Advanced Monte Carlo dose calculations for eye plaque brachytherapy

by

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Abstract

Eye plaque brachytherapy is a radiation therapy technique designed for treating ocular cancers. Current clinical radiation dose evaluations for eye plaque brachytherapy involve considerable assumptions, resulting in inaccurate doses. This thesis aims to advance dose evaluations for eye plaque brachytherapy by implementation of detailed models for various photon and beta plaques in the `egs_brachy` Monte Carlo code. Calculated dose distributions are verified against published results. These plaque models and dose distributions will be distributed to the medical physics community. Dose distributions are calculated for different scenarios (clinical assumptions; plaque in water; plaque in realistic eye model). These results confirm the importance of model-based calculations for photon plaques and show that modelling tissues should be considered for beta plaques. Finally, dose distributions for photon and beta plaques are compared, demonstrating the potential of model-based dose evaluations in future studies.

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Statement of originality

This thesis contains the large majority of work which was undertaken during the author's Master's degree at Carleton University. Many of the results in the following pages form the basis for a future database of geometry and dose distribution files and an accompanying paper. All calculations using `egs_brachy` in this work were done by the author.

Dr. Rowan Thomson supervised the author and contributed to every step of this work. She was the source of the initial project idea, and offered guidance throughout the work, including the final thesis revisions.

Stephen Deering and Dr. Marc Chamberland created initial eye plaque geometries for use with `egs_brachy`. Their designs were used as a point of comparison and a source for ideas in the design of the plaque geometries presented here. The model of the human eye used for calculations throughout this work was developed by Marielle Lesperance.

Contents

Abstract	ii
Acknowledgments	iii
Statement of originality	iv
Table of contents	vii
List of tables	viii
List of figures	x
Nomenclature and notation	xi
1 Introduction	1
1.1 Ocular cancer	1
1.2 Radiation therapy and dose	2
1.3 Eye plaque brachytherapy	2
1.4 Treatment planning	4
1.4.1 Motivation for advanced dose calculations	7
1.5 Model-based and Monte Carlo dose calculations	13
1.6 Thesis purpose	14
1.7 Thesis outline	16

2	Methods	17
2.1	Eye plaque model geometries	18
2.1.1	Photon plaque geometries	20
2.1.2	Beta plaque geometries	27
2.2	Phantom geometry	31
2.2.1	Voxelized eye phantom model	32
2.3	Dose calculations	35
2.3.1	Simulation parameters	36
2.3.2	Dose normalization	37
2.4	Calculation specifications	38
2.4.1	Uncertainties	40
3	Results	42
3.1	Photon plaque results	42
3.1.1	Comparison of D_{TG43} , $D_{w,w}$, and $D_{m,m}$	54
3.2	Beta plaque results	58
3.2.1	Comparison of CIA beta plaque models	64
3.3	Photon and beta plaque comparisons	66
4	Discussion	71
4.1	Verification of egs_brachy and eye plaque models	71
4.1.1	Photon plaque verification	71
4.1.2	Beta plaque verification	73
4.2	Comparison of D_{TG43} , $D_{w,w}$, and $D_{m,m}$	75
4.3	Comparison of photon and beta plaque dose distributions	79
5	Conclusions and outlook	81

Appendix	84
A Photon plaque seed positions	84
References	89

List of Tables

2.1	Photon plaque dimensions	25
2.2	^{125}I spectrum	26
2.3	^{103}Pd spectrum	27
2.4	Beta plaque dimensions	29
2.5	Scoring region dimensions	32
2.6	Eye phantom composition	34
2.7	Photon and beta transport parameters	37
2.8	Source parameters	38
3.1	D_{TG43} and ratios at points of interest	47
3.2	$D_{w,w}$ and ratios at points of interest	52
3.3	Eye structure dose comparison between D_{TG43} , $D_{w,w}$, and $D_{m,m}$. . .	57
3.4	Points of interest comparison between COMS 16 mm and CCB . . .	69
3.5	Eye structure dose comparison between COMS 16 mm and CCB . . .	70
A.1	Seed position coordinates	85

List of Figures

1.1	TG43 geometry	6
1.2	Photon mass-energy absorption coefficients of eye tissues	11
1.3	Electron stopping powers of eye tissues	12
2.1	Eye coordinate systems	18
2.2	Plaque models in model of eye	19
2.3	COMS seed positions	21
2.4	COMS eye plaque geometric parameters	22
2.5	COMS plaque regions	23
2.6	Photon brachytherapy seed geometries	28
2.7	Beta plaque geometry definition	29
2.8	CIA plaque geometry model	31
3.1	D_{TG43} central axis depth-dose curves: ^{125}I	45
3.2	D_{TG43} central axis depth-dose curves: ^{103}Pd	46
3.3	$D_{w,w}$ central axis depth-dose curves: ^{125}I	50
3.4	$D_{w,w}$ central axis depth-dose curves: ^{103}Pd	51

3.5	A three-dimensional $D_{w,w}$ comparison of egs_brachy and BrachyDose	53
3.6	Photon plaque isodose contours	55
3.7	Photon plaque $D_{w,w} / D_{TG43}$ and $D_{m,m} / D_{TG43}$ ratios	56
3.8	Beta plaque central axis relative depth-dose curves	60
3.9	CCB central axis dose per decay with depth	61
3.10	Beta plaque lateral axis dose profiles: Planar, CCA, CCB, and CCC .	62
3.11	Beta plaque lateral axis dose profiles: CCD, CGD, CCX, and CXS .	63
3.12	CIA plaque central axis and notched axis doses	65
3.13	CIA plaque isodose contours	65
3.14	Depth-dose comparisons between COMS 16 mm and CCB beta plaque	67
3.15	isodose contour comparison between COMS 16 mm and CCB beta plaque	68

Nomenclature and notation

AAPM	American Association of Physicists in Medicine
CLRP	Carleton Laboratory for Radiotherapy Physics
COMS	Collaborative Ocular Melanoma Study
$\dot{D}(r, \theta)$	Dose rate
$D_{m,m}$	Dose with plaque in voxelized, tissue accurate eye phantom
D_{TG43}	Dose from superposition of seeds in water
$D_{w,w}$	Dose with plaque in voxelized water phantom
$F(r, \theta)$	2D anisotropy function
$g_L(r)$	TG43 radial dose function
$G_L(r, \theta)$	TG43 geometry function
Gy	Gray [joule / kilogram]
K_{col}	Collision kerma
MBDCA	Model based dose calculation algorithm
MC	Monte Carlo
R_{CSDA}	Continuous slowing down approximation range
S_K	Air-kerma strength
S_K^{hist}	Air-kerma strength per history
S/ρ	Stopping power
TG43	AAPM Task Group No. 43
TG129	AAPM Task Group No. 129
TG186	AAPM Task Group No. 186
Z_{eff}	Effective atomic number
Λ	Dose rate constant

μ_{en}/ρ	Mass-energy absorption coefficient
ρ	Mass density [g / cm ³]
$\phi_{an}(r)$	1D anisotropy function
Φ_E	Photon fluence differential in energy
(x, y, z)	Eye plaque coordinate system
(X, Y, Z)	Eye centre coordinate system

Chapter 1

Introduction

1.1 Ocular cancer

There are several forms that cancer can take within the eye including ocular melanoma, intraocular lymphoma and retinoblastoma. There are 2500 reported cases of uveal melanoma per year in the United States, making it the most common form of ocular cancer¹. Uveal melanoma arises from the uveal layer of the eye, which is the tissue layer that contains the iris, the choroid, and the ciliary body. Uveal melanoma is usually diagnosed through a clinical examination with biomicroscopy and ophthalmoscopy and can be further confirmed using ultrasonography, fundus fluorescein angiography, or optical coherence tomography². Previously, the most common treatment for ocular melanoma was enucleation, which is a full removal of the eye. While enucleation offers a high level of local tumor control, loss of the eye is not a desired outcome. In the pursuit of treatments which allow for vision retention, radiation therapy offers numerous choices. Radiation therapy techniques used for ocular cancer treatments include external beam therapy of various types such as: photon^{3,4}, proton^{5,6,7,8}, helium-ion⁹, and Gamma Knife^{10,11,12}. Eye plaque brachytherapy is a technique that was specifically developed for ocular cancer treatment and is the focus of this thesis.

1.2 Radiation therapy and dose

Radiation therapy is a cancer treatment that uses ionizing radiation to deliver ‘dose’ to a cancerous region in an attempt to damage the cancerous cells¹³. Dose is an important quantity in radiation physics that gives the energy deposited in a medium by ionizing radiation per unit mass. The unit for dose is the Gray (Gy), which is given by joules per kilogram in SI units [$\text{Gy} = \frac{\text{J}}{\text{kg}}$]. If enough dose is delivered to the cancerous cells then they will experience cell death via different mechanisms such as apoptosis (a form of programmed cell death), mitotic catastrophe (the primary form of cell deaths from ionizing radiation, caused by abnormal entry to mitosis), and senescence (a state where damaged cells no longer replicate)^{14,15}. Unfortunately, by the same mechanisms radiation therapy can damage and kill healthy tissue. When performing radiation therapy treatments care is taken to prevent excess damage to healthy tissues; this is where treatment planning and dose evaluation come in. During treatment evaluation, dose calculations are made to determine an appropriate radiation treatment for the patient, where an appropriate treatment attempts to deposit the majority of energy in the cancerous tissue and limits the energy deposited to healthy tissues. For treatment evaluation to be as effective as possible, advancing the accuracy and availability of dose evaluation methods used for treatment evaluation is crucial.

1.3 Eye plaque brachytherapy

Brachytherapy is a radiotherapy (another term for radiation therapy) cancer treatment that involves radioactive sources which are placed near or in the area to be treated¹⁶. Brachytherapy sources traditionally come in the form of ‘seeds’. These seeds are small (a few mm in length) and contain a radioisotope such as ^{125}I , ^{103}Pd

or ^{131}Cs . The photons emitted by brachytherapy seeds tend to have a much lower energy than external beam treatments. For example, ^{125}I has a mean photon energy of 28 keV and ^{103}Pd a photon energy of 21 keV compared to typical external beam energies that are over 1 MeV. The lower energy of these treatments allows for a steep drop off in photon fluence and gives the ability to deposit energy local to the treatment target while avoiding excessive dose to organs at risk further from the seeds.

Eye plaque brachytherapy is a specific type of brachytherapy used to treat ocular cancers. Eye plaque brachytherapy was developed with the purpose of retaining vision while treating cancer. The first recorded attempt at using brachytherapy for eye treatments was using radon seeds in 1930^[17]. Since then many different sources have been used including ^{60}Co ^[18], $^{106}\text{Ru}/^{106}\text{Rh}$ ^[19, 20, 21], ^{125}I ^[22, 23, 24, 25], ^{103}Pd ^[26, 24, 25, 27], ^{198}Au ^[28], $^{90}\text{Sr}/^{90}\text{Y}$ ^[21], and ^{131}Cs ^[24, 25]. Eye plaques are typically metallic plaques that come in various sizes and configurations. The plaques contain a radionuclide source in the form of multiple brachytherapy seeds or a film coated in a radioactive substance. The eye plaques are sutured onto the eye over the cancerous region and left in place for the duration of treatment time (which is generally on the order of 100 hours). The most common sources in use today (^{125}I , ^{103}Pd , ^{131}Cs , $^{106}\text{Ru}/^{106}\text{Rh}$) are lower energy photon sources or beta sources.

The plaques themselves come in various designs, but a standard design was introduced during a large scale study that began in 1985 known as the Collaborative Ocular Melanoma Study (COMS)²⁹. COMS was a multi-institutional clinical trial that used standardized plaque designs and treatment methods to examine the effectiveness of eye plaque brachytherapy treatments. The COMS study concluded that eye plaque brachytherapy is comparable to enucleation for tumor control³⁰. A recent study concluded that eye plaque brachytherapy is more effective than proton therapy

for tumor control³¹.

The COMS plaque designs are not the only plaque designs in use. Various other plaque models are used in clinics around the world. There are also silver plaque designs that utilize beta radiation as the active method of energy deposition. These plaques are more commonly used in Europe. The most popular beta plaque variants are $^{106}\text{Ru}/^{106}\text{Rh}$ designs created by Eckert & Ziegler BEBIG of Berlin, Germany. The mean energy of $^{106}\text{Ru}/^{106}\text{Rh}$ beta particle emissions is 1.41 MeV and the maximum energy is 3.54 MeV which result in approximate ranges of about 7 mm and 15 mm in soft tissue respectively based on the assumption of continuous energy loss as the electrons traverse the medium (this assumed range is known as R_{CSDA} , where CSDA stands for continuous slowing down approximation). The short range of electrons in tissue allows for a high deposition of energy in the tumor region and limited deposition of energy in surrounding tissues and organs at risk³².

1.4 Treatment planning

Today, most dose calculations for treatment evaluation of photon eye plaque brachytherapy are done using a protocol created by the American Association of Physicists in Medicine (AAPM) Task Group 43 (TG43)^{33,34}. The TG43 protocol includes various assumptions that make the calculation of dose easier. Among these assumptions is that the patient is made purely of water, the presence of the eye plaque is ignored, and any interseed effects (interaction of photons emitted from a seed with other nearby seeds) that may occur are also ignored. When using the TG43 protocol eye plaque dose calculations are often done such that a dose of 85 Gy is prescribed to the tumor apex³⁵.

The TG43 formalism gives two possible equations, one for one dimensional (1D)

(referring to the fact that the source is in effect treated as a ‘point’ source) calculations and one for two dimensional (2D) (referring to the fact that the source is in effect treated as a ‘line’ source) calculations. The generalized 1D brachytherapy dose rate formula given by the TG43 protocol is,

$$\dot{D}(r, \theta) = S_K \Lambda \frac{G_L(r, \theta)}{G_L(r_0, \theta_0)} g_L(r) \phi_{an}(r) \quad (1.1)$$

and for the 2D case,

$$\dot{D}(r, \theta) = S_K \Lambda \frac{G_L(r, \theta)}{G_L(r_0, \theta_0)} g_L(r) F(r, \theta) \quad (1.2)$$

where r is the distance to the point of interest from the centre of the source; r_0 is the given protocol reference distance of 1 cm; θ is the polar angle at the point of interest with respect to the longitudinal axis of the source; θ_0 is a protocol reference of 90° and defines the transverse plane of the source; S_K is the air-kerma strength given by $S_K = K_\delta(d)d^2$ (the air-kerma rate in vacuo at distance d multiplied by that distance squared); Λ is the dose-rate constant in water, $\frac{\dot{D}(r_0, \theta_0)}{S_K}$; $G_L(r, \theta)$ is the geometry function that gives an inverse square-law correction to the dose rate; $g_L(r) = \frac{\dot{D}(r, \theta_0)}{\dot{D}(r_0, \theta_0)} \frac{G_L(r_0, \theta_0)}{G_L(r, \theta_0)}$ is the radial dose function that accounts for the effects of photon attenuation and scattering on the dose rate with respect to the radius r ; and $F(r, \theta)$, called the 2D anisotropy function, where $F(r, \theta) = \frac{\dot{D}(r, \theta)}{\dot{D}(r, \theta_0)} \frac{G_L(r, \theta_0)}{G_L(r, \theta)}$ accounts for the angular dependent attenuation and scattering effects on the dose rate. In the 1D case $F(r, \theta)$ is replaced by $\phi_{an}(r)$, the 1D anisotropy function which is $\frac{\int_0^\pi \dot{D}(r, \theta) \sin(\theta) d\theta}{2\dot{D}(r, \theta_0)}$ or the ratio of solid angle weight dose rate to the on the traverse plane at the same distance. For reference, r and θ are defined in Figure 1.1.

However, TG43 is a dose calculation protocol specific to photon brachytherapy

Figure 1.1: Schematic diagram of the geometry and variables used for the TG43 formalism.

and is not appropriate for beta plaque dose calculations. Beta plaque treatment planning and dose evaluation do not have a standardized protocol to follow and various methods may be used when planning a treatment. The calculation of dose for beta plaques is often done using 1D methods along the axis defined from the centre of the plaque and therefore only considers depth and cannot account for organs at risk outside of the sclera. These 1D calculations are done with knowledge of the plaque source activity to calculate time needed to reach the prescription dose at tumor apex. 2D or 3D calculations in the clinic are limited to hand calculations or Plaque Simulator, a commercially available but non-FDA approved eye plaque treatment planning system created by Eye Physics, LLC of Los Alamitos, CA (which may also be used for photon eye plaque dose calculations).

1.4.1 Motivation for advanced dose calculations

The assumptions employed by the TG43 approach create discrepancies between calculated and delivered doses, this can be seen both from a physics stand point and from the conclusions of recent works on the subject of model-based dose calculations. Various reports and studies recommend including a model-based approach in treatment planning and evaluation including the AAPM reports TG129^[36] and TG186^[37].

To understand why the assumptions of TG43 break down an understanding of photon interactions with matter is needed. Photon interactions with matter come in one of four types: the photoelectric effect, Compton scattering, Rayleigh scattering, and pair/triplet production. Photon interactions with matter result in an attenuation of incoming photons which can lower photon fluence (number of photons crossing a perpendicular area) and energy fluence (amount of energy crossing a perpendicular

area) as the photons traverse a given medium. At low energies photon interactions with matter are dominated by the photoelectric effect. Compton scattering is the second most important type of photon interaction at low energy levels and is the result of a photon interacting with a charged particle (such as an electron). Compton scattering is inelastic and results in a loss of energy and a change of direction for the photon and a gain of some recoil energy by the electron. Pair/triplet production is an interaction where a photon of high enough energy causes the creation of an electron-positron pair. Pair/triple production does not occur at the low energies of photon brachytherapy seeds. Rayleigh scattering is only a minor contributor to photon interactions with matter. Rayleigh scattering is elastic and the incident photons do not lose energy (and so do not deposit any energy in a medium), though their direction changes. Compton scattering is only linearly dependent on Z_{eff} (the effective atomic number of a medium) and does not see a great increase in occurrence with change in Z_{eff} ; however, the dominating photoelectric effect is highly Z_{eff} dependent (its cross-section per atom is related to $\frac{Z^n}{h\nu^3}$ where n can vary between 4 and 5)³⁸. The high Z_{eff} dependence of the photoelectric effect establishes that at low energies the probability of a photon interaction in matter occurring becomes highly dependent on material composition and small change in composition could greatly affect the final dose. The increased dependence on material composition is especially relevant for eye plaques where the plaques themselves act as shielding and are made out of much higher Z_{eff} components than the tissue media being irradiated.

To understand where the probabilities of photon interactions in matter and their dependence on Z_{eff} come into play it is important to note how absorbed dose is calculated for low energy photons. Typically when considering low energy photon sources like brachytherapy seeds, absorbed dose is assumed to be equal to the collision KERMA (Kinetic Energy Released per unit MAss). Kerma is the quantity used to

discuss energy transferred to charged particles in a medium by non-ionizing particles such as photons. Kerma has both a radiative kerma component (energy transferred that results in loss to radiative interactions) and a collision kerma component (energy transferred that results in local energy deposition through electron ionization and excitation). The assumption of equivalency between absorbed dose and collision kerma is viable because the range of electrons at such low energies is very short in comparison to the considered scale. At low energies absorbed dose for a spectrum of photons is given by the general definition of collision kerma,

$$D \simeq K_{col} = \int_0^{E_{max}} E \Phi_E \left(\frac{\mu_{en}(E)}{\rho} \right)_m dE \quad (1.3)$$

where Φ_E is the photon fluence differential in energy ($\frac{d\Phi}{dE}$), and $\left(\frac{\mu_{en}(E)}{\rho} \right)_m$ is the mass-energy absorption coefficient for the photon energy E in medium m . These quantities are all integrated up to the maximum energy of the incident photon spectrum. Photons deposit energy in a medium through the photon interactions discussed previously, and because of this the mass-energy absorption coefficient, $\left(\frac{\mu_{en}}{\rho} \right)$ depends on the probabilities of photon interactions which in turn depend on Z_{eff} at low energies.

The TG43 formalism treats the patient as an infinite body of water, where as real human tissue and organ compositions can differ markedly from water and each other. To see the effect change in medium can have on $\left(\frac{\mu_{en}}{\rho} \right)_m$ refer to Figure 1.2 which shows the ratios of $\left(\frac{\mu_{en}}{\rho} \right)_m$ for different eye tissues to $\left(\frac{\mu_{en}}{\rho} \right)_{water}$. TG43 also does not take into account the shielding effects of the eye plaque or photon attenuation caused by the high Z materials that seeds are often made of (e.g. titanium and lead are components of the ^{103}Pd Theragenics TheraSeed 200, titanium and silver are components of the ^{125}I OncoSeed 6711, etc.). The limitations of the TG43 approach

motivate the development and application of advanced dose calculation algorithms.

Charged particles (such as electrons) do not interact with matter in the same way that nonionizing particles (such as photons) do. An incident electron may interact with matter in one of three ways: an elastic collision with the matter, an inelastic collision with bound electrons, or a radiative interaction with the nucleus. Elastic collisions result in no energy loss for the incident electron and so do not deposit energy in the medium. Inelastic collisions with a bound electron result in a transfer of energy to that electron which can result in excitation or ionization. Radiative interactions with the nucleus result in the deceleration or slowing down of the incident electron through the electric field of the nucleus, which then emits bremsstrahlung radiation in the form of a photon. The loss of energy in the incident electron is characterized by the total stopping power, S_{tot} . S_{tot} is the average energy loss per unit distance of a charged particle as it traverses some medium m . Like the mass-energy absorption coefficient, stopping power tends to be presented as a ratio with medium mass density, or mass stopping power, $\left(\frac{S}{\rho}\right)_m$. Figure 1.3 shows the ratio of stopping power in various eye tissues to that in water, $\left(\frac{S}{\rho}\right)_{water}^m$. While the differences seen in $\left(\frac{S}{\rho}\right)_{water}^m$ are not as large as those seen in $\left(\frac{\mu_{en}}{\rho}\right)_{water}^m$, their existence does give motivation to check the effects the inclusion of eye tissues has on beta plaque dose calculations.

Figure 1.2: Ratios of mass-energy absorption coefficient in different eye tissues to mass-energy absorption coefficient in water at increasing photon energies calculated using the EGSnrc application ‘g’.

Figure 1.3: Ratios of total electron stopping power in different eye tissues to total electron stopping power in water at increasing electron energies. Stopping powers are calculated using NIST ESTAR³⁹.

1.5 Model-based and Monte Carlo dose calculations

Model-based dose calculation algorithms (MBDCAs) are a class of advanced dose calculation algorithms. MBDCAs take into account detailed models of shielding, applicators, seeds, and human tissue when calculating dose. Of the model-based methods that have been published, one of the most promising and the focus of this work is dose calculation through Monte Carlo simulation.

Monte Carlo dose calculations utilize detailed geometries of applicators and sources which are placed into voxelized patient models (where voxels are volume elements). The voxels are assigned compositions and mass densities to create an accurate model of the treatment area. A variety of Monte Carlo codes are capable of calculating eye plaque brachytherapy doses including BrachyDose^{40,41,42,43}, the MCNP series of codes^{44,24,45,42}, PENELOPE^{46,47}, and GEANT4^{48,49} and many results are available in the literature from these codes.

Recent simulation studies including those by Melhus and Rivard²⁴ using MCNP5, and Thomson et al.^{41,50} using BrachyDose give dose results using fully loaded plaques in water. These results found a reduction of dose at tumor apex (depth 5 mm) of up to 11% for ¹²⁵I seeds and 19% for ¹⁰³Pd seeds when compared to calculations emulating the TG43 approach. Further, Rivard et al.⁴² gave dose results from three different brachytherapy TPSs (Pinnacle, BrachyVision, and Plaque Simulator) and two different MC codes (MCNP5 and BrachyDose) for a modelled fully loaded COMS 16 mm plaque in water and compared them to calculations emulating the TG43 approach. This comparison found that when aiming for a prescription dose of 85 Gy the actual delivery is 76 Gy for ¹²⁵I seeds and 67 Gy for ¹⁰³Pd seeds.

Other simulation studies have shown that the use of modified COMS plaques or non-standard plaques creates dose differences as well. Using Plaque Simulator,

Astrahan et al.⁵¹ showed differences of up to 11% when comparing a COMS plaque with a gold alloy seed guide to the traditional Silastic insert. A 2014 work by Lesperance et al.⁵² developed five non-standard plaque models for use in BrachyDose and found significant dose differences not only when compared to TG43 emulating calculations, but also between the plaque models (up to 12% at the opposite side of the eye). Another 2014 work by Lesperance et al.⁵³ utilized an anatomically realistic eye model along with plaque models to calculate doses with BrachyDose. The work found considerable differences when comparing doses calculated with the eye model to those with a plaque in water, signaling a need to advance model-based calculations even further.

This work utilizes the Monte Carlo code, `egs_brachy`. `egs_brachy` is a new open source, fast, and efficient Monte Carlo code created specifically for brachytherapy dose calculations by members of the Carleton Laboratory for Radiotherapy Physics (CLRP) research group⁵⁴ and implemented within the EGSnrc family of radiation transport tools. `egs_brachy` is paired with the open-source `egs++` geometry package, which is used to create models of sources, plaques, and patients.

1.6 Thesis purpose

With the need for model-based calculations in treatment planning comes a need for methods to do those calculations. This thesis aims to demonstrate the need for modelling both the plaque and the eye for both photon and beta plaques. The literature does not show any previous attempts to investigate the impact of eye tissues on beta plaque dose distribution compared to water based calculations. This work also aims to verify `egs_brachy` and plaque models developed in this work as tools for model-based calculations against previously published results. In this process a dataset of eye plaque geometries and dose distributions is developed that will be

distributed with `egs_brachy` and on an internet database. This distribution will facilitate future research into eye plaque brachytherapy and the use of model-based calculations for clinical dose evaluation. The created geometries include all sizes of COMS plaques, four models based on non-standard photons plaques, most of the standard BEBIG beta plaque catalog, and a method for creating a more accurate model for BEBIG beta plaques which contain a cutout notch. The development of plaques and the results calculated herein enable a comparison between photon and beta plaque dose distributions and their doses to various eye structures.

In summary, the context and purpose of this thesis are as follows: Current clinical approach to dose calculations has been found inadequate. The TG43 formalism does not accurately emulate the physics behind photon plaque brachytherapy, nor does it compare favourably to MBDCA results found in the literature. Further, a standardized calculation method for beta plaque brachytherapy does not even exist. The work done in this thesis aims to contribute to the advancement of dose evaluation for treatment and research through the use of Monte Carlo calculations. The purposes of this thesis are to:

- (i) Demonstrate the need for model-based dose calculations**
- (ii) Create a database of geometries and dose distributions for eye plaque brachytherapy**
- (iii) Verify `egs_brachy` dose distributions against other published results**
- (iv) Compare dose distributions of photon and beta plaques**

1.7 Thesis outline

Chapter 1 has introduced the ideas behind eye cancer treatment and current dose evaluation, as well as why model-based calculations are being investigated. Next, **Chapter 2** explores the methods used to model the source, plaque, and phantom geometries and obtains accurate dose calculations from the Monte Carlo simulations using those geometries. **Chapter 3** reports the results of these simulations and verification for photon plaques, then beta plaques, and finally a comparison between the two. **Chapter 4** discusses the verification of `egs_brachy` and the plaque models, the comparison of model based calculations to those which emulate the TG43 protocol, and the differences between photon and beta dose distributions. **Chapter 5** discusses conclusions drawn from the results of the thesis as well as future work that could be enabled by it. Lastly, an **Appendix** includes information on the seed positions for all photon plaques presented in this work.

Chapter 2

Methods

This chapter describes the methods used to create eye plaque dose distributions with `egs_brachy` version v2017.09.15. The chapter starts with the development of the geometries used for this work including the eye plaque applicator, seed sources, phantoms, and scoring regions. Following an introduction to the geometries is a summary of the parameters used for simulations and the methods for calculation and normalization of dose. Finally, a description of the steps taken to verify the doses against other Monte Carlo codes is included.

Two Cartesian coordinate systems are used when discussing the plaque and phantom geometries, as well as dose distributions: the eye plaque coordinate system and the eye centre coordinate system. The former has an origin located at the inner sclera on the z axis and the centre of the plaque face on the xy plane, and the latter has an origin located at the centre of the eye. These coordinate systems are based on those used in the Collaborative Ocular Melanoma Study (COMS) where the eye was assumed to have a diameter of 2.46 cm and a sclera 1 mm thick. The eye plaque coordinate system can be converted to the eye centre coordinate system by a translation of +1.13 cm along the z axis. To simplify notation the eye plaque coordinate system will use the variables (x, y, z) and the eye centre coordinate system (X, Y, Z) . Figure 2.1 shows the location of the two coordinate systems.

Figure 2.1: Schematic diagram of eye coordinate systems (not to scale), (a) shows the eye plaque coordinate system (x, y, z) with the origin at the inner sclera; (b) shows the eye centre coordinate system (X, Y, Z) where the origin is at the centre of the eye. A shift of $z+1.13$ cm to convert from (x, y, z) to (X, Y, Z) is needed. In both cases y/Y is coming out of the page.

2.1 Eye plaque model geometries

egs_brachy utilizes the open source egs++ geometry package for the creation of simulation geometries. This work involves the creation of many different plaque (both photon and beta) models with egs++. The models will be distributed through an internet database and eventually included in a future version of egs_brachy. This section includes the geometric descriptions of 10 to 24 mm COMS, “no lip - silastic”, “short lip - acrylic”, “COMS thin - acrylic”, “stainless steel - acrylic”, CCA, CCB, CCC, CCD, CXS, CCX-Y-Z, CGD, and CIA plaques. Each of these plaque models (with CCB representing the beta plaques) can be seen placed on the nasal side of a model of the right eye in Figure 2.2.

Figure 2.2: A cross section of an eye model at $Y=0.025$ cm. The model is of the right eye with the plaque located on the nasal side. Each of the subfigures includes a different plaque type. The COMS 16 mm photon plaque and the CCB beta plaque were selected to represent the COMS design standard and BEBIG plaque designs respectively. Some of the important structures of the eye are labeled in subfigure (a).

2.1.1 Photon plaque geometries

Standard COMS plaques

In 1985 the Collaborative Melanoma Study (COMS) created a collection of standardized eye plaques. These eye plaques consist of a gold alloy backing and a silicone-polymer insert. Commercially, these materials are known as Modulay and Silastic respectively. For the backing a medium composed of gold, silver, copper, and palladium (mass fraction composition is 0.77, 0.14, 0.08, and 0.01 respectively) with a mass density ρ of 15.8 g cm^{-3} is used. For the insert the medium is given a composition of hydrogen, carbon, oxygen, silicon, and platinum (mass fraction composition is 0.063, 0.249, 0.289, 0.399, and 0.00005 respectively) with $\rho = 1.12 \text{ g cm}^{-3}$. The radioactive source for these eye plaques consists of standard brachytherapy seeds that are fitted into the Silastic insert. Each plaque size has a different set of positional indentations in the insert for fitting the seeds (e.g. the 16 mm diameter COMS plaque has indentations for 13 seeds within its insert). Seed positions for COMS 10 mm to COMS 22 mm are taken from the AAPM Task Group 129 report³⁶ and are included in Appendix A. For more detailed information on seed positions refer to Table A.1, and Figures 2.3 and 2.4. The original set of eye plaques ranged from 12 mm to 20 mm in diameter in 2 mm increments. The 10 mm, 22 mm, and 24 mm⁵⁵ plaques were added to the range in the following years.

The plaque designation (e.g. COMS 16 mm) gives the diameter of the plaque insert at its widest point in the xy plane. The backing adds 1 mm to the overall diameter (e.g. COMS 16 mm is 17 mm in total diameter). The inner and outer radius of curvature for the backing are 14.55 mm and 15.05 mm respectively for all plaque sizes. The radius of curvature of the insert where it rests against the eye is

Figure 2.3: Schematic diagram of COMS plaque seed positions (not to scale). Seed positions are for a COMS 16 mm plaque. x_c , y_c and z_c are the x , y , and z coordinates for the seed centres, where z comes out of the page.

12.30 mm, once again for all plaque sizes. Each plaque size has a different height, which is given in Table 2.1.

For the purposes of modelling the COMS plaques, they are divided up into three ‘regions’ as shown in Figure 2.5. Region 1 consists of curved backing on both the inside and the outside. Region 2 is a small region where the inside of the plaque is still curved but the outside has become a straight lip. For Region 3 the backing is all lip, while the insert ends in the curved indentation meant to sit against the eye.

The COMS 24 mm plaque was not a part of the original study and was developed by McCauley Custinger et al.⁵⁶ for the Mayo Clinic in 2015. It was created to treat larger tumors than treatable by available COMS plaques. Geometry information can be found in Table 2.1 and seed positions in Table A.1. Not all plaques used in the

Figure 2.4: Schematic diagrams of COMS plaque geometry parameters (not to scale). Subfigure (a) shows the eye plaque coordinate system where the z-axis is along the plaque central axis and the origin is at the point where the z-axis intersects with the inner sclera. x_c , y_c and z_c are the x , y , and z coordinates for the seed centres, where the y -axis is out of the page. Subfigure (b) shows the geometry parameters of the plaque; R_o is the outer radius of curvature of the plaque backing; R_i is the inner radius of curvature of the plaque backing; r_i is the inner radius of the plaque backing; r_o is the outer radius of the plaque backing; l_i is the inner lip length of the plaque backing; l_o is the outer lip length of the plaque backing; and h is the total height of the plaque.

Figure 2.5: Schematic diagram of the three regions which make up the COMS plaque geometry in egs++ (not to scale). Region 1 has wholly curved backing, Region 2 has curved inner backing and straight outer backing, and Region 3 has both inner and outer backing that are straight and includes the indent placed against the eye.

clinic follow the COMS plaque guidelines perfectly and often use different materials or sizes than the standard. Four plaque models, described below, based on those found in a paper by Lesperance et. al. are provided as an example of other potential models⁵².

“Short lip - acrylic”

This plaque has a shorter lip than the standard COMS 16 mm plaque. While the backing is still Modulay, the insert is acrylic (all acrylic inserts are given a mass fraction composition of 0.6 carbon, 0.32 oxygen, and 0.08 hydrogen, and mass density 1.19 g cm^{-3}). The short lip causes the seed centres to be closer to the eye with $r_{seed} = 1.295 \text{ cm}$. More detailed information on seed positions is available in Table A.1, and for the plaque geometry in Table 2.1. Figure 2.2 gives a visual representation

of this plaque in a patient model.

“COMS - thin acrylic”

This plaque still has a Modulay backing, but has a thin acrylic insert that the seeds are attached to, rather than inserted into. It was designed to approximate plaques like those used in a few different experimental studies. In one work seeds were affixed to a thin acrylic insert placed in a standard COMS backing⁵⁷. Another similar plaque uses a thin reusable seed guide to affix the seeds to an acrylic layer in the same positions as the standard COMS plaque⁵¹. Similarly, the OSU-NAG plaque is a Modulay backed plaque with fewer seeds that are also glued onto the backing⁵⁸. The plaque in this work is modelled with the same seed positions as the standard COMS 16 mm plaque. More detailed information on seed positions is available in Table A.1 and for the plaque geometry in Table 2.1. Figure 2.2 gives a visual representation of this plaque in a patient model.

“No lip - Silastic”

This plaque still has a Modulay backing with the standard Silastic insert, but the backing has no collimating lip. Plaques used by both Puusaari et al.⁵⁹ and Nag et al.⁶⁰ use a gold backing with no collimating lip, though the latter affixes the seeds directly to the backing. For the plaque model presented here the backing is Modulay like the other COMS plaques rather than gold. The seeds are placed at the same position as for the short lip plaque with seed centres $r_{seed} = 1.295$ cm. More detailed information on seed positions is available in Table A.1 and for the plaque geometry in Table 2.1. Figure 2.2 gives a visual representation of this plaque in a patient model.

Table 2.1: Geometric dimensions for all photon plaques where r_{seed} is the radius from eye centre to a seed centre; R_o is the outer radius of curvature of the plaque backing; r_i is the inner radius of the plaque backing; r_o is the outer radius of the plaque backing; l_i is the inner lip length of the plaque backing; l_o is the outer lip length of the plaque backing; and h is the total height of the plaque as defined in Figure 2.4.

Plaque	r_{seed} (cm)	R_o (cm)	r_i (cm)	r_o (cm)	h (cm)	l_i (cm)	l_o (cm)
COMS 10 mm	1.370	1.505	0.50	0.55	0.38	0.24	0.28
COMS 12 mm	1.370	1.505	0.60	0.65	0.43	0.25	0.28
COMS 14 mm	1.370	1.505	0.70	0.75	0.49	0.26	0.29
COMS 16 mm	1.370	1.505	0.80	0.85	0.57	0.28	0.31
COMS 18 mm	1.370	1.505	0.90	0.95	0.67	0.30	0.33
COMS 20 mm	1.370	1.505	1.00	1.05	0.79	0.34	0.36
COMS 22 mm	1.370	1.505	1.10	1.15	1.00	0.40	0.42
COMS 24 mm	1.370	1.505	1.20	1.25	1.23	0.55	0.57
“Short lip - acrylic”	1.295	1.410	0.80	0.85	0.46	0.15	0.30
“No lip - silastic”	1.295	1.410	0.80	-	0.47	-	-
“COMS - thin acrylic”	1.370	-	0.80	0.85	0.57	0.28	0.31
“Stainless steel - acrylic”	1.345	1.505	0.80	0.90	0.56	0.21	0.26

“Stainless steel - acrylic”

This plaque has a thicker (1 mm) stainless steel backing (mass fraction elemental composition of 0.9905 iron, 0.015 carbon, 0.05 manganese, and 0.03 sulfur with mass density 7.9 g cm^{-3}) instead of the standard Modulay, and uses an acrylic insert. This plaque is representative of the ROPES eye plaque⁶¹ as well as a plaque developed by Karolis et al.⁶² The seed centres are again slightly different from the COMS 16 mm plaque with $r_{seed} = 1.345 \text{ cm}$. More detailed information on seed positions is available in Table A.1 and for the plaque geometry in Table 2.1. Figure 2.2 gives a visual representation of this plaque in a patient model.

Source models

For photon plaques the sources are standard brachytherapy seeds. `egs_brachy` comes bundled with various seed models created with `egs++` and these are used as the source geometries for the photon plaque simulations in this thesis. Two different seed types are used, an ^{125}I seed model based on the Amershan OncoSeed 6711 seed^{63,64} and a ^{103}Pd seed model based on the Theragenics Co. TheraSeed 200 (T200) seed^{65,66,67}. The `egs_brachy` models of these seeds were benchmarked in Chamberland et al.⁵⁴ in 2016.

The 6711 seed (shown in Figure 2.6) has an outer container of titanium ($\rho = 4.54 \text{ g cm}^{-3}$) with a length of 3.75 mm, a diameter of 0.800 mm, walls that are 0.0700 mm thick, and hemi-spherical end caps that are 0.375 mm thick. Within this titanium tube is a silver rod ($\rho = 10.5 \text{ g cm}^{-3}$) of length 2.80 mm, a radius of 0.250 mm, and end points have radius 0.175 mm. This silver rod is coated in a $1.75\mu\text{m}$ thick AgI/AgBr isotope (elemental composition 0.536 Ag, 0.284 Br, 0.18 I, and $\rho = 6.20 \text{ g cm}^{-3}$). The coated silver rod is left free to move within the titanium cylinder⁶⁴. Table 2.2 gives the tabulated line spectrum used to initialize photons in the ^{125}I seeds^{68,69}.

Table 2.2: Spectrum used for ^{125}I brachytherapy seeds taken from NCRP Report 58^{68,69}.

Energy (MeV)	Intensity (%)
0.0272017	0.397
0.0274723	0.741
0.031	0.257
0.0354919	0.0668

The T200 seed shown in Figure 2.6 has a more complex geometry than the 6711 seed. The outer container is a titanium tube with wall thickness 0.0560 mm

and outer diameter 0.826 mm. This tube has end cups rather than end caps that have a thickness of 0.040 mm. The end cups are a hemispherical shell with a hollow cylinder of length 0.334 mm attached, and both have an inner radius 0.306 mm. The combination of tube and end cups has a length of 4.500 mm. Within the tube are two graphite cylinders ($\rho = 2.265 \text{ g cm}^{-3}$) coated in radioactive Pd ($\rho = 12.02 \text{ g cm}^{-3}$). The two cylinders are separated by a lead cylinder⁶⁶ ($\rho = 11.34 \text{ g cm}^{-3}$). Refer to Table 2.3 for the tabulated line spectrum used to initialize photons in the ^{103}Pd seeds^{69, 70}.

Table 2.3: Spectrum used for ^{103}Pd brachytherapy seeds taken from NNDC 2.6^{70, 69}.

Energy (MeV)	Intensity (%)
0.002700	0.0873
0.020074	0.2240
0.020216	0.4250
0.022699	0.0354
0.022724	0.0685
0.023172	0.0164
0.039748	6.83E−4
0.053290	3.00E−7
0.062410	1.04E−5
0.241880	5.00E−9
0.294980	2.80E−5
0.357450	2.21E−4
0.443790	1.50E−7
0.497080	3.96E−5

2.1.2 Beta plaque geometries

The $^{106}\text{Ru}/^{106}\text{Rh}$ beta plaque geometries are all bowl-shaped plaques made of silver. A very thin layer ($\sim 0.1\mu\text{m}$) of the active source coats the plaque under a 0.1 mm silver window. The plaque itself is modelled in egs++ by using a cone of the appropriate opening angle to cut it out from a silver shell ($\rho = 10.5 \text{ g cm}^{-3}$). The angles

Figure 2.6: Geometry of brachytherapy seeds used in this work. Subfigure(a) shows Amershan OncoSeed 6711 ^{125}I seed; blue is titanium casing and red is silver rod coated in radioisotope. Subfigure (b) shows Theragenics Co. TheraSeed 200 ^{103}Pd seed; orange is the titanium casing, grey is lead cylinder, red are graphite cylinders coated in radioisotope.

used for each plaque type as well as their radii and other geometric information can be found in Table 2.4. Figure 2.7 is a schematic diagram that shows how the plaques are constructed given the information in Table 2.4, while Figure 2.2 gives a visual representation of the CCB plaque in a patient model.

For the beta plaques the source is modelled using a radionuclide source model (known as the EGS_RadionuclideSource). The radionuclide model uses Evaluated Nuclear Structure Data Files (ENSDFs) from the Laboratoire National Henri Becquerel (LNHB) to model the full decay spectrum of a nuclide.

Figure 2.7: Schematic diagram showing geometry parameters for BEBIG beta plaques (not to scale). R_i and R_o are the radii of curvature for the inner and outer plaque; D_a is the diameter of the active source; D_o is the diameter of the full plaque; h is the height of plaque; and the opening angle θ is the angle used to cut the plaque from a spherical shell.

Table 2.4: Geometric dimensions for all beta plaques. R_i and R_o are the radii of curvatures for the inner and outer plaque; r_i is the radius of the active source; r_o is the radius of the full plaque; h is the height of plaque; and the opening angle θ is the angle used to cut the plaque from a spherical shell.

Plaque type	R_i (cm)	R_o (cm)	D_a (cm)	D_o (cm)	h (cm)	θ ($^\circ$)
CCA	1.20	1.30	1.30	1.53	0.33	36.04
CCB	1.20	1.30	1.78	2.02	0.54	50.82
CCC	1.30	1.40	2.25	2.48	0.80	62.34
CCD	1.20	1.30	1.55	1.79	0.43	43.51
CGD	1.30	1.40	1.99	2.23	0.61	52.79
CCX-Y-Z	1.20	1.30	0.95	1.16	0.23	26.49
CXS	1.20	1.30	0.77	1.16	0.23	26.49

Two versions of beta plaque with cutout notch

Beta plaques that contain a cutout notch are more complex and require a slightly different approach. The base plaque is designed in the same way as for the symmetric plaques (e.g. the plaque presented here, the CIA plaque, has the same base dimensions as the CCA symmetric plaque seen in Table 2.1). Two different methods are implemented to create the cutout.

The first emulates the work of Hermida-Lopez⁴⁷ (with geometrical values gained from personal communication, Hermida-Lopez 2018) and uses a bounding cylinder geometry to fill the cutout portion of the plaque with water. The cylinder is of radius 0.7 cm and located at a midpoint of (0.915, 0, -1.2) cm in eye centre coordinates. The sharp corners of the plaque left by the cylinder are then cutoff with a clipping plane at 0.43 cm on the x-axis. This results in the geometry of Figure 2.8(a). This version of the CIA plaque will henceforth be referred to as the CIA-HL model in this thesis.

The second method uses the `egs++` geometry package to round the ‘corners’ rather than clipping them off. This is done by splitting the plaque into two regions at 0.32 cm along the x-axis. One region, which contains the majority of the plaque, uses the same bounding cylinder from the first method to create a notch. The other smaller region uses two small cylinders to cut off the area around the ‘corners’. These cylinders are 0.175 cm in radius and are located at (0.32, 0.55, -1.2) and (0.32, -0.55, -1.2) cm in eye centre coordinates. The resultant model can be seen in Figure 2.8(b). The second version will henceforth be referred to as the CIA model.

Figure 2.8: (a) CIA-HL plaque model with points of notch clipped off; (b) CIA plaque model with points of notch rounded out.

2.2 Phantom geometry

All simulations in this work are done within a ‘phantom,’ which is a representation of the patient created by assigning media to different regions of a geometry in which radiation will be transported. For the simulations in this work the phantom is either a $(30 \times 30 \times 30)$ cm³ cube of water centered at $(0, 0, 0)$ cm or a more realistic representation of the eye detailed in Section 2.2.1. The phantom contains a dose scoring grid of voxels (volume elements) where the dose is calculated, the region outside of this scoring region is made much larger than the scoring region to allow scattered particles to return to the scoring region. The dimensions and locations within the eye coordinate system of scoring regions used in this work can be seen in Table 2.5. All water in this work is given $\rho = 0.998$ g cm⁻³ as recommended in the TG43 protocol³⁴.

Table 2.5: Dimensions of voxelized scoring regions for simulations scoring in water (D_{TG43} and $D_{w,w}$).

Source type	$x_{min}(cm)$	$x_{max}(cm)$	$y_{min}(cm)$	$y_{max}(cm)$	$z_{min}(cm)$	$z_{max}(cm)$	voxel vol. (cm^3)
Photon	-1.275	1.275	-1.275	1.275	-0.075	2.475	0.000125
Beta	-1.575	1.575	-1.575	1.575	0.025	3.175	0.000125

2.2.1 Voxelized eye phantom model

Lesperance et al.⁵³ created a voxelized eye phantom model for use with BrachyDose in 2014. The model was converted to the .egsphnt format, which `egs_brachy` can utilize for more accurate dose calculations. The composition and dimensions of structures within the eye vary based on several factors such as age and existing eye conditions, so the eye phantom uses averaged dimensions and compositions for ocular structures to create a realistic model of a human eye (specifically the right eye). The eye model is inserted into an ICRP reference computational phantom of a human head⁷¹ and a coordinate system is created such that the eye centre is at the origin (0, 0, 0) cm. The structures modelled include the sclera, the cornea, the iris, the lens, the aqueous humor, the vitreous humor, the optic nerve, and a nasal uveal tumor.

The sclera is modelled as a sphere that has an outer diameter of 24 mm^{72,73,74}. The inner diameter varies as an ellipsoid where the sclera has a thickness of 0.3 mm on the plane between anterior and posterior halves of the eye and a thickness of 1 mm at the anterior and posterior poles⁷². The cornea is the region between two spheres of radii 7.8 mm and 6.5 mm⁷². The outer cornea intersects with the outer sclera to form a circle with radius 5.85 mm⁷². The thickness of the cornea is 0.52 mm where both the inner cornea and outer cornea intersect with the x-axis⁷². The iris is the region between two cylinders of diameters 5.0 and 11.7 mm⁷³. These cylinders

are cut off by two planes to give the iris a thickness of 0.6 mm in the x direction⁷³. The anterior surface of the iris rests at the intersection between the cornea and the inner sclera. The lens region is formed from two spheres of radii 6 mm and 10 mm for the posterior and anterior surfaces respectively⁷². The intersection of the two spheres gives the lens a diameter of 9.5 mm in the yz plane at its widest point⁷². The aqueous humor is contained within the region between the inner cornea and the anterior of the lens. The vitreous humor takes up the rest of the eye within the inner sclera. The optic nerve is modelled as three cylindrical segments that increase from 1.5 mm in diameter at the optic disc to 2.5 mm in mid-sclera, and 3.5 mm at the outer sclera⁷². The length of the optic nerve included in the model is 1.1 cm. Finally, a tumor of medium size (with a maximum tumor height of 5 mm from the inner sclera in the z direction) is modelled against the inner sclera, along the nasal side of the eye. The tumor is modelled as a segment of a sphere with a base diameter of 12 mm in the y direction.

Scleral and corneal tissue compositions are taken from Meek⁷⁵, and their density was taken to be that of ‘muscle skeletal 2’ from Woodard and White⁷⁶ since the actual densities were not available in literature. The composition and density for the iris are also from the ‘muscle skeletal 2’ medium in Woodard and White⁷⁶ as the iris is mostly composed of muscle⁷⁷. The optic nerve, once again lacking literature data, was taken to be ‘brain white matter’ from Woodard and White⁷⁶. Vitreous density and compositions are taken from Tumram⁷⁸, aqueous from Berman⁷⁹, and lens from the ICRU⁸⁰. The tumor was assigned the density and composition of ‘melanoma 1’ from Maughan et al⁸¹. The compositions and densities used for the eye phantom are listed in Table 2.6.

Table 2.6: Mass fraction and density for the elemental composition of eye structures used in the eye phantom⁵³.

Eye structure	Elemental mass fraction									Density (g/cm ³)
	H	C	N	O	Na	P	S	Cl	K	
Sclera	0.0970	0.1696	0.0499	0.6831	-	-	0.0003	-	-	1.050
Cornea	0.1016	0.1199	0.0364	0.7411	-	-	0.0009	-	-	1.050
Iris	0.1020	0.1430	0.0340	0.7100	0.0010	0.0020	0.0030	0.0010	0.0040	1.050
Lens	0.0960	0.1950	0.0570	0.6460	0.0010	0.0010	0.0030	0.0010	-	1.070
Aqueous	0.1110	0.0010	-	0.8810	0.0030	-	-	0.0040	-	1.0035
Vitreous	0.1109	-	-	0.8804	0.0038	-	-	0.0045	0.0003	1.0071
Opt.nrv.	0.1060	0.1940	0.0250	0.6610	0.0020	0.0040	0.0020	0.0030	0.0030	1.040
Tumor	0.0940	0.2120	0.0560	0.6150	0.0025	0.0051	0.0064	0.0039	0.0051	1.040

2.3 Dose calculations

For the photon plaque simulations the default tracklength estimator scoring method⁸² (explained below) of `egs_brachy` is used. As discussed in Section 1.4.1 this method first assumes that for the purposes of low energy photons any charged particles generated will have a very short range compared to the length scales of interest. The short range of the charged particles allows the assumption of local energy deposition. Taking that assumption into account charged particle equilibrium can also be assumed. This allows for absorbed dose to be approximated as collision kerma. The tracklength dose equation differs from the generalized collision kerma equation discussed previously and is given as,

$$D^j = K_{col}^j = \frac{\sum_i E_i t_i \left(\frac{\mu_{en}}{\rho} \right)_i}{V_j} \quad (2.1)$$

where D^j is the dose to voxel j , K_{col}^j is the collision kerma to the same voxel; E_i is the energy of photon i ; $\left(\frac{\mu_{en}}{\rho} \right)_i$ is the mass-energy absorption coefficient for that energy; t_i is the length of the i^{th} photon's path through the voxel; and V_j is the volume of the voxel⁸³.

For beta plaques the main particles of interest are electrons, collision kerma is quantity for non-ionizing particles, so for ionizing particles like electrons the equality of dose and collision kerma cannot be made and absorbed dose must be calculated. In `egs_brachy` a high enough electron cutoff energy is needed to allow for electron transport and therefore interaction scoring.

For both photon and beta dose calculations, statistical uncertainties are calcu-

lated using history-by-history (where a history is one of the initial, statistically independent particles) statistics^{83,84} to avoid difficulties that arise in uncertainty calculation for batch submitted simulations. The formula for history-by-history statistics is^{83,84},

$$S_{\bar{D}} = \sqrt{\frac{1}{N-1} \left(\frac{\sum_{i=1}^N D_i^2}{N} - \left(\frac{\sum_{i=1}^N D_i}{N} \right)^2 \right)} \quad (2.2)$$

where $S_{\bar{D}}$ is the uncertainty for the mean dose D over all batches; N is the number of histories; and D_i is the quantity of dose in the i^{th} batch. All simulations in this work were performed using 10^{10} histories.

Near to the plaque there can be voxels that contain both the medium of the plaque and the medium of the scoring region, which could result in improper scoring in that region. This can be dealt with using a volume correction. In `egs_brachy` the volume correction is a simple Monte Carlo calculation that determines how much of the volume of a voxel with media intersection is actually phantom. This volume information is used to correct the volume where dose will be scored. The main input for the volume correction is the density of points, which gives the number of points per cubic centimeter which will be checked during the MC calculation. Volume corrections for simulations in this work use a density of points equal to 10^8 per cm^3 .

2.3.1 Simulation parameters

Several parameters are needed for `egs_brachy` to calculate dose. By default, `egs_brachy` utilizes a dataset file (`brachy_xcom_1.5MeV.muendat`) of mass-energy absorption coefficients for various media at different energies, as well as a file that contains material compositions and densities (`material.dat`). The photon cross sections used to generate the mass-energy absorption coefficients for `egs_brachy` are from the XCOM

photon cross section database⁸⁵. Radiation transport cutoffs for photon plaques and beta plaques are given in Table 2.7.

Table 2.7: Transport parameters for photon and beta plaque simulations, where Global ECUT determines an energy below which electrons are not simulated and Global PCUT does the same for photons. Source ECUT determines an energy below which electrons are not generated by the source, but does not stop secondary particles from being simulated below that energy, and Source PCUT is the same for photons.

Source Type	Global ECUT	Global PCUT	Source ECUT	Source PCUT
Photon	1.512 MeV	0.001 MeV	1.512 MeV	0.001 MeV
Beta	0.522 MeV	0.010 MeV	0.522 MeV	0.010 MeV

2.3.2 Dose normalization

egs_brachy outputs dose in the form of dose per history, where the number of histories is the number of initial, statistically independent particles. To normalize this output to absolute dose several parameters must be taken into account. For a low dose rate photon source like those used in this work the formula for egs_brachy to convert from dose per history to absolute dose is as follows⁸³,

$$D_{tot}^j = [S_k]_{max} \frac{\bar{D}^j}{S_K^{hist}} \tau (1 - e^{-\lambda t_{tot}}) \quad (2.3)$$

where D_{tot}^j is the absolute dose to voxel j ; $[S_k]_{max}$ is the initial maximum air-kerma strength of the chosen brachytherapy sources; $\bar{D}^j$ is the dose per history output by egs_brachy; S_K^{hist} is the air-kerma strength per history for the sources; τ is the mean lifetime of the sources; λ is the decay constant of the sources; and t_{tot} is the total treatment time for the procedure. The values used for the ^{125}I and ^{103}Pd sources in this work can be found in Table 2.8.

Table 2.8: Parameters used for calculation of absolute dose with ^{125}I and ^{103}Pd seeds. The unit of air kerma strength U is given by $[\frac{\text{cGy cm}^2}{\text{h}}]$.

Source	$[S_k]_{max} (U)$	$S_K^{hist} (E^{-14} \frac{U}{hist})$	$\tau (d)$	$\lambda (d^{-1})$	$t_{tot} (h)$
^{125}I	4.572	3.769	41.17	0.01167	100
^{103}Pd	3.879	6.428	11.78	0.0408	168

However, not all results in literature are presented in terms of absolute dose and so a different normalization may be used. Many of the photon plaque calculations in this work are scaled such that D_{TG43} is 85 Gy at the prescription dose point of the tumor apex (0, 0, 0.5) cm. This was done to compare to the MCNP5 calculations of Melhus and Rivard²⁴.

For beta plaques, when comparing to most of the previous calculations (eg. those using PENELOPE⁸⁶) normalization is done to 100% at (0, 0, 0.1) cm. For comparisons to the calculations from Sommer’s GEANT4 paper⁴⁹ the dose is instead normalized to Gy per decay. This is done by dividing dose per history by the particle fluence per history.

When comparing the photon and beta plaque results to each other $D_{w,w}$ is scaled to 100% at the tumor apex ((0.25, 0.25, -0.675) cm in eye centre coordinates), $D_{m,m}$ is then scaled using the same value.

2.4 Calculation specifications

The dose calculations in this work include three possible model configurations. First, simulations are done for photon plaques which are meant to emulate TG43 (referred to as D_{TG43}). D_{TG43} calculations are done with seeds in a water phantom (see Table 2.5) and no plaque model is included. The dose is calculated for each seed, one at a time and then summed through superposition to avoid interseed effects (attenuation of photons emitted by one seed caused by another seed). `egs_brachy`

has a superposition run mode that makes this a simple task. Second, dose calculations which take into account the plaque geometry and involve radiation transported and scored in water are represented by $D_{w,w}$ ⁸⁷. For photon plaques these simulations calculate dose with seeds inserted into a model of the plaque as described in Section 2.1.1. The plaque is in turn inserted into a water phantom. For beta plaques these simulations calculate dose with the full silver plaque model and source film in a similar water phantom (see Table 2.5). Both D_{TG43} and $D_{w,w}$ calculations set the origin at the eye plaque coordinate origin of the inner sclera. Last, dose calculations using the anatomically realistic voxelized eye phantom from Lesperance et al⁵² where radiation travels through eye media and is scored in eye media are referred to as $D_{m,m}$. $D_{m,m}$ calculations are done with both photon and beta plaques as well and the origin set at the eye centre.

All calculations in this thesis were done using egs_brachy version v2017.09.15 on the Carleton physics research server cluster of 644 cores. Each simulation was run with 10^{10} histories. For photon plaques the calculations of D_{TG43} and $D_{w,w}$ took between 21 and 35 CPU hours, while calculations of $D_{m,m}$ can take as many as 50 CPU hours. For beta plaques calculations take much longer due to the need for interaction scoring. $D_{w,w}$ can take up to 720 CPU hours and $D_{m,m}$ 750 CPU hours.

Dose results are extracted from along the central axis for each of these calculations. In addition to central axis doses, values for four points of clinical interest in the eye at eight different eye plaque positions are extracted. The points of interest include the fovea, optic disc, lens, and lacrimal gland which are located at (-1.13, 0, 0) cm, (-1.06, 0.4, 0) cm, (0.77, 0, 0) cm, and (0.77, 0.82, 0.82) cm respectively, where the coordinates are with respect to the eye centre. The eight plaque positions are:

#1: on the temporal side of the eye, centered on the equator

- #2: on the nasal side of the eye, centered on the equator
- #3: on the superior side of the eye, centered on the equator
- #4: on the inferior side of the eye, centered on the equator
- #5: on the temporal side of the eye, posterior from the equator
- #6: on the nasal side of the eye, posterior from the equator
- #7: on the superior side of the eye, posterior from the equator
- #8: on the inferior side of the eye, posterior from the equator

For beta plaques, dose profiles along the lateral axis at different depths with respect to the eye plaque (0.1 cm, 0.2 cm, 0.3 cm, 0.4 cm, 0.6 cm, and 0.8 cm) are extracted.

Comparisons ΔD_{LOCAL} and ΔD_{GLOBAL} ^{88,89} are used as an overall comparison metric, where $\Delta D_{LOCAL} = \frac{D_A(r) - D_B(r)}{D_B(r)}$ and $\Delta D_{GLOBAL} = \frac{D_A(r) - D_B(r)}{D_B(r_{ref})}$ and r_{ref} is given by dose at the reference tumor apex point (0, 0, 0.5) cm within the eye plaque coordinate system. These results are given for both photon seed types.

2.4.1 Uncertainties

Statistical uncertainties

For the purposes of discussing statistical uncertainties it should be stated that the tumor apex is located at $z = 0.5$ cm, and the eye centre and opposite side of the eye are at $z = 1.23$ cm and $z = 2.46$ cm respectively. The statistical uncertainties for the `egs_brachy` photon plaque dose calculations presented here are less than 0.1% at the tumor apex and up to 0.4% at the opposite side of the eye. For beta plaques the statistical uncertainties are less than 0.3% within the tumor and 1% up to the eye centre. Across the lateral points the largest uncertainties are 2.4%, but the majority are below 1%. The `BrachyDose` calculations from the CLRP eye

plaque database have statistical uncertainties similar to those of the `egs_brachy` calculations, specifically 0.1% within the tumor and 0.5% up to the opposite side of the eye⁴³. The BrachyDose dose calculations in the Rivard 2011 paper are reported with statistical uncertainties up to 0.1% within the tumor and 0.5% at the opposite side of the eye⁴². The MCNP5 dose calculations are reported to have statistical uncertainties up to 0.9% within the tumor and 2.0% at the opposite side of the eye^{24, 42}. The PENELOPE dose calculation statistical uncertainties are reported as generally less than 2%⁴⁷. The GEANT4 dose calculation statistical uncertainties are reported as 0.12% within the tumor and 0.4% up to the eye centre⁴⁹.

Other uncertainties

Sources of uncertainty other than statistical uncertainties are important to consider and in fact likely account for a greater uncertainty in the calculations than statistics. The TG43U1³⁴ update includes an uncertainty analysis for Monte Carlo brachytherapy calculations that concludes a 2% uncertainty in both photon cross sections and seed geometries near the seed. There are uncertainties to consider in the plaque models as well.

The suture eyelets on the plaques have not been modelled, though they are small and out of the way and so unlikely to have a large effect on dose. For photon plaques the seeds are modelled to be completely enveloped in the insert material. In reality the seeds are slotted into the insert, so small air pockets would exist. The radionuclide coated film that is the source for beta plaques is not geometrically modelled. The beta plaques are modelled as a single piece. The film is on the order of 0.1 μm thick and its exclusion is unlikely to have large effects. It is also assumed that the beta plaque sources have a uniform activity distribution, which BEBIG certificates show may not be the case.

Chapter 3

Results

In this chapter, doses calculated in `egs_brachy` are presented for both the photon and beta plaques. These doses are shown alongside MC calculated doses from previous studies using MCNP5^{24,42}, BrachyDose^{42,43}, PENELOPE⁴⁷ and GEANT4⁴⁹. For photon plaques most dose data are presented using both ¹²⁵I and ¹⁰³Pd seeds and include calculations of D_{TG43} , $D_{w,w}$, and $D_{m,m}$. For beta plaques dose calculations are made with the source nuclide ¹⁰⁶Ru and both $D_{w,w}$, and $D_{m,m}$ are presented. Various graphs and tables comparing `egs_brachy` photon and beta plaque results conclude the chapter. In some places, two doses (D_a , D_b) are compared via their relative difference, defined as $|1 - D_a/D_b| \times 100\%$ (e.g. $D_a = 76$ Gy and $D_b = 85$ Gy, $|1 - 76/85| \times 100\% = 10.6\%$).

3.1 Photon plaque results

D_{TG43} depth-dose curves along the plaque central axis are given for COMS 10 mm to COMS 22 mm (and with both 6711 ¹²⁵I seeds and T200 ¹⁰³Pd seeds) in the subfigures of Figures 3.1 and 3.2. The doses in the majority of these figures are scaled to 85 Gy at the reference tumor apex ((0, 0, 0.5) cm in the eye plaque coordinate system) to match the common prescription dose used with the TG43 protocol. The doses in the subfigures (a) of Figures 3.1 and 3.2 are calculated to be absolute doses

and are done with a COMS 16 mm plaque (see Section 2.3 for the parameters used to determine absolute dose for the two different seed types). The depth range of the prescription dose scaled figures is limited by the amount of data presented for MCNP5²⁴ and only reach the eye centre. The absolute dose calculations in Rivard et al. have points at further depths and therefore allow for a more complete depth-dose curve from the inner sclera to the opposite sclera⁴².

For D_{TG43} (Figures 3.1 and 3.2), when using ^{125}I seeds, the relative difference between `egs_brachy` and MCNP5 at depths up to the centre of the eye is less than 0.8% for the best agreement (COMS 10 mm) and up to 2.1% for the worst agreement (COMS 22 mm). The agreement tends to get better as plaque size decreases. When using ^{103}Pd seeds the relative difference between `egs_brachy` and MCNP5 at depths up to the centre of the eye is less than 1.9% for the best agreement (COMS 22 mm) and up to 3.3% for the worst agreement (COMS 20 mm). There does not appear to be a trend between plaque size and agreement when using the ^{103}Pd seeds. The largest statistical uncertainties for the `egs_brachy` dose results are 0.14% for ^{125}I and 0.17% for ^{103}Pd .

For the absolute D_{TG43} depth-dose curves (subfigure (a) of both Figures 3.1 and 3.2) the relative differences between `egs_brachy` and BrachyDose calculated doses using ^{125}I are less than 0.5% up to the tumor apex and 0.7% up to the opposite side of the eye. Between `egs_brachy` and MCNP5, 1.4% is the largest relative difference for both tumor apex and opposite side of the eye. When comparing absolute doses calculated with ^{103}Pd , `egs_brachy` and BrachyDose disagree by only 0.5% up to the tumor apex and 0.6% up to the opposite side of the eye. `egs_brachy` and MCNP5 disagree by 2.2% and 3.4% for up to tumor apex and the opposite side of the eye respectively.

The absolute doses for D_{TG43} (at the points of interest discussed in Section 2.4) given in Table 3.1 and show a relative difference between `egs_brachy` and `BrachyDose` of less than 1% for all points and for both seed types. For ^{125}I the relative difference between `egs_brachy` and `MCNP5` are generally within 2% and there are only a few outliers. For ^{103}Pd these outliers have a relative difference of between 2% and 9%.

Figure 3.1: Central axis depth-dose curves for D_{TG43} calculations using ^{125}I seeds. Dose curve (a) is normalized to absolute dose (see 2.3.2) and dose curves (b) to (h) are scaled to 85 Gy at (0, 0.5) cm. Uncertainties are too small to be plotted.

Figure 3.2: Central axis depth-dose curves for D_{TG43} calculations using ^{103}Pd seeds. Dose curve (a) is normalized to absolute dose (see 2.3.2) and dose curves (b) to (h) are scaled to 85 Gy at (0, 0.5) cm. Uncertainties are too small to be plotted.

Table 3.1: D_{TG43} doses for egs_brachy at different plaque locations and points of interest. Ratios are given between egs_brachy and both BrachyDose and MCNP5. Both ^{125}I and ^{103}Pd results are shown.

plq. pos.	point of interest	^{125}I			^{103}Pd		
		D_{eb} (Gy)	$\left(\frac{D_{\text{eb}}}{D_{\text{BD}}}\right)$	$\left(\frac{D_{\text{eb}}}{D_{\text{MCNP5}}}\right)$	D_{eb} (Gy)	$\left(\frac{D_{\text{eb}}}{D_{\text{BD}}}\right)$	$\left(\frac{D_{\text{eb}}}{D_{\text{MCNP5}}}\right)$
#1	fovea	16.4	1.00	1.01	10.9	1.00	1.04
	opt. dsc.	11.3	1.00	0.99	6.8	1.00	1.02
	lens	21.7	1.00	1.01	15.6	1.00	1.06
	lac. gld.	39.8	1.00	1.03	30.7	1.00	1.04
#2	fovea	16.4	1.00	1.01	10.9	1.00	1.04
	opt. dsc.	27.6	1.00	1.01	20.7	1.00	1.04
	lens	21.7	1.00	1.01	15.6	1.00	1.06
	lac. gld.	7.2	1.00	1.04	3.9	1.01	1.04
#3	fovea	16.4	1.00	1.01	10.9	1.00	1.04
	opt. dsc.	16.5	1.00	1.02	10.9	1.00	1.05
	lens	21.7	1.00	1.01	15.6	1.00	1.06
	lac. gld.	39.8	1.00	1.03	30.8	1.00	1.04
#4	fovea	16.4	1.00	1.01	10.9	1.00	1.04
	opt. dsc.	16.4	1.00	1.02	10.9	1.00	1.04
	lens	21.7	1.00	1.01	15.6	1.00	1.06
	lac. gld.	7.3	1.01	1.04	3.9	1.00	1.03
#5	fovea	73.6	0.99	1.01	66.5	0.99	1.09
	opt. dsc.	31.0	1.00	1.01	23.9	1.00	1.06
	lens	12.3	1.00	0.99	7.7	1.00	1.04
	lac. gld.	13.5	1.00	1.03	8.4	1.00	1.03
#6	fovea	27.1	1.00	1.00	20.2	1.00	1.04
	opt. dsc.	59.2	1.00	1.01	51.5	1.00	1.07
	lens	15.9	0.99	0.99	10.7	1.00	1.02
	lac. gld.	6.4	1.00	1.02	3.3	1.01	1.12
#7	fovea	73.6	0.99	1.01	66.5	0.99	1.09
	opt. dsc.	60.2	1.00	1.00	51.1	1.00	1.02
	lens	12.3	1.00	0.99	7.7	1.00	1.04
	lac. gld.	13.5	1.00	1.01	8.4	0.99	1.06
#8	fovea	73.6	0.99	1.01	66.5	0.99	1.09
	opt. dsc.	60.3	1.00	0.98	51.1	1.00	1.02
	lens	12.3	1.00	0.99	7.7	1.00	1.04
	lac. gld.	6.1	1.00	1.02	3.0	0.99	1.04

The depth-dose curves for simulations of $D_{w,w}$ using both ^{125}I and ^{103}Pd are given in Figures 3.3 and 3.4. The majority of these central axis sub-figures are scaled using the value needed for D_{TG43} to be 85 Gy at the prescription point (0, 0, 0.5) cm; however, once again, sub-figures (a) of Figures 3.3 and 3.4 are absolute dose calculations (see Section 2.3 for the parameters used to determine absolute dose for the two seed types). The egs_brachy doses and ratios to other MC codes are given in Table 3.2 for the different points of interest discussed in Section 2.4.

For $D_{w,w}$ using ^{125}I seeds, the largest disagreement between egs_brachy and BrachyDose at depths up to the tumor apex is less than 0.3% to 0.6% for all plaque models, and up to the eye centre is also less than 0.3% to 0.6% for all plaque models. The best agreement is with the COMS 12 mm plaque and the worst with COMS 22 mm plaque, but there does not appear to be a trend relating to plaque size. When using ^{103}Pd seeds the relative difference between egs_brachy and BrachyDose at depths up to the tumor apex ranges from less than 0.3% to 0.8% for all plaque models, and at depths up to the eye centre is less than between 0.4% and 1.0% across all plaques. In this case, the best agreement is with the COMS 10 mm plaque and the worst is the COMS 20 mm plaque. It should be noted that very few central axis doses have greater than 0.001% relative difference between egs_brachy and BrachyDose when using ^{103}Pd seeds.

When comparing $D_{w,w}$ with MCNP5 using ^{125}I seeds, relative differences are at most 0.5% to 1.8% up to the tumor apex and 0.7% to 2.4% up to the eye centre across all plaque models. The best agreement exists with the COMS 10 mm and the worst with the COMS 22 mm. Here the differences do appear to increase with the plaque size. With ^{103}Pd seeds dose relative differences range between 0.9% and 1.9% up to the tumor apex and between 2.0% and 3.4% up to the eye centre, once again for all plaque models. The worst agreement is with the COMS 14 mm and the best

the COMS 16 mm.

For the absolute $D_{w,w}$ depth-dose curves (subfigure (a) of both Figures 3.3 and 3.4) the relative differences between doses calculated with egs_brachy and BrachyDose using ^{125}I are less 0.3% up to the tumor apex and 0.9% up to the opposite side of the eye. Between egs_brachy and MCNP5 the largest relative difference is 2.4% for both tumor apex and opposite side of the eye. When comparing absolute doses calculated with ^{103}Pd , egs_brachy and BrachyDose disagree by 0.5% up to the tumor apex and 1.4% up to the opposite side of the eye. egs_brachy and MCNP5 disagree by 2.2% and 3.4% up to tumor apex and up to the opposite side of the eye respectively.

For both D_{TG43} and $D_{w,w}$ agreement between egs_brachy and MCNP5 is comparable to agreement between BrachyDose and MCNP5. In some cases the agreement seems to be related to plaque size, but not often enough to be considered a strong trend. Agreement also tends to be stronger with ^{125}I seeds over ^{103}Pd seeds. Larger disagreements are commonly at dose locations that are at a further depths and/or are in the penumbra region of the plaque. This is shown in the dose ratios at points of interest in Table 3.2 (e.g. the lacrimal gland for positions #1 and #3 of the plaque).

Figure 3.3: Central axis depth-dose curves for $D_{w,w}$ calculations using ^{125}I seeds. Dose curve (a) is normalized to absolute dose (see 2.3.2) and dose curves (b) to (h) are scaled such that a D_{TG43} equivalent would be 85 Gy at $(0, 0, 0.5)$ cm. Uncertainties are too small to see.

Figure 3.4: Central axis depth-dose curves for $D_{w,w}$ calculations using ^{103}Pd seeds. Dose curve (a) is normalized to absolute dose (see 2.3.2) and dose curves (b) to (h) are scaled such that a D_{TG43} equivalent would be 85 Gy at $(0, 0, 0.5)$ cm. Uncertainties are too small to see.

Table 3.2: $D_{w,w}$ doses for eggs_brachy at different plaque locations and points of interest. Ratios are given between eggs_brachy and both BrachyDose and MCNP5. Both ^{125}I and ^{103}Pd results are shown.

plq. pos.	point of interest	^{125}I			^{103}Pd		
		D_{eb} (Gy)	$\left(\frac{D_{\text{eb}}}{D_{\text{BD}}}\right)$	$\left(\frac{D_{\text{eb}}}{D_{\text{MCNP5}}}\right)$	D_{eb} (Gy)	$\left(\frac{D_{\text{eb}}}{D_{\text{BD}}}\right)$	$\left(\frac{D_{\text{eb}}}{D_{\text{MCNP5}}}\right)$
#1	fovea	12.7	0.99	0.98	7.7	1.00	1.03
	opt. dsc.	9.0	1.00	0.99	5.0	0.99	0.99
	lens	17.9	1.00	0.99	12.0	1.01	1.06
	lac. gld.	6.6	0.98	1.21	3.7	0.98	1.27
#2	fovea	12.7	0.99	0.98	7.7	1.00	1.03
	opt. dsc.	20.7	1.00	0.97	13.5	1.00	1.02
	lens	17.9	1.00	0.99	12.0	1.01	1.06
	lac. gld.	5.6	0.99	0.99	2.9	1.01	1.04
#3	fovea	12.7	0.99	0.98	7.7	1.00	1.03
	opt. dsc.	12.8	1.00	1.00	7.7	0.99	1.01
	lens	17.9	1.00	0.99	12.0	1.01	1.06
	lac. gld.	6.5	0.96	1.18	3.7	0.98	1.26
#4	fovea	12.7	0.99	0.98	7.7	1.00	1.03
	opt. dsc.	12.8	1.00	1.00	7.8	1.00	1.02
	lens	17.9	1.00	0.99	12.0	1.01	1.06
	lac. gld.	5.7	1.00	1.00	2.9	1.00	1.03
#5	fovea	58.7	1.00	1.00	45.3	1.00	1.09
	opt. dsc.	23.0	1.00	0.97	15.1	1.00	1.01
	lens	10.1	1.00	0.97	6.0	1.01	1.00
	lac. gld.	9.1	1.00	1.02	4.8	1.00	1.02
#6	fovea	20.4	1.00	0.96	13.2	1.00	1.01
	opt. dsc.	45.7	1.00	0.99	33.8	1.00	1.05
	lens	13.2	1.01	0.99	8.3	1.00	1.01
	lac. gld.	5.1	1.00	1.00	2.6	1.01	1.10
#7	fovea	58.7	1.00	1.00	45.3	1.00	1.09
	opt. dsc.	47.1	1.00	0.97	33.8	1.00	1.00
	lens	10.1	1.00	0.97	6.0	1.01	1.00
	lac. gld.	9.1	1.00	0.98	4.9	1.00	1.03
#8	fovea	58.7	1.00	1.00	45.3	1.00	1.09
	opt. dsc.	47.0	1.00	0.97	33.7	1.00	1.01
	lens	10.1	1.00	0.97	6.0	1.01	1.00
	lac. gld.	4.9	1.00	1.00	2.4	1.00	1.04

Comparisons between `egs_brachy` and `BrachyDose` 3D dose distributions using a COMS 16 mm eye plaque for both ^{125}I and ^{103}Pd seeds are given in Figure 3.5. These comparisons are in the form of ΔD_{LOCAL} and ΔD_{GLOBAL} discussed in Section 2.4. For ^{125}I , ΔD_{LOCAL} is not strongly skewed but is shifted as to be centered at -0.6%, showing a small systematic difference between the two codes. ΔD_{GLOBAL} shows a skew, but is peaked at -0.1%. For ^{103}Pd there is a larger skew in ΔD_{LOCAL} than for ^{125}I , the peak is at $\sim 0.5\%$ showing a similar systematic difference but in the opposite direction. ΔD_{GLOBAL} shows a smaller skew than for ^{125}I , the natural peak seems to be at 0.02%, but a sharp single bin peak occurs at 0%. These comparisons show a small overall difference between the two codes that is discussed in Chapter 4.

Figure 3.5: A 3D $D_{w,w}$ comparison between `egs_brachy` and `BrachyDose`. Subfigures (a) and (b) were obtained using ^{125}I seeds. Subfigures (c) and (d) were obtained using ^{103}Pd seeds.

3.1.1 Comparison of D_{TG43} , $D_{w,w}$, and $D_{m,m}$

Figure 3.6 shows isodose contours for different photon plaque models, using COMS 16 mm as a reference for the COMS standard. Profiles of D_{TG43} , $D_{w,w}$, and $D_{m,m}$ are shown for each of the COMS 16 mm, “COMS - thin acrylic”, “short lip - acrylic”, “no lip - silastic”, and “stainless steel - acrylic” plaques. These isodose contours are scaled such that D_{TG43} is 85 Gy at the tumor apex ((0.25, 0.25, -0.675) cm in eye centre coordinates). It is shown that D_{TG43} is consistently higher than $D_{w,w}$ or $D_{m,m}$. In contrast, $D_{m,m}$ varies between being higher than $D_{w,w}$ and lower than $D_{w,w}$ depending on the medium being traversed, but is still generally lower than D_{TG43} .

Figure 3.7 gives plaque central axis dose ratios for $D_{w,w}$ and $D_{m,m}$ over D_{TG43} . For the purposes of comparison to Lesperance et al.⁵², the ratio of egs_brachy calculated $D_{w,w}$ over D_{TG43} is shown alongside their equivalent BrachyDose results. Relative differences in the ratios appear to be statistical and are between 0.4% and 0.8% up to the tumor apex and between 1.6% and 2.3% up to the opposite side of the eye. The smallest difference up to the tumor is for the “no lip - silastic” plaque and the largest differences are the COMS 16 mm and “stainless steel - acrylic”. Up to the opposite side of the eye the smallest difference is for the “short lip - acrylic” plaque and the largest is for the “COMS - thin acrylic” plaque.

Minimum, maximum, and average D_{TG43} , $D_{w,w}$, and $D_{m,m}$ in different eye structures for a COMS 16 mm plaque loaded with ^{125}I seeds are given in Table 3.3. This table also includes the ratios of $D_{w,w}$ and $D_{m,m}$ with D_{TG43} for those eye structures.

Figure 3.6: Isodose contours of D_{TG43} , $D_{w,w}$, and $D_{m,m}$ calculated using ^{125}I seeds: COMS 16 mm, “COMS - thin acrylic”, “short lip - acrylic”, “no lip - silastic”, and “stainless steel - acrylic”.

Figure 3.7: Dose ratios are given in (a) $D_{w,w} / D_{TG43}$, and (b) $D_{m,m} / D_{TG43}$ for five different plaque models fully loaded with ^{125}I seeds: COMS 16 mm, “COMS - thin acrylic”, “short lip - acrylic”, “no lip - silastic”, and “stainless steel - acrylic”. All plaques are loaded with ^{125}I seeds.

Table 3.3: D_{TG43} , $D_{w,w}$, and $D_{m,m}$ to structures in the eye (the lens, tumor, sclera, optic nerve, vitreous fluid, aqueous humor, iris, and cornea) using a COMS 16 mm plaque model loaded with ^{125}I seeds.

eye struct.	value	D_{TG43}	$D_{w,w}$	$D_{m,m}$	$\frac{D_{w,w}}{D_{TG43}}$	$\frac{D_{m,m}}{D_{TG43}}$
lens	Min	12.80	10.40	8.74	0.81	0.68
	Max	39.34	32.41	28.61	0.82	0.73
	Avg	22.45	18.42	16.01	0.82	0.71
tumor	Min	92.19	80.94	87.30	0.88	0.95
	Max	289.69	247.89	274.78	0.86	0.95
	Avg	169.40	147.03	160.96	0.87	0.95
sclera	Min	6.44	5.23	4.07	0.81	0.63
	Max	337.25	283.25	250.47	0.84	0.74
	Avg	38.37	30.34	26.25	0.79	0.68
opt. nrv.	Min	8.05	1.29	1.36	0.16	0.17
	Max	24.70	18.06	17.85	0.73	0.72
	Avg	13.86	6.17	6.18	0.45	0.45
vitreous	Min	6.74	5.49	5.21	0.81	0.77
	Max	180.22	155.01	165.55	0.86	0.92
	Avg	29.38	24.43	25.30	0.83	0.86
aqueous	Min	8.94	7.20	6.69	0.81	0.75
	Max	61.14	48.59	50.61	0.79	0.83
	Avg	20.08	15.78	15.74	0.79	0.78
iris	Min	9.82	7.88	7.20	0.80	0.73
	Max	41.24	32.17	32.45	0.78	0.79
	Avg	20.00	15.76	15.36	0.79	0.77
cornea	Min	9.77	7.74	6.15	0.79	0.63
	Max	31.15	19.64	17.58	0.63	0.56
	Avg	15.95	11.52	9.78	0.72	0.61

3.2 Beta plaque results

Doses calculated in `egs_brachy` using ^{106}Ru beta plaques are presented in several different forms. First, Figure 3.8 shows depth-dose curves along the plaque central axis. These curves are presented alongside doses calculated with PENELOPE by Hermida-Lopez⁴⁷ and are given for eight different plaque varieties (with geometries discussed in Section 2.1.2): CCA, CCB, CCC, CCD, CGD, CXS, CCX-Y-Z, and a simplified planar model of the CCB plaque. These depth-dose curves give $D_{w,w}$ and are scaled to 100% at the point (0, 0, 0.1) cm in the eye plaque coordinate system, the same scaling used in the PENELOPE paper⁴⁷. The relative dose difference between `egs_brachy` and PENELOPE at depths up to the tumor apex is less than 2.2% for the CXS plaque. All other plaques have relative dose differences less than 1.4%. The relative dose differences get substantially larger past the tumor where doses are low (less than 10 Gy) and reach up to 13%. The best agreement is with the CCB plaque, which has a relative dose difference of less than 4.5% up to 1 cm.

Next, Figure 3.9 is a central axis depth-dose curve for the CCB plaque given in units of pGy per decay. The pGy per decay depth-dose curve is presented alongside doses calculated by Sommer et al.⁴⁹ using a Monte Carlo code known as GEANT4 and a reference dataset from the BEBIG ^{106}Ru beta plaque user manual that is averaged over 174 measurements and has a combined uncertainty of 20%⁹⁰. Comparing pGy per decay for `egs_brachy` and GEANT4 there is at most a 0.7% disagreement up to the tumor apex. Up to 1 cm, the largest disagreement is 4.4%. Comparing `egs_brachy` doses to the BEBIG reference dataset shows a disagreement of up 7.5% at a depth of 1 cm.

For each plaque type, dose profiles of an axis lateral to the central axis are shown in Figures 3.10 and 3.11. These profiles are given at different depths along the z axis

ranging from 0.1 cm to 0.8 cm in the eye plaque coordinate system. The profiles are presented alongside PENELOPE results from the same work as the depth-dose curves⁴⁷ and are scaled to 100% at the same point (0, 0, 0.1) cm. Looking at the lateral profiles at different depths most relative dose differences are less than 3%. Higher relative differences are found at locations in the penumbra region of the plaque, at far depths, or where they intersect the plaque.

Figure 3.8: Central axis depth-dose curves for eight different beta plaque models. `egs_brachy` results are presented alongside PENELOPE results. Doses are relative to 100% at the point (0, 0, 0.1) cm. Uncertainties are too small to see.

Figure 3.9: A comparison of the central axis dose per decay with depth between egs_brachy, GEANT4, and an averaged dataset from BEBIG certificates. Uncertainties are too small to see.

Figure 3.10: Dose profiles at different depths comparing `eggs_brachy` and `PENELOPE`. Doses are relative to 100% at the point (0, 0, 0.1) cm. Uncertainties are too small to see.

Figure 3.11: Dose profiles at different depths comparing egs_brachy and PENELOPE. Doses are relative to 100% at the point (0, 0, 0.1) cm. Uncertainties are too small to see.

3.2.1 Comparison of CIA beta plaque models

Some of the beta plaques have a cutout notch to fit around structures of the eye. Two versions of the CIA beta plaque are simulated here, CIA-HL and CIA. For a definition of these two plaque geometries refer to Section 2.1.2. Figure 3.12 presents dose distribution for the two models alongside the PENELOPE doses for the same plaque calculated by Hermida-Lopez⁴⁷. For the majority of points both central axis depth-dose curve and x-axis profiles have very good agreement (within a disagreement of only 0.1%) between the CIA-HL and the CIA. There are a few points where they have a 1 – 3% disagreement and even fewer with quite large disagreement (20% or greater). All of these points are located at far depths or further out on the x-axis or both. The points of greatest disagreement are shown in Figure 3.13, where isodose contours of CIA-HL and CIA are given. The isodose contours in this figure show large discrepancies near the plaque on the positive x-axis. When comparing to PENELOPE, both versions of the egs_brachy CIA plaque result in doses in good agreement (2% or less disagreement) for most points, only at further depths and distances along the x-axis do the disagreements reach 3% – 6%.

Figure 3.12: Plots showing doses using different CIA models. Lines give the egs_brachy doses using the CIA-HL model, 'o's give the egs_brachy doses using the CIA model, and 'x's give the doses calculated with PENELOPE by Hermida-Lopez⁴⁷. (a) Shows the depth-dose curve along the plaque central axis. (b) Shows dose profiles along the x-axis (where the cutout notch is located) at different depths of the central axis. Uncertainties are too small to see.

Figure 3.13: isodose contours for the two models, CIA and CIA-HL. Solid lines are CIA, dashed lines are CIA-HL.

3.3 Photon and beta plaque comparisons

To compare a photon-emitting plaque to a beta-emitting plaque, the COMS 16 mm and CCB plaques were chosen for their similar size and as commonly-used and studied designs of the two plaques types. Subfigures (a) and (b) of Figure 3.14 show a comparison of plaque central axis depth-dose curves for $D_{w,w}$ and $D_{m,m}$ respectively. Figure 3.15 shows isodose contours of $D_{w,w}$ and $D_{m,m}$ for both the photon plaque and the beta plaque. Both Figures show generally higher doses near to the plaque for the beta plaque than for the photon plaque. Table 3.4 gives the dose $D_{w,w}$ at the points of interest laid out in Section 2.4 for both plaque types and shows much lower doses to the points of interest for the beta plaque in most cases. Table 3.5 gives minimum, maximum, and average doses to ocular structures within the eye for both plaque types and both $D_{w,w}$ and $D_{m,m}$, as well as a ratio between $D_{m,m}$ and $D_{w,w}$ for the beta plaque to highlight the differences caused by introducing the eye phantom.

Figure 3.14: Depth-dose curves for both the COMS 16 mm photon-emitting plaque and the CCB beta-emitting plaque, (a) gives $D_{w,w}$, and (b) $D_{m,m}$. Uncertainties are too small to see.

Figure 3.15: Isodose contours for comparison of the COMS 16 mm photon plaque and the CCB beta plaque. Subfigure (a) gives $D_{w,w}$ isodose contours for COMS 16 mm and CCB plaques, subfigure (b) gives $D_{m,m}$ isodose contours for COMS 16 mm and CCB plaques, and subfigure (c) gives $D_{w,w}$ and $D_{m,m}$ isodose contours for CCB beta plaque.

Table 3.4: $D_{w,w}$ calculated using a COMS 16 mm plaque loaded with ^{125}I seeds and a CCB plaque at different plaque locations and points of interest. $D_{w,w}$ is scaled to 100% at the tumor apex $(X, Y, Z) = (0.25, 0.25, -0.675)$ cm.

plq. pos.	point of interest	COMS 16 mm	CCB	$\frac{D_{CCB}}{D_{COMS}}$
#1	fovea	17.06	1.08	0.06
	opt. dsc.	12.11	0.19	0.02
	lens	24.08	2.99	0.12
	lac. gld.	8.9	3.35	0.38
#2	fovea	17.06	1.08	0.06
	opt. dsc.	27.92	9.81	0.35
	lens	24.08	2.99	0.12
	lac. gld.	7.59	0.11	0.01
#3	fovea	17.06	1.08	0.06
	opt. dsc.	17.19	1.06	0.06
	lens	24.08	2.99	0.12
	lac. gld.	8.7	3.28	0.38
#4	fovea	17.06	1.08	0.06
	opt. dsc.	17.23	1.07	0.06
	lens	24.08	2.99	0.12
	lac. gld.	7.68	0.11	0.01
#5	fovea	79.02	263.75	3.34
	opt. dsc.	31.01	14.25	0.46
	lens	13.65	0.18	0.01
	lac. gld.	12.25	0.4	0.03
#6	fovea	27.44	9.24	0.34
	opt. dsc.	61.42	132.38	2.16
	lens	17.84	0.54	0.03
	lac. gld.	6.9	0.11	0.02
#7	fovea	79.02	263.75	3.34
	opt. dsc.	63.25	149.44	2.36
	lens	13.65	0.18	0.01
	lac. gld.	12.27	0.43	0.04
#8	fovea	79.02	263.75	3.34
	opt. dsc.	63.27	149.57	2.36
	lens	13.65	0.18	0.01
	lac. gld.	6.56	0.1	0.02

Table 3.5: Minimum, maximum, and average $D_{w,w}$ and $D_{m,m}$ for different eye structures (the lens, tumor, sclera, optic nerve, vitreous fluid, aqueous humor, iris, and cornea) calculated using both a COMS 16 mm plaque loaded with ^{125}I seeds and a CCB beta plaque. $D_{w,w}$ is scaled to 100% at the tumor apex $(X, Y, Z) = (0.25, 0.25, -0.675)$ cm, $D_{m,m}$ is scaled using the same value.

eye struct.	value	$D_{w,w}$		$D_{m,m}$		$\left(\frac{D_{m,m}}{D_{w,w}}\right)$
		COMS	CCB	COMS	CCB	CCB
lens	Min	12.77	0.19	10.73	0.17	0.91
	Max	39.79	30.05	35.13	29.58	0.98
	Avg	22.62	4.62	19.66	4.39	0.95
tumor	Min	99.37	99.83	107.18	93.88	0.94
	Max	304.35	427.06	337.36	418.98	0.98
	Avg	180.52	283.43	197.62	274.50	0.97
sclera	Min	6.42	0.08	5.00	0.08	1.00
	Max	347.77	463.11	307.52	454.73	0.98
	Avg	37.26	58.61	32.23	57.56	0.98
opt. nrv.	Min	1.59	0.09	1.67	0.09	0.98
	Max	22.17	4.97	21.91	4.60	0.93
	Avg	7.58	0.67	7.59	0.62	0.93
vitreous	Min	6.74	0.08	6.40	0.08	0.95
	Max	190.32	402.32	203.25	400.69	1.00
	Avg	30.00	26.50	31.06	26.01	0.98
aqueous	Min	8.83	0.11	8.21	0.10	0.95
	Max	59.65	126.65	62.13	130.67	1.03
	Avg	19.38	6.44	19.32	6.45	1.00
iris	Min	9.68	0.12	8.84	0.11	0.92
	Max	39.50	34.23	39.85	34.45	1.01
	Avg	19.34	4.63	18.86	4.53	0.98
cornea	Min	9.51	0.12	7.56	0.11	0.87
	Max	24.11	10.14	21.59	9.97	0.98
	Avg	14.15	1.29	12.01	1.22	0.95

Chapter 4

Discussion

4.1 Verification of `egs_brachy` and eye plaque models

The verification of `egs_brachy` and plaque models against other Monte Carlo calculations in Chapter 3 shows overall good agreement. While discrepancies exist in the doses calculated by the various Monte Carlo codes, they can be explained through small differences in how the codes operate, how the geometries are defined, and what data they use to make calculations. The source of cross sections and material composition data, the difference in volume correction algorithms, differences in geometry of models, and the assumptions made in the dose calculation methods all impact the final dose outcome for a given MC code and therefore result in a small discrepancy in calculated doses between codes.

4.1.1 Photon plaque verification

Comparisons between `egs_brachy` and `BrachyDose` show agreement that is generally within 1% across all plaque models. Figure 3.5 shows a minor systematic difference between `egs_brachy` and `BrachyDose` of about 0.6% for ^{125}I seeds and 0.5% for ^{103}Pd seeds. This systematic uncertainty may arise from the differences in the scoring

routines or modelling of geometries between the two codes. For other possible sources of the systematic discrepancy see ‘**Other uncertainties**’ in Section 2.4.1.

When looking at the $D_{w,w}$ results reported as absolute dose there are a few points of interest with large discrepancies ($\sim 20\%$) between `egs_brachy` and MCNP5. These points are shown in Table 3.2 and are located at the lacrimal gland for plaque positions #1, #2, and #3. Looking at those same points for D_{TG43} in Table 3.1 it is shown that the large discrepancies seen in $D_{w,w}$ are no longer present. The implication is therefore that the introduction of the plaque model creates these discrepancies. This follows logically, as the lacrimal gland is located in the far penumbra region of the plaque for those positions (the penumbra is the dose to areas where radiation would either have to pass through the shielding or scatter into it to reach). It can be concluded that these discrepancies are likely caused by differences in the plaque geometry between `egs_brachy` and MCNP5. In Melhus and Rivard’s 2008 and 2011 papers they confirm overall good agreement between MCNP5 and BrachyDose. `egs_brachy` shows very strong agreement with BrachyDose and similar agreement with MCNP5.

It is expected that D_{TG43} results would compare more favourably than $D_{w,w}$ as there is less room for discrepancy (if there is no plaque there can be no discrepancies in plaque geometry or composition), and this is the case when comparing `egs_brachy` and BrachyDose results. However; the results show similar or worse discrepancies between `egs_brachy` and MCNP5 for D_{TG43} . A possible explanation for this is in the level of accuracy in emulating TG43. The exact configuration of the ‘homogeneous’ simulations for the 2008²⁴ and 2011⁴² MCNP5 works is left somewhat ambiguous. In a personal communication with Melhus it was confirmed for the ‘homogeneous’ simulations in the 2008 paper, while the seeds were modelled in water without a plaque, interseeds effects were also modelled. `egs_brachy` has a superposition mode

that simulates each seed individually and then combines the results to obtain a total dose, allowing for simulations to be done without interseed effects. The addition of interseed effects would affect the final doses calculated using MCNP5 and could result in larger discrepancies when comparing when comparing TG43 doses (D_{TG43}) than expected.

4.1.2 Beta plaque verification

There is solid agreement when comparing to beta plaque doses calculated by Hermida-Lopez using PENELOPE. The largest discrepancies are observed at locations that are either at depths near or past the R_{CSDA} of 7 mm for the electron mean energy, 1.41 MeV, and/or within the penumbral region of the plaque. There are also a few outlier points where the PENELOPE calculations have a much higher dose than those calculated in egs_brachy. These outliers are observable in subfigure (b) of Figure 3.10 (the CCA plaque) and subfigure (a) of Figure 3.11 (the CCD plaque). These points are located along the lateral axis at a depth of 0.1 cm. The dose is near to the plaque and located in voxels where the plaque intersects the scoring region.

These differences could be attributed to a differences in scoring methods, volume corrections or the modelling of geometries between PENELOPE and egs_brachy. Other than the outlier points most disagreements are less than 3% across all profiles for all plaques. The statistical uncertainties reach 2%-2.5% with egs_brachy and 1-2% with PENELOPE at the far points which have 3% disagreement (a combined uncertainty of 3.2%), meaning most disagreement can be explained through statistical uncertainties. The larger differences observed at points toward the eye centre are past the range of the mean electron energy, so little energy is deposited, resulting in low doses. With such low doses a small difference in dose value can result in a large

difference in dose ratio. For example, at the same points most of the photon plaques have central axis doses of greater than 30 Gy, while the beta plaque doses can be on the scale of a few (1-4) Gy.

There is a stronger agreement when comparing `egs_brachy` with the GEANT4 doses calculated by Sommer et al.⁴⁹ in Figure 3.9. Up to the tumor apex there is only a 0.7% disagreement and, while the disagreement does reach 4.4% by 1 cm, the actual dose difference at that point is only 0.13 Gy per decay. A stronger agreement between `egs_brachy` and GEANT4 doses is not unexpected as both use a model of the nuclide decay in their calculation⁴⁹, while the PENELOPE doses were calculated using an ICRU spectrum⁴⁷. When comparing with the BEBIG reference dataset disagreements of up to 7.5% with `egs_brachy` are observed. The GEANT4 results have similar discrepancies with the BEBIG reference. In either case the reported combined uncertainty for the BEBIG measurements is 20%, accounting for far greater than 7.5% disagreements.

When looking at the CIA plaque which contains a cut-out notch, two different versions of the plaque were developed for this work. The ‘CIA-HL’ plaque is based on the work of Hermida-Lopez, and the ‘CIA’ plaque is developed in this work. The dose differences between the dose distributions calculated using the two different versions are not large, but are observable. Figure 3.12 shows that the central axis depth-dose curve and the x-axis profiles for the two versions are within a 0.01% disagreement for most points. At a few points they have 2 – 3% disagreement or even greater disagreements of up to 20%.

The larger disagreements are shown more clearly in the isodose contours of Figure 3.13 where a quite large discrepancy between CIA-HL and CIA can be observed near the plaque in the area of the cutout notch. This discrepancy shows higher doses

near to the cutout for CIA over CIA-HL. The rounded region allows for the active source to cover a larger area of the plaque, and thus increases dose in that region. The disagreement between CIA and CIA-HL suggests that the more accurate CIA version should be used over the more simplistic CIA-HL version.

As seen in the depth-dose and cutout axis curves in Figure 3.12, comparing to PENELOPE shows the two `egs_brachy` CIA plaque versions result in doses that are in solid agreement. The rounded CIA model disagrees with PENELOPE at the same points and for similar percentages that it disagrees with the CIA-HL. This shows that the CIA-HL geometry created to mirror Hermida-Lopez is functioning properly as a clone, and confirms that rounding the edges of the cutout makes a difference in the dose calculations.

For BEBIG beta plaques accurate information of their physical geometry is difficult to find, especially with plaques that contain a cutout. This uncertainty in physical geometry could influence the difference seen in beta plaque dose distributions between various Monte Carlo codes.

4.2 Comparison of D_{TG43} , $D_{w,w}$, and $D_{m,m}$

As introduced in Section 1.4, the probability of photon interaction at low energies is highly dependent on Z_{eff} for the medium being traversed. This gives the expectation of large discrepancies between D_{TG43} and $D_{w,w}$. Reports such as TG129³⁶ and TG186⁸⁷ have strongly recommended that clinics with the means include calculation of $D_{w,w}$ in their dose evaluation process. Other studies such as Melhus and Rivard's 2008 work²⁴ and Rivard et al's 2011 work⁴² with MCNP5 and BrachyDose have shown results confirming the existence of large discrepancies between the two calculation types. This conclusion is supported by comparing the depth-dose curves

of Figures 3.1 and 3.2 to those in Figures 3.3 and 3.4, as well as by looking at the ratios in Figure 3.7 and Table 3.3.

Figure 3.6 shows isodose contours of D_{TG43} alongside $D_{w,w}$ for different photon plaque models. The isodose contours show that D_{TG43} is consistently higher than $D_{w,w}$ for standard COMS 16 mm plaques as well as all the other plaque models except the “COMS - thin acrylic”, which has a higher $D_{w,w}$ than D_{TG43} close to the plaque. That D_{TG43} is greater than $D_{w,w}$ shows the possibility of prescribing an underdose to a patient. For example, the results show a difference of 11% at the tumor apex between $D_{w,w}$ and D_{TG43} ; if $D_{TG43} = 85$ Gy is prescribed, then $D_{w,w}$ would only be 75.65 Gy. More extreme differences in D_{TG43} and $D_{w,w}$ can be observed in Table 3.3. The minimum dose to the optic nerve shows a 84% difference and the average dose to the optic nerve shows a 55% relative difference. These large discrepancies in dose calculations show a problematic lack of accuracy in clinical dose evaluations.

The isodose contours in Figure 3.6 and the ratios in subfigure (b) of Figure 3.7 show that differences between D_{TG43} and $D_{w,w}$ exist for each of the photon plaque models used in this thesis. It is interesting to note that $D_{w,w}$ calculated using the standard COMS plaque has a greater disagreement along the plaque central axis with D_{TG43} than all of the other photon plaque models. Understanding the geometry of the different plaques allows for an understanding of the differences in disagreement. The “COMS - thin acrylic” plaque would result in less photon attenuation than the standard COMS plaque as the seeds are just affixed to an acrylic layer on the backing, so the majority of photons leaving the plaque would not have to go through the medium of the insert. Because the “short lip - acrylic” plaque has a shorter lip there is less medium between the seeds and the eye and, therefore; less photon attenuation. The “no lip - silastic” plaque has the same height as the “short lip - acrylic” plaque, so the seeds are closer to the eye than in the standard COMS plaque.

The lipless plaque also has no high Z_{eff} lip to attenuate the photons. The “stainless steel - acrylic” plaque’s backing is stainless steel rather than a gold alloy, resulting in a much lower Z_{eff} for the shielding of the plaque and less photon attenuation compared to the standard COMS plaque.

Lesperance et al.⁵³ show that the addition of realistic tissue compositions and structures when calculating dose in the eye with photon plaques makes a significant difference. The ratios in Table 3.3 and Figure 3.7 and the isodose contours in Figure 3.6 show that similar differences observed in BrachyDose are achieved in `egs_brachy` as well. For example, Lesperance et al. report differences in average dose to the lens of 29%, average dose to the vitreous of 14%, and maximum dose to the optic nerve of 27% when comparing $D_{m,m}$ and D_{TG43} . Table 3.3 shows that with `egs_brachy` dose differences in the same structures come to 29%, 14% and 28% respectively.

Table 3.3 shows a larger or similar difference in dose between $D_{m,m}$ and D_{TG43} to the ocular structures when compared to $D_{w,w}$ and D_{TG43} . One of the most obvious differences between Figure 3.7(a) ($D_{w,w}$ and D_{TG43} central axis ratios) and Figure 3.7(b) ($D_{m,m}$ and D_{TG43} central axis ratios) is the behaviour within the tumor region. Within the tumor region $D_{m,m}$ is much higher than in $D_{w,w}$ and then drops off immediately after the tumor apex. Interestingly the tumor is the one structure where $D_{m,m}$ is much closer to D_{TG43} when using the standard COMS 16 mm plaque.

$D_{m,m}$ has a 95% agreement with D_{TG43} for the minimum, maximum and average dose to the tumor as shown in Table 3.3. Figure 1.2 shows at the average photon energy of 28 keV for ^{125}I , the mass-energy absorption coefficient is $\sim 10\%$ higher for tumor than water. With a greater mass-energy absorption coefficient more photon attenuation occurs in the tumor medium. More photon attenuation leads to more energy being deposited in the tumor medium than would be in water. $D_{m,m}$ includes

attenuation from the plaque model and interseed effects, while D_{TG43} does not, and so is still less than D_{TG43} in the tumor region.

Other eye structures can be analyzed in a similar manner. At 28 keV the cornea, lens, and sclera have mass-energy absorption coefficients that are respectively, 8%, 9%, and 12% less than water. For each of these three structures $D_{m,m}$ is less than $D_{w,w}$. The lower $D_{m,m}$ concurs with the idea that a lower mass-energy absorption coefficient results in less photon attenuation and consequently, less energy deposition. The optic nerve has a mass-energy coefficient almost equivalent to water and so therefore $D_{w,w} \sim D_{m,m}$, which concurs with the idea that they should have equivalent photon attenuation. The iris, aqueous, and vitreous have mass-energy absorption coefficients that are approximately 2%, 5%, 6% more than water respectively, which suggests that $D_{m,m}$ should be higher than $D_{w,w}$ for these three eye structures. The max doses in each case are high for $D_{m,m}$, but only for vitreous is the average dose of $D_{m,m}$ greater than $D_{w,w}$. Photons must cross other media before reaching the iris or aqueous, so it is possible the attenuation in those upstream media result in less photons reaching the iris and aqueous, resulting in less energy deposition than expected.

Figure 3.15 and Table 3.5 allow for comparison of $D_{w,w}$ and $D_{m,m}$ for beta-emitting plaques. Subfigure (c) of 3.15 shows isodose contours for $D_{w,w}$ and $D_{m,m}$ using a CCB beta plaque. The figure shows that $D_{w,w}$ is consistently slightly higher than $D_{m,m}$ for a beta plaque. Table 3.5 shows that when comparing $D_{w,w}$ and $D_{m,m}$ for a beta plaque, the difference in average dose for each eye structure is between 0% – 7%. Looking at the stopping power ratios in Figure 1.3 explains these differences. The ratios of stopping powers in medium to water show that each of the media of the eye have slightly lower stopping powers than water (up to 2% lower with the tumor). Stopping power characterizes energy loss of a incident charged particle. A

lower stopping power allows less energy to be deposited by a charged particle as it traverses the medium. Less energy deposition in the medium results in less dose, which contributes to $D_{m,m}$ being consistently lower than $D_{w,w}$.

Clinically, dose evaluations are expected to be within a few percent of the actual dose that will be delivered. This is particularly relevant for organs at risk (OARs), where doses should be under a certain tolerance value to minimize damage to healthy tissues. Table 3.3 shows large discrepancies between D_{TG43} and both $D_{w,w}$ and $D_{m,m}$ that are far greater than a few percent. An example is the optic nerve, where D_{TG43} is different from $D_{m,m}$ by 55%. For other OARs, the differences are closer to 20%, and thus are clinically significant. D_{TG43} emulates clinical dose calculations, so the differences discussed show major clinical relevance.

4.3 Comparison of photon and beta plaque dose distributions

As electrons are charged particles, they have a short range when traveling through most media. They undergo a large number of interactions and lose energy quickly. This is shown generally in Figure 3.14. Comparing beta plaques to photon plaques, the beta plaques have a higher dose within the tumor region that quickly drops off to nearly zero dose along the plaque central axis. isodose contours of a standard COMS 16 mm plaque and a CCB beta plaque are given in Figure 3.15. The isodose contours show how different the shape of the dose distribution is between photon and beta plaques. They also confirm how quickly dose calculated using a beta plaque drops off, not just along the central axis, but across the distribution.

Unfortunately, the short range of beta electrons in tissue leads to a “weakness”

of the beta plaque. If a tumor is too ‘tall’ then the beta electrons may not be able to deposit an acceptable amount of energy throughout the entire tumor. Table 3.4 shows that this behaviour of electrons results in much lower doses to organs at risk than photon plaques for most plaque positions. It also shows, however, that with certain plaque placements the beta plaque will deliver much higher doses when the beta plaque is placed near the organ at risk. Table 3.5 shows minimum, maximum, and average doses to the tumor and the surrounding organs at risk. The table shows that the beta plaque offers lower minimum dose to the lens, sclera, optic nerve, and vitreous fluid; lower average doses except to the sclera and lower maximum doses to the lens and optic nerve for both $D_{w,w}$ and $D_{m,m}$. Within the tumor the beta plaque has higher maximum, minimum, and average doses than the photon plaque. The difference is larger for photon plaques, a logical conclusion to make based on Figures 1.2 and 1.3. These figures show a larger difference in mass-energy absorption coefficients between various eye tissues and water (with a largest difference of $\sim 10\%$ between tumor and water) than in stopping powers between the same tissues and water (the largest difference being $\sim 2\%$ between tumor and water).

Chapter 5

Conclusions and outlook

Current clinical photon eye plaque brachytherapy dose calculations are done using the TG43 formalism. The TG43 formalism for dose calculation involves several assumptions. All media are considered to be water rather than the high Z materials of a plaque or the varying tissues of the human body. Additionally, the interseed effects, which account for a notable amount of photon attenuation, are ignored.

For beta plaques the situation is more ambiguous. No accepted standard for dose calculations and treatment planning has been put in place. Dose calculations for beta-emitting plaques can vary in accuracy and methodology. Questions as to the accuracy of doses calculated using the TG43 formalism have led to various studies on the effects of taking a model-based approach instead (modelling both the plaque and tissue compositions during dose calculation) and have concluded that there is a significant difference between TG43 calculations and model-based calculations. An investigation into various model-based dose calculation algorithms has ensued and one of the most prominent candidates is the use of Monte Carlo-based dose calculations.

Through comparisons of D_{TG43} , $D_{w,w}$, and $D_{m,m}$ this work has demonstrated the need for model-based calculations (purpose (i), section 1.6). This importance of more advanced model-based eye plaque brachytherapy dose calculations has been demon-

strated many times, but implementation within the clinic will require access to well documented and thorough algorithms, plaque geometries, and material composition data. The plaque models presented in this work are developed within an open source geometry package (egs++) that works in conjunction with an open source Monte Carlo code (egs_brachy) and will be distributed with future releases of egs_brachy (purpose (ii), section 1.6). Through comparisons to other doses calculated by other Monte Carlo codes this work has verified egs_brachy and the plaque models against published results (purpose (iii), section 1.6). The availability of these plaque models will allow for future studies to investigate further aspects of eye plaque dose distributions and encourage the adoption of advanced model-based calculations by treatment planning systems and the clinics that use them. The variety of plaque models will also give a baseline for creation of more plaque models to be implemented within egs_brachy. Future studies will be able to investigate the difference between the various plaque types, their effects among the various ocular structures, and comparisons to other treatment modalities. Using the eye phantom, doses to volume such as those presented in Table 3.5 could be used to more accurately investigate the doses in ocular structures at chosen prescription doses. Such investigations could provide exploration of changes to prescription dose, including possible dose de-escalation. An example de-escalation study found decreased toxicity with equivalent tumor control using TG43 dose calculations but only a 69 Gy prescription dose⁹¹.

Model-based Monte Carlo calculations could also enable investigation into radiobiology for eye plaque brachytherapy. In 2012 Gagne et al.⁹² developed a method for evaluating model-based Monte Carlo dose distributions and the resultant biologically-effective dose distributions for photon plaques. While there is currently no analogous study for beta plaques, the geometries in this work could be used to accomplish one. Gagne et al.⁹³ also investigated the radiobiological effects of implant duration and

the choice of radionuclide for photon plaques. The study found that ^{103}Pd offered a radiobiological benefit when compared to ^{125}I and ^{131}Cs ; however, the magnitude of uncertainties eclipsed the significance of the differences. Similar studies could be done with `egs_brachy` and the plaques in this work to confirm or deny their results or to investigate the same effects from beta-emitting plaques.

This thesis work utilizes the `egs_brachy` Monte Carlo code and the `egs++` geometry package to create and verify various plaque models (both photon and beta-emitting) and investigate the use of a realistic eye phantom. The model-based approach also enables a comparison between a photon-emitting plaque and a beta-emitting plaque to be performed with the goal of showcasing the versatility of research that the plaque models enable (purpose (iv), section 1.6). The created plaques will be distributed with future releases of `egs_brachy` as well as through a website database maintained by the CLRP. It is hoped they will be utilized in future studies, as a baseline for future plaque models, and to encourage the implementation of model-based dose calculations by treatment planning system developers and clinics.

Appendix A

Photon plaque seed positions

The photon plaque designs in this thesis each have a different number of seeds and positions for those seeds. The seed positions for each of these plaque designs are presented in Table [A.1](#). Seed positions are given in terms of x_c , y_c , and z_c , which are the x , y , and z coordinates of the seed centres with respect to the eye plaque coordinate system. The angle of the seed θ which is measured from the x-axis is also included.

Table A.1: Seed centre and seed angle coordinates for all photon plaques developed for this work. Seed angles are in relation to the x-y plane where 0° is a seed lying along the x-axis.

Plaque size	seed number	x_c (cm)	y_c (cm)	z_c (cm)	θ (rad)
COMS 10 mm	1	-0.230	-0.230	-0.201	2.36
	2	0.230	-0.230	-0.201	0.78
	3	0.230	0.230	-0.201	5.50
	4	-0.230	0.230	-0.201	3.93
	5	0.000	0.000	-0.240	4.71
COMS 12 mm	1	-0.372	-0.27	-0.16	2.51
	2	0.142	-0.437	-0.16	1.26
	3	0.46	0	-0.16	6.28
	4	0.142	0.437	-0.16	5.03
	5	-0.372	0.27	-0.16	3.77
	6	0	-0.27	-0.213	1.57
	7	0	0.27	-0.213	4.71
	8	0	0	-0.24	4.71
COMS 14 mm	1	-0.59	0	-0.106	3.14
	2	-0.295	-0.511	-0.106	2.09
	3	0.295	-0.511	-0.106	1.05
	4	0.59	0	-0.106	0
	5	0.295	0.511	-0.106	5.24
	6	-0.295	0.511	-0.106	4.19
	7	-0.41	0	-0.177	3.14
	8	0	-0.41	-0.177	1.57
	9	0.41	0	-0.177	0
	10	0	0.41	-0.177	4.71
	11	0	-0.21	-0.224	1.57
	12	0	0.21	-0.224	4.71
	13	0	0	-0.24	4.71
COMS 16 mm and “COMS - thin acrylic”	1	-0.568	-0.273	-0.087	2.69
	2	-0.14	-0.614	-0.087	1.8
	3	0.393	-0.493	-0.087	0.89
	4	0.63	0	-0.087	0
	5	0.393	0.493	-0.087	5.39
	6	-0.14	0.614	-0.087	4.48
	7	-0.568	0.273	-0.087	3.59
	8	-0.45	0	-0.164	3.14
	9	0	-0.45	-0.164	1.57
	10	0.45	0	-0.164	6.28
	11	0	0.45	-0.164	4.71
	12	0	-0.18	-0.228	1.57
	13	0	0.18	-0.228	4.71

Plaque size	seed number	x_c (cm)	y_c (cm)	z_c (cm)	θ (rad)
COMS 18 mm	1	-0.77	0	-0.003	3.14
	2	-0.544	-0.544	-0.003	2.36
	3	0	-0.77	-0.003	1.57
	4	0.544	-0.544	-0.003	0.78
	5	0.77	0	-0.003	6.28
	6	0.544	0.544	-0.003	5.5
	7	0	0.77	-0.003	4.71
	8	-0.544	0.544	-0.003	3.92
	9	-0.62	0	-0.092	3.14
	10	-0.31	-0.537	-0.092	2.09
	11	0.31	-0.537	-0.092	1.04
	12	0.62	0	-0.092	6.28
	13	0.31	0.537	-0.092	5.23
	14	-0.31	0.537	-0.092	4.19
	15	-0.318	-0.318	-0.164	2.36
	16	0.318	-0.318	-0.164	0.78
	17	0.318	0.318	-0.164	5.5
	18	-0.318	0.318	-0.164	3.92
	19	0	-0.2	-0.225	1.57
	20	0	0.2	-0.225	4.71
	21	0	0	-0.24	4.71
COMS 20 mm	1	-0.81	-0.29	0.06	2.79
	2	-0.43	-0.75	0.06	2.09
	3	0.15	-0.85	0.06	1.40
	4	0.66	-0.55	0.06	0.70
	5	0.86	0.00	0.06	6.28
	6	0.66	0.55	0.06	5.58
	7	0.15	0.85	0.06	4.89
	8	-0.43	0.75	0.06	4.19
	9	-0.81	0.29	0.06	3.49
	10	-0.65	-0.15	-0.07	2.91
	11	-0.29	-0.60	-0.07	2.02
	12	0.29	-0.60	-0.07	1.12
	13	0.65	-0.15	-0.07	0.02
	14	0.52	0.42	-0.07	5.60
	15	0.00	0.67	-0.07	4.71
	16	-0.52	0.42	-0.07	3.82
	17	-0.38	-0.28	-0.16	2.51
	18	0.15	-0.45	-0.16	1.26
	19	0.47	0.00	-0.16	6.28
	20	0.15	0.45	-0.16	5.02
	21	-0.38	0.28	-0.16	3.77
	22	0.00	-0.23	-0.22	1.57
	23	0.00	0.23	-0.22	4.71
	24	0.00	0.00	-0.24	4.71

Plaque size	seed number	x_c (cm)	y_c (cm)	z_c (cm)	θ (rad)
COMS 22 mm	1	-0.92	0.00	0.12	3.14
	2	-0.65	-0.65	0.12	2.35
	3	0.00	-0.92	0.12	1.57
	4	0.65	-0.65	0.12	0.78
	5	0.92	0.00	0.12	6.28
	6	0.65	0.65	0.12	5.50
	7	0.00	0.92	0.12	4.71
	8	-0.65	0.65	0.12	3.93
	9	-0.72	-0.13	-0.03	2.97
	10	-0.25	-0.69	-0.03	1.92
	11	0.47	-0.56	-0.03	0.87
	12	0.72	0.13	-0.03	6.11
	13	0.25	0.69	-0.03	5.06
	14	-0.47	0.56	-0.03	4.01
	15	-0.31	-0.37	-0.15	2.27
	16	0.37	-0.31	-0.15	0.70
	17	0.31	0.37	-0.15	5.41
	18	-0.37	0.31	-0.15	3.84
	19	0.02	-0.22	-0.22	1.48
	20	-0.02	0.22	-0.22	4.62
	21	0.00	0.00	-0.24	4.62
COMS 24 mm	1	0.00	0.00	-0.24	1.57
	2	0.00	0.19	-0.23	1.57
	3	0.00	-0.19	-0.23	4.71
	4	0.44	0.09	-0.16	2.95
	5	0.06	0.45	-0.16	1.7
	6	-0.41	0.19	-0.16	0.44
	7	-0.31	-0.33	-0.16	5.53
	8	0.22	-0.39	-0.16	4.21
	9	0.47	0.45	-0.08	2.37
	10	-0.06	0.64	-0.08	1.48
	11	-0.54	0.35	-0.08	0.58
	12	-0.61	-0.20	-0.08	5.96
	13	-0.22	-0.61	-0.08	5.07
	14	0.34	-0.55	-0.08	4.17
	15	0.64	-0.08	-0.08	3.27
	16	0.83	0.07	0.04	3.05
	17	0.59	0.59	0.04	2.36
	18	0.07	0.83	0.04	1.66
	19	-0.48	0.68	0.04	0.96
	20	-0.80	0.21	0.04	0.26
	21	-0.75	-0.35	0.04	5.85
	22	-0.35	-0.75	0.04	5.15
	23	0.21	-0.80	0.04	4.45
	24	0.68	-0.48	0.04	3.75
	25	0.91	0.45	0.21	2.69
	26	0.41	0.93	0.21	1.99
	27	-0.28	0.98	0.21	1.29
	28	-0.84	0.57	0.21	0.59
	29	-1.01	-0.11	0.21	3.04
	30	-0.71	-0.73	0.21	5.42
	31	-0.07	-1.01	0.21	4.78
	32	0.60	-0.82	0.21	4.08
	33	0.98	-0.25	0.21	3.38

Plaque size	seed number	x_c (cm)	y_c (cm)	z_c (cm)	seed angle (rad)
“Short lip - acrylic” and “No lip - silastic”	1	-0.568	-0.273	-0.012	2.688
	2	-0.140	-0.614	-0.012	1.798
	3	0.393	-0.493	-0.012	0.890
	4	0.630	0.000	-0.012	0.000
	5	0.393	0.493	-0.012	5.393
	6	-0.140	0.614	-0.012	4.486
	7	-0.568	0.273	-0.012	3.595
	8	-0.450	0.000	-0.089	3.142
	9	0.000	-0.450	-0.089	1.571
	10	0.450	0.000	-0.089	6.283
	11	0.000	0.450	-0.089	4.712
	12	0.000	-0.180	-0.153	1.571
	13	0.000	0.180	-0.153	4.712
“Stainless steel - acrylic”	1	-0.568	-0.273	-0.058	2.688
	2	-0.140	-0.614	-0.058	1.798
	3	0.393	-0.493	-0.058	0.890
	4	0.630	0.000	-0.058	0.000
	5	0.393	0.493	-0.058	5.393
	6	-0.140	0.614	-0.058	4.486
	7	-0.568	0.273	-0.137	3.595
	8	-0.450	0.000	-0.137	3.142
	9	0.000	-0.450	-0.137	1.571
	10	0.450	0.000	-0.137	6.283
	11	0.000	0.450	-0.137	4.712
	12	0.000	-0.180	-0.203	1.571
	13	0.000	0.180	-0.203	4.712

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