidelines. They are of water-equivalent material and provide an excellent spatial resolution for 2D dosimetry. EBT3 type films consist of a sensitive polyethylene terephthalate (PTP) layer coated by a thin layer of protecting emulsion yielding a total thickness of 0.3 mm. Upon irradiation, the sensitive layer turns dark without the need for postirradiation film development. They show no significant energy dependence with regard to the dose response of a $^{106}$Ru beta ray emitter. $^{17,32}$

For calibration, films were cut into pieces of 30 $\times$ 30 mm$^2$ and exposed to a $^{60}$Co photon beam at a depth of 4 cm in a polymethyl methacrylate (PMMA) phantom under fixed SSD conditions in a 10 $\times$ 10 cm$^2$ reference field. Additional 10 cm water-equivalent material was used for backscatter. Films were irradiated in ten dose steps between 0 and 25 Gy. One piece of film was left unirradiated for background reading. The films were then digitally read out with an EPSON Perfection V700 scanner (image resolution of 0.169 mm/pixel) 24 h after exposure, to account for postexposure darkening. $^{33}$ For further evaluation only the red component of the RGB image, which shows the highest sensitivity, was used. To obtain a calibration curve, the mean optical densities (OD) of an area of 20 $\times$ 20 mm$^2$ in the center of the films were determined. From these OD values, the net optical densities (netOD) were calculated according to the procedure proposed by Devic et al., $^{34}$

$$\text{netOD} = \log_{10} \frac{SV_{BG}}{SV}. \quad (1)$$

$SV_{BG}$ is the average pixel gray value from the red color channel of the un-irradiated film, and $SV$ is the averaged scan value in the center region of interest of the irradiated calibration films.

For film handling, general recommendations for radiochromic film dosimetry were followed. $^{31}$ Special care was put on the film cutting procedure. A sharp scalpel and hard metal surface were used for the manual film cutting. Additionally, the cutting device was inclined to minimize crimping on that side of the film. Bending of the film was found to be the main cause for separation of the film protective layers and was therefore avoided at all. By doing this, the whole of the film except the first 0.5–1.0 mm could be used to extract dosimetric information. To avoid penetration of water, a thin layer of nail polish was carefully applied on the film edges.

The films were calibrated up to doses of about 25 Gy and irradiated up to doses between 15 and 18 Gy. This dose level...
allowed achieving adequate dose levels even at distances of 12 mm from the plaque’s surface. No significant saturation effects were observed for this calibrated dose range.

A dedicated polystyrene eye phantom was developed in-house. Polystyrene has near water-like properties and is commonly used as phantom material. The phantom consists of a spherical cap with a 12 mm radius and a height of 7 mm, to fit to the shape of the $^{106}$Ru plaques, and several interchangeable slabs (see Fig. 1). The base of the phantom was built from horizontal slabs to allow for film measurements at different depths, each with a square cross-section of $30 \times 30 \text{ mm}^2$ and thicknesses of 1–3 mm.

The phantom can be split into two segments in order to insert an EBT3 film parallel to the central axis $z$ (see Fig. 1). This enables 2D depth dose measurements from the eye plaque surface down to a maximum depth of 22 mm, representative for a human eye. To account for possible backscatter and to allow for a robust assembly of the polystyrene slabs, the eye phantom was embedded in PMMA. In order to avoid air pockets, the whole eye phantom assembly can be filled with water.

Experimental values for reference dose rates were exclusively obtained with EBT3 films, as these were the only detectors available for this study that can be calibrated. To account for the nonwater equivalence of the polystyrene phantom, the dose rate was corrected to absorbed dose to water using reference parameters from ICRU report 72.

Film results were compared to absolute dose rates from MC simulations as well as to the dosimetric reference data provided by the manufacturer. 2D dose distributions measured parallel and perpendicular to the central axis were compared with MC results using dose difference maps. To investigate the reproducibility of the setup, both horizontal and vertical measurements were repeated three times with individually cut film pieces, applying the same read-out protocol each time.

### 2.C. Measurement comparison with high spatial resolution point detectors

For relative dose measurements, a newly developed single crystal diamond detector (PTW-Freiburg, Germany) and a $p$-type silicon diode (PTW-Freiburg, Germany) were used. The microdiamond detector has a $1 \mu \text{m}$ thick sensitive layer with a volume of $0.004 \text{ mm}^3$ and is thus suitable for measuring along steep dose gradients. While having identical outer dimensions of $7 \text{ mm}$ in diameter, the Si-diode has a slightly larger sensitive volume of $0.03 \text{ mm}^3$ ($30 \mu \text{m}$ thickness, $1 \text{ mm}^2$ circular).

Despite the larger sensitive thicknesses compared to films, these detectors provide distinct advantages over ionization chambers. The effective point of measurement does not only depend on the water equivalent thickness of the detector coating and the thickness of the sensitive volume, respectively, but also on the offset of the $7 \text{ mm}$ diameter detectors and the curved inner surface of the plaque. This may introduce uncertainties which we estimate to be less than 10% at the reference depth of $2 \text{ mm}$. In accordance with Soares et al., the effective point of measurement was considered to be the center of the sensitive volume of each detector [see also Fig. 2(a)]. Consecutively, the net effective points of measurements resulted in a depth of $1.80 \text{ mm}$ (diamond) and $1.87 \text{ mm}$ (Si-diode).

For the measurements, the plaques were carefully placed in a paraffin wax mold with the active concave side face up. They were reproducibly aligned with respect to an orthogonal laser beam system in a linear accelerator bunker. The detectors of identical outer dimensions were each mounted on the robotic step motor of a standard scanning water phantom (IBA Dosimetry, Schwarzenbruck, Germany). The high precision motor enabled high resolution dose profile acquisition in steps of $1 \text{ mm}$. The whole setup was surrounded by water. In the setup shown in Fig. 2(a), depth dose profiles along the central axis, from the surface to a depth of $10 \text{ mm}$ in $1 \text{ mm}$ steps, were determined. Figure 2(b) illustrates off-axis profile measurements perpendicular to the central axis across the plaque’s rim. The off-axis measurements were also used to confirm the centered position of the detectors on the central axis for (a) by determining the dose minimum corresponding to the center of the plaque. All measurements were repeated five times in individually set up sessions.

### 2.D. Monte Carlo simulations using MCNP6

All experimental setups described in Secs. 2.B and 2.C were simulated using MCNP6, a MC code that can be used for simulating neutron, photon, and electron transport. The purpose of the MC simulations was to reproduce the performed measurements in order to validate the underlying physical dose calculation model, which is intended to be used for treatment planning as a later stage. After confirming the findings of previous studies, the $^{106}$Ru beta spectrum as well as the prompt gamma emissions of $^{106}$Ru/$^{106}$Rh were disregarded and the final dose calculations were performed using the $^{106}$Rh beta energy spectrum as given in ICRU Report 72. For the simulation of electrons as well as the secondary photons produced by the electron transport in the media (E, P mode) MCNP’s, ITS option was chosen since its accuracy for tissue equivalent materials was confirmed in numerous studies. Furthermore, the ESTEP default option was used. The geometries of the three investigated plaque types (CCB, CCA, and COB) were modeled based on the data made available by the manufacturer for our specific plaques, identified by serial numbers.

Concerning the relative dose measurements with the diamond detector and the silicon diode, the absorbed dose in water was calculated at the respective detector positions as indicated.

![Schematic illustration of the polystyrene eye phantom. It enables horizontal off-axis measurements at different depths in the $x$-$y$ plane and vertical measurements by inserting a properly cut film piece (highlighted in red) in the $x$-$z$ plane.](image-url)
in Fig. 2 for each specific experiment. In order to compare the absolute dose rate with the data provided by BEBIG, the absorbed dose to water was calculated in volumes that had the identical geometry, dimension, and position as the scintillation detectors used by the manufacturer to determine their dosimetric reference data. By applying the stated reference activity of the manufacturer, the respective MC results were converted to an absolute dose rate corresponding to a specific date.

The dosimetric reference data from BEBIG are based on measurements performed in water. Although electron transport in soft tissue and water hardly differs, additional simulations were performed with soft tissue as surrounding medium to assess systematic differences. For this investigation, even two different soft tissue definitions were considered, i.e., the elemental compositions listed in the Oak Ridge National Laboratory’s technical report ORNL TM-8381 (Ref. 41) and the one published in ICRU Report No. 44.42

In order to determine how well the MC simulations can reproduce the experiments, all measurements performed with radiochromic films were simulated by specifically modeling the polystyrene eye phantom, including its material composition. As described before, mesh tallies were superimposed in the respective planes with the same spatial resolution as the film scan. Despite the water-like properties of polystyrene, this approach was chosen in order to minimize systematic uncertainties regarding the comparison of experiment and simulation. Dependent on the specific simulation, between 3 and $20 \times 10^6$ starting particles were used.

2.E. Evaluation of 2D dose distributions

Plaque nonuniformity is an important parameter to characterize the homogeneous activity distribution for radioactive plaques in brachytherapy. Tests for source nonuniformity were performed on a circle of radius $r$ according to published ICRU guidelines,30

$$U_{\text{ICRU}} = \frac{D_{\text{max}} - D_{\text{min}}}{D_{\text{avg}}} \times 100\%, \quad (2)$$

$D_{\text{min}}$ and $D_{\text{max}}$ describe the minimum and maximum absorbed dose rates, respectively, whereas $D_{\text{avg}}$ is the average dose rate. The source nonuniformity $U_{\text{ICRU}}$ should not exceed a maximum of 30%.43 For concave plaques, the measurement of source uniformity was particularly challenging because it contains both the effects of an inhomogeneous source and those of varying distance when the plaque is not perfectly leveled. Therefore, an additional setup was used in which the plaques were placed on a thin PMMA slab with circular cutouts to fit the different plaque types as shown in Fig. 2(c). A piece of film was irradiated for the best possible leveled positioning of the plaques to obtain the pure source nonuniformity index.

Vertical film measurements were used to investigate the penumbra generated by the transition between active $^{106}$Ru layer and inactive rim. This region changes with depth and indicates the usable area of differently sized plaques. If the plaques show a low nonuniformity index across the plaque surface, one can consider vertical measurements from a single upright film piece to be representative for all vertical cross-sections through the central axis.

From these 2D dose distributions, the plaque specific dosimetric characteristics in the penumbra region can be assessed. The dose reduction across the inactive edge of the plaque can be expressed by plotting equidistant dose profiles with respect to the plaque’s surface. Such profiles determined along a constant radius give an idea of the active width of the plaque including its variation as opposed to the geometrical width of the different plaque types. From such measurement results it is possible to estimate dosimetric margins, tumor coverage, and doses to adjacent organs at risk.

Figure 3 illustrates this conceptual idea in more detail. 2D dose distributions were superimposed on model geometries of dome-shaped tumors in order to obtain dosimetric margins and coverage of a broad range of representative tumor volumes. Target coverage was evaluated as follows: 100% isodose was defined at the tumor apex which reflects the simple treatment planning of eye plaque therapy if a minimum target dose is prescribed.44,45 Three- or two dimensional dose calculation and dose volume histogram based evaluation are not widely available for plaque therapy of uveal melanoma, but might contribute to establish DVH-based dose–response relationships for tumors and organs at risk in the future.

The representative tumor model illustrated in Fig. 3 consists of a circular segment of radius $R_{\text{tumor}}$ constrained by the tumor base diameter $d_{\text{base}}$ and the apex height $h_{\text{apex}}$.

**Fig. 2.** Measurement setup of diamond/diode measurement. (a) Illustration of the effective point of measurement when considering the offset $z_{\text{eff}}$ due to the curvature of the plaque and the water-equivalent depth $z_{\text{eq}}$ of the sensitive volume. This allows for measurements along the central axis in $z$-direction and across the plaque in $x$-direction. (b) Ophthalmic plaques used in this study are shown in scale. The radioactive area is marked hatched. The diameters are 15.3, 20.2, and 19.8 mm for CCA, CCB, and COB, respectively. (c) Nonuniformity index measurements were performed in water placing the plaque on a PMMA slab with a dedicated cut-out.
Equation (3) describes the radius of the segment of the circle in which $r_{\text{eye}}$ is the radius of the eye and $\alpha_0$ the angle defined by the tumor base (in one direction from the central axis). From this, one can derive the angle $\beta_n$ defined by points $P_n$ lying on the equidistant profiles with radius $r_n \ (n = 2 \text{ or } 3 \text{ mm})$ at the surface of the tumor,

$$
\beta_n = \frac{r_n^2 - R_{\text{tumor}} + (r_{\text{eye}} + R_{\text{tumor}} - h_{\text{apex}})^2}{2(r_{\text{eye}}(\cos \alpha_0 - 1) + h_{\text{apex}})}.
$$

Equation (4) was used to describe the extension of the tumor from the central axis of the respective points $P_n$ as a multivariate function depending on the shape of the tumor for different base diameters and apex heights, respectively. By converting the angles $\beta_n$ into radian via $eta_n = \beta_n/r_n$, this yields a measure of the tumor extent $b_n$ at depths of 2 and 3 mm. From vertical film measurements, equidistant profiles were obtained and used to describe the extent of dose coverage (100% isodose) attributed to a certain apex height.

3. RESULTS

The statistical error ($k = 2$) of all MC simulations referred to in Secs. 3.A and 3.B was $\leq 1.4\%$.

3.A. Percent depth dose on central axis

Results of the relative depth dose along the central axis, normalized at a depth of 2 mm, are shown in Fig. 4 for all three investigated types of plaques. Film data consist of three independent measurements of the same plaque and two measurements of different plaques. Film measurements were weighted according to the number of measurements performed on the same plaque and finally averaged values for three different plaques were used for dosimetric comparisons. Microdiamond and diode measurements were repeated five times on the same plaque.

The data obtained from the different measurement techniques show good agreement with the MC derived data as well as with the manufacturer’s specification. The film measurements showed deviations of less than 6% for depths up to 7 mm from the plaque surface for CCB and CCA type data. For the COB type plaque, deviations are typically higher. This can also be seen in detail in Fig. 5(a), in which percent depth dose data from the specifications of different plaques of the same type were averaged and plotted with the resulting standard deviation ($k = 2$). These were compared to the averaged film measurements and plotted in terms of relative deviations. The deviations increase significantly for depths larger than 7 mm (CCB and CCA) and 5 mm (COB).
The diode (diamond) detector results resulted in dosimetric deviations of less than 5% (7%) for the first 7 mm (5 mm). At larger depths, deviations increased up to 40%. The reproducibility of measurements is shown in Table I. The given values refer to the lowest precision within the 95% confidence level found among the detectors over the whole depth dose curve.

3.B. Lateral dose distribution and source nonuniformity

The off-axis profiles determined with the microdiamond and diode showed an agreement within 10% compared to the MC simulations for points located at distances ≤8 mm off-axis. These profile measurements also confirmed that the central position of the detector is crucial.

The evaluation of the source nonuniformity yielded mean indices of 15.51%, 16.81%, and 14.69% for the CCB type plaque. CCA type plaques yielded values of 15.79%, 18.51%, and 13.45% and COB values of 17.67%, 20.15%, and 18.50%. The indices are well within the recommended limits of 30%.

3.C. Reference absolute dose rate

Reference dose rates were determined at a depth of 2 mm in accordance with the report of The Netherlands Commission on Radiation Dosimetry. The values were directly taken from the netOD of the vertical film strips. The averaged dose rates over three separate measurements are summarized in Table II together with MC derived data and the manufacturer’s specifications.

EBT3 film measurements were, in general, in good agreement with the manufacturer’s reference data with deviations (k = 2) of 0.4% ± 2.7% (CCB), 1.4% ± 4.5% (CCA), and 4.6% ± 2.9% (COB), respectively.

The largest deviations of about 4.6% were found for the COB type plaque. In this case, the large experimental uncertainty might originate in a systematic offset in the alignment of the plaque and EBT3 film because it was aimed to determine the dose along the cross-section through the optical nerve sparing notch [see Fig. 1(b)]. For the smallest plaque, i.e., the CCA type, a standard deviation (k = 2) of 4.5% was achieved which indicated reproducible results between three independent measurements.

The deviations between the MC simulation and the BEBIG data were higher than the ones obtained when comparing with film data. Therefore, a more detailed analysis was performed in a depth interval between 1 and 10 mm. The absolute dose rate differences in mGy/min between mcν6 and BEBIG are displayed in Fig. 5(b) as the difference from the BEBIG value in every measuring point. As BEBIG data are associated with an uncertainty of about 20% (k = 2) whereas the statistical error of the simulation is negligible (see Table I), these deviations were considered to be acceptable. At a depth of 1 mm, the simulation results for the CCB, CCA, and COB plaques reached only 85%, 89%, and 80% of values specified by the manufacturer. This divergence however quickly vanishes and drops below 5% with respect to the normalized dose rates at distances of 2.5, 1.8, and 3 mm for CCB, CCA, and COB, respectively.

Table II. Absorbed dose rates and uncertainties (k = 2) for three plaque models (CCB, CCA, and COB) on the central axis at a depth of 2 mm obtained with EBT3 film, mcν6 and from the BEBIG data.

| Type | CCB | CCA | COB |
|------|-----|-----|-----|
| Avg. (mGy/min) | SD | Avg. (mGy/min) | SD | Avg. (mGy/min) | SD |
| Measured | 19.91 | 0.54 | 16.12 | 0.72 | 20.61 | 0.60 |
| mcν6 | 17.78 | 0.12 | 15.16 | 0.08 | 17.25 | 0.08 |
| BEBIG | 19.84 | 1.98 | 15.86 | 1.59 | 19.68 | 1.97 |

Repeated simulations of the absolute dose rate with the soft tissue definitions of Oak Ridge National Laboratory and ICRU did not show any significant difference in the volume around the plaque that was considered to be relevant for the dosimetry of $^{106}$Ru plaques, i.e., the area up to 10 mm from the plaque along the central axis, when compared to the results obtained assuming water as medium.

3.D. Dose difference maps of vertical 2D distributions

Two dimensional dose distributions for all three plaque types are shown in Figs. 6(a)–6(c).

Film measurements were compared to MC simulations by means of absolute dose difference maps and maps showing deviations between averaged film measurements from different plaques of the same type and MC simulations [see Figs. 6(d)–6(i)]. In order to align 2D maps, translational shifts of up to 2 pixels in vertical and up to 6 pixels in lateral direction were applied (pixel width 0.169 mm) by a single rigid transformation.

For the absolute dose difference maps [Figs. 6(d)–6(f)], the largest deviations of up 150 cGy between simulation and measurement were found close to the plaque surface within the first 1.5 mm, corresponding to about 15% dose difference with respect to the reference dose at 2 mm depth. At distances between 2 and 7 mm, the absolute dose difference decreases to <50 cGy (5%) and vanishes to 0 cGy beyond that depth. Increased differences in the dose penumbra region around the edges of the plaques can be either explained by a rotational misalignment (which was not corrected to conserve the dosimetric information of the relatively coarse dose grid) or a mismatch between simulated $^{106}$Ru layer and actual extent of the measured plaque. Difference in the penumbra regions was especially observed for the COB type plaque.

Fig. 6. Vertical dose distributions of film measurements for the three applicators with an activity of 4.9 MBq (CCB), 3.1 MBq (CCA), and 4.1 MBq (COB) irradiated for 600, 612, and 607 min, respectively [(a)–(c)]. Absolute dose difference maps between MC calculation and film measurements are shown in [(d)–(f)]; relative deviations for averaged film measurements vs MC in [(g)–(i)]. The last two rows show a comparison of film measurements of the same plaque [(j)–(l)] and between different plaques of the same type [(m)–(o)] for all three investigated plaque types. For all comparisons, a margin of about 0.5 mm was excluded from the cutting edge.
After averaging the film measurements for three different plaques of the same type the influence of source inhomogeneities could be minimized, but still remain prominent for the COB type plaque [Figs. 6(g)–6(i)]. Local relative deviations remained within 5% up to 7 mm from the plaque surface for CCB and CCA type plaques and up to 5 mm for COB.

Results on the reproducibility of the 2D film measurements are shown in Figs. 6(j)–6(l) for two measurements on the same plaque. The influence of source inhomogeneity among plaques of the same plaque type was investigated by comparing film measurements of different plaques [Figs. 6(m)–6(o)]. When comparing relative deviations between two film measurements of different plaques, a good agreement in the central region was observed up to 5–7 mm depth. For film measurements of the same plaque, good agreement was not only observed in the central region up to 7 mm depth, but was found for a wider region in lateral direction as well. At larger depth, relative deviation increased up to 20%–40% as the absorbed doses approached 0 cGy for both comparisons. Discrepancies in the high dose region (>10% of reference dose) appeared primarily for the interplaque comparison close to the plaque edges and were most pronounced for the COB type. This effect can be explained most likely by minor differences in the 106Ru layer extent close to the edges in different plaques of the same type, and again a rotational misalignment of the plaque between different measurements. The comparison of film measurements of the same plaque showed a generally better agreement, especially in lateral direction.

3.E. Penumbra characteristics and tumor coverage

The dosimetric characteristics along equidistant “radial” profiles are shown on the left side in Fig. 7 for both CCA and CCB type plaques. These profiles were in turn used to estimate the tumor extension $x_{\text{tumor}}$ in lateral direction from the center at point $P_n$ ($n = 2$ or 3 mm) that can be treated by considering different scenarios of tumor apex height and basal diameters, respectively. Typically, the dose $D_{\text{CTV}}$ is prescribed to the tumor apex. Hence, with increasing apex height the overall area that is covered by this prescription dose ($D_{\text{CTV}} = 100\%$) is decreased as represented by the area $x_{\text{100\%}}$ enclosed under the dose profile in Fig. 7. However, at the same time the geometric extension of the tumor increases. Any points beyond this area receive less than 100% of the prescribed dose. The difference between the tumor extension $x_{\text{tumor}}$ and the overall margin $x_{\text{100\%}}$ illustrates the specific margin $x_{\text{margin}}$ that guarantees complete tumor coverage. Alternatively, this can be understood as the tolerable shift of the plaque without compromising full tumor coverage with respect to the 100% prescription isodose, thus simulating an error in surgical positioning. Finally, margin matrices were derived that describe $x_{\text{margin}}$ (in mm) for various scenarios of apex heights and base diameters for CCA and CCB type plaques. Typically, margins increase with decreasing apex height and base diameter.

Negative values in Fig. 7 indicate an underdosage at the respective points. Such scenarios were found for especially large basal diameters ($d_{\text{base}} > 11$ mm) for the small CCA type plaque. On the other hand, a typically good coverage was found for these scenarios for the larger CCB type plaque, with margins well above 1 mm.

4. DISCUSSION

Multidimensional dosimetric measurements for eye plaques as used in brachytherapy for uveal melanoma have been
described in other studies. In our study, a recent type of radiochromic film was used and a purpose built phantom enabled measurements in several planes. A reproducible method for eye plaque dosimetry was established that allowed evaluating the dose characteristics. Such dosimetric information can be directly used in a simple treatment planning procedure.

The most noticeable dosimetric deviation was observed between MC simulations and reference values specified by the manufacturer in the vicinity of the plaque, with dose differences ranging from −20% (for COB) to −10% (for CCA). BEBIG data for an individual plaque were measured with scintillators, which might overestimate the dose due to Cherenkov radiation effects. The Cherenkov effect vanishes once the velocity of the charged particle drops below the speed of light in water, which corresponds to an electron energy of about 0.26 MeV assuming specific relativity. The mean electron energy of 0.52 MeV was determined by MC simulations at the most distant measurement point of 10 mm, indicating that a potential effect stemming from Cherenkov radiation would affect the entire depth dose range. Since the observed deviations between measurements and MC simulations only occur at depths below 2 mm, effects from Cherenkov radiation were ruled out. A more likely explanation for these large deviations is uncertainties with respect to size, extension, and homogeneity of the active layer of the plaque. Also, the comparison of the simulations with the measurements of the 2D vertical planes (see Fig. 7) showed divergent results close to the plaque (in a depth of 1–2 mm). Another possible source of error could be the assumption of the effective point of measurement to be the center of the sensitive volume of each detector.

The reproducibility of the film measurements was validated for the same plaque on the central axis and in 2D. Deviations between consecutive measurements of the same plaque were lower than for measurements of different plaques of the same type, especially in lateral direction. This suggests that, while there are considerable uncertainties associated with the difficulty of 

Overall, a good agreement between measured and calculated multidimensional 1D and 2D dose distributions was found. The dosimetric information allowed describing the penumbra region of the plaques in terms of lateral dose falloff, the usable area of the investigated plaque models as a function of depth along the central axis. Moreover, from such dosimetric information margins and coverage values can be extracted for the different types of plaques and applied in treatment planning as illustrated by the following considerations. According to the authors’ knowledge, such an extrapolation of the 2D dosimetric characteristics of plaques and the margins has not been addressed previously in such detail.

A tumor base diameter (e.g., determined from funduscopy) and an apex height of a dome-shaped tumor (e.g., determined from ultrasound imaging) were used as input parameters to determine the plaque size and prescribed dose. The size of the plaque (15.3 mm for CCA and ~20 mm for CCB and COB) was chosen with respect to the clinical target volume (CTV) which usually exceeds the gross tumor volume (GTV) by 1–2 mm on each side. In addition, a margin of 1 mm was added to account for dosimetric uncertainties. According to this protocol, one can evaluate the suitability of different sized plaques to fully cover a target volume in terms of its CTV dimension from the matrices presented in Fig. 7. For example, a tumor with a CTV of 5 mm apex height and 13 mm base diameter yields a margin of −0.06 mm at a depth of 3 mm (P3mm) and 0.45 mm at 2 mm depth (P2mm) when overlaid with the experimentally determined dose distribution of a CCA type plaque. This means that P3mm cannot be completely covered by the prescribed 100% isodose and P2mm would not allow for any margin. With a CCB type plaque, the same CTV can be treated with margins well above 1 mm. This exemplifies that even if the physical diameter of the plaque matches or exceeds the CTV diameter, it is not guaranteed that the entire tumor volume receives the full prescribed dose. So far, a perfectly dome-shaped target volume was assumed but other, more bulky tumor geometries that have been described need to be addressed in more detail from a geometric point of view.

The observed interplaque differences emphasize the importance of these dosimetric safety margins. Sources for inhomogeneities among different plaques of the same type may be the heterogeneous source material as well as uncertainties in determining the edge of the active layer in the rim of the plaque. This may lead to local dose deviations not only close to the plaque surface but also in regions around the central axis. However, while the relative local discrepancies can be large, the introduction of a 1 mm dosimetric margin would in most cases be sufficient.

As seen in Fig. 5(a), but also in the 2D dose map evaluation in Fig. 6, the local deviation between film measurements and BEBIG data (film and MC, respectively) increases significantly for larger distances. While the absolute differences in absorbed doses at these distances remain rather small due to the low dose rates, for very large tumor apex heights these uncertainties must be considered when estimating the dose to adjacent healthy tissue.

While tumor control, recurrence rate, and overall survival are the most important aspects in cancer management, the outcome of 

EBT3 films and an in-house developed phantom enabled multidimensional dosimetric measurements for 

Profiles determined along the central axis and those determined off-axis showed good agreement with values specified.
by the manufacturer, with MC simulations and point detectors well within the experimental uncertainties or tolerances given by the manufacturer.

Dosimetric information from such multidimensional film measurements can be utilized for treatment planning, i.e., to assess target coverage and to establish margin concepts for representative tumor models.

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