Vocal cord irradiation: experimental dosimetry and comparison to treatment planning computations

Joris Hautekiet

Supervisor: Prof. dr. ir. Carlos De Wagter
Counsellor: Annemieke De Puysseleyr

Master's dissertation submitted in order to obtain the academic degree of
Master of Science in Biomedical Engineering

GE17
Chairman: Prof. dr. ir. Carlos De Wagter
Faculty of Engineering and Architecture
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Preface

Hereby I would like to sincerely thank my promoter professor Carlos De Wagter for presenting the topic and all the guidance during my thesis. To Annemieke De Puysseleyr who showed me the wonderful world of film dosimetry and counselled me very well in writing this thesis, Merci. Thank you, dr. Leen Paelinck, for creating my clinical treatment plan. Thank you Prof. Barbara Vanderstraeten and all others of the Physics team at UZ for guiding me when I got lost in their software.

Special thanks to my friends, family and parents, who crossed their fingers multiple times during my entire education. It worked.

Joris Hautekiet

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Summary

This thesis deals with the determination of the dose absorbed in and near the vocal cords during irradiation with 6 MV photon beams. A comparison is made between two dose computations of two different treatment planning systems for 4 set-ups of phantom geometries. The Collapsed Cone Convolution algorithm of Pinnacle and the XVMC Dose Engine of Monaco are hereby used to calculate the dose distribution in a phantom, which modelled the larynx and the vocal cords. The radiochromic EBT3 film was used in the phantom to experimentally determine the absorbed dose at the tissue-air interfaces and the surrounding region.

The phantom was irradiated with a one field configuration for three different phantom and film set-ups, for one of those set-ups, also an IMRT plan was applied.

All the measurements in our phantom show that the XVMC Dose Engine can better predict the dose in the regions of the air cavity surrounding the modelled vocal cord than the CCC algorithm. This is in accordance with literature. In the measurements, radiochromic film has shown to be a good and reliable dosimeter.

Key words

radiochromic EBT3 film, vocal cord dosimetry, Monaco XVMC Dose Engine, CCC algorithm
VOCAL CORD IRRADIATION: EXPERIMENTAL DOSIMETRY AND COMPARISON TO TREATMENT PLANNING COMPUTATIONS
Hautekiet Joris
Supervisors: Prof. dr. ir. Carlos De Wagter, ir. Annemieke De Puysseleyr

Abstract - Purpose: To compare dose calculations of 6 MV photon beams of the CCC algorithm of Pinnacle and the XVMC Dose Engine of Monaco in the glottic region. Methodology: Radiochromic film measurements in different phantom set-ups will be compared with the aid of percentage depth dose (PDD) curves and gamma analyses with the criteria of ‘3% dose difference/3 mm distance to dose’. Results: Gamma analyses show that Monaco complies more with the criteria, e.g. 85.29% against 99.05% and 10.85% against 75.02%. Dose differences between Pinnacle and radiochromic film measurements of more than 10% were observed. Conclusion: Monaco can better predict the dose in the glottic region.

Keywords - radiochromic EBT3 film, vocal cord dosimetry, Monaco XVMC Dose Engine, CCC algorithm

I. INTRODUCTION

Radiotherapy is used for the treatment of low staged glottic carcinoma. The vocal cords or glottis are part of the larynx, superior the trachea and they are surrounded by air. A mean male glottic diameter of 18 mm, a length of 10-15 mm and a thickness of 5 mm were found in literature [1, 2, 3].

The vocal cords are typically treated with photon beams, the conventional treatment consisted of 2 parallel opposing beams which had a good local control rate, but intensity modulated radiotherapy (IMRT) is now indicated because it spare the normal tissue. Treatment planning systems (TPS) are used to calculate the prescribed doses, depending on the used algorithm, it is know that overestimations in region with an air cavity occurs [4, 5].

Because of the air-tissue interfaces, there is a loss of the transient charged particle equilibrium. This happens in a build-up region, the lateral charged particle disequilibrium is also significant at air-tissue interfaces [6].

Figure 1: Coronal section of the larynx.

In order to evaluate the dose computations in the region of the vocal cords, a phantom is used in which the radiochromic film EBT3 (International Speciality Products Ashland Inc., Gafchromic®) is placed to measure the absorbed dose. This is compared to two dose algorithms of two commercial TPS: The Collapsed Cone Convolution (CCC) algorithm of Pinnacle® (version 9, Philips Healthcare, Best, The Netherlands) and the XVMC Dose Engine of Monaco (version 3.20.01, Elekta, Atlanta, USA).

The Pinnacle software contains multiple dose algorithms, in this thesis, the CCC algorithm is used. It calculates the dose by convolving an energy deposition kernel with the total energy released per mass (TERMA). This TERMA is calculated using the patient density representation obtained from CT data. The kernel is once generated using Monte Carlo methods and has a scaling factor to incorporate heterogeneities in the medium. It is a relative fast algorithm, it uses superposition and not a Fourier transformation to solve the convolution [8].

The XVMC Dose Engine of Monaco is modified Monte Carlo algorithm, it uses a virtual source model and transmission filter to analytically described the origin of mod-elled particles. The Monte Carlo part is based on the Voxel Monte Carlo code. Monte Carlo method means relying on repeated random sampling in order to obtain useful simulations. Because it is very time consuming in clinical practice, some simplifications are applied in order to decrease calculation time in Monaco [7].

II. MATERIALS AND METHODS

This study concerned 4 different measurements. A summary of the different parameters is shown in table 1. The phantom was irradiated with one field configuration for three different phantom and film set-ups, for one set-up also an IMRT plan was applied.

PHANTOM

The phantom is made of polystyrene slabs (ρ of 1.04 g/cm³, Zeff of 5.7 [9]) hold together with two threaded rods.

Figure 2: Transverse cross-section of the phantom.

It is a cylindrical phantom with a circular air cavity (diameter 20 mm) 10 mm below the surface. The narrowing of the vocal cords in modelled by 4 sections with
a thickness of 2 mm in which the width ‘Q’, indicated at figure 2, changes from 20 mm (= full circle) to 17 mm, 14 mm, 11 mm and 8 mm. Two solid sections (‘Q’=0mm) with a total thickness of 4 mm were chosen for geometry A, one section with a thickness of 1 mm was chosen for geometry B, they model the vocal cords when closed. Geometry A is better to model the high staged tumours, while geometry B better represents the thin structure of the unaffected vocal cords.

<table>
<thead>
<tr>
<th>Measurement number</th>
<th>1</th>
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<th>3</th>
<th>4</th>
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<tr>
<td>Thickness of vocal cords</td>
<td></td>
<td></td>
<td></td>
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<tr>
<td>Geometry A: 4 mm</td>
<td>x</td>
<td>x</td>
<td>x</td>
<td>x</td>
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<tr>
<td>Geometry B: 1 mm</td>
<td>x</td>
<td>x</td>
<td>x</td>
<td>x</td>
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<tr>
<td>Film configuration</td>
<td></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>1 film caudal</td>
<td>x</td>
<td>x</td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td>5 stacked films caudal</td>
<td>x</td>
<td>x</td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td>1 film cranial</td>
<td>x</td>
<td>x</td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td>Beam set-up</td>
<td></td>
<td></td>
<td></td>
<td></td>
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<tr>
<td>field size 6x6 cm², one beam,</td>
<td>x</td>
<td>x</td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td>gantry angle 90°</td>
<td>x</td>
<td>x</td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td>IMRT with 6 beams directions</td>
<td>x</td>
<td>x</td>
<td>x</td>
<td>x</td>
</tr>
</tbody>
</table>

Table 1: Overview of measurement parameters.
*polystyrene + stacked films.

**Radiochromic film**

Radiochromic film does not require any chemical processing after the exposure, unlike radiographic film. The active component is a special dye which gets polymerized when exposed to radiation [9]. The EBT3 is a symmetrical film, which has at both sides of the active layer a clear polyester [10]. The lot number was A07251101. Due to the polymerization a colouration from green to blue occurs. The assumption of film dosimetry is that the absorbed dose in the film is reflected in the resulting optical density (OD) of that film. The OD is the reduction of the intensity of light passing through the film. By obtaining a calibration curve absolute dosimetry is possible.

In order to obtain a precise and accurate dose distribution, a good protocol is required. To correct for inhomogeneities of the film the netto OD is used. For the film scanning, the flat bed scanner Epson® Expression 10000XL (Seiko Epson Corporation, Nagano, Japan) is used as recommended by the film manufacturer. The films are scanned in positive transmission mode, with no color corrections at a resolution of 100 dpi. To obtain the netto OD (defined in eq. 1) a pre- and post-irradiation scan need to be taken. The red channel is used, as recommended by the manufacturer for the used dose range. $I_0$ represents the intensity of the film-attenuated light, $I_0$ the intensity of the undisturbed light.

$$netto \ OD = \log \frac{I_0}{I_{post}} - \log \frac{I_0}{I_{pre}} \quad (1)$$

The films pieces are cut and marked in two opposing corners and at the side in order to match the pre- and post-irradiation scan. The films are placed on the scanner bed with the aid of a template in order to minimize variation in positioning. They were scanned 9 times, but only the last 3 images are used to make an average.

After the irradiation, the procedure is repeated and the further analysis in MATLAB® (version 7.12, The MathWorks Inc., Massachusetts, USA) is done. The averaged pre- and post-irradiation read out was converted into the OD of the red channel, this is the most sensitive for the used dose range. The scans are corrected for sensitivity ununiformities over the scanner bed, they are matched according to their corner marks and the netto OD was obtained by applying a calibration curve.

For the calibration curve, 7 pieces (3.5 x 3.5 cm²) were placed between two layers of 10 cm of polystyrene plates, they were irradiated with a 6 MV photon beam, source surface distance (SSD) of 90 cm and a field size of 10 x 10 cm², 0, 29, 70, 128, 201, 290 and 396 Monitor Units (MU) were delivered. Their netto OD is fitted to a third degree polynomial to obtain the dose as function of netto OD. A calibration curve was made for each measurement.

Each measurement was repeated three times and the three dose distribution was averaged.

As this study only concerned the dose distribution in and near the vocal cords, templates (fig. 3) are used in order to minimize the amount of unused film. The template is cut from radiochromic film and thus has the same thickness as the measurement films. It also helps to avoid air gaps and to have a good consequent positioning.

When the measurement film was positioned in the template, the isocenter was marked to compare the measured dose distribution with the calculations.

**Film configuration and beam set-up**

**Isocenter**

In figure 4, an overview of the positioning and the used coordinate system is shown.

In figure 5, the exact positioning of the isocenter used in all measurements is shown. It is situated 6 mm caudal to the solid slab and 5 mm posterior the air cavity. This way the beam central axis does not contain the film plane as indicated by literature [11, 5].

The phantom was positioned always positioned along this isocenter. With the aid of the positioning lasers and the marking lines applied to the outer surface of the phantom an accurate positioning was obtained.
Figure 4: Overview of the relative positioning of the phantom to the linear accelerator.

Film configuration

In total 3 different film configurations are used, they are shown in green in figure 6. Measurement 1: One caudal, 2: 5 stacked films divided into positions A and B, 3 and 4: one caudal and one cranial.

Beam set-up

In measurements 1, 2 and 3, the phantom was irradiated with one 6 MV photon beam and a field size of 6 x 6 cm$^2$, the gantry was set at 90°. 200 MU were given. In all the measurements the phantom and calibration films were irradiated with the Elekta Synergy linear accelerator with the MLCi2 (Elekta, Crawley, West-Sussex, UK) at the Ghent University Hospital.

In measurement 4, a clinical class solution comprising a coplanar step-and-shoot IMRT treatment, was applied. It was optimized in Monaco for a randomly selected patient, diagnosed with a cT1a glottic carcinoma. The plan consisted of 6 different beam directions with in total 39 segments. 349.88 MU were given to the phantom.

Dose calculations

The CT images of the phantom were acquired on a Toshiba Aquilion CT scanner (Toshiba Medical Systems, Tokyo, Japan). A tube potential and tube current of 120 kVp and 300 mA were used respectively. The TCOT Reconstruction algorithm and a slice thickness of 1 mm were used.

The CT data were imported in Pinnacle, the densities of the polystyrene were manually overwritten to 1.000 g/cm$^3$ and those of the air cavity to 0.000 g/cm$^3$ because the reconstruction of the phantom was not accurate enough for the sharp edges and specific dimensions of the phantom. The beam was modelled at the described isocenter. The minimal dose grid of 1 mm was used (voxel size of 1 mm$^3$). The dose was calculated with the CCC algorithm.

The IMRT treatment plan was imported in Pinnacle, the overridden densities and the minimal dose grid of 1 mm were used again.

The CT data was also imported in Monaco, here the relative electron densities for the polystyrene were set to 1.000 and those of the air cavity to the minimal of 0.010. The beam was modelled and set to the correct isocenter. The dose grid and variance settings were 1 mm and 1% respectively. 200 MU were delivered and the dose to medium was calculated.

The IMRT treatment plan was created in Monaco, the dose was calculated in the ‘QA mode’ of Monaco were the phantom with the overridden relative electron densities was loaded. A dose grid of 1 mm and a variation of 2% were used this time.

Comparison methods

Besides dose distributions in the transversal plane and percentage depth dose (PDD) curves (Dose along the black dotted lines, which is averaged over 11 pixels and scaled to the maximum dose calculated by Monaco along that line.), a gamma analysis is used. For every pixel, the minimum gamma is calculated using equation 2. A tolerance level of 3% is used for the dose difference (DD) and 3 mm for the distance to dose (DtD.

$$\gamma = \sqrt{\frac{\text{Dose Difference}}{\text{Tolerance DD}}} + \frac{\text{Distance to Dose}}{\text{Tolerance DtD}}$$  \hspace{1cm} (2)

III. Results and discussion

On the dose distribution of measurement 1 (fig. 7), which is centred along the isocenter, a grey rectangular is indicated, to which the gamma analysis is restricted. In the region of the air cavity the absorbed dose is lower. The PDD curve of measurement 1 (fig. 8) shows the rebuild-up region at X = -0.5 cm and the dose drop for the measurement and the dose calculation of Monaco. The effect of the presence of the air cavity is negligible for Pinnacle, where only a small kink in the curve can be observed. Pinnacle overestimates the dose hereby with 5%.
Figure 7: Dose distribution in cGy of measurement 1. Transversal cross-section, the beam at 90°, region of grey rectangular is used for the gamma analyses, the dotted black lines indicates the position of the PDD. The dotted white lines show the caudal cross-section.

Figure 8: PDD of measurement 1 (y=-1.5cm), the dose differences between Monaco and the film measurement remains below the 2%. Pinnacle does not adapt the dose to the presence of the air cavity.

In fig.9 the gamma analysis for measurement 1 is shown. Monaco performs here better.

Figure 9: Gamma analysis for measurement 1. 85.29% of the pixels values passes the criteria in Pinnacle while for Monaco this is 99.05%.

Figure 10: Dose distributions for measurement 2, the color code is the same as in figure 7. A is the average of the 2 most caudal films of the stack, B is the average of the 3 more cranial. The dose calculations of A equal those for measurement 1.

In measurement 2 (fig. 10) it can be seen that the influence of the air cavity on the dose is lower for position B (not exposed to the air).

The dose calculations of Monaco show very clearly the secondary build-up (or rebuild-up) zones (dose increase of 1.6%). The gamma analysis (fig. 11) shows again that Monaco performs better in calculating the dose in the region and that most of the exceeding of Pinnacle are situated at the air-tissue interface in position A. Fig. 12 shows the dose differences along the black dotted lines indicated in dose distribution. The difference of Monaco with the film measurement in the region of the air cavity remains below the 2%.

Figure 11: Gamma analysis of measurement 2. Blue means they passed the criteria. From left to right: 78.17%, 98.53%, 88.49% and 97.87%.

Figure 12: Differences along the indicated black lines in figure 10. Pinnacle overestimates the doses with almost 5%, while Monaco calculates a lower dose in the air cavity region than the ones of the film.
Measurement 3 was performed using phantom geometry B, so both films were exposed to the air. In the dose distributions (fig. 13) we see that in the most caudal film both adjacent sections appear because of a dose reduction at the edges. For the caudal film measurement there is a dose drop with 4% over 2 mm when the beam enters the air cavity and 2 rebuild-up zones with respectively a dose increase of 3% and 2% over 2 mm at X = -0.4 and -1.0 cm (see fig. 14). This is also visible in the dose distribution of Monaco. The relative maxima of Monaco, however, are 2 mm more to the left than those of the film. After the air cavity the curves of Pinnacle and Monaco show a good agreement with each other (difference below 1.5%).

In the cranial PDD we see also that the trends of the curve of Monaco and the measurement comply very well, but that there is an overall difference of 3.96%.

The gamma analysis showed that only 50.59% (caudal) and 22.53% (cranial) of the pixels passed the criteria for Pinnacle, for Monaco this was respectively 95.70% and 75.37%.

In Measurement 4, the IMRT plan was applied. The dose distribution can be seen in fig. 15. Because we no longer have 1 beam, 2 PDD profiles are shown, one along the X-axis (y=-1.5 cm) (fig. 16) and one along the Y-axis (x=0 cm) (fig. 17).

The caudal PDD along the X-axis shows a very good agreement of the dose measured with film and those calculated with Monaco between x=-0.5 cm and 0.5 cm. Pinnacle shows a small dose drop in that region, but overestimates the whole region of the air cavity with ~7%. The dose calculated with Monaco seems to have an almost constant difference of 4.18%.

At the caudal PDD along the Y-axis calculates Monaco between Y = [-2.0cm; -0.5cm] an almost homogeneous dose, while the measurement shows for that region a dose gradient of 4% over the 1.5 cm.
If we look at the dose differences of measurement 1 and 2 we see that the result of 1 and 2B are very similar: Pinnacle gives a higher dose (up to 5%). Monaco clearly predicts the dose within the 1.8% at position A and B. When those results are compared against those of measurement 3, we can still assume that Monaco is better at predicting the dose in the region of the air cavity, but the difference between Monaco and the film measurements increases to 3-5% at the cranial position. Pinnacle however overestimates the dose with minimally 5% in the region of the air cavity. At the cranial position, the difference even exceeds 8%.

The results of measurement 3 are very similar to those of measurement 4. Even with a more relevant treatment plan, Monaco performs more accurate if we take our film measurement as reference. Like already mentioned, the curves of the film and Monaco have a similar trend at the cranial PDDs, they differ only with ~4%.

All the gamma analyses show that the calculations of Monaco comply better with the criteria of 3%/3mm.

The results of 3 and 4 seems to indicate an offset between the calculations and the measurements. Pinnacle and Monaco calculate the dose in voxels (here 1mm³), while the absorbed dose in the film is measured for a much thinner region. The film thickness is only 0.285 mm.

IV. Conclusion

All measurements show that the XVMC Dose Engine of Monaco better predicts the dose in the regions of the air cavity surrounding the modelled vocal better than the CCC algorithm of Pinnacle which calculates a higher dose in those regions.

Radiochromic film has shown to be a reliable dosimeter, but care has to be taken when its absorbed dose is compared against dose distribution obtained with a wider dose grid than the film thickness.

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<td>4.26</td>
<td>PDDs of measurement 4 at y = -1.5 cm</td>
<td>48</td>
</tr>
<tr>
<td>4.27</td>
<td>PDDs of measurement 4 at x = 0 cm</td>
<td>49</td>
</tr>
</tbody>
</table>
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Chapter 1

Radiotherapy

This thesis deals with the determination of the dose absorbed in and near the vocal cords during irradiation. Radiotherapy is mainly used for the treatment of cancer, both in curative and palliative settings. Depending on the type of cancer, radiotherapy can be combined with surgery or chemotherapy. The goal of radiotherapy is to damage the DNA of the tumour cells by exposing them to ionizing radiation. This will result directly (by the photons itself) or indirectly (through the formation of highly reactive radicals) into the breakage of DNA strands. When there is a single strand break, the DNA could be repaired, which is not the case for a double strand break. If the DNA damage is severe and can no longer be repaired, the tumour cell could have no longer the ability to divide and may die [1, 2].

The success of this process depends on many factors. The healthy tissue can also be damaged by ionizing radiation, luckily not every (tumour) cell has the same radiosensitivity, repair mechanisms, repopulation activity, vascularization etc. In general, normal healthy cells recuperate better, allowing to enhance normal tissue sparing while maintaining tumour control by fractionating the dose. It is of great important for the treatment and for research that the amount of radiation is well known. In radiotherapy the prescriptions are based on the (absorbed) dose expressed in Gray (Gy) [2, 3].

The vocal cords are typically treated with external radiotherapy. They are usually irradiated with a linear accelerator in which electrons are accelerated by microwaves up to a certain energy (in the range of 4-25 MeV). The X-rays are produced by the bremsstrahlung released during the collision of the electrons with a target within the linear accelerator. Those X-rays have a continuous spectrum determined by the energy of the accelerated electrons and the specific machine components and settings. This photon beam quality is typically expressed in Megavolts (MV)[1].
Chapter 2

Vocal cords

2.1 Anatomy and functioning of the vocal cords

The vocal cords or glottis are a part of the voice apparatus or the larynx. (See figure 2.1.) They are an important component in the voice ability, this paired structure appears pearly white due to the lack of blood vessels. It is largely composed of elastic fibers [4].

The vocal cords are partly responsible for the change in sound and pitch of your voice. It is believed that the wide range of phonation frequencies in humans are a result of the multi-layered structure and corresponding material properties of the vocal cords [5]. (See figure 2.2b.) While air moves out, the vocal cords open and close partially to produce different sounds and frequencies. Next to producing sound the larynx also protects the airways while swallowing by tightly closing the epiglottis and the vocal cords. In this way food and fluids are prevented from entering the lungs. The last function of the vocal cords is to open during breathing in order to let air in and out of the airways [6].

Regarding radiotherapy, it is important to mark the presence of the trachea inferior the larynx. This anatomical structure can be seen as a cylindrical concavity filled with air. Also the region above and beneath the vocal cords can be seen as a cavity filled with air.

The mean inner coronal diameter of the cricoid cartilage (below the thyroid cartilage see figure 2.1) is for men (group of 9) 18,26 mm (Standard deviation (SD) = 1,23 mm) and for women (group of 7) 12,79 mm (0,31 mm) [7]. In another study the cricoid and glottic diameters were obtained during a postmortem examination, the mean male cricoid diameter was 17 mm (SD = 1,5 mm), the female was 13 mm (1 mm). The mean male glottic diameter was 18 mm (1,5 mm), the female was 14 mm (1 mm) [8]. The vocal folds themselves are about 10-15 mm in length and have a thickness of 5 mm [9], where this thickness probably relates to the thickness at the base.

The mean inner diameter of the trachea, measured in the age group of 50 and 59 years, depends on the plane and sex. For men the coronal tracheal diameter is 19,2 mm (SD = 2,3 mm) and the sagittal one is 20,4 mm (2,6 mm). The mean coronal and sagittal tracheal diameters for women are smaller, respectively 16,5 mm (1,6 mm) and 17,0 mm (2,0 mm) [10].
Figure 2.1: Sagittal section of the larynx, in which the relative positions of the larynx and the vocal cords in the trachea are visible [6].

Figure 2.2: Coronal section of larynx with the vocal cords which shows the multi-layered structure in figure 2.2b.
2.2 Vocal cord cancer

Cancers that start in the region of the glottis, are called glottic cancers. They are often reported together with the supraglottic and subglottic cancers as laryngeal cancer. The laryngeal cancer incidence for Belgium from 2009 until 2011 is listed in table 2.1 from which you can see that men will form the largest group of patients; for each woman diagnosed with laryngeal cancer, there are 5 men. The share of laryngeal cancer in the total amount of cancer incidence is relatively small, maximum only 1.59% [11].

According to a report in 2008 of the French Sanitary Surveillance Institute, half of laryngeal cancers involve the vocal folds. It can happen that only one fold is affected. The majority of glottic cancer patients is diagnosed with squamous cell carcinoma. In 90% of the cases of glottic tumours, the type is malpighian carcinoma mostly in a well or moderately differentiated form [12].

Usage of tobacco is the main risk factor for glottic cancer. In comparison to other laryngeal cancers, alcohol is less implicated. However, due to mucosal irritation, the carcinogens of the smoke are able to penetrate deeper. These risk factors could explain the difference of incidence rate between the sexes as they are more common in men. By changing habits, however, women will become more at risk [6, 12].

Compared with other cancers in the head and neck region, the survival rate of laryngeal carcinoma is high [14]. One reason is that early diagnosis, due to an early change in voice quality, provides an important survival advantage and a glottic cancer produces an early change in voice quality. The hoarseness warns the patient and the physician. The second reason for the high survival rate of cancers starting at the true vocal cords is that the lymphatics are sparse. This is beneficial because only in 4-6% there is metastatic spread. A supraglottic carcinoma is often detected later and appears at a more advanced stage because there is no early change in voice quality [14].

Glottic cancer is classified according to the criteria developed by the International Union Against Cancer (IUCC) into an TNM classification (See table 2.2), the tumour and the degree of extension is hereby scored. When TNM is prefixed with a ‘c’, the classification is clinically based and with a ‘p’ anatomopathologically.

Depending on the size, location and stage of the tumour, different treatments are possible. While it is important to control the tumour, conservation of the voice quality or ability is an important factor too. Surgical laser excision and radiation therapy have a good probability of

<table>
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<tbody>
<tr>
<td></td>
</tr>
<tr>
<td>men</td>
</tr>
<tr>
<td>women</td>
</tr>
</tbody>
</table>
Chapter 2. Vocal cords

Table 2.2: UICC TNM Classification for the Glottis, 7th edition [6, 12].

<table>
<thead>
<tr>
<th>T</th>
<th>staging criteria for carcinoma of the larynx with glottis as primary tumour</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>No evidence of primary tumour</td>
</tr>
<tr>
<td>1a</td>
<td>Tumour limited to the one vocal cord(s) (may involve anterior or posterior commissure) with normal mobility</td>
</tr>
<tr>
<td>1b</td>
<td>Tumour limited to both vocal cord(s) (may involve anterior or posterior commissure) with normal mobility</td>
</tr>
<tr>
<td>2</td>
<td>Tumour extends to supraglottis and/or subglottis and/or with impaired vocal cord mobility</td>
</tr>
<tr>
<td>3</td>
<td>Tumour limited to the larynx with vocal cord fixation and/or invasion of paraglottic space and/or inner cortex of the thyroid cartilage.</td>
</tr>
<tr>
<td>4a</td>
<td>Moderately advanced local disease</td>
</tr>
<tr>
<td>4b</td>
<td>Very advanced local disease</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>N</th>
<th>spread to nearby lymph nodes in the neck</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>No regional lymph node metastasis</td>
</tr>
<tr>
<td>1</td>
<td>Metastasis in a single ipsilateral lymph node, 3 cm or less in greatest dimension</td>
</tr>
<tr>
<td>2a</td>
<td>Metastasis in a single ipsilateral lymph node, 3 cm &lt; greatest dimension ≤ 6 cm</td>
</tr>
<tr>
<td>2b</td>
<td>Metastases in multiple ipsilateral lymph nodes, greatest dimension &lt; 6 cm</td>
</tr>
<tr>
<td>2c</td>
<td>Metastases in bilateral or contralateral lymph nodes, greatest dimension &lt; 6 cm</td>
</tr>
<tr>
<td>3</td>
<td>Metastasis in a lymph node, &gt; 6 cm in greatest dimension</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>M</th>
<th>metastasis</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>No distant metastasis</td>
</tr>
<tr>
<td>1</td>
<td>Distant metastasis</td>
</tr>
</tbody>
</table>

restoring normal voice quality, but may only be indicated as the only exclusive treatment in low staged (T1,T2). More persistent tumours could require corpectomy or vertical hemilaryngectomy. Tumours that cause fixation of the vocal cords require chemotherapy followed by radiation therapy. Total laryngectomy is indicated in more advanced staged glottic tumours (T4), or with tumours with subglottic extension[12, 14].

2.3 Vocal cord irradiation

If the lesion is strictly glottic (T1), in which the lymphatics are spare, only the target tumour volume is involved in the radiation. It can be irradiated with 2 opposing parallel fields of 5 x 5 cm₂ or 6 x 6 cm₂ [12]. In case of supraglottic or subglottic extension the radiation can be extended to some lymph-node areas [12].

For the treatment of locally advanced laryngeal cancer(T4), an anterior supravacuicular field was added to the set up of two lateral fields, but it did not allow the sparing of normal tissue.
Also 3D conformal radiotherapy led to poor patient quality of life [16].

Those conventional methods obtained good local control rates but the thyroid and paratoid glands, thyroid and cricoid cartilages, carotid arteries and the swallowing muscles among other structures are also exposed to high radiation doses, which could lead to an increased probability of complications [13].

Intensity modulated radiotherapy (IMRT) is now indicated to modulate the received dose and to spare normal tissue [12, 15]. Another advantage of IMRT is that tumour dose escalations become possible because of the possibility to define different dose levels within the target volume. The main concern for the usage of IMRT has been that no adequate dose to a small primary cancer would be delivered, where it cannot be missed by conventional methods. Another concern but one that is also applicable to the conventional approaches, is that of the limitations of the dose calculation algorithms. This can result in over- or underdosage because of the presence of tissue-air interfaces [13]. Image guided radiotherapy (IGRT) and adaptive radiotherapy (ART) even allow to visualize the shrinkage and to replan during the therapy.

Those new techniques may be effective in patients with locally advanced laryngeal carcinoma for anatomic and functional laryngeal preservation, but severe dysphagia and aspiration restrict the quality of life. It is most likely due to excessive radiation dose to the muscles of the pharynx [16].
Chapter 3

Dosimetry of vocal cords

This chapter will handle the difficulties in measuring and calculating the doses to a small structure with a tissue-air interface. In clinical practice, a treatment planning system (TPS) is responsible for the accurate calculation of dose. Treatment planning systems from different companies use different models and algorithms. In this thesis, Pinnacle³ (version 9, Philips Healthcare, Best, The Netherlands) and Monaco (version 3.20.01, Elekta, Atlanta, USA) were used. Radiochromic film will be described in his function as an absolute dosimeter.

3.1 Electronic disequilibrium

The basic principles of photon and electron interactions and dose delivery are not the scope of this thesis, information about it can be found in literature [1, 2]. However, the principle of electronic disequilibrium or charged particle equilibrium (CPE) is of special interest as it is an important factor at tissue-air interfaces and in build-up regions.

CPE is the condition in which the assumption is made that the energy carried in and out of a volume by charged particles (CP) is equal. The absorbed dose depends on the energy deposited in that volume, which can be described as the difference in the energy entering ($\epsilon_{in}$) and exiting ($\epsilon_{out}$) the volume [1]:

$$\epsilon = \sum \epsilon_{in} - \sum \epsilon_{out}$$  

(3.1)

$$= \sum (\epsilon_{in})_{photons} - \sum (\epsilon_{out})_{photons} + \sum (\epsilon_{in})_{CP} - \sum (\epsilon_{out})_{CP}$$  

(3.2)

In equation 3.2 the last two terms cancel each other out when CPE is assumed.

This assumption is only valid if the photon fluence is constant. (See figure 3.1(a)) The photon fluence however, decreases with depth and thus also the number of charged particles which can enter the deeper layer. Physically a CPE does not exist at any point. A transient CPE exists where the dose at a certain point can be thought of as the spatially shifted collision kerma. (See figure 3.2(b)).

In the build-up region, the dose increases -instead of decreasing exponentially- because of the imbalance of the charged particles. The accumulation of dose ends when the secondary charged particles (mostly electrons) balance with the attenuation of the photons [1].
Figure 3.1: Absorbed dose as a function of depth. Physically a CPE does not exist as the photon fluence decreases with depth. [2]

Lateral charged particle disequilibrium is significant in a few situations and can affect the dose distributions when (i) high photon beam energies are used, (ii) for small field sizes, (iii) for low density media and (iv) at air-tissue interfaces [1]. For this thesis (iv) is definitely present.

3.2 Calculation algorithms

The accuracy of a TPS depends i.a. on the type of method and the algorithms it uses. The first generation of computerized methods are called correction-based methods. Tabulated data, for different field sizes, generated in water are stored in the TPS and are the basis for calculating the dose. They are also referred to as beam library systems. By Clarkson summation or other similar techniques the dose is calculated and corrections are performed for beam modifiers, the patient outline and heterogeneities. Those techniques are relative fast compared with the model-based methods but have a low accuracy [17].

Model-based methods model the beam, also based on data taken from specific measurements, into a photon energy fluence and calculate the dose by using convolution or superposition techniques with a specific kernel. This way the radiation transport is modelled more explicitly [17].

Another approach to predict the dose is to simulate every physical interaction of the photons and electrons in a Monte Carlo (MC) algorithm, an accurate dose calculation is hereby proven in literature. Parts of this concept are used in Monaco.
3.2.1 Collapsed Cone Convolution of Pinnacle

In the Pinnacle\(^3\) 3D TPS multiple algorithms can be chosen, this thesis focuses on the collapsed cone convolution (CCC) algorithm. The calculations are done in different steps. The first part models the incident energy fluence coming from the accelerator head. In the next part, the patient density representation is used for the computation of the total energy released per mass (TERMA). For the third part the TERMA is convolved with a kernel and the superposition is taken. Electron contamination is incorporated in the last part [18].

The first part handles the computation of a energy fluence distribution, the photon fluence model needs to be adjusted for the different components of accelerators head, e.g. the flattening filter [18].

TERMA represents the energy of the primary photons imparted to secondary charged particles and retained by the scattered photons in a unit mass. It can be thought of as the energy lost out of the primary beam in a unit mass or thus TERMA \(T\) at a certain point \(r\) is the negative of the divergence of the energy fluence \(\Psi\). See equation 3.3, the lost energy of the fluence distribution equals the attenuated energy fluence (equation 3.4) [1].

\[
T(r) = -\frac{1}{\rho(r)} \nabla \cdot \Psi(r) \quad (3.3)
\]

\[
= \frac{\mu}{\rho}(r) \Psi(r) \quad (3.4)
\]

With \(\mu\) the attenuation factor. Thus in the second part of the model, a projection of the incident energy fluence plane through a CT (Computed Tomography) patient representation is made, the attenuation is handled using the mass attenuation coefficients. Pinnacle claims that patient heterogeneities are taken into account with the density dependence of the mass attenuation coefficient [18]. With aid of this projection the TERMA is computed making use of a ray-tracing technique in which they calculate the attenuated energy fluence for each voxel in the path of a ray.

The TERMA is then convolved with a kernel in order to calculate the absolute dose per energy fluence in a medium [19]. Those kernels describe the fractional energy imparted per unit volume at a single point[20]. Depending on the implementation of the convolution algorithm, the kernels are referred to as e.g. dose spread functions or energy desposition kernels [1]. Such a kernel can be a combination of a series of mono-energetic energy deposition kernels which are once generated using Monte Carlo methods [18]. The CCC algorithm uses a kernel with a scaling factor for both the electron part of the kernel and the photon part in order to scope with heterogeneities in the medium. For the electron part the ratio of mass stopping powers of the medium to water is used, and for the photon, the ratio of mass attenuation coefficients of the medium to water [20]. By a superposition each dose deposition contribution is added and the dose deposition for the patient volume is calculated. When multiple beams contribute to the dose, they are calculated independently and added together with the correct beam weight in
order to compute the entire dose distribution [18]. The calculation time of CCC algorithms only increases with $\sim N^3$ ($N$, the number of voxels along one side of the calculation volume) while others have $N^6$. This makes it a relative fast algorithm [21, 1]. The Fourier convolution (Solving the convolution in the frequency domain.) can be compared with respect to computation time but does not allow kernel scaling and the associated accuracy advantage [1].

### 3.2.2 XVMC Dose Engine of Monaco

Monaco is a commercially available treatment planning system. For the final dose calculations, after the optimization which uses a fast pencil beam calculation, a modified Monte Carlo (MC) calculation is used, the XVMC Dose Engine. This dose calculation can be described in three parts: the virtual source model (VSM), the transmission filter and the patient model [22]. The X-ray Voxel Monte Carlo (XVMC) is used as the basis framework, the XVMC code is modified from the Voxel based Monte Carlo (VMC) code, initially developed by Kawrakow and Fippel in order to calculate electron beam doses [22, 23].

The Monte Carlo method means relying on repeated random sampling in order to obtain useful simulations [24]. Monaco uses a pseudo random number generator to simulate the particles. The weight of each virtual source, determined by a modelling process, is hereby respected. The position of the particle, its energy, its source (primary/secondary photon or electron) are all factors that needs to be determined [22].

**Virtual source model**

The virtual source model is implemented to decrease the computation time. The VSM acts as the source of the three types of radiation dose carriers: primary photons, the secondary photons (scattered) and the electron contamination. The VSM is based on in-water measurements, specific for each machine. The flattening filter is present by an analytic representation, the energy dependency is hereby included. Also the leaves and jaws are represented, they are modelled through three dimensional transmission filters [22].

**Transmission filter**

The transmission of the particles through the primary collimator, the leaves and the jaws is a very time consuming process when MC calculations are performed, due to the highly attenuating materials. In order to decrease computation time, probability-based transmission maps are used to track the created particles. They are based on divergent projections and the filter does not modify the energies of the particles, nor generate secondary particles or alter the incident angle. If the transmission value of the leaves and/or jaws are lower than or equal to 0.011, they are modelled as fully absorbing. A 3D representation of the probability transmission filter is implemented in order to improve the penumbra. A random number is used here to calculate if a particle passes through the filter. The filter does not modify the energy of the primary photons
nor does it generate secondary particles. This is because the secondary photon source bypasses the filter [22].

Patient model and XVMC

The last part is the patient model which uses the XVMC method. The patient CT data are converted to relative electron density via tabular input and then into mass densities by a function. These mass density data are used in the MC calculation to model the transport of photons and electrons. The size of the voxels is very important. It influences the calculation time but also the dose accuracy that can be obtained in high gradient regions. Depending on field size ranges of 1 up to 5 mm are typical [22].

XVMC use MC calculations but it applies some simplifications in order to decrease calculation time in Monaco. Only rectangular geometries generated from the CT data are used. The energy ranges where the XVMC model can be applied is limited from 1 to 25 MeV. It is only valid for low Z materials, defined by the density range of 0-3 g/cm$^3$. Those limitations apply because some of the used equations would otherwise introduce errors due to the simplifications made in it or they are made to decrease calculation time; An energy cut-off for electrons is set to be 500 keV. An energy cut-off for photons is also included and set to 50 keV, when a photon is below this limit the dose is added to the current voxel location [22].

The probabilities of the photon interactions are based on the attenuation coefficients and electron densities from the ICRU Report 46. Probability density functions sample the distance between 2 interaction points. The interaction type is determined by the ratio of the interaction attenuation coefficient and the total attenuation coefficient. The considered interaction types are Compton interaction, pair production and the photoelectric effect.

In a homogeneous region, Monaco assumes that the electron transport is independent as it takes place far enough each other. The electron history for 2 mono-energetic beams are reused. A translation is made to other locations in order to decrease calculation time. To translate that history to a heterogeneous medium the path lengths and scattering angles has to be adapted taking the different density in mind [22].

3.3 Radiochromic film

Radiation sensitive film is one of the oldest detection method of X-rays. Prof. Goodspeed made already in 1890 a picture of 2 coins, but only in 1895 Wilhelm Röntgen discovered the origin of the film exposure. The X-ray film of his wife’s hand became famous [25]. They used radiographic film, which consists of a radiosensitive emulsion containing silverhalide crystals and a base. Ionizing radiation initiates the formation of an elemental silver atom. The film needs to be developed. In this process only the remaining silverions which are in a crystal where elemental silver is present, will get reduced. The undeveloped crystals are washed away during the fixation and only the irradiated crystals remain behind and appear black [1].
In this thesis however, radiochromic film is used for the experimental dosimetry part. Usage of radiochromic film was first described in 1988 [26].

3.3.1 General properties of radiochromic film

Radiochromic film exists of a thin active radiochromic layer, surrounded by layers of laminating polyester. The active layer contains a special dye which gets polymerized when exposed to radiation. Radiochromic film does not need a chemical developer and fixer (like radiographic film) nor any thermal or physical developmental procedure. It is self-developing.

For this thesis the measurements were done with a radiochromic film (International Speciality Products (ISP) Ashland Inc. Gafchromic® R⃝, type EBT3, Lot #A07251101). This is a symmetrical film, which consists of an active substrate layer of approximately 30 μm, laminated at both sides with a clear polyester. The approximate total thickness is 285 μm. The active monomer in this film is the lithium salt of pentacosa-10,12-diynoate [27].

Due to the polymerization, a colouration, for EBT3 from yellow to green, occurs. Radiochromic film dosimetry is based on the assumption that the absorbed dose in the film is reflected in the resulting optical density (OD) of that film. The OD is defined as follows:

\[ OD = \log_{10} \frac{I_0}{I} \]  

Where \( I_0 \) represents the intensity of the undisturbed light and \( I \) the intensity of the film-attenuated light. The 2D OD map can be converted into a dose distribution by using a calibration curve. In order to do absolute dosimetry it is important to eliminate systematic errors: corrections of the inhomogeneity of the film itself and a correction map for the scanner bed. A calibration curve for the dose conversion need to be part of the applied protocol.

A short summary of the most important features for a film dosimeter is hereby presented, based on [1, 2, 28].

Accuracy

Accuracy stands for the ability to measure the dose correctly. It makes it one of the most important features when applying absolute dosimetry. It can be limited by systematic and stochastic errors. To reduce stochastic errors it is recommended to do multiple measurements to minimize this influence [1]. In order to limit systematic errors the usage of a good well-thought protocol can help. Measurements will always have inaccuracies, this is characterized as uncertainty. In radiotherapy they divide them in 2 types: type A which can be evaluated statistically and type B which rely on intelligent estimations of the uncertainties related with the measurement.
Chapter 3. Dosimetry of vocal cords

Precision

The consistency of a measurement can be expressed as the precision. Can the results be reproduced under similar conditions? The fluctuations around a mean is a good indicator of precision, where a high precision is associated with a small standard deviation.

Measurement range

The lowest dose which can still be detected determines the lower detection limit. Natural background fluctuations and instrument noise have an influence in this respect. In radiotherapy, it is of smaller interest because of the relatively high doses [1]. For the upper limit the saturation of dose readings will have a high impact. Radiochromic film has a wide dose range [28]. Ashland claims that the EBT3 has a dose range from 1cGy to 40 Gy [29].

Dose response

The measured parameter should be linked with the dose. Ideally the reading of an additional given dose should be independent of any dose which is already read out. For radiochromic film, the dose is not perfect linear with the OD. A calibration curve needs to be made. Ashland indicates to use the red channel of an RGB image for doses up to 8 Gy as the OD is depended on the wavelength of the used light [29].

Dose rate response

The delivery rate of dose should not be able to influence the registration of dose. This is important because linear accelerators deliver dose in short pulses. It is found that the radiochromic films EBT and EBT2 are relatively dose rate independent [30].

Energy dependence

The dose response should be independent of the photon energy spectrum and the beam quality. This is however not realistic, therefore the dosimetry system has to be calibrated at the same radiation beam quality. It is also important that the effective atomic number of the film is similar to the material under investigation [28].

Spatial resolution

We are interested at the dose in a specific region, as dose is a point quantity, we want a good dose distribution with a good spatial resolution. This will depend on the intrinsic resolution of the film and the subsequent processes. For radiochromic film this is only determined by the scanning. A big advantage of a film dosimeter is the mapping ability; the dose over a region can be measured and compared, while with the usage of a traditional ionization chamber only 1 point measurement is allowed.
Ease of handling

The development of radiographic film is much more complex compared with radiochromic film. Radiochromic film requires no chemical processing, but and because of the post-irradiation colouration there needs to be a minimum amount of time between the irradiation and the digitalization of the film. A film dosimeter is however not reusable and for each measurement, new film is needed.

It is important to have a good protocol in order to obtain accurate absolute doses. Like already mentioned systematic errors should be eliminated. The non-uniformity in film response, the directional dependence of scanning and non-uniformity for the scanner bed, are a few examples to consider.

3.3.2 Value of radiochromic film dosimetry around air cavities

Tumours situated in the region of an air-tissue interface could get an under dosage, due to the fact that some TPS overestimate the dose in those regions. This is indicated in different papers [31, 32, 33, 34]. Film measurements where the film contains the central beam axis, however, show an underestimation of the dose. A study of Suchowerska explains this by potential air gaps that are introduced in slab phantoms. They propose to angle the beam by $2^\circ$ to the plane of the film or adapting the phantom design [35].

The value of radiochromic film dosimetry around air cavities was investigated by L. Paelinck in 2003 where the measurements were compared against Monte Carlo simulations. The study uses a polystyrene phantom containing an air cavity. The film intersected centrally the air cavity. Two cases were considered. (1) The film contained the beam central axis. (2) By offsetting the cavity an angle of $\sim 1.1^\circ$ between the beam direction and the film plane was ensured. An underestimation of the dose by the film measurements behind the cavity was detected when the film contained the central beam axis. In the second case, the underestimation disappeared [36].
Chapter 4

Methodology

4.1 Experimental dosimetry

The experimental determination of absorbed dose was performed by radiochromic films positioned in a polystyrene slab phantom.

4.1.1 Phantom

Material

The phantom is made of polystyrene or poly(1-phenyl-ethylene) with a density of 1.04 g/cm\(^3\). (See table 4.1 for other characteristics.) Measurements in polystyrene are done in clinical practice as a user and time friendly substitute for measurements in water. It is relatively inexpensive and the good machinability of polystyrene slabs allows the creation of anthropomorphic phantoms. Polystyrene is a good insulator, this can cause charge accumulation in the material which could alter the dose registration. For MV photon beams the charge accumulation is only of concern in the build-up region.[1]

Geometry A

The geometry of the polystyrene phantom served as a simplistic model of the larynx region. For this purpose, a cylindrical phantom with a cylindrical cavity was chosen. The dimensions of the

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Description</th>
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<tr>
<td>Composition ((C_8H_8)_n)</td>
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<tr>
<td>Density</td>
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</tr>
<tr>
<td>Weight composition</td>
<td>H: 0.077</td>
</tr>
<tr>
<td></td>
<td>C: 0.923</td>
</tr>
<tr>
<td>Effective atomic number</td>
<td>5.7</td>
</tr>
</tbody>
</table>

Table 4.1: Characteristics of polystyrene[1].
Figure 4.1: Overview and transverse cross-section of the phantom. The width of the air cavity Q changes along the z-axis (cranial-caudal), this can be seen in figure 4.2. The 2 symmetric extremal circles are there for the threaded rods in order to hold the slabs together. [Dimensions are expressed in mm.]

The trachea and vocal cords were based on literature, as described in chapter 2.1. The geometry includes a narrowing which models the vocal cords (see figure 2.2b). The dimensions of the narrowing are based on data used in the study of Xue et al.[5].

More practically, it is a slab phantom where all the different sections are held together with plastic threaded rods. All the slabs are oriented in the transversal plane. The shape of each cross-section is cut laser-guided in plates of polystyrene. The thickness of the plates varies between 1 mm - 2 mm - 5 mm - 10 mm - 20 mm. The cylindrical phantom is placed on two foam pads for stability reasons. The foam pads are made of styrofoam, extruded polystyrene, with a density of 0.02-0.07 g/cm\(^3\).

The total length of the phantom is ~25 cm. In figure 4.1 and 4.2 the chosen dimensions in mm are indicated. Figure 4.1 shows one cross-section, it has a diameter of 140 mm and at the top a circular cutout of 20 mm in diameter to model the trachea. The size of Q varies between 20 (=full circle) to 17 mm - 14 mm - 11 mm - 8 mm at the position of the narrowing, each slice in the narrowing has a thickness of 2 mm. The total length of the narrowing is then 8 mm. (See figure 4.2.) On the cranial edge of this narrowing, two full transversal slabs (without cutout, with a total thickness of 4 mm) model the vocal cords when closed.

**Geometry B**

The second geometry only differs in the thickness of the two full transversal slabs cranial to the narrowing. Instead of two slabs of 2 mm, one full slab of 1 mm is chosen, in order to better represent the thin structure of the unaffected vocal cords (while geometry A is better to model stages T2-T4).
4.1.2 Radiochromic film measurement

The characteristics of the used radiochromic film EBT3, can be found in section 3.3. Here, the protocol used in this thesis is explained.

It is important to have a good protocol and to be consequent when radiochromic film is used for absolute dosimetry in order to minimize the uncertainty of the measured dose.

1. Cutting of film for the measurements and the calibration curve.
2. Pre-irradiation scanning of the film.
3. (A dosimetry check of the linear accelerator.)
4. Irradiation of the film pieces for the calibration.
5. Irradiation of the phantom with the measurement film.
6. $\Delta t$: waiting for at minimum 36 hours.
7. Post-irradiation scanning of the film.
8. Further analysis in Matlab.

As this study only concerns the dose distribution in and near the vocal cords, templates (see figure 4.3) are used in the phantom in order to minimize the amount of unused film. The template is cut from radiochromic film and thus has the same thickness and composition as the measurement films. This helps to avoid air gaps and to have a consistent and good positioning. These templates can be reused for all the different measurements.

A number of measurement films needs to be cut in isosceles trapezoid with sides of 3 and 5 cm and with a height of 5 cm. The orientation of the film needs to be respected as part of the
protocol. This has to be done as the scanning is film orientation angle dependent. Because we use the netto transmission difference of the film, two marks need to be made in order to match the pre- and post-irradiation scan for every film piece. The films are marked in two opposing corners and at the side.

Seven squares with sides of 3.5 cm need to be cut in order to obtain the calibration curve. Also here it is needed to indicate the orientation of the sheet on each piece.

For the film scanning or digitization, the flat bed scanner Epson® Expression 10000XL (Seiko Epson Corporation, Nagano, Japan) is used. In order to minimize variation in positioning between measurements the films are placed on the scanner bed with the aid of a template. Each film is scanned 9 times, but only the last 3 images are used for the analysis. The first scans are aberrant, as the measured transmission might depend on the temperature of the scanner light source [30]. The films are scanned in positive transmission mode, with no color corrections at a resolution of 100 dpi. The software that is used for the scanning is EPSON scan v3.04D (Seiko Epson Corp., Nagano, Japan).

The output of the linear accelerator was checked using a calibrated ionisation chamber and electrometer. It is irradiated with a 6 MV photon beam with a field size of 10x10 cm$^2$ with 200 monitor units (MU) at a depth of 10 cm of polystyrene (plates of 30x30 cm$^2$), source surface distance (SSD) is 90 cm. The measurement is corrected for room temperature and pressure, it is repeated 3 times and the mean used. This is only done for the measurements 1 and 2.

In order to obtain a calibration curve, the 7 pieces of film are irradiated. The film is placed on top of a 10 cm layer of polystyrene plates and on top of the film again 10 plates of 1 cm are placed. SSD is 90 cm. The film is irradiated with 0, 29, 70, 128, 201, 290 and 396 MU and a field size of 10 x 10 cm$^2$. 1 MU corresponds at the isocenter, 10 cm below the water surface, to 1 cGy for a 10 x 10 cm$^2$ for a source-to-surface distance of 90 cm.
For the irradiation the film is placed in the template (see figure 4.3). The location of the template depends on the measurement. The template is fixed with tape to the phantom and the film is gently fixed in the right bottom corner with 3M-Micropore tape. The lines of the isocenter are marked at every side in order to process the film more accurately. The rest of the phantom is put in place and the whole phantom is tightened to avoid that air gaps influence the measurement.

Post-exposure polymerization requires to wait at least 36 hours after the exposure. The films are kept in envelopes to avoid light contamination. The pre-irradiation scanning procedure is repeated with the same settings for the post-irradiation scanning.

The analysis of the obtained data of the scanner is processed in MATLAB\textsuperscript{R} (version 7.12, The MathWorks, Inc., Massachusetts, USA). Like already mentioned, the films for the calibration and the measurements were scanned 9 times. But only the last 3 images were used to average the read out. This read out was converted into the optical density of the red channel. A correction was made for sensitivity ununiformities over the scanner bed. The pre- and post-irradiated scanned films were matched according to their corner marks and the netto OD was obtained. The calibration data were fitted to a third degree polynomial to obtain the dose as function of netto OD. The dose for the measurement films was obtained by applying this calibration curve. Each measurement was repeated 3 times, thus for each measurement condition 3 sets of film were obtained. Every measurement film was centered according to the isocenter laser lines. The last step was to take the average of those 3 repetitions for each film position.

4.1.3 Set-up

In this study 4 measurements were done and compared with dose calculations of 2 treatment planning systems. The measurements differ in the thickness of the modelled vocal cords, the film configuration and in the beam set-up. See table 4.2 for an overview. For the largest thickness, two film configurations were used. For the smallest thickness, two treatment plans were used.

The isocenter remained at the same position during all the measurements; On the symmetry axis, 5 mm posterior the air cavity and 6 mm inferior the first solid slice, see figure 4.4. The isocenter is indicated at the outside surface of the phantom in order to have a good and accurate positioning.

For all the measurements, the phantom was irradiated with the Elekta Synergy linear accelerator with the MLCi2 (Elekta, Crawley, West-Sussex, UK) at the Ghent University Hospital.

For measurement 1, phantom geometry A and one film is used caudal of the full slab. For the exact positioning of the film see figure 4.5.

For measurement 2 also the phantom geometry A is used, but 5 layers of film are used instead of one. Those 5 layers substitute the most caudal of the 2 solid slabs. For the exact positioning of the films see figure 4.6.

The main difference with the previous measurements is that only one slice of 1 mm (instead of two slices of 2 mm) + the narrowing is used to model the vocal cords in measurement 3. Two
Table 4.2: Overview of measurement parameters

<table>
<thead>
<tr>
<th>Measurement number</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thickness of modelled vocal cords</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Geometry A: 4 mm</td>
<td>x</td>
<td>x</td>
<td></td>
<td></td>
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<tr>
<td>Geometry B: 1 mm</td>
<td></td>
<td></td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td>Film configuration</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1 film caudal</td>
<td>x</td>
<td>x</td>
<td>x</td>
<td></td>
</tr>
<tr>
<td>5 stacked films caudal</td>
<td></td>
<td>x</td>
<td></td>
<td></td>
</tr>
<tr>
<td>1 film cranial</td>
<td></td>
<td></td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td>Beam set-up</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1 beam, 6x6 cm², gantry angle 90°</td>
<td>x</td>
<td>x</td>
<td></td>
<td></td>
</tr>
<tr>
<td>IMRT plan with 6 beam directions</td>
<td></td>
<td></td>
<td></td>
<td>x</td>
</tr>
</tbody>
</table>

Figure 4.4: Schematic representation of the set up. The isocenter is indicated in red. The beam direction and field used in the first three measurements, are indicated in blue. The coordinate system that is used in this thesis is represented in the left bottom corner. [Dimensions are expressed in mm.]

film positions were used, one caudal and one cranial that slice, as illustrated in figure 4.7.

The phantom was each time irradiated with a photon beam of 6 MV and a field size of 6x6 cm² at the isocenter. Gantry was set at 90°, the collimator rotation angle was at 0°. 200 MU were given. The whole set up is visualised in figure 4.4.
Chapter 4. Methodology

Figure 4.5: Narrowing and film position for measurement 1. The position of the film is indicated with a green line. [Dimensions are expressed in mm.]

Figure 4.6: Narrowing and film positions for measurement 2. The positions of the 5 film layers are indicated in green. [Dimensions are expressed in mm.]

Figure 4.7: Narrowing and film positions for measurements 3 and 4. The positions of the films are indicated with green lines [Dimensions are expressed in mm.]

The goal for measurement 4 was to see if the differences that would appear in the previous measurements also were present in a more clinical treatment plan. The phantom set up of the third measurement is used (see figure 4.7). A clinical class solution comprising a coplanar
step-and-shoot IMRT treatment, was applied. The IMRT plan was created in Monaco and optimized for a randomly selected patient with vocal cord cancer. The patient had a low differentiated invasive squamous cell glottic carcinoma, staged as cT1a cN0 M0 (right glottis). Some parameters of the IMRT plan are listed in table 4.3. The phantom was positioned according the described isocenter.

<table>
<thead>
<tr>
<th>Table 4.3: Some parameters of the IMRT plan</th>
</tr>
</thead>
<tbody>
<tr>
<td>Beam</td>
</tr>
<tr>
<td>------</td>
</tr>
<tr>
<td>1</td>
</tr>
<tr>
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</tr>
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<tr>
<td>4</td>
</tr>
<tr>
<td>5</td>
</tr>
<tr>
<td>6</td>
</tr>
<tr>
<td>$\Sigma$</td>
</tr>
</tbody>
</table>

4.2 Dose calculations

The isocenter as defined in figure 4.4 and commented in section 4.1.3 was used for all the calculations in the two treatment planning systems.

4.2.1 CT-scan

The CT images of the phantom (fig. 4.8) were acquired on a Toshiba Aquilion CT scanner (Toshiba Medical Systems, Tokyo, Japan). A tube potential and tube current of 120 kVp and 300 mA were used respectively. The TCOT Reconstruction algorithm and a slice thickness of 1 mm were used.

4.2.2 Pinnacle

The doses in Pinnacle are calculated for a cuboid of 65 mm x 80 mm x 40 mm (XYZ).

Measurements 1-3: One beam Dose calculations in Pinnacle are done with the CCC algorithm. The CT data of the phantom are imported. The phantom was manually contoured and the density of the polystyrene was overwritten with 1.000 g/cm$^3$, this was done because the CT reconstruction of the phantom was not accurate enough for the sharp edges and specific dimensions of the phantom. The density inside the contour of the air cavity was set to 0.000 g/cm$^3$. A photon beam of 6 MV with a field size of 6x6 cm$^2$ was used. The gantry was set at 90°. The minimal dose grid of 1 mm was used (= voxel size of 1 mm$^3$). 200 MU were given.
Figure 4.8: CT images of phantom. Here only 16 transverse sections are shown, they have a slice thickness of 1 mm. On the center bottom a midsagittal is shown and in the right bottom corner a coronal cross-section, they are differently scaled.

The first 2 measurements had the same thickness for the modelled vocal cords (see table 4.2), and thus their dose calculation are equal. The third differed only in the fact that the air cavity was prolonged to obtain geometry A. The beam parameters remained the same.
**Chapter 4. Methodology**

**Measurement 4: IMRT plan**  The IMRT treatment plan was thus created and optimized in Monaco, it consisted of 6 beams with different segments, see table 4.3. The plan was imported in Pinnacle. The overridden densities of measurement 3 and the minimal dose grid of 1 mm were used. The plan was calculated with the CCC algorithm of Pinnacle.

**4.2.3 Monaco**

**Measurements 1-3: One beam**  For the dose calculations in Monaco the CT images were imported, also here the polystyrene phantom was contoured and the corresponding relative electron density was overwritten with 1.000. The air cavity was also contoured and the relative electron density set to the minimum of 0.010. One 6 MV photon beam of 6x6 cm$^2$ was simulated with the gantry at 90°. A dose grid of 1 mm and the variance settings were set to 1%. The dose was set to 200 MU. Dose to medium was calculated.

Also here the relative electron density was rewritten for the third measurement, so that the vocal cords were only simulated by the narrowing and 1 mm of polystyrene. The beam parameters remained the same.

**Measurement 4: IMRT plan**  The IMRT treatment plan (see table 4.3) was optimized for a randomly selected patient. The dose was calculated in the 'QA mode’ of Monaco, where the phantom with the contours and overridden relative electron densities (1.000 for the polystyrene phantom and the minimum of 0.010 for the air cavity) was loaded. Also here the dose to the medium was calculated. The minimal dose grid of 1 mm was used and the variance settings were set to 2%. In total 349.88 MU were delivered.

**4.3 General comparison methods**

In this thesis, the radiochromic film measurements will be used to compare the two algorithms.

The computed dose distributions were compared against the film measurements using a gamma analysis. A gamma analysis uses the dose difference between a pixel with position \((i,j)\) and a reference \((k,l)\) and the distance to dose with respect to the reference pixel for the calculation of a gamma value as shown in equation 4.1. For every pixel \((i,j)\) a \(\gamma(i,j)\) is calculated and the minimum of all those gammas is given to the reference pixel \((k,l)\). This is repeated for the whole dose distribution. For this analysis, the tolerance for the dose difference (DD) was 3% and the tolerance for the distance to dose (DtD) was 3 mm.

\[
\gamma(i,j) = \sqrt{\frac{(Dose \ Difference)^2}{tolerance \ DD^2}} + \frac{Distance \ to \ Dose^2}{tolerance \ DtD^2} \tag{4.1}
\]

Percentage dose depth (PDD) curves are calculated by averaging the curves over 11 pixels along the indicated black dotted line in the dose distributions in order to minimize statistical fluctuations. They are normalized to the maximum dose along the PDD calculated by Monaco.
Chapter 5

Results

5.1 Measurement 1

The results of the radiochromic film measurement are shown in figure 5.1a, the mean standard deviation (SD) on the measurement data is 1.7% for the area indicated with the gray rectangular (2.54 cm x 3.81 cm). The white dots indicate the geometry of the nearest phantom slice, for measurement 1, this is the slice caudal to the film. The shape of the film is explained in the previous chapter. Figures 5.1b and 5.1d respectively show the dose calculated with the CCC algorithm of Pinnacle and the XVMC Dose Engine algorithm of Monaco. (Whenever 'Pinnacle' is used in this chapter, the CCC algorithm is used, the same is true for 'Monaco' and the XVMC Dose Engine, unless stated otherwise.)

Figure 5.1a is the reference for this comparison for the first measurement. In the region of the air cavity the dose drops down (from the right to the left) and increases again quite fast in the rebuild-up. At the top of the film the penumbra of the beam becomes visible.

In the dose distribution of Pinnacle (fig. 5.1b), there is only the top of the air cavity a very small dose extremum along the x-axis. Overall there is no secondary dose build-up.

Figure 5.1d demonstrate the dose calculated with Monaco. The dose is affected more abruptly by the air cavity than in Pinnacle. The dose increases again after the region of the cavity, so there is again a dose build-up region.

The gamma analyses are shown next to their dose distributions in the figures 5.1c and 5.1e. Those figures show us that the agreement between our reference and the Monaco calculation is much better than that between the calculation of Pinnacle and our reference. All the pixels which exceed the tolerance of 3mm/3% and thus having a $\gamma > 1$ are light coloured in figures 5.2a and 5.2b. For the Pinnacle calculation 85.29% of the pixel values passes that criteria, while for the Monaco calculation 99.05% of the pixel values passes. The disagreement of the Pinnacle dose is mostly situated in at the surface of the air-phantom interface. Also in the region after the air cavity there are more excesses of the tolerance levels, thus pixels having a gamma $\gamma > 1$. The first half (positive side) for the region 90.74% $< 1$, while for the second half (negative side) this is 65.36%.
In figure 5.3 the PDD curve along the X-axis at $Y = -1.5$ cm is shown. The coordinate system of figure 5.1 is used. The maximum dose calculated by Monaco along the PDD was here 257.12 eGy.

It is clear that the effect of the presence of the air cavity is limited for the dose distribution of Pinnacle. Around the edge ($X = 0.4$ cm) a difference in the slope of the dose can be seen. At the border of our measurement film the difference with Monaco increases a bit. Also at the end of the air cavity ($x = -0.4$ cm) Monaco gives lower doses than the film measurement.

The mean differences between the PDDs are shown in figure 5.4. The errorbars indicate the SD over the dose differences in the indicated region. Monaco remains within 1.5% of the film measurement, Pinnacle overestimate the dose in the region around the vocal cords in this case over the whole line.
(a) Dose (cGy) measured with radiochromic film. Transversal cross-section, the beam came from 90° (right), region of grey rectangular is used for the gamma analyses (c and d), the dotted black line indicates the position of the PDD of figure 5.3. At every side the position markers indicate the isocenter lines.

(b) Dose (cGy) calculated with Pinnacle

(c) Gamma analysis (3%/3mm) of Film-Pinnacle

(d) Dose (cGy) calculated with Monaco

(e) Gamma analysis (3%/3mm) of Film-Monaco

Figure 5.1: Dose distributions and gamma analysis of measurement 1. The gamma analysis was restricted to the rectangular region indicated in panel a. The isocenter is situated along the axis’ origin.
(a) Pinnacle: All pixels with $\gamma > 1$ are light.

(b) Monaco: All pixels with $\gamma > 1$ are light.

(c) Histogram of the gamma values. For Pinnacle 85.29\% passes the criteria, while for the Monaco this is 99.05\%.

Figure 5.2: Overview gamma analysis for measurement 1
Figure 5.3: Percentage dose depth curve (PDD) of measurement 1 at Y = -1.5 cm, indicated with the black dotted lines of figures 5.1a, 5.1b and 5.1d.

Figure 5.4: Dose differences divided per region along the PDD curve. In the region of the air cavity the Pinnacle calculation and the film measurement differ with 4.7%. Monaco shows for the regions around the air cavity an overdosage while before and behind an underdosage is seen.
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5.2 Measurement 2

The doses of the second measurement are shown in figure 5.5. As explained in the previous chapter, the difference with measurement 1 is that one slice of the phantom is changed into an equal number of film layers. The dose distributions on position A, the mean of layer 1 and 2 or thus the two most caudal, should in that matter correspond with those of measurement 1. Position B, the mean of film layers 3, 4 and 5, does not have an immediate contact with the air cavity.

In figure 5.6 the dose distributions for position A are shown. The mean SD on the measurement data is 2.16% for the area indicated with the gray rectangular. However the dose calculations for position A are equal to figures 5.1d and 5.1b, they are again shown to interpret the gamma analysis more easily. The doses and gamma analyses between measurement 1 and measurement 2A matches very well.

The mean SD on the measurement data for position B (fig. 5.7a) is 2.01% for the area indicated with the gray rectangular. The presence of the air cavity can be no longer be seen very clearly. In the Pinnacle calculation (fig. 5.7b) no effect of the presence of the air cavities can be seen, this is in contrast with Monaco (fig. 5.7d) where the shape is still very well silhouetted. The gamma analyses (figures 5.7c and 5.7e) are very similar to those of position A.

The percentage of the pixel values that passes the criteria of 3%/3mm is for Pinnacle 78.17% (A) and 88.49% (B) and for Monaco 98.53% (A) and 97.87% (B). In the figures 5.8a and 5.8b you can see that there is a better level of agreement between the calculations of Pinnacle and the measurement in the slice that is surrounded by 2 completely solid structure (B) than the position with the exposed to air at the caudal side (A).

The PDDs for the third measurement are shown in figure 5.9. The indicated errorbars represent also the differences between the different film layers. Figures 5.9a is as expected similar to figure 5.3 and the same comments of section 5.1 about it are also valid here. At position B, the films were not directly situated at an air-tissue interface and nor Pinnacle, nor the measured dose show a secondary build-up. Monaco, however shows a rebuild-up of 1.6% and also dose drop of 3% when reaching the the region of the air cavity at X = 0.4 cm.

The difference between Pinnacle and Monaco for both positions in the region of the air cavity varies between 3 to 5%, see figure 5.10. Pinnacle shows a higher dose than measured with film for the whole film length. The differences of Monaco with the film measurement in the region of air cavity remains below the 2%.
Figure 5.5: Division of film layers in position A and B for measurement 2. The 5 layers of film are laterally divided in two groups. The two most caudal film layers are combined in group A and three more cranial layers are combined in group B.
Chapter 5. Results

(a) Dose (cGy) measured with radiochromic film. The transversal cross-section at position A, the beam at 90°, region of grey rectangular is used for the gamma analyses (c and d), the dotted black lines indicates the position of the PDD of figure 5.9a.

(b) Dose (cGy) calculated with Pinnacle

(c) Gamma analysis (3%/3mm) of Film-Pinnacle

(d) Dose (cGy) calculated with Monaco

(e) Gamma analysis (3%/3mm) of Film-Monaco

Figure 5.6: Dose distributions and gamma analysis of measurement 2 position A. For the gamma analysis, the region indicated in subfigure a is used. The dose calculations for this position are the same as measurement 1, but are shown to better interpret the gamma analysis.
(a) Dose (cGy) measured with radiochromic film. Transversal cross-section at position B, the beam at 90°, region of grey rectangular is used for the gamma analyses (c and d), the dotted black lines indicates the position of the PDD of figure 5.9b.

(b) Dose (cGy) calculated with Pinnacle

c) Gamma analysis (3%/3mm) of Film-Pinnacle

(d) Dose (cGy) calculated with Monaco

(e) Gamma analysis (3%/3mm) of Film-Monaco

Figure 5.7: Dose distributions and gamma analysis of measurement 2 position B. For the gamma analysis, the region indicated in subfigure a is used.
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(a) All pixels with $\gamma > 1$ are light coloured.

(b) Histogram of gamma distributions.

Figure 5.8: Overview gamma analyses of measurement 2
(a) The PDD of measurement 2 at position A. This is as expected similar to figure 5.3 and the same comments of section 5.1 about it are also valid here.

(b) The PDD of measurement 2 at position B. These films were not directly situated at an air-tissue interface. Nor Pinnacle, nor the measured dose show a secondary build-up. Monaco, however shows a rebuild-up of 1.6% and also dose drop of 3% when reaching the the region of the air cavity at $X = 0.4$ cm. $D_{\text{max}} = 250.40$ cGy.

**Figure 5.9:** PDDs of measurement 2 at $Y = -1.5$ cm.
Figure 5.10: Dose differences divided per region along the PDD curve for measurement 2. The difference between Pinnacle and Monaco in the region of the air cavity varies between 3 to 5.
5.3 Measurement 3

In measurement 3 there is only 1 mm of polystyrene between the two film layers and every film is situated at the air-tissue interface as could be seen in figure 4.7.

The dose distributions of the most caudal one can be seen in figure 5.11. In the caudal film measurement (fig. 5.11a), with a mean SD of 1.51% for the area indicated with the gray rectangular, the shapes of the superior and inferior polystyrene section can be seen. Those shapes represent a dose reduction. The shapes are more distinct at the negative half-plane. At X = 0.4 cm, along the horizontal black dotted line, the dose drops with 4% over 2 mm, two rebuild-up regions can be distinguished in the region of the air cavities, for the first at X = -0.4 cm, the dose increases with 3% over 2 mm and at X = -1 cm, a minor increase of 2% also over 2 mm. See also the comments about the PDD curves.

In the dose distribution calculated with Pinnacle (fig. 5.11b), nor the shapes of the adjacent sections, nor the rebuild-up regions are visible. However, at the top left corner (-0.65 cm; -2.30 cm), there is a dose extremum of 214 cGy, but this represents only 101% of the dose 3 mm to the right (-0.35 cm; -2.30 cm). The dose distribution of Monaco (Fig. 5.11d) demonstrates again the shapes of the air cavities like the film film measurement. The two rebuild regions visible on the film measurements can also here be distinguished.

In figure 5.12 the dose distributions of the most cranial are shown. In the film measurement (fig. 5.12a), with a mean SD of 2.03% for the area in the gray rectangular, a rebuild-up zone is visible along the X-axis in the region of the air cavity.

For this position, Pinnacle (fig. 5.12b) shows 2 dose extrema: again one relative maximum of 215 cGy in the top left ‘corner’ of the cavity (X = -2.4 cm; Y = -2.3 cm) and one relative minimum of 206 cGy around the edge at the bottom left side of the cavity (X = -0.9 cm; Y = -1.2 cm). Monaco’s dose distribution (fig. 5.12d) shows again the circular region of the air cavity and a rebuild-up zone. At the bottom of the circular air cavity, the dose drop is extended 2 mm posteriorly. This could be a consequence of the lateral electronic disequilibrium. This effect is, however, not visible in the film measurement.

The PDDs at the caudal and cranial position are shown in figure 5.13. For the caudal position the agreement between Monaco and the measurement in the region of the air cavity seems to be quite good, the mean between X = [-0.3 cm; 0.3 cm] is 1.08%, for [-0.7 cm; -0.4 cm], 0.74% and for [0.4 cm; 0.7 cm] 0.11% (See also fig. 5.15). Monaco and the film measurement show, like already mentioned, both rebuild-up zones. Monaco increases with 1.3% over 2 mm at X = -0.5 cm and with 3.3% over 4 mm at -1 cm. At the extremities, Monaco seems to show a higher dose than the measurement, but no measurement data is available at those distances.

There seems to be a better level of agreement between the PDD of Monaco and the PDD of the film for the caudal position than for the cranial, where the difference between Monaco and the film is also bigger, see figure 5.15. Pinnacle shows only in the cranial position a clear rebuild-up zone, the doses increased there with 1% over 2 mm. In the region of the cavity the
(a) Dose (cGy) distribution measured with radiochromic film. Transversal cross-section at the caudal position, the beam at 90°, region of grey rectangular is used for the gamma analyses (c and d), the dotted black lines indicates the position of the PDD of figure 5.13a.

(b) Dose (cGy) calculated with Pinnacle.

(c) Gamma analysis of (3%/3mm) Film-Pinnacle.

(d) Dose (cGy) calculated with Monaco.

(e) Gamma analysis of (3%/3mm) Film-Monaco.

Figure 5.11: Dose distributions and gamma analysis of measurement 3 (caudal position).
(a) Dose (cGy) distribution measured with radiochromic film. Transversal cross-section at the cranial position, the beam at 90°, region of grey rectangular is used for the gamma analyses (c and d), the dotted black lines indicates the position of the PDD of figure 5.13b.

(b) Dose (cGy) calculated with Pinnacle.

(c) Gamma analysis (3%/3mm) of Film-Pinnacle.

(d) Dose (cGy) calculated with Monaco.

(e) Gamma analysis (3%/3mm) of Film-Monaco.

Figure 5.12: Dose distributions and gamma analysis of measurement 3 (cranial position).
slope of the dose reduction is more negative than outside the cavity. The difference outside the cavity between Pinnacle and Monaco is maximum 2%, in the air cavity this increase to 5%.

The position of the relative dose maxima of the rebuild-ups zones in the PDDs are different for Monaco than for the film measurement. At the caudal position we a secondary build-up zone at the caudal position at X = -0.4 cm and -1 cm. At the cranial position we saw only one at X = -1 cm. The maximum appeared always ~2mm later in the Monaco calculation than in the film measurement.

Figures 5.14 and 5.15 show an overview of the presented data. From the histogram it is clear that there are gamma values above 2.5, this is no surprise as the criterium was 3%/3mm and the PDDs showed dose differences of 9% (See figure 5.15.) between the film measurement and the dose calculated with Pinnacle.
Chapter 5. Results

(a) The PDD at the most caudal position. Dmax = 257.01 cGy.

(b) The PDD at the most cranial position. Dmax = 257.82 cGy.

Figure 5.13: PDDs of measurement 3 at y = -1.5 cm. Indicated at the dose distributions with the dotted black lines.
Figure 5.14: Histogram of the gamma analyses of measurement 3.

Figure 5.15: Dose differences along the PDD for measurement 3.
5.4 Measurement 4

In measurement 4, the phantom was irradiated with the IMRT plan. Also here, there were 2 film positions: caudal and cranial. In figure 5.16, the dose distributions and gamma analyses for the caudal position are shown. From the dose distribution of Pinnacle and Monaco (fig. 5.16b and 5.16d), it seems that the region around the air cavity will get an almost homogeneous dose. The highest intensity is more situated at the right, which comply with the fact that the plan was optimized for a glottic carcinoma of the right vocal cord.

In the dose distribution of Monaco the shapes of the adjacent polystyrene sections, shown with the white dotted lines, are clearly visible because of the dose difference (8.3% over 3mm at (X = 0 cm; Y = -0.2 cm), this is in contrast to the dose of Pinnacle (3.2%) and those of the film (4.5%).

The mean SD on the measurement data (fig. 5.16a) is 1.50% for the area indicated with the gray rectangular. At the top left corner of the measurement film a gradient of almost 56 cGy over 1 cm is visible. At the bottom of the region of the air cavity, the dose is lower than its surroundings, see comments PDDs.

The corresponding gamma analysis of Pinnacle with the film measurement (fig. 5.16c) shows that the gradient in the top left corner more or less complies with the criteria of 3%/3mm, but overall only 13.26% of the gamma values are smaller than 1, while this is 72.65% for the calculation of Monaco.

The mean SD of the cranial film measurement, figure 5.17a, is 1.51%, again indicated for the area with the gray rectangular. On the figures of 5.17, only the circular shape of the air cavity is visible in the dose distribution of the film and of Monaco. In Pinnacle no dose reduction is seen in the figure. The amount of pixel values of the gamma analysis Film-Pinnacle (fig. 5.17c) complying with the criteria is even lower for the cranial position than for the caudal position, only 10.85%. Also now they are mostly situated in the region of top left corner. In other part of the circular cavity region, they cross the value of 2.5.

In the gamma analysis Film-Monaco (fig. 5.17e), there seems to be a quite good level of agreement, 72.65% comply the criteria of 3%/3mm, compared with the values of Film-Pinnacle.

In figure 5.18, the histograms of the gamma values are shown. Especially the dose of Pinnacle at the cranial position scores very bad.

The PDD along the X-axis at the caudal position (fig. 5.13) shows a very good agreement of the dose measured with film and those calculated with Monaco between $X = -0.5$ cm and 0.5 cm. Pinnacle shows a small dose drop in that region, but overestimates the whole region of the air cavity with $\sim 7\%$.

At the extremities the dose differences between Pinnacle and Monaco are very small. The dose calculated with Monaco seems to have an almost constant difference ($\sim 4\%$) with the measurement in the PDD at the cranial position. Pinnacle does not seems to take the presence of
the air cavity very well into account.

Because this plan contains 6 beam directions instead of 1, PDDs along the Y-axes at X = 0 cm are made. This position is also indicated at the dose distributions of figures 5.16 and 5.17. Again the mean of 11 pixel values was used. The region of the air cavity is indicated with the vertical gray dotted lines.

At the caudal position Monaco 5.20a calculates a homogeneous dose in $Y = [-2.0 \text{ cm}; -0.5 \text{ cm}]$, while the measurement shows there a dose gradient of 4\% over 1.5 cm.

Again at the cranial position, this was also for the PDD along the X-axis, the dose calculated with Monaco seems to have an almost constant difference of $\sim4\%$ with the measurement in the PDD. At $Y = 1.2 \text{ cm}$, thus 1.7 cm posterior the air cavity, Monaco and Pinnacle differ respectively 4.5\% and 5.3\% of the film measurement.
(a) Dose (cGy) measured with radiochromic film. Transversal cross-section at the caudal position, phantom irradiated with an IMRT plan, region of grey rectangular is used for the gamma analyses (c and d), the dotted black lines indicates the position of the PDDs of figures 5.19a and 5.20a.

(b) Dose (cGy) calculated with Pinnacle.

(c) Gamma analysis (3%/3mm) of Film-Pinnacle.

d) Dose (cGy) calculated with Monaco.

(e) Gamma analysis (3%/3mm) of Film-Monaco.

Figure 5.16: Dose distributions and gamma analysis of measurement 4 (caudal position).
Chapter 5. Results

(a) Dose (cGy) measured with radiochromic film. Transversal cross-section at the cranial position, phantom irradiated with an IMRT plan, region of grey rectangular is used for the gamma analyses (c and d), the dotted black lines indicates the position of the PDDs of figures 5.19b and 5.20b.

(b) Dose (cGy) calculated with Pinnacle.

(c) Gamma analysis (3%/3mm) Film-Pinnacle.

(d) Dose (cGy) calculated with Monaco.

(e) Gamma analysis (3%/3mm) Film-Monaco.

Figure 5.17: Dose distributions and gamma analysis of measurement 4 (cranial position).
Figure 5.18: Histogram of the gamma analyses of measurement 4.
(a) The horizontal PDD at the caudal position. $D_{\text{max}} = 240.04$ cGy.

(b) The horizontal PDD at the cranial position. $D_{\text{max}} = 240.65$ cGy.

**Figure 5.19:** PDDs of measurement 4 at $y = -1.5$ cm, indicated at the dose distributions with the horizontal dotted black line.
(a) The vertical PDD at the caudal position. $D_{\text{max}} = 236,15 \text{ cGy}$.

(b) The vertical PDD at the cranial position. $D_{\text{max}} = 236,24 \text{ cGy}$.

Figure 5.20: PDDs of measurement 4 at $x = 0 \text{ cm}$, indicated at the dose distributions with the vertical dotted black line.
5.5 Comparison between the measurements

Measurement 1 vs Measurement 2
If we look at the dose differences of measurement 1 (fig. 5.4) and 2 (fig. 5.10), both set-ups had solid slabs with a total thickness of 4 mm, we see that the result of measurement 1 and 2B (at the caudal position) are very similar: Pinnacle gives a higher dose (up to 5%) with the biggest difference in the center of the region of the air cavity. Monaco clearly predicts the dose within the 1.8% at position A and also at B.

Measurements 1 and 2 vs Measurement 3
When we compare those results against those of measurement 3 (fig. 5.15), we still can assume that Monaco is better at predicting the dose in the region of the air cavity, but the difference between Monaco and the radiochromic film measurement increases to 3-5% at the cranial position. Pinnacle however overestimates the dose with minimally 5% in the region of the air cavity at the cranial position, the difference exceeds even the 8%.

Measurement 3 vs Measurement 4
The differences between the film measurements and the calculations of measurement 3 are very similar to those of measurement 4. Even with a clinically more relevant treatment plan, Monaco performs more accurate if we take our film measurement as reference. At the cranial positions, the trend of the curves lines of the calculation of Monaco and the film measurement are very similar, the differences seems to be a constant, for measurement 3 the mean over \( X = [-1.4 \text{ cm}; 1.4 \text{ cm}] \) is 3.96% (SD = 1.21%) and for measurement 4, 4.18% (SD = 0.99%).

Gamma analyses of Measurements 1, 2, 3 and 4
In table 5.1, all the percentages of gamma values that comply with the criteria, are summarised. The main trend is that those of Pinnacle are low compared with those of Monaco.

<table>
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<tr>
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<th>1</th>
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<tr>
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<tr>
<td>caudal</td>
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<td>78,17%</td>
<td>50,59%</td>
<td>13,26%</td>
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<tr>
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<tr>
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<td>98,53%</td>
<td>95,70%</td>
<td>72,65%</td>
</tr>
<tr>
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<td>97,87%</td>
<td>75,37%</td>
<td>75,02%</td>
</tr>
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</table>

Table 5.1: Overview of the gamma values complying the criteria of 3%/3mm.
5.6 Discussion

Comments on the measurement data and the comparison

The first measurement shows that there is a good accordance between the radiochromic film measurements and the calculated Monaco dose distributions. However at the lateral extremities of the film, where the film is surrounded by the solid phantom, Monaco gives a higher dose in all the measurements. This maybe indicates a systematic error.

In the vertical PDDs of measurement 3 and 4, this offset shows also up, indicating that there is indeed a systematic error. In the case that the offset is a systematic error in the radiochromic measurements, normalising the data to a point posterior and more lateral the air cavity (X = 1 cm; Y = 1,5 cm), could maybe help in predicting the dose difference of Pinnacle and Monaco as it is expected that the dose calculation would be more accurate away the air cavity. Nevertheless it is clear that Monaco follows the trends of the film measurement much better than Pinnacle which always calculates a higher dose. This can be expected as literature shows that CCC algorithms are not able to predict the interface dose accurately when lateral electronic disequilibrium occurs [31].

The fact that the doses of the films are based on the differences between two scans, makes them vulnerable to small deviations as their effect could be increased. The described protocol was however, applied to our best efforts.

Comments on the phantom and set-up

In our set-up there was a smaller angle (0,34° for the caudal position and 0,40° for the cranial position) between the beam direction and the plane of the film than those that were used in literature ([36],[35]): 2° and 1,1°. Radiochromic film containing the beams axis was proven to show an under dosage. This could maybe be eliminated by setting the isocenter at a distance of 3,5 cm or 2 cm more caudal from the first film. That way the appropriate angle could be assured, but the films would also be near the border of the conventionally used field size of 6 x 6 cm² and the relation with clinical practise, where the isocenter is near the vocal cors, would be omitted.

The shape and dimension of the film were chosen because we were mainly interested in the dose in the region of the air cavity. But to have a good idea of the precision of the measurements, a greater measurement area would have been practical and could maybe help in interpreting the data.

The vocal cords are modelled according to dimensions found in literature, a narrowing and solid slice with a certain thickness was hereby used to model 2 closed vocal cords. Both the ‘solid’ characteristic of the vocal cords and the ‘thickness’ could be discussed. During normal breathing the vocal cords are slightly open, creating an opening between the two parts of the air cavities of the phantom would maybe be a better model. The thickness of the solid slice was
already changed in this study from 4 mm to 1 mm. 1 mm is also the lower limit of the dose grid of both treatment planning systems.

**Comments on the dose calculations**

Pinnacle and Monaco calculate the dose in voxels of 1 mm$^3$, while the absorbed dose in the film (with a thickness of 0.285mm) is measured by the radiochromic film.

The dose represented in a voxel by the treatment planning system could interpreted as the mean absorbed dose of the region inside that voxel or as the absorbed dose at the specific location. Either way, a direct comparison with a film positioned at the 'border’ or 'in’ the voxels could give important differences as the measurements in this thesis occurs at a air-medium interface. The bigger differences for the cranial position than those for caudal position (in measurement 3 and 4) could hereby maybe be explained, the same for the bigger differences of position A, which is situated at an air-medium interface, while B was more surrounded.

The importance of the slice thickness is hereby also stressed.

**Comments on the secondary (and tertiary) build-up zones**

In the PDD of measurement 2 position A the secondary build-up or rebuild-up zone is visible, Monaco shows for both position A and B such secondary build-up zone. The build-up zone calculated in Monaco takes longer to come to its maxima than the film measurement. This is also the case for measurement 3 as discussed in the that section. Maybe this is a spatial accuracy error or maybe is Monaco not very accurate at the air-medium interface. The distance of the maxima of Monaco to those of the measurement is in measurement 3 $\sim$ 2 mm or thus the size of 2 voxels in the dose calculation of Monaco.
Chapter 6

Conclusion

The goal of this thesis was to compare treatment planning computations with regard to vocal cord irradiation, against radiochromic film measurements. All the measurements in our phantom show that the XVMC Dose Engine of Monaco can better predict the dose in the regions of the air cavity surrounding the modelled vocal cord than the CCC algorithm of Pinnacle. This is in accordance with literature.

In the measurements, radiochromic film has shown to be a good and reliable dosimeter, but care has to be taken when its absorbed dose is compared against dose distributions obtained with a wider dose grid than the film thickness.
Bibliography


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