Evaluation of the pencil beam algorithm and pencil beam redefinition algorithm for bolus electron conformal therapy dose computation

James Kavanaugh
Louisiana State University and Agricultural and Mechanical College

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EVALUATION OF THE PENCIL BEAM ALGORITHM AND PENCIL BEAM REDEFINITION ALGORITHM FOR BOLUS ELECTRON CONFORMAL THERAPY DOSE COMPUTATION

A Thesis

Submitted to the Graduate Faculty of the Louisiana State University and Agricultural and Mechanical College in partial fulfillment of the requirements for the Degree of Master of Science

in

The Department of Physics and Astronomy

by

James Alexander Kavanaugh
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Abstract

**Purpose:** This study aimed (1) to demonstrate the ability of mixed beam (MB) therapy, intensity modulated x-ray therapy (IMXT) optimized on top of a bolus electron conformal therapy (ECT) plan, to deliver a uniform dose to the planning target volume (PTV) in a homogenous phantom and (2) to evaluate the accuracy of the pencil beam algorithm (PBA) and pencil beam redefinition algorithm (PBRA) in calculating the dose distributions for the bolus ECT and MB plans.

**Methods:** PTVs and critical structures from parotid and post-mastectomy chest wall patients were modeled in a 27cm φ polystyrene phantom. Bolus ECT treatment plans using 20MeV and 16MeV beams, respectively, used .decimal p.d software to conform the 90% isodose surface to the distal PTV surface. IMXT was optimized over the bolus ECT dose distributions to homogenize dose to the PTV (ECT:IMXT weightings of 2.08:1 and 1.73:1 for the parotid and chest wall, respectively). Multiple in-phantom radiographic film measurements of the dose delivery were acquired for each of the five transverse and one sagittal planes for the bolus ECT, IMXT, and MB plans. The bolus ECT component of the MB dose distribution was computed using the Pinnacle PBA and an in-house PBRA. Calculated and measured dose distributions were compared. Acceptability criteria for dose points in the bolus ECT, IMXT, and MB dose distributions was set at either ±4% dose difference or ±2mm distance to agreement.

**Results:** Measured dose distributions (36 planes) had an average precision of less than 1% or 1mm. Results for the parotid ECT, parotid MB, chest wall ECT, and chest wall MB showed pass rates (either criteria) for the PBA of 98.9%, 97.5%, 97.3%, and 95.7%, respectively. Pass rates for the PBRA were 98.9%, 97.2%, 98.7%, and 98.0%, respectively. The IMXT dose component of the MB plans showed pass rates of 93.7% and 95.8%, respectively.

**Conclusions:** This study provided confidence that MB therapy planned using the Pinnacle system can be delivered accurately for a homogenous phantom. Also, the PBA and PBRA calculated dose distributions have similar, clinically acceptable accuracy for the bolus ECT and MB plans in homogenous material.
Chapter 1: Introduction

1.1 Background and Significance

1.1.1 Properties of Electron Dose Distributions

Electron therapy is a common radiation modality for treating superficial planning target volumes (PTVs) located within 6cm of the external surface of a patient (Hogstrom 2003). The characteristic sharp distal dose falloff, finite penetration, and variable therapeutic range ($R_{90}$) of electron distributions allow for a uniform dose delivery to the PTV while limiting dose to nearby distal critical structures, often producing favorable plans for superficial tumors when compared to photon dose plans. Some of the unique physical characteristics of electron distributions are illustrated in Figure 1.1.

![Figure 1.1: Depth dose curves for three different energy electron beams. As the incident electron energy increases, so does the surface dose ($D_s$), practical range ($R_p$), and therapeutic range ($R_{90}$).](image)

Electrons produce a surface dose ($D_s$) that is 70%-90% of the maximum dose in the electron range of 6 – 20 MeV. The therapeutic depth ($R_{90}$), which increases as the beam energy increases, is selected for optimal coverage of the PTV. The incident electron energy increases $R_{90}$, allowing for deeper coverage of a PTV. The sharp distal falloff ($R_{10} - R_{90}$) and finite dose range ($R_p$) increases as beam energy increases, which allow for healthy tissue distal to a PTV to be spared significant dose.
### 1.1.2 Principles of Electron Treatment Planning

A significant portion of chest wall, parotid gland, nose, and ear PTVs can be treated fully or with a large fraction utilizing traditional electron beams (Tapley 1976, Vaeth and Meyer 1991). Electron treatments utilize the sharp distal dose falloff to spare nearby, distal critical structures, including lung tissue and the heart for chest wall treatments, and the brain, spinal cord, eyes and optic chiasm for head and neck treatments. Traditional electron therapy treatment plans select the necessary incident electron energy to treat the deepest part of the PTV to $R_{90}$, in order to deliver at least 90% of the prescribed dose to the entire treatment volume.

For patient cases with a relatively uniform PTV that has a maximum depth falling between the therapeutic depth of two electron energies, a constant thickness bolus can be placed on the skin surface to decrease the penetration depth within the patient. However, in cases that exhibit a variable depth across the PTV, the energy required to treat the maximum depth results in needless dose to tissue distal to the shallower PTV regions (Figure 1.2a). These cases can benefit from electron conformal therapy (ECT), in which the energy (or range) and sometimes intensity of the electron beam is modulated to shape the 90% isodose surface to the distal PTV surface, resulting in lower dose to distal normal tissue (Hogstrom 2003), as seen in Figure 1.2b.

![Figure 1.2: (a) Healthy tissue distal to the lateral regions of the PTV absorb significant dose for non-conformal electron therapy (Low et al. 1995). The target volumes are shown in red, with the right and left lung outlined in blue. The 90% isodose line, equal to 50cGy is represented by the bold, black line. (b) Using bolus electron conformal therapy reduces the dose to the distal healthy tissue.](image-url)
1.1.3 Electron Conformal Therapy

Superficial tumors in the head, neck, and chest wall often exhibit PTV’s with variable depth (Perkins et al. 2001, Kudchadker et al. 2003). Traditional electron therapy treatments would deposit unnecessary dose to healthy tissue distal to the target volume. Electron conformal therapy (ECT) provides alternative, clinically-applicable techniques to conform the dose distribution (e.g. 90% isodose) to the distal surface of the PTV. ECT treatment plans produce the conformal dose distribution through modulation of the energy or range in the plane perpendicular to the central axis of the incident electron beam (Hogstrom et al. 2003). Energy modulation can be conducted continuously through the use of an electron bolus or discretely with multiple segmented electron fields of varying energies. Current clinical application of electron conformal therapy is limited to segmented-field ECT and bolus ECT.

1.1.3.1 Segmented-Field Electron Conformal Therapy

Segmented-field ECT modulates the energy and intensity of the electron beam by dividing it into multiple abutted fields, each with its own energy and weighting. The set of abutted fields share a common virtual source position in order to reduce hot/cold spots in the abutment region (Hogstrom et al. 2003), however utilizing different electron energies still produces dose heterogeneity between field segments (Zachrisson and Karlsson 1996). Segmentated-field ECT treatments can be manually forward planned on a 3-D treatment planning system; however this is a tedious, time-intensive process. To expedite the planning process, Perrin (2008) developed an automated forward planning process, i.e. a computer-based segmented-field planning algorithm. Presently, the individual fields can be shaped using multiple Cerrobend® inserts (Richert 2006) or with the use of an x-ray/electron multi-leaf collimator (MLC) (Zachrisson and Karlsson 1996). The dose in the abutment region can be homogenized through a variable source-to-collimator distance (SCD) for electron cutouts (Richert 2006) or by edge modulation of the electron MLC (Eley 2009). While segmented-field ECT offers significant promise for providing modulated electron therapy, the current lack of commercially available planning and delivery
systems (e.g. electron MLC makes) it difficult to implement clinically. An additional disadvantage of segmented-field ECT results from clinical electron beams being spaced at interval of 2-4 MeV, limiting the treatment depth, $R_{90}$, to discrete steps of 0.6 -1.2cm.

1.1.3.2 Bolus Electron Conformal Therapy

In contrast to segmented-field ECT, bolus ECT modulates the range (energy) of the electron beam by adding varying thicknesses of wax bolus upstream of the patient surface. To conform $R_{90}$ to the distal surface of the PTV, bolus thickness is the least where the distal edge of the PTV is deepest, while the bolus is the thickest upstream of shallow regions of the PTV. The bolus design process can be optimized through the use of a set of forward planning design operators (Low et al. 1991). These operators depend on the plan geometry to determine the thickness of the electron bolus along fan lines extending from the virtual source to the phantom surface, with off-axis thickness set at $x$ and $y$ coordinates in the plane perpendicular to the beam direction as shown in Figure 1.3a. The bolus calculation geometry used by Low et al., is further defined in Figure 1.3b.

The bolus design is a multi-step process consisting of a sequence of operators. It begins with the application of a creation operator, which provides the initial estimate of the bolus thickness along fan lines a distance $\Delta$ inside the PTV. In the present work, the creation operator based on the physical depth ($P(\Delta,R_t)$), where $\Delta$ is the lateral applied margin and $R_t$ is the desired therapeutic range ($R_{90}$). The relatively small PTV depth, $d_{i,j}$, at the edge of a PTV can result in large bolus thickness and sharp gradient at the edges of the field, which can produce additional electron scatter and unacceptable high doses within the PTV. The lateral applied bolus margin ($\Delta$) (Figure 1.3) reduces the effect of bolus build-up along the lateral PTV edges by limiting the bolus thickness calculation to at least $\Delta$ distance inside the PTV. The physical depth creation operator, given by Equation 1.1, is a function of the desired therapeutic range, $R_t$, water equivalent density of the bolus material, $\rho_w$, and the PTV thickness along fan line $i$, $j$, $d_{i,j}$ (Low and Hogstrom 1994).
Figure 1.3: Bolus electron conformal therapy design parameters. (a) The off-axis coordinate grid for bolus construction, including the PTV lateral applied margin, collimator edge, and bolus edge (projected to isocenter). The electron fan beam, shown in b., illustrates the initial creation of the proximal bolus surface and the flat extension of the bolus in the lateral applied margin, $\Delta$. (Low et al. 1991).

\[ b_{ij} = \left( \frac{1}{\rho_b} \right) [(R_t)_{ij} - d_{ij}] \]  

(1.1)

Modification operators are applied to the initial bolus design to improve conformity to the PTV and to minimize dose heterogeneities within the PTV due to the irregular proximal bolus surface (Low et al. 1991). The isodose shift operator, $I$, corrects for any regions where $R_{90}$ does not conform to the distal PTV surface. The difference between $R_{90}$ and the distal PTV surface is determined along each fan line, and the effective thickness is then added (or subtracted) from the bolus surface. Since the isodose shift operator only accounts for information along the fan line, it is known as a “one-dimensional” operator.
x and y directions, resulting in them being known as “two-dimensional” operators. The height smoothing operator, $S_h(\mu, \eta)$, applies a weighted Gaussian to smooth the proximal bolus surface. The smoothing operator depends on the weighting coefficient ($\mu$) and the lateral distance ($\eta$) over which the weighting coefficient is applied. Small values of $\mu$ have little effect in smoothing the bolus surface whereas large values of $\mu$ will over smooth the surface, resulting in poor coverage of the distal PTV surface.

The final group of bolus operators, extension operators, defines the bolus beyond the lateral limits of the PTV. The height of the bolus ($H_h$) just inside the lateral bolus margin $\Delta$ is extended out a set distance beyond the lateral edges of the collimator. This operator is necessary to produce a usable dose distribution along the edges of the electron field.

### 1.1.4 Commercial Implementation of Bolus ECT

Clinical utilizations of bolus ECT (Low et al. 1995, Perkins et al. 2001, Kudchadker et al. 2002) have reported that only a few of the bolus design operators were required (useful) for patient planning. The physical depth creation operator, Gaussian smoothing and isodose shift modification operators, and bolus height extension operators were found to be the most useful for the bolus planning process. A typical sequence might be physical depth creation, isodose shift, smooth bolus height, isodose shift, smooth bolus height, and height extension operators.

Recently, .decimal, Inc. released a commercial product, p.d software. In its present state (see Appendix C), the software utilizes only the creation (physical depth), Gaussian height smoothing and bolus height extension operators.

### 1.1.5 Improving Dose Homogeneity for Bolus ECT

Kudchadker et al. (2002) documented a dose spread within the PTV of up to 20% due to the irregularities of the bolus surface, compared to the minimal dose variation of 10% (90%-100%) produced with electrons incident on a smooth water surface. Also, they showed that the dose heterogeneity can
be restored (90% - 100%) using intensity modulation of the incident electron beams (IMET); however, IMET is not commercially available in clinical settings. Alternatively, Blasi (2009) showed intensity modulated x-ray therapy (IMXT) can be mixed with conventional electron beams to improve dose uniformity similar to that normally found in IMXT, and preliminary calculations have shown the same possibility for mixing IMXT with bolus ECT.

1.1.6 Intensity Modulated X-Ray Therapy

Compared to electrons, photons deposit dose through a much larger volume of tissue, with the gradual exponential falloff delivering some dose to a significant portion of healthy tissue distal to the target volume. Additionally, healthy tissue proximal to the target will receive a larger dose than the PTV; hence, multiple beams are required when treating with megavoltage x-rays. IMXT is the use of multiple, intensity-modulated beams from multiple directions, with the intended purpose to minimize dose heterogeneity within the PTV, while maximizing the sparing of surrounding normal tissue. The use of IMXT provides the advantage of a homogenous dose distribution, steep distal dose gradients, and an excellant PTV conformity when compared to traditional 3D x-ray therapy (Khan 2010). However, by utilizing multiple beam angles and dose escalation, IMXT also contributes addition low dose to the entire volume of healthy tissue (Figure 1.4).

1.1.7 Mixed Beam Radiation Therapy

Historically, the aim of mixed beam radiation therapy has been to harness the advantages of the component modalities while minimizing each modality’s detrimental effects. For example, mixing an x-ray beam with an electron beam can result in decreased skin dose and a more homogenous depth dose through the PTV (Tapley 1976).
Figure 1.4: IMXT (a), Bolus ECT (b), and mixed beam (c) 2D dose distributions treating head and neck cancer. The PTV is given by the red color wash in all three plans. (a) The increased low dose to healthy tissue is illustrated by the extension of the yellow line across most of the patient for the IMXT plan, while maintaining tight dose conformity and a homogenous dose in the PTV. (b) The 90% isodose conforms to the PTV while sparing distal healthy tissue from significant dose; however, the PTV has hot spots of up to 116% of the given dose. (c) Mixing the two modalities results in a homogenous plan that spares healthy tissue, compared to either component utilized individually.

Optimizing IMXT over a bolus ECT dose distribution seeks to homogenize the PTV dose distribution (IMXT) while maintaining a low dose deposited distal to the PTV (bolus ECT) (Weinberg et al. 2008, Mu et al. 2004). Preliminary studies have shown that mixing IMXT with electron therapy can improve dose homogeneity within the PTV, while reducing integral dose to the surrounding healthy tissue (Surucu et al. 2009).

1.1.8 Relevance of Accuracy of Dose Algorithm

Most modern electron beam treatment planning systems implement a variation of the Hogstrom pencil beam algorithm (PBA) (Hogstrom, Mills and Almond 1981) as implemented in 3D by Starkschall et al. (1993) to calculate 3D electron dose distributions. It has been well documented that the PBA can produce inaccuracies in regions adjacent to or behind edges of tissue heterogeneities deep
to the surface (Hogstrom and Steadham 1996, Shiu and Hogstrom 1991a, Hogstrom et al. 1984), which has led to the more accurate pencil beam redefinition algorithm and Monte Carlo dose algorithms.

As the objective of bolus ECT is to produce a highly conformal, homogenous dose distribution to the PTV, it is important that the electron dose computation algorithm provides an accurate model calculated dose. The accuracy of the calculated dose distribution is increasingly important for mixed beam therapies, because the IMXT planning system uses the calculated bolus ECT dose distribution to homogenize the dose within the PTV.

In the present study, the PBA as originated by Shiu and Hogstrom (1991b) and refined by Boyd et al. (1998, 2001b) was utilized. This algorithm has been shown highly accurate by comparing with measured dose distributions in a water phantom with regularly shaped heterogeneities by Boyd et al. (2001a). It has also been shown to agree with calculations in four different patient sites (Boyd 2001).

1.2 Purpose and Objectives

The purpose of this study is to evaluate the accuracy of the planned dose distributions for bolus ECT and mixed beam therapy (bolus ECT and IMXT). There are three primary objectives associated with this purpose.

1. Compare the accuracy of the PBA (Pinnacle) and PBRA dose planes for bolus ECT in a homogeneous phantom. Both are expected to give similar results, although this is not expected for future studies in heterogeneous phantoms or patients.

2. Compare the accuracy of the collapsed cone convolution algorithm (Pinnacle) for IMXT plans optimized on top of a bolus ECT plan. Results are expected to approximate previous IMXT results.

3. Evaluate the accuracy of the PBA and PBRA for mixed beam (bolus ECT + IMXT) dose planes in a homogeneous phantom. The accuracy of the dose planes will be evaluated by comparing with film measurements in a homogeneous cylindrical phantom with patient-like PTVs.
1.3 Hypothesis

At least 98% of dose points for the Bolus ECT and Bolus ECT + IMXT measured dose distributions will be within ±4% of the calculated dose in the low gradient regions and ±2mm distance to agreement in the high gradient regions of both the PBA and PBRA calculated dose distributions for head and neck and post-mastectomy patient-like PTVs modeled in a cylindrical high-impact white opaque (HIWO) polystyrene phantom.

1.4 Specific Aims

Aim 1: Develop in-phantom bolus ECT, IMXT, and mixed beam treatment plans.

• Relevant ROIs are modeled from existing patient data and modified to fit the phantom.

• Specific plans are chosen based on utility of bolus ECT to treat the PTV.

Aim 2: Measure dose distributions for the parotid and chest wall bolus ECT, IMXT, and mixed beam treatment plans.

• Acquire three to four in-phantom film measurements in five transverse planes and one sagittal plane.

• Evaluate accuracy of the film measurements by calculating standard error of each set of dose points and comparing the intersection of sagittal and transverse slices.

Aim 3: Evaluate the accuracy of the PBA and PBRA by comparing calculated dose distributions to the set of parotid and chest wall measurements.

• Compare five transverse and one sagittal calculated planar dose distributions to the corresponding film measurements for the bolus ECT, IMXT, and mixed beam plans.

• Accuracy is determined with the criteria of ±4% dose difference or ±2mm distance to agreement for each dose points.
2.1 Aim 1: Develop In-Phantom Bolus ECT, IMXT, and Mixed Beam Treatment Plans

The purpose of this aim is to develop ECT, IMXT, and mixed-beam treatment plans for parotid and chest wall PTVS, which have been modeled in a polystyrene cylindrical phantom. The phantom allows the measurement of dose distributions using radiographic film, which can be compared to dose calculations generated in this aim.

2.1.1 CT Scan of the HIWO Phantom

A high-impact white opaque (HIWO) polystyrene cylindrical phantom was selected because of its ability to perform film dosimetry measurements in multiple planes (Chi et al. 2005). The phantom had a diameter of 27cm and a length of 37.4cm, allowing it to mimic the skin surface of a chest wall and parotid treatment plan. The phantom had two film cassettes, one in a sagittal and one in a transverse phantom plane. Each cassette had a central slot to fit a single bare radiographic film. Schematics and an image of the phantom are shown in Figure 2.1.

The CT image set of the HIWO polystyrene phantom was acquired on a large bore GE LightSpeed CT scanner (General Electric Medical Systems). The central sagittal plane of the film phantom was aligned to the central vertical plane of the CT scanner. Images were acquired using a helical scanning technique, with a pitch and slice thickness of 0.938mm and 2.5mm, respectively. The parotid image set incorporated 121 transverse slices to make an image length of 30cm, while the chest wall image set consisted of 141 transverse slices and a length of 35cm. CT scanner current and potential values were set to 400mA and 120 kVp, respectively. These values are standard for CT imaging at Mary Bird Perkins Cancer Center (MBPCC) for patient CT scans used for treatment planning. Once the scan was completed, the image set was transferred to the Pinnacle v8.0 treatment planning system to create the ECT, IMXT, and mixed beam treatment plans.
2.1.2 Modeling of Patient-Like Parotid and Chest Wall ROIs

The parotid and chest wall PTVs used to create the phantom bolus ECT plans and dose distributions were transferred from pre-existing clinical patient treatment plans to the HIWO phantom. To produce measurement data consistent with clinical bolus ECT plans, contours for the parotid bolus plan were modeled from six 2D CT slices presented by Kudchadker et al. (2003), five of the six are shown in Figure 2.2. The chest wall PTV was modeled after an existing patient previously treated at MBPCC; five of the six CT slices shown in Figure 2.3. The proximal surface of each PTV was adjusted to fit the cylindrical phantom surface; however, the distal PTV edge retained similar shape and scale of the original patient PTV contours.
Figure 2.2: (a, c, e, g, i) Transverse parotid PTV contour from Kudchadker et al. (2003) spaced 3cm apart. (b, d, f, h, j) Parotid PTV contours (red contour) modeled after contours (a, c, e, g, i) in HIWO phantom. The phantom contours have been rotated counterclockwise ~8° so that the electron beam is vertical and centered on the sagittal cassette. Also, the fabricated bolus is shown placed on the phantom. The purple contour represents the spinal cord while the blue represents the lung.
Figure 2.2 (continued): (a, c, e, g, i) Transverse parotid PTV contour from Kudchadker et al. (2003) spaced 3cm apart. (b, d, f, h, j) Parotid PTV contours (red contour) modeled after contours (a, c, e, g, i) in HIWO phantom. The phantom contours have been rotated counterclockwise ~8° so that the electron beam is vertical and centered on the sagittal cassette. Also, the fabricated bolus is shown placed on the phantom. The purple contour represents the spinal cord while the blue represents the lung.
Figure 2.3: (a, c, e, g, i) Transverse chest wall PTV contours, spaced 3cm apart, from a patient treated at MBPCC. (b, d, f, h, j) Chest wall PTV contours modeled after contours (a, c, e, g, i) in HIWO phantom. The phantom contours have been rotated counterclockwise by ~30° so that the electron beam is vertical and centered on the sagittal film cassette. The phantom PTV did not model the part of the patient PTV that included the 1cm Superflab bolus. The electron bolus is shown lying on the phantom.
Figure 2.3 (continued): (a, c, e, g, i) Transverse chest wall PTV contours, spaced 3cm apart, from a patient treated at MBPCC. (b, d, f, h, j) Chest wall PTV contours modeled after contours (a, c, e, g, i) in HIWO phantom. The phantom contours have been rotated counterclockwise by ~30° so that the electron beam is vertical and centered on the sagittal film cassette. The phantom PTV did not model the part of the patient PTV that included the 1cm Superflab bolus. The electron bolus is shown lying on the phantom.

2.1.3 Bolus Creation and ECT Treatment Planning

The bolus design process requires plan information, which included PTV and external contour structure files, beam orientation, modifiers, energy, and SSD. These parameters were set in the Pinnacle TPS (Philips Electronics North America Corporation, Andover, MA) and then exported to the .decimal p.d (.decimal, Inc., Sanford, FL) software. Both the parotid and chest wall plans were created for the Varian 2100EX (Varian Medical Systems, Inc., Palo Alto, CA), which has five discrete electron energies (6 MeV, 9
MeV, 12 MeV, 16 MeV, 20 MeV) and two x-ray energies (4 MV and 10 MV). Bolus ECT plans utilized a single electron beam to treat the entire PTV. The minimum energy was selected to deliver the physician’s prescribed dose to the 90% isodose line located at the deepest distal edge of the PTV.

The parotid and chest wall plans required electron energies of 20 MeV and 16 MeV, respectively. Isocenter was set along the phantom mid-sagittal slice and 5cm superior to the phantom surface, producing a pre-bolus SSD of 105cm, which created the necessary clearance for the bolus between the phantom and electron cone. Beam angles were determined so that beam direction was approximately perpendicular to the distal PTV surface along the central axis (CAX). The phantom PTVs were rotated to allow for 180° gantry orientation in both the parotid (~8° ccw rotation) and chest wall ECT (~30° ccw rotation) plans. The dose calculation grid contained 2.5mm voxels.

An electron block was created to conform the lateral edges of the uniform dose region of the electron beam to the PTV. The block was designed around the beams-eye view (BEV) projection of the PTV to isocenter with a 1cm margin added to ensure the entire PTV was inside the penumbra and received 90% of the given dose. Figure 2.4 illustrates the block geometry, with the block construction process being discussed in section 2.1.5.

Figure 2.4: BEV of (a) parotid and (b) chest wall PTV and block. The outer edges fit the 20x20cm² and 25x25cm² applicators, respectively.
After finalizing the ROI structures and beam characteristics, the plan and structure files were exported from the TPS to the DICOM folder. From there, the plan and structure files were transferred to the .decimal p.d planning computer via file transfer protocol (FTP).

Bolus design was completed using .decimal p.d software, which utilized the physical depth creation, Gaussian smoothing modification, and height extension operators. In the p.d program, the beam energy and ROIs representing the external surface and PTV were selected from the transferred Pinnacle data. Additional input parameters used within the bolus design operators included the inner margin (Δ), outer margin, distal and proximal height extension, and the full width half at maximum (FWHM) of the smoothing Gaussian function. Bolus construction parameters for both the parotid and chest wall cases are defined in Table 2.1. Details of the p.d algorithm are included in Appendix C.

<table>
<thead>
<tr>
<th>Bolus Parameter</th>
<th>Parotid ECT</th>
<th>Chest Wall ECT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Beam Energy (MeV)</td>
<td>20</td>
<td>16</td>
</tr>
<tr>
<td>( R_x , (cm) ) (x = 90% in our design)</td>
<td>6.0</td>
<td>5.3</td>
</tr>
<tr>
<td>Distal Bolus Surface Extension: Dist. Beyond Target (cm)</td>
<td>0.0</td>
<td>2.0</td>
</tr>
<tr>
<td>Proximal Bolus Surface Extension (cm)</td>
<td>Max + 0.5</td>
<td>Max + 0.5</td>
</tr>
<tr>
<td>Smoothing FWHM</td>
<td>30mm</td>
<td>30mm</td>
</tr>
<tr>
<td>Inner Margin ( \Delta ) (cm)</td>
<td>0.5</td>
<td>0.5</td>
</tr>
<tr>
<td>Outer Margin (cm)</td>
<td>2.5</td>
<td>2.5</td>
</tr>
</tbody>
</table>

Once the digital bolus contour was created in the p.d software, it was transferred back to the Pinnacle TPS and inserted into a copy of the original treatment plan. Proper alignment between the distal bolus and proximal phantom surfaces was verified in each CT slice. The density of the bolus material was set to 0.92 g/cm\(^3\) (Low and Hogstrom 1994), and the dose distribution was recalculated using the Pinnacle PBA. Quality of the bolus was determined by analyzing the accuracy of the 90% isodose contour to conform to the distal PTV surface. For the first iteration of the bolus design, the 90% isodose contour differed from the distal PTV surface by as much as ±1 cm, usually along the lateral edges of the PTV and regions exhibiting sharp gradients, as seen in Figure 2.5a. To improve the conformity, a new copy of the PTV ROI was created to be used as the PTV within the p.d design software. The copied
ROI, shown as PTVext in Figure 2.5b, was increased in regions where the 90% isodose line fell short of the distal PTV surface, decreased in regions where the 90% isodose surface contour extends deeper than the distal PTV surface, and held constant wherever the 90% isodose contour accurately conformed to the distal PTV surface, with an action threshold of 2mm.

These modifications were manually completed for each CT slice, with the resulting structure and plan exported once again to the p.d software. A new bolus was designed using the same parameters as the first iteration, with the exception of PTVext representing the PTV. The digital bolus contour was imported back to the TPS, and the conformity of the 90% isodose dose surface to the distal PTV surface was once again assessed. This process was continued for approximately 10 iterations until $R_{90}$ satisfactorily conformed to the entire distal PTV surface. This manual iterative process was required because p.d software does not yet incorporate the isodose shift operator of Low et al. (1991). The final digital bolus file was uploaded to the .decimal website.

Fabrication of the bolus was completed at the .decimal facilities in Sanford, FL using blue machinable wax. The .decimal bolus fabrication process used a series of control points to ensure accurate milling of the wax bolus from the digital design. Once fabricated the bolus was shipped to MBPCC.

Figure 2.5: (a) The red contour represents the PTV while the green line is the 90% Isodose line. To extend the 90% Isodose line to cover the entire PTV, an additional ROI (PTVext) was used to model the PTV in subsequent bolus design iterations. (b) The PTVext, seen as the yellow line, is compared with the original PTV (red line).
2.1.4 Verification of Bolus

The fabricated bolus was visual inspected upon delivery to detect any gross errors produced in the fabrication process. Then, dosimetric quality assurance was conducted to assess the accuracy of the fabricated bolus dose distribution compared to the planned bolus dose distribution. Small grooves were carved in the flat surface of the bolus outside the radiation beam at increments of 2.5cm for the parotid plan and 2.0cm for the chest wall plan, with the central groove corresponding to the transverse plane containing isocenter. Radio-opaque fiducial markers were placed in these grooves in order to accurately index the fabricated bolus CT image set to the original bolus CT image set. The CT image set of the phantom with the fabricated bolus was created using the same image parameters as discussed in 2.1.1. Prior to the CT scan, the rotational position of the bolus was set by aligning the proximal outer surface of the bolus parallel to the horizontal using a digital level (M-D Smart Tool, Oklahoma City, OK) with an accuracy of 0.1°. The fabricated bolus image set was imported into a copy of the existing digital bolus plan and aligned to the digital bolus contour; then, the two image sets were fused. ROIs were copied from the primary digital bolus image set to the secondary fabricated bolus image set allowing the ROIs to be accurately placed within the fabricated bolus coordinate system.

A new dose plan was created using the fabricated bolus image set and the copied ROIs were imported over from the fused plan. Electron beam isocenter was placed 5cm above the surface of the phantom, centered along the mid-sagittal phantom plane and in the transverse plane marked by the central opaque marker. Beam parameters were assigned as discussed in section 2.1.3. The accuracy of the 90% dose surface to conform to the distal surface of the PTV was qualitatively assessed with the bolus being accepted if the prescribed dose satisfactorily covered the PTV.

2.1.5 Fabrication of Cerrobend Block

Beam shaping electron blocks designed by the TPS were fabricated onsite using Cerrobend®, which was poured around a foam block that defined the aperture shape. The outer edge of the block
was defined by a steel frame, allowing it to be properly placed in the selected applicator size. For both the parotid and chest wall cases, the aperture was determined by the projection of the PTV onto a plane perpendicular to beam direction and located at isocenter. An additional margin of 1cm was added around the PTV projections for both the parotid and chest wall plans.

To ensure accuracy, a computer-controlled hot wire, Compu•cutter • system (Huestis Medical, Bristol, RI), was used to cut foam blocks with diverging edges corresponding to the edge of the beam. The Compu•cutter • system uses four geometric inputs (Figure 2.6): (1) electron block thickness, (2) the source-to-tray distance (STD), defined as the distance from the virtual radiation source to the bottom of the insert tray, (3) the source-to-axis distance (SAD), defined as the distance from the radiation source to isocenter, and (4) the source-to-film distance (SFD), defined as the distance from the radiation source to a perpendicular plane in which the field shape is defined (the bottom of the insert tray for this work). It should be noted that in constructing the block the radiation source was defined as the virtual source resulting in a SAD of 90cm (Shiu et al. 1994). Thus the SFD and STD were defined to be 85cm.

![Figure 2.6: Schematic of parameters used by Compu•cutter • to cut electron block.](image-url)
Compu•cutter ® uses a four-axis servo-controlled hot nichrome wire to cut the diverging field shapes into foam blocks. The software accepts a “burn thickness” parameter to represent the measured width of material lost along the path of the hot wire (70°C). The field edge was then extended by half the “burn thickness”. In this study a 4mm wire with a corresponding 0.97mm burn thickness was used.

The two largest contributors to error in the block fabrication process include digitizing the aperture into the Compu•cutter ® software and positioning the foam cutout within the steel frame. The accuracy of the digitization process is a function of the RF digitizer and is quoted as 0.5mm. Accuracy in positioning the foam block is dependent on the user’s ability to properly align the x and y axes on the steel frame to the axes on the foam block. To reduce the positioning uncertainty, the to-scale BEV printout was attached to the top of the foam cutout. The uncertainty in positioning the foam block is estimated at 0.5mm, resulting in a total maximum estimated uncertainty of 0.7 mm.

2.1.6 IMXT Treatment Planning

A mixed beam plan to reduce dose heterogeneities within the PTV was generated by optimizing an IMXT plan on top of a bolus ECT dose distribution using the following procedure. In the Pinnacle TPS the bolus ECT plan was copied, and limited fractions of IMXT were optimized over the bolus ECT dose distribution to reduce dose heterogeneities within the PTV. Both the parotid and chest wall plans used a nine beam IMXT field arrangement. All beams shared a common isocenter, located along the mid-sagittal plane of the phantom, 2cm deep from the superior phantom surface, and in the same transverse plane as the ECT isocenter. Beams in the chest wall plan were evenly spaced at 40° increments ranging from 20° to 340°, while the parotid plan had beams following 32° increments from 52° to 308°, with 180° representing the direction of the electron beam for the ECT plan.

The density of the electron bolus was overridden and set equal to air to remove the bolus from the IMXT dose calculation. Modifications to the bolus density resulted in the ECT dose distribution, a necessary component in the IMXT planning process, being deleted. The ECT dose was recreated by
copying the plan. Trial binary file containing the dose information from the ECT plan to the IMXT plan, with both plans maintaining identical dose grid parameters.

Mixed beam fraction was determined using equation 2.1. The parotid plan had a 1.7:1 (ECT:IMXT) dose fractionation, with the 95% ECT isodose equaling 190cGy, while the chest wall plan had a 2.1:1 (ECT:IMXT) dose fractionation, with 95% ECT isodose equaling 202.5cGy. To limit under dosing the edges of the PTV, a PTV extension of 0.9cm was created and assigned a maximum dose of 300cGy. Additional constraints for critical structures, seen in Figure 2.2, were determined from ratios of critical structure dose limits to traditionally prescribed dose for similar IMXT head and neck plans. Table 2.2 contains dose constraints used for the parotid and chest wall IMXT plans. Beams were optimized using direct machine parameter optimization (DMPO). Parameters for the optimization process are shown in Table 2.3.

\[
ECT:IMXT = D_{ECT,95\%} \left( D_{MB,100\%} - D_{ECT,95\%} \right) \tag{2.1}
\]

\[
D_{ECT,95\%} = 95\% \text{ of ECT dose, where } 90\% \text{ is set as the prescribed bolus ECT dose.}
\]

\[
D_{MB,100\%} = 100\% \text{ of mixed beam dose, as set as the dose at a reference point in the HDR.}
\]

Table 2.2: IMXT planning constraints for the mixed beam plan. PTV Ext is a 1cm extension around the PTV. Ring represents the remaining healthy tissue volume.

<table>
<thead>
<tr>
<th>Parotid IMXT Constraints</th>
<th>Chest Wall IMXT Constraints</th>
</tr>
</thead>
<tbody>
<tr>
<td>ROI</td>
<td>Constraint Type</td>
</tr>
<tr>
<td>PTV</td>
<td>Uniform Dose</td>
</tr>
<tr>
<td>PTV</td>
<td>Maximum Dose</td>
</tr>
<tr>
<td>Spinal Cord</td>
<td>Maximum Dose</td>
</tr>
<tr>
<td>Lung</td>
<td>Maximum DVH</td>
</tr>
<tr>
<td>Ring</td>
<td>Maximum Dose</td>
</tr>
<tr>
<td>PTV Ext</td>
<td>Maximum Dose</td>
</tr>
</tbody>
</table>

Table 2.3: Parameters used in mixed beam treatment planning.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Parotid</th>
<th>Chest Wall</th>
</tr>
</thead>
<tbody>
<tr>
<td>Optimization Method</td>
<td>DMPO</td>
<td>DMPO</td>
</tr>
<tr>
<td>Minimum MU/Segment</td>
<td>3</td>
<td>3</td>
</tr>
<tr>
<td>Maximum # Segments</td>
<td>128</td>
<td>112</td>
</tr>
<tr>
<td>Minimum Segment Area (cm²)</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>Leaf/Jaw Overlap (cm)</td>
<td>0.2</td>
<td>0.2</td>
</tr>
<tr>
<td>Beam Splitting (cm)</td>
<td>2</td>
<td>2</td>
</tr>
</tbody>
</table>
2.2  Aim 2: Measure Dose Distributions for the Parotid and Chest Wall Bolus ECT, IMXT, and Mixed Beam Treatment Plans

The purpose of this aim is to generate, using film dosimetry, a set of measured dose distributions for the bolus ECT plan, the mixed beam plan, and the IMXT component of the mixed beam plan, hereafter referred to as the IMXT plan. The measured dose distributions, which consist of planar dose distributions in five transverse and one sagittal plane, provide a thorough sampling of the full 3D dose distribution. Each planar dose distribution is measured multiple times, allowing calculations of mean dose and standard error of the mean dose at each dose point in the plane.

2.2.1  Output Quality Assurance

Prior to all film measurements, beam dose output was determined using a 0.6 cm$^3$ PTW Farmer chamber, following the protocol set by AAPM Task Group 51 (Almond et al. 1999). For electrons, the ion chamber was placed at 100cm SSD and at reference depths (4.1cm for 16 MeV and 5.1cm for 20 MeV), while photons used a SAD setup of 10cm depth and 90cm SSD. All output measurements were taken in Plastic Water (Elimplex) using a 10cm x10cm field size and 10cm of Plastic Water placed downstream for backscatter. Water equivalent depth was determined from the relative stopping power of the phantom, which was 0.974 (Khan et al. 1991). Temperature was measured within the Plastic Water ion chamber cavity prior to measurements. Local pressure was determined using a mercury barometer.

Results of the output measurements allowed for a specific number of delivered monitor units to be accurately converted to dose. For electron beams, the machine is calibrated to 1.000 cGy/MU. Hence, if on the day of measurement the output is 1.01 cGy/MU, the dose for each calibration measurement (taken at $d_{max}$) was determined by multiplying the delivered monitor units by the 1.01 cGy/MU, allowing all measurements to account for daily output fluctuations.

Because the dose calculations assumed that the output at the time of measurement was 1.000cGy/MU, the measured dose distributions should be scaled down by the inverse of the factor used to scale the dose in the calibration film. In other words, there was no need to make any corrections for
daily output. In the present study this was not discovered until late in the analysis. Due to this oversight, the measured data is higher than then expected by 0.3% - 0.4% for electron measurements and ranges from -0.6% to 0.5% for IMXT measurements and IMXT mixed beam components. Table 2.4 gives the difference between the measured and calculated dose distribution for each modality.

Table 2.4: Output correction factors applied to each measurement modality.

<table>
<thead>
<tr>
<th>Treatment Modality</th>
<th>Output Correction</th>
</tr>
</thead>
<tbody>
<tr>
<td>Parotid ECT</td>
<td>100.4</td>
</tr>
<tr>
<td>Parotid IMXT</td>
<td>100.1</td>
</tr>
<tr>
<td>CW ECT</td>
<td>100.3</td>
</tr>
<tr>
<td>CW IMXT</td>
<td>99.4</td>
</tr>
</tbody>
</table>

2.2.2 XV and EDR2 Radiographic Film Calibration and Validation

This study used both Kodak X-OMAT V and Kodak EDR2 radiographic film (Carestream Health, Inc., Rochester, NY), both of which are double sided with an active AgBr/AgI crystal emulsion. Kodak XV film was originally chosen for this project due to its limited energy dependence for electron energies used in radiotherapy (Dutreix and Dutreix 1969). Kodak XV film is a suitable dosimeter of high resolution dosimetric data in a 2D plane for doses of 0cGy to 80cGy, corresponding to optical densities of 0 to 4 (Pai et al. 2007). However, before all the measurements were completed, Kodak XV film was discontinued as a product by the manufacturer (Kodak), necessitating a change to a different radiographic film (RGF). Kodak EDR2 was selected as a replacement due to its large linear dose range (0 to 350 cGy) and its clinical applicability for IMXT and electron dosimetric evaluations (Gerbi and Dimitroyannis 2003, Childress, White and Rosen 2005). Kodak EDR2 film exhibits a small energy dependence on electron beam energies but show a significant difference for x-ray beam energies (Ahmad et al. 2006), as shown by the calibration curves in Figure 2.7. Hence, in the present study, Kodak X-OMAT V was used for bolus ECT measurements, while Kodak EDR2 was used for IMXT and mixed beam measurements. Both EDR2 and XV films with dimensions of 10” x 12” were used in this study.
Intensity to dose calibration curves were measured with each set of film measurements. All calibration curves consisted of a set of twelve different dose points (dose, OD), measured perpendicular to the beam direction. For each measurement, a single film was cut into two equally sized pieces with one film piece placed in the sagittal cassette with the white paper from the packet along the side of the cassette with the registration markings. The second piece of film was placed back in the film jacket and stored in a light tight film box. The cassette was sealed using black photo tape. Electron calibration measurements were taken at $d_{\text{max}}$ (2.0cm for 20 MeV, 3.3cm for 16 MeV) using a 10cm x 10cm field and an SSD of 100cm. Photon calibration films were taken at a depth of 10cm with an SSD of 90cm.

![Graph](image-url)

Figure 2.7: Optical Density (OD) to dose response of EDR2 to different energies for electrons and photons. OD can differ by as much as 0.3 between 20MeV electrons and 4 MV photons within the linear region (0 to 350cGy) of the OD curve.

For photon treatment plans, parallel film measurements calibrated with a perpendicular calibration set resulted in an under representation of the absolute dose (Cheek et al. 2008). In these cases, the difference between the measured and actual dose can range from 2% to 10% (Suchowerska et al. 2001). It was noted by Cheek et al. (2008) that the HIWO cylindrical polystyrene phantom used in
this study produced a dose difference of ~2%. Hence, all transverse IMXT film measurements and the IMXT component of mixed beam film measurements were increased by 2%.

The combination of film type, radiation modalities, and energies resulted in five unique calibration data sets. The film calibration measurements for each set were scanned using a transmission type scanner (VIDAR DosimetryPRO Advantage, Vidar Systems Corporation, Hendon, Virginia). The conversion of transmitted intensity to dose was accomplished by fitting the data with a piece-wise polynomial function using RIT version 5.2 software (Colorado Springs, CO). This fitted curve was tested to ensure accurate mapping of intensity to dose for points falling between two calibration measurements.

EDR2 and XV film measurements were compared against Pinnacle calculated dose distributions for simple cases to validate the accuracy of the film to reproduce accepted 4MV photon data and 20 MeV electron data. A 10cm x 10cm field was used to treat films in the sagittal cassette within the phantom for both a 20 MeV electron beam and a 4 MV photon beam. The electron beam had an SSD of 100cm, while the photon beam was treated at an SSD of 90cm. A gantry angle of 180° was used to deliver a total of 200 MU was delivered for the 4MV measurement while a gantry angle of 181° was used to deliver 200 MU for the 20 MeV beam. Calculated data was exported from the Pinnacle v8.0 TPS and registered to the measured data using the RIT software. Depth dose profiles comparisons are shown in Figure 2.8 and Figure 2.9 for the 4MV EDR2 and 20 MeV XV data, respectively. The 4MV data exhibits deeper dose penetration depths due to a slight air gap of around 0.8mm in the film cassette. The deeper penetration/higher dose is noticeable for beamlets projected down the film opening of the transverse cassette. The deeper penetration is not an issue for bolus ECT measurements as electrons have already achieved side scatter equilibrium in the bolus.
Figure 2.8: Measured (dashed) versus calculated (solid) depth dose curve for the 4 MV EDR2 measurements. The measured data has a deeper penetration due to an air gap of ≈0.8mm in the film cassette. The blue represents the difference between the measured and calculated dose.

Figure 2.9: Measured versus calculated depth dose curve for the 20 MeV XV measurements. The blue line represents the difference between the measured and calculated dose. The difference between the measured and calculated dose in the tail region is likely due to a poor background film.
2.2.3 ECT Film Measurements

Two to four film measurements were taken in each of five transverse and one sagittal phantom planes for both the parotid and chest wall treatment plans, as represented in Table 2.5. Both the sagittal and transverse film cassettes accepted bare radiographic film, requiring the transfer of the film to and from the cassette to occur within a completely dark environment. To design the film to fit the cassette, a single 10”x12” film was removed from the sealed jacket and secured within the cassette.

Table 2.5: Number of measurements taken for each plane of the parotid and chest wall ECT, IMXT, and mixed beam plans.

<table>
<thead>
<tr>
<th>Parotid</th>
<th>Chest Wall</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Measurement</strong></td>
<td><strong>ECT</strong></td>
</tr>
<tr>
<td><strong>Plane</strong></td>
<td></td>
</tr>
<tr>
<td>Transverse:</td>
<td></td>
</tr>
<tr>
<td>+5.0cm</td>
<td>4</td>
</tr>
<tr>
<td>+2.5cm</td>
<td>3</td>
</tr>
<tr>
<td>Central Axis (CAX)</td>
<td>3</td>
</tr>
<tr>
<td>-2.5cm</td>
<td>3</td>
</tr>
<tr>
<td>-5.0cm</td>
<td>3</td>
</tr>
<tr>
<td>Mid-Sagittal</td>
<td>3</td>
</tr>
</tbody>
</table>

A single-sided razor was used to remove excess film extending beyond the outer cassette edge. For each measurement, a single sheet of white paper, taken from the film package and cut to fit the cassette dimensions, was placed between the bare film and the inner phantom surface containing the registration marks. The purpose of the paper was to tighten the fit of the film in the cassette to eliminate any significant air gap in the cassette, which had caused some dosimetric artifacts as previously described by Dutreix and Dutreix (1969). It was later observed that the paper also reduced dosimetric variations produced by Cherenkov radiation in the registration markings, as discussed in section 2.2.7. Two layers of black photo tape (ProArt, PRO-5360-1) were then placed along the cassette’s surface that exposes the film edge to ambient light, removing unwanted light contamination.

The cassette was inserted into the HIWO phantom, which was placed on the treatment couch. Phantom rotation in the plane of the couch surface (about the vertical axis) was set using the couch’s “exact bar”. For transverse measurements, the y-axis for the light field was aligned parallel to the mid-
sagittal phantom plane with the x-axis aligned to the film opening in the cassette. For both treatment plan setups (parotid and chest wall), vertical positioning was set to an SSD of 105cm using the optical distance indicator (ODI). This configuration served as the base setup and produced central axis (CAX) transverse measurements (Figure 2.10a). To measure the other transverse planes, the table was shifted longitudinally from the CAX, as illustrated in Figure 2.10a-e.

After the phantom had been accurately positioned, the treatment table was no longer moved and the electron bolus was placed on the phantom surface. Markings on the outer edges of the bolus surface were aligned to the axes of the light field. Due to the cylindrical symmetry of the phantom, the rotation of the bolus was set using a bubble level by ensuring the unmilled anterior bolus surface was level. The electron applicator was attached to the gantry and the custom block was placed within the applicator. A qualitative check of the custom field size projection onto the bolus surface was performed prior to each measurement. For all measurements in the negative transverse planes, the phantom was rotated 180° (Figure 2.10d,e). This rotation was necessary for bolus placement as the transverse film measurement plane was located 5.0cm from the edge of the phantom. Comparisons of transverse CAX measurements performed for both phantom orientations showed no variation in measurement.

Figure 2.10: Alignment of HIWO film cassette for measurements. Images depict the phantom and bolus along the mid-sagittal plane. The red line represents the placement of the film for transverse measurements. Transverse measurements are taken by shifting the phantom longitudinally while keeping the bolus in a fixed position, relative to the beam. (a) Transverse CAX Measurement (Base Position). (b) Superior transverse 2.5cm measurement position. (c) Superior transverse 5.0cm measurement position.
Figure 2.10 continued: Alignment of HIWO film cassette for measurements. Images depict the phantom and bolus along the mid-sagittal plane. The red line represents the placement of the film for transverse measurements. Transverse measurements are taken by shifting the phantom longitudinally while keeping the bolus in a fixed position, relative to the beam. (d) Inferior 2.5cm measurement position. Note that the phantom has been rotated by 180° along the vertical axis. (e) Inferior transverse 5.0cm measurement position. (f) Sagittal measurement position.

For sagittal measurements, the y-axis of the light field was again aligned to the mid-sagittal phantom plane, with the x-axis aligned exactly half the distance from the superior and inferior edges of the sagittal cassette. Vertical positioning was set to 105 cm using the ODI. The bolus was positioned using the same procedure discussed for transverse ECT measurements (Figure 2.10f).

To maintain optical densities under 2.0, the bolus ECT plans were irradiated with 50 MU to an approximate dose of ≈44cGy at a point 1cm deep along the intersection between the transverse CAX and sagittal measurement planes.

2.2.4 IMXT Film Measurements

A total of two to four film measurements were taken in the same measurement planes as discussed in section 2.2.3, shown in Table 2.5. For all measurements, lateral position was determined by aligning the y-axis for the light field to the mid-sagittal phantom axis. Vertical positioning was set to an SSD of 98.0 cm for both treatment plans using the ODI with the gantry at 180°. Proper alignment was further maintained using the “exact bar”, increasing the reproducibility of phantom positioning by constraining the rotation of the phantom on the couch. For transverse measurements, the x-axis
aligned to the film opening in the transverse cassette. This configuration served as the base setup and produced central axis (CAX) transverse measurements. To measure the other transverse planes, the table was shifted longitudinally by a set distance from the CAX, which was verified by ruler measurement. For sagittal measurements, longitudinal position was determined by aligning the x-axis of the light field exactly half the distance from the superior and inferior edges of the sagittal cassette. The gantry was rotated through the full range to verify no risk of collision with the phantom or couch.

The IMXT treatment plan was transferred from Pinnacle TPS and delivered through MOSAIQ record and verify software system. The parotid plan received ≈200cGy while the chest wall plan received ≈250cGy.

2.2.5 Mixed Beam Measurements

Phantom alignment for mixed beam treatments was the exactly the same as described in 2.2.3 and 2.2.4. A total of two to four film measurements were taken in each of the film planes discussed in section 2.2.3. Delivery order of IMXT and bolus ECT dose was swapped for each film to maintain efficiency in the measurement process, with no observed effect on the film measurement results.

Kodak EDR2 radiographic film has a modality and energy-dependent response to radiation dose (c.f. optical density to dose curves shown in Figure 2.7) (Gerbi and Dimitroyannis 2003). Using a single modality/energy calibration curve will produce inaccurate absolute dose results for the mixed beam response plans. It can be shown that the measured mixed beam dose distribution \( D_{ij}^{MB} \) is,

\[
D_{ij}^{MB} = \left[ W_{ij}^{ECT} \frac{1}{D^{ECT}(OD_{ij}^{MB})} + W_{ij}^{IMXT} \frac{1}{D^{IMXT}(OD_{ij}^{MB})} \right]^{-1}
\]

Where \( D^{ECT}(OD_{ij}^{MB}) \) is the mixed beam measurement converted to dose using the electron calibration and \( D^{IMXT}(OD_{ij}^{MB}) \) is the mixed beam measurement converted to dose using the photon calibration. In equation 2.2, the weighting factors are determined from the calculated ECT and IMXT dose matrices.
Equation 2.2 can be proven true for a linear OD versus dose relationship in equations 2.6 and 2.7 and following the scheme outlined in Figure 2.11. Because the film response curves are approximately linear it was assumed that these equations were suitable for our derivation.

\[ D_{ij}^{MB, CALC} = D_{ij}^{ECT, CALC} + D_{ij}^{IMXT, CALC} \]

2.5

The mixed beam calibration weighting process was validated by comparing the dose response of films exposed to known ratios of photon and electron radiation. Four films were exposed to 300 cGy each, with the ratios of radiation dose shown in Table 2.6. Using the mixed beam technique, the resulting measured dose for each was 300 cGy ± 2%, shown in Table 2.6. It should be noted that while the delivered dose was the same in all cases, the optical density differs depending on the relative weighting of electron and photon components.

Table 2.6: Results for mixed beam EDR2 OD\(_{NET}\) to dose test (all dose in cGy).

<table>
<thead>
<tr>
<th>Photon Dose:Electron Dose</th>
<th>300:0</th>
<th>200:100</th>
<th>100:200</th>
<th>0:300</th>
</tr>
</thead>
<tbody>
<tr>
<td>Measured Dose</td>
<td>293.5</td>
<td>296.9</td>
<td>305.5</td>
<td>304.3</td>
</tr>
<tr>
<td>% Dose Difference</td>
<td>-2.2%</td>
<td>-1.0%</td>
<td>+1.8%</td>
<td>+1.4%</td>
</tr>
</tbody>
</table>
2.2.7 Cherenkov Radiation Effect

In the present study it was necessary to account for the effect of Cherenkov radiation on bare radiographic film used to measure dose. The age and prior use of the HIWO polystyrene film phantom produced several radiation induced or chemically induced (e.g. film or oily hand contact) stains along the inner surface of the film cassette abutting the bare film (stains were only seen on the lid side of the cassette, which was handled every time a new film was inserted). These stains were darker than the unaffected polystyrene (Figure 2.12) and apparently absorbed a significant amount of the Cherenkov radiation produced in the phantom. Previous measurements (Perrin, Hogstrom and Cheek 2007) showed that approximately 20% of the film response in this HIWO phantom is due to Cherenkov radiation. The stains resulted in decreased doses ranging from 1% - 8% lower in the film regions abutted to them. If it is assumed that both sides of the cassette contributed equal amounts of Cherenkov radiation, then a 1%-8% drop in film response correlates to a range in attenuation of 10%-80% of the Cherenkov light from stains on one side of the cassette.
A correction factor to account for the Cherenkov radiation effect was determined by perpendicularly irradiating a bare film within the cassette (i.e. component order starting from upstream consisted of half of the cassette with the oily surface, film, white paper from film jacket, and other half of cassette) using a large, uniform field ($25\times25\text{ cm}^2$ applicator for electrons), often referred to as a “flood field” and comparing its dose distribution ($D_{ij}^{\text{flood}}$) with that measured using a film protected completely from Cherenkov radiation (i.e. consisted of the bare film surrounded by the black inner surface of the film packet with the white paper removed, inside the cassette). Both exposures were for the same number of MU (dose $\approx 200\text{ cGy}$) with the film at $d_{\text{max}}$ (3.3cm for the 16 MeV and 2.1cm for 20 MeV) and the plastic water surface at 100cm SSD. 4 MV x-ray measurements were taken at a depth of 10cm with a 90cm SSD.

After converting OD to dose and registering the film, the correction factor was determined by:

$$C_{ij} = \left[ \frac{D_{ij}^{\text{flood,exp}}}{D_{ij}^{\text{flood,jacket}}} \right] / \left[ \frac{D_{ij}^{\text{flood,exp}}}{D_{ij}^{\text{flood,jacket}}} \right]$$

Where $D_{ij}^{\text{flood,exp}}$ are the dose values for the first irradiation geometry above and $D_{ij}^{\text{flood,jacket}}$ are the dose values for the second irradiation geometry above (with black surface abutting film). The $ij = \text{center}$
subscript indicates that the values are normalized to values near the center of the cassette where there is uniform Cherenkov contribution and where readings were taken for the film OD vs dose calibration curve.

Calculated and measured registered dose distributions were converted to the MATLAB file format and exported as dose matrices. The Cherenkov masking factor was applied to all measurements prior to data analysis, as shown in Equation 2.9.

\[ D'_{ij} = D_{ij} \times C_{ij} \]  

\( D'_{ij} \) = Cherenkov corrected dose.  
\( D_{ij} \) = Uncorrected dose.  
\( C_{ij} \) = Cherenkov correction factor.  
\( ij \) = dose point location (i,j)

The magnitude of \( C_{ij} \) values can be seen in Figure 2.13, which shows the results of the 2D 20 MeV EDR2 mask.

![2D Cherenkov correction mask for EDR2 20 MeV electron beam.](image)

Figure 2.13: 2D Cherenkov correction mask for EDR2 20 MeV electron beam. The hot spots correlate to stains seen in Figure 2.12.

### 2.2.8 Film Processing

All radiographic film were processed using the SRX-101A medical film processor (Konica Minolta Medical & Graphing, Inc, Wayne, NJ). To ensure accurate results, films were stored for a minimum of 3
hours after exposure prior to developing (Childress and Rosen 2004). The processor was allowed to warm up for a minimum of 45 minutes prior to use, ensuring a temperature of 93°F for the entire processing session. One background film and one film irradiated to approximately 1 OD (200cGy for EDR2 and 25cGy for XV) were developed at the beginning and end of each session and compared to verify development consistency across the entire batch of processed films. A typical development session consisted of developing films in the following order: three blank unprocessed films, a single film with a 10x10cm² field with an OD of ≈1.0 (150 MU for EDR2 and 30 MU for XV), all calibration films, all measurement films, a single film with a 10x10cm² field with an OD of ≈1.0, and a single blank unprocessed film. To reduce the chance of artifacts, the plastic film guide between the developer and fixer was cleaned between processing each film. This cleaning process increased the development time per film to ≈6 minutes. Film measurements were processed with the superior edge first, eliminating streaking artifacts produced along all trailing film edges.

2.2.9 Digitization

All films were scanned using a transmission type scanner (VIDAR DosimetryPRO Advantage, Vidar Systems Corporation, Hendon, Virginia) and digitized into 178 μm x 178 μm dose points. The scanner uses a white light emitting diode source and a solid state detector array consisting of 89 μm x 89 μm charge-coupled device (CCD) detector elements. The 16-bit grayscale readings of 4 CCD (2x2) detectors were averaged to produce each OD point. The measured values were registered to the Pinnacle calculated planar dose values as described in Section 2.3.2.

The variation in response across the scanner was evaluated by scanning a film, flipping it around the vertical axis, and scanning a second time. The resulting digital scans were co-registered with the variation in scanner response is shown in Figure 2.14. Variation in scanner response was less than 1% across all detectors.
2.3 Aim 3: Evaluate the Accuracy of the PBA and PBRA by Comparing Calculated Dose Distributions to the Set of Parotid and Chest Wall Measurements

2.3.1 Electron Dose Calculation Algorithms

Electron beam dose calculations were made using the Pinnacle PBA and an in-house implementation of the PBRA. The pencil-beam algorithm of Hogstrom et al. (1981), as implemented in 3D by Starkschall et al. (1993), has been incorporated into the Pinnacle TPS. The Pinnacle PBA has been commissioned for the Varian 2100EX according to Hogstrom and Steadham (1996) and the Pinnacle User’s Manual by our clinical physics group. Similarly, the PBRA has been commissioned by our research physics group.

2.3.1.1 Pencil Beam Algorithm CT Lookup Table

Electron PBA dose calculations require a look-up table to accurately convert CT Hounsfield Units (HU) to relative linear collisional stopping and linear scattering powers (Hogstrom et al. 1981).
v8.0 TPS contains an extra step by converting the HU to density before converting the density to the relative stopping and scattering powers. For the HIWO polystyrene phantom it was necessary to modify the standard clinical Pinnacle HU to density and density to stopping and scattering power tables used for patients. The ranges of HU for the HIWO polystyrene phantom (~980-1080) and wax bolus (~880-930) were determined from the phantoms CT image set (Figure 2.15). Look-up tables in the Pinnacle TPS were modified to map the HIWO polystyrene and wax bolus relative stopping and scattering power ratios to the density of water (Table 2.7b) and the density was converted to the correct range of HU values (Table 2.7a). Using these tables, the stopping and scattering powers (±FWHM) for the wax bolus relative to water were 0.952 ± 0.005 and 0.729 ± 0.004, respectively (Low and Hogstrom 1994). Relative stopping and scattering powers for the HIWO polystyrene, determined from the phantom’s density (ρ = 1.054gcm⁻³) and mass stopping and scattering powers from AAPM TG-25 (Khan et al. 1991), were 1.021 ± 0.005 and 0.868 ± 0.004, respectively. These tables were used to modify the NewElectronStoppingPower8.0.db file in the Pinnacle TPS.

![Figure 2.15: Histogram of the Hounsfield values for the entire CT image volume of the phantom and bolus. The average and FWHM HU values for the phantom and bolus are represented by the peaks on the right and left, respectively.](image_url)
Table 2.7: The Hounsfield number to density conversions (a) and density to S & T conversions (b) used in Pinnacle. Note that a mean CT value for machinable wax of 908 equates to a density of 0.952, which corresponds to $s_{water}^{Wax}$ and $t_{water}^{Wax}$ of 0.952 and 0.729.

<table>
<thead>
<tr>
<th>Hounsfield Unit</th>
<th>Density (ρ) [g-cm$^{-3}$]</th>
<th>Density (ρ) [g-cm$^{-3}$]</th>
<th>Relative Stopping Power ($S_{rel}$)</th>
<th>Relative Scattering Power ($T_{rel}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0.000</td>
<td>0.000</td>
<td>0.000</td>
<td>0.000</td>
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<tr>
<td>895</td>
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<td>1892</td>
<td>1.609</td>
<td>1.422</td>
<td>1.422</td>
<td>1.863</td>
</tr>
<tr>
<td>12000</td>
<td>11.300</td>
<td>12.700</td>
<td>12.700</td>
<td>88.000</td>
</tr>
</tbody>
</table>

(a) (b)

2.3.1.2 Pencil Beam Redefinition Algorithm

Dose distributions using the PBRA were calculated using in-house “research” software. A copy of the bolus ECT PBA treatment plan served as the PBRA treatment plan. The PBRA program requires the plan.trial, plan.points, plan.roi, ImageSet_0.img, and ImageSet_0.header files, extracted from the Pinnacle PBA plan, to ensure identical geometric and planning conditions. These files were used in the PBRA program to produce a dose distribution equal in size to the dose grid in the PBA plan. To view the PBRA dose distribution, a copy of the PBA plan in the Pinnacle TPS was made and the PBRA output dose file replaced the plan.Trial.binary file within this plan.

The PBRA uses HU values, relative stopping power, and relative scattering power tables equivalent to those utilized by the Pinnacle PBA. For this study, all dosimetric parameters (MU, beam energy, gantry angle, isocenter, etc.) were defined in the Pinnacle TPS and remained identical between the PBA and PBRA plans. As discussed in section 2.1.5, the virtual source to surface distance is approximately 90cm; however, Pinnacle TPS hard codes the virtual source as 100cm, which was the value used in both PBA and PBRA calculations. This should have little impact on the dose calculation as the phantom was set to $\approx$102cm SSD. The maximum distance the collimator edge is off axis was 11cm, which results in the radiation field edge lying 0.9mm outside the calculated field edge.
2.3.2 Registration of Calculation to Measurement

2.3.2.1 Calculated Data Export

Calculated planar dose distributions corresponding to each measurement plane (Table 2.5) were exported from Pinnacle to RIT as ASCII files. Separate planar PBA and PBRA dose distributions were exported for the ECT, IMXT, and mixed beam treatment plans. Each planar dose file consisted of a 720 x 800 matrix with 0.5mm dose point dimensions. The plane center was defined as the apex of the circular cross section for transverse planes and the center of the top edge of the sagittal cassette for sagittal planes. These planar dose files were read into the RIT software and subsequently registered with the corresponding planar film measurements.

2.3.2.2 Registration Process

For a given measurement plane, the calculated planar dose ASCII file was opened in the RIT software as a target image and a custom registration template consisting of five registration points (P1-P5 in Figure 2.16) was applied. These registration points were a set distance (X,Y) from the planar dose isocenter located at the apex of the transverse plane or the top center of the sagittal plane. The transverse and sagittal cassettes were carefully marked with ink dots at the same locations relative to the planar dose isocenter. Processed films were then aligned in the cassette, and a small pin-prick hole was made at the registration locations (ink demarcations on the inner surface of the cassette could be seen through the developed film). Each film was then digitized (Section 2.2.9) and separately opened as a reference image in the RIT software. The marked holes on the measured films were registered to the corresponding template locations in the calculated planar dose files. Once registered, the film measurement data were converted from 0.179mm x 0.179mm dose points to dose points representing the average dose over a 0.5mm x 0.5mm area of the film measured data.
2.3.3 Determination and Validation of Measured Dose and Estimation of Error

Multiple measurements taken in the same plane for the same radiation conditions were averaged at each dose point location. Both the mean and standard error in the mean were calculated.

\[ \overline{D'_{ij}} \text{, the mean corrected dose, is given by} \]

\[ \overline{D'_{ij}} = \frac{1}{N} \sum_{k=1}^{N} D'_{ijk} \]

\[ \text{where } D'_{ijk} = \text{Cherenkov corrected dose at point i,j in film k and } N = \# \text{ of film measurements.} \]

\[ \sigma_{\overline{D'_{ij}}} \text{, the standard error of the corrected mean dose, is given by} \]

\[ \sigma_{\overline{D'_{ij}}} = \left[ \frac{1}{N} \sum_{k=1}^{N} \left( \frac{D'_{ij,k} - \overline{D'_{ij}}}{N-1} \right)^2 \right]^{1/2} \]

\[ \%\sigma_{\overline{D'_{ij}}} \text{, the percent standard error of the mean dose, is given by} \]

\[ \%\sigma_{\overline{D'_{ij}}} = \frac{\sigma_{\overline{D'_{ij}}}}{D_{Norm}} \times 100\% \]

\[ \text{where, } D_{Norm} = \text{Dose normalization value for the given modality, as seen in Section 2.3.4.} \]
\( \sigma_{ij}^{\text{dist}} \), the standard error for distance to agreement (mm) in the HGR is given by

\[
\sigma_{ij}^{\text{dist}} = \frac{\sigma \overline{D}_{ij}}{|\nabla D_{ij}|}
\]

where, \( \nabla D_{ij} \) = Dose gradient at dose point \( ij \).

The percent standard errors of the mean, averaged for all dose points in a 1cm\(^2\) region, were superimposed on isodose plots of the mean dose for all measurement planes. Also, additional depth dose plots illustrating the central axis dose for all measurements in a single plane were produced. Confidence in the accuracy of the data was accomplished by comparing depth dose data along the intersections of sagittal and transverse measurements.

### 2.3.4 Evaluation of the Accuracy of Calculated Dose Distributions

Calculated dose distributions for ECT, IMXT, and mixed beam plans were compared to their respective measured dose distributions. For each treatment plan, both the calculated and measured dose distributions in all planes were normalized to a single dose value. This value was determined from the high dose, low gradient region of the CAX measurement plane at a point 1cm deep along the mid-sagittal plane. Dose normalization values for all six treatment plans are given in Table 2.8.

<table>
<thead>
<tr>
<th>Treatment Plan</th>
<th>Dose Normalization Value (cGy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Parotid ECT</td>
<td>44.5</td>
</tr>
<tr>
<td>Chest Wall ECT</td>
<td>43.6</td>
</tr>
<tr>
<td>Parotid IMXT</td>
<td>200.0</td>
</tr>
<tr>
<td>Chest Wall IMXT</td>
<td>250.0</td>
</tr>
<tr>
<td>Parotid Mixed Beam</td>
<td>300.0</td>
</tr>
<tr>
<td>Chest Wall Mixed Beam</td>
<td>299.9</td>
</tr>
</tbody>
</table>

Table 2.8: Normalization values used to normalize both measured and calculated dose distributions

To assess the accuracy of the dose calculation, the ECT and mixed beam 2D dose distributions were divided into three regions defined by the gradient of the dose distributions: high-dose (low-gradient) region (HDR), high-gradient region (HGR), and low-dose (low-gradient) region (LDR). These regions are defined with respect to relative dose in Table 2.9.
Table 2.9: Dosimetric regions of the ECT and mixed beam measurement planes and evaluation criteria.

<table>
<thead>
<tr>
<th>Region</th>
<th>ECT Measurements</th>
<th>Mixed Beam Measurements</th>
<th>Criteria</th>
</tr>
</thead>
<tbody>
<tr>
<td>High-dose region (HDR)</td>
<td>Depth &gt; R_{80}</td>
<td>Depth &gt; R_{90}</td>
<td>±4% Dose</td>
</tr>
<tr>
<td>High-gradient region (HGR)</td>
<td>R_{80} &gt; Depth &gt; R_{10}</td>
<td>R_{90} &gt; Depth &gt; R_{30}</td>
<td>±2mm DTA</td>
</tr>
<tr>
<td>Low-dose region (LDR)</td>
<td>R_{10} &gt; Depth</td>
<td>R_{30} &gt; Depth</td>
<td>±4% Dose</td>
</tr>
</tbody>
</table>

A passing criteria of ±4% dose difference or ±2mm distance to agreement (DTA) was applied between the calculated and measured values. Dose difference was analyzed at each dose point using equation 2.10.

\[
% \text{DoseDiff}_{ij} = \left( \frac{D^c_{ij} - D^{\text{cal}}_{ij}}{D^c_{ij}} \right) \times 100\%
\]

2.12

The DTA was calculated at each dose point by determining the distance from the dose point center to the nearest contour of the same value in the 2D planar dose matrix. Contours were set using the contourc function in MATLAB, which determined specific dose values along the edges of dose points and then linearly interpolated between each value (Figure 2.17a). The DTA function determined the distance from the measured dose point to each calculated linear isodose segment and set the DTA as the smallest of these values (Figure 2.17b).

![Calculated Dose Distribution](image)

Figure 2.17: Determination of DTA between measured and calculated dose distributions. In (a), the contourc function determined the location of the value 7 along the edges of the dose points in the calculated dose distribution. Isodose lines (dashed line) were produced by linearly interpolating between each boundary value.
b. Measured Dose Distribution

Figure 2.20 continued: Determination of DTA between measured and calculated dose distributions. Part (b) gives the measured dose matrix for the same region as part a. The shortest distance from the center of the measured dose points (value of 7 in (b)) to each calculated isodose segment of the same value is calculated and the DTA is set as the shortest distance, represented as 1.3mm.
Chapter 3: Results and Discussion

3.1 Aim 1: Develop in-phantom bolus ECT, IMXT, and mixed beam treatment plans.

This section presents the Pinnacle calculated dose distribution for the ECT, IMXT, and mixed beam treatment planes used in this study. Each dose distribution is plotted in either transverse or mid-sagittal planes that correspond to where the measurements were made. This section shows only the PBA calculated dose distributions in the mid-transverse and sagittal planes. PBA dose distributions in all 5 transverse planes can be found in Appendix B, where calculations are compared with measurements. PBRA dose calculations for the 5 transverse planes are found in Appendix B as well, with PBRA calculated dose distributions in the sagittal plane for the ECT and mixed beam plans can be found in Figures 3.44, 3.69, 3.80, and 3.101. For orientation purposes the bolus cross-section in the plane of measurement is demarcated, although it is present only for the electron dose component.

In Figures 3.1–3.6, the PTV is seen as an orange color wash region. Also, the ECT and mixed beam dose distributions show the uncollimated (dashed line) and collimated (solid line) edges of the electron beam. Note that the dark blue 90% isodose line closely conforms to the distal side of both the parotid and chest wall PTV.

Mixed beam, ECT, and IMXT dose distributions for central axis and sagittal planes representing both the chest wall and parotid panes are represented in Figures 3.1–3.6. All six Figures contain the bolus and PTV (seen as an orange region), while the ECT and mixed beam dose distributions also show the uncollimated (dashed line) and collimated (solid line) edges of the electron beam. The blue 90% isodose line closely conforms to the distal side of both the chest wall and parotid PTV.

ECT, IMXT, and mixed beam dose distributions for the parotid are seen in Figures 3.1, 3.2, and 3.3, respectively. Similar dose distributions for the chest wall are shown in Figures 3.4, 3.5, and 3.6, respectively. Note the dose heterogeneities present in the ECT planes, evident by the range of dose from greater than 100% (red) to 90%. On the other hand, note the dose homogeneity of the mixed
beam plan, evident by the speckled 100% isodose islets. The IMXT component of the mixed beam plane also results in a sharper dose gradient around the PTV, but a larger volume of low dose (10%-30%).

Figure 3.1: (a) Parotid bolus ECT CAX and (b) sagittal Pinnacle calculated dose distributions.
Figure 3.2: (a) Parotid IMXT CAX and (b) sagittal Pinnacle calculated dose distributions.
Figure 3.3: (a) Parotid mixed beam and (b) sagittal Pinnacle calculated dose distributions.
Figure 3.4: (a) Chest Wall bolus ECT CAX and (b) sagittal Pinnacle calculated dose distributions.
Figure 3.5: (a) Chest wall IMXT CAX and (b) sagittal Pinnacle calculated dose distributions.
Figure 3.6: (a) Chest wall mixed beam CAX and (b) sagittal Pinnacle calculated dose distributions.
3.2 Aim 2: Measure Dose Distributions for the Parotid and Chest Wall Bolus ECT, IMXT, and Mixed Beam Treatment Plans

For this aim results of measured dose distributions for the transverse central axis and mid-sagittal planes for the ECT, IMXT, and mixed beam plans are presented. Results are highlighted in the text, and measured data for all five transverse planes are provided in Appendix A. Isodose plots of the mean dose are shown, and the percent standard error of the mean dose (HDR and LDR regions) and mean DTA in mm (HGR, grayscale), averaged over 1cm² regions in the data set, are superimposed on the isodose curves.

To appreciate the measured precision, depth-dose curves along a vertical line at the -5mm lateral position in the related 2D isodose sets are compared for multiple film measurements. Also, to increase confidence in the data accuracy, depth doses at the intersections of the sagittal plane with the 5 transverse planes are compared.

3.2.1 Parotid Case

Sections 3.2.1.1 through 3.2.1.3 highlight results for the transverse central axis and mid-sagittal plane for ECT, IMXT, and mixed beam parotid plans. All depth-dose curves comparing multiple film measurements correspond to a red vertical dashed line at the -5mm lateral position in the related 2D isodose data sets. The percent standard error of the mean dose (HDR and LDR regions) and mean DTA in mm (HGR regions, grayscale), averaged over 1cm² regions of the data set are superimposed on the isodose curves.

3.2.1.1 Parotid Electron Conformal Therapy Measurements

The CAX depth dose curves for the 3 RGF measurements in the sagittal plane and the 3 RGF measurements in the CAX planes are compared in Figures 3.8 and 3.10, respectively. Results show a maximum percent dose difference of ~1.3% in the high dose region for both the CAX and sagittal planes. The maximum DTA difference in the high gradient region was approximately 1.5mm in the CAX plane and 1.0mm in the sagittal plane.
The average measured dose distribution for the sagittal and CAX planes are shown in Figures 3.7 and 3.9, respectively. All parotid ECT isodose measurements were normalized to 44.5 cGy, a value obtained from the Pinnacle TPS calculated dose at a point 2.1cm deep from the surface of the bolus along the line representing the intersection of the CAX and mid-sagittal planes. For all parotid ECT measurement planes, the percent standard error of the mean is typically less than 1% in the HDR and LDR, while the standard error of the mean DTA is typically less than 0.35mm in the HGR. These results indicate very high precision of the measured data.

Figure 3.7: Parotid ECT sagittal 2D isodose plot, normalized to 44.5 cGy.

Figure 3.8: Parotid ECT sagittal depth dose of three film measurements along the red dashed line in Figure 3.7.
Figure 3.9: Parotid ECT CAX 2D isodose plot, normalized to 44.5 cGy.

Figure 3.10: Parotid ECT CAX depth dose of three film measurements along the red dashed line in Figure 3.9.

Figure 3.11: Parotid ECT depth dose along the intersection between the CAX and sagittal measurement planes seen as the red dashed line.
The percent standard error of the mean and standard deviation of the standard error of the mean for the HDR, HGR, and LDR was calculated for all six measurement planes (Table 3.1). Values for the HDR and LDR were normalized to 44.5 cGy.

The standard error in the distance to agreement for the HGR was at least 30% larger in the CAX plane than the other measurement planes. This increase is attributed to a slight difference (~0.5mm) in the registration process for one CAX film. The mean percent standard error in the Inferior 2.5 cm plane was 30%-40% higher than all the other planes. One of the film measurements was ~1% different than the other film measurements in the Inferior 2.5cm plane, resulting in a larger percent standard error.

Table 3.1: Parotid ECT measurement statistics. Mean percent standard error and standard deviation of the mean percent standard error in the three parotid ECT dose regions of each measurement plane.

<table>
<thead>
<tr>
<th>Measurement Plane</th>
<th>High Dose Region (HDR)</th>
<th>High Gradient Region (HGR)</th>
<th>Low Dose Region (LDR)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$\sigma_{err}$ (%)</td>
<td>$\sigma_{\sigma_{err}}$ (%)</td>
<td>$\sigma_{err DT \Delta}$ (mm)</td>
</tr>
<tr>
<td>Central Axis (CAX)</td>
<td>0.50</td>
<td>0.15</td>
<td>0.29</td>
</tr>
<tr>
<td>Superior 2.5 cm</td>
<td>0.40</td>
<td>0.13</td>
<td>0.23</td>
</tr>
<tr>
<td>Superior 5.0 cm</td>
<td>0.83</td>
<td>0.22</td>
<td>0.24</td>
</tr>
<tr>
<td>Inferior 2.5 cm</td>
<td>0.90</td>
<td>0.26</td>
<td>0.22</td>
</tr>
<tr>
<td>Inferior 5.0 cm</td>
<td>0.47</td>
<td>0.12</td>
<td>0.19</td>
</tr>
<tr>
<td>Sagittal</td>
<td>0.48</td>
<td>0.13</td>
<td>0.17</td>
</tr>
</tbody>
</table>

Precision of the each transverse measured data set was further evaluated by comparing the depth dose curves corresponding to the intersection line between the transverse and sagittal planes. The depth dose along the intersection between the sagittal and CAX measurements is shown in Figure 3.11 and correlates to the -5mm position in both Figures 3.7 and 3.9.

Differences between the sagittal and CAX transverse depth dose curves in the HDR are as great as 3.5% relative to 44.5cGy. The maximum expected dose difference in this region, as determined from Table 3.1, is 1.7%. This larger difference in dose can likely be attributed to individual dose points having a greater standard error than represented in Table 3.1. It should also be noted that variations in film measurements can be 2%-3% between films from the same batch. The application of a Cherenkov mask
increases the noise in the measurement by multiplying each dose measurement by normalized mask. This increased noise is particularly noticeable in the regions with higher dose.

3.2.1.2 Parotid Intensity Modulated X-ray Therapy Measurements

The CAX depth-dose curves for the 4 RGF measurements in the sagittal plane and the 3 RGF measurements in the CAX planes are compared in Figures 3.13 and 3.15, respectively. Results show a maximum percent dose difference of ~3.3% in both the CAX and sagittal planes.

The average measured dose distribution for the sagittal and CAX planes are shown in Figures 3.12 and 3.14, respectively. All parotid IMXT isodose measurements were normalized to 200 cGy. For all parotid IMXT measurement planes, the percent standard error of the mean is typically less than 1%, indicating very high precision of the measured data.

Mean standard error and standard deviation values are similar for the transverse measurement planes (Table 3.2). The standard error in the sagittal plane is lower due to smaller standard error values in the low dose regions located outside the beam (Figure 3.12).

Precision of the each transverse measured data set was further evaluated by comparing the depth dose curves corresponding to the intersection line between the transverse and sagittal planes. The depth dose along the intersection between the sagittal and CAX measurements is shown in Figure 3.16 and correlates to the -5mm position in both Figures 3.12 and 3.14.

Differences between the sagittal and CAX transverse depth dose curves are as great as 2.6% relative to 200cGy. The maximum expected dose difference, as determined from Table 3.2, is 1.4%. This larger difference in dose can likely be attributed to individual dose points having a greater standard error than represented in Table 3.2. Furthermore, the mean standard error for both the sagittal and transverse planes are likely an underestimation for the high dose regions that exhibit the greatest difference between the sagittal and transverse depth dose curves.
Figure 3.12: Parotid IMXT measured sagittal 2D isodose plot.

Figure 3.13: Parotid IMXT sagittal depth dose of three film measurements (Red line in Figure 3.12).

Figure 3.14: Parotid IMXT measured CAX 2D isodose plot.
Figure 3.15: Parotid IMXT CAX depth dose of three film measurements (Red line in Figure 3.14).

Figure 3.16: Parotid IMXT comparing depth dose along the intersection between the CAX and sagittal measurement planes.

Table 3.2: Parotid IMXT measurement statistics. Mean standard error of the mean and standard deviation of the mean standard error of the mean for each measurement plane of the parotid IMXT plan.

<table>
<thead>
<tr>
<th>Measurement Plane</th>
<th>$\bar{\sigma}_{err}$ (%)</th>
<th>$\sigma_{err}$ (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Central Axis (CAX)</td>
<td>0.47</td>
<td>0.25</td>
</tr>
<tr>
<td>Superior 2.5 cm</td>
<td>0.48</td>
<td>0.28</td>
</tr>
<tr>
<td>Superior 5.0 cm</td>
<td>0.41</td>
<td>0.25</td>
</tr>
<tr>
<td>Inferior 2.5 cm</td>
<td>0.45</td>
<td>0.3</td>
</tr>
<tr>
<td>Inferior 5.0cm</td>
<td>0.43</td>
<td>0.29</td>
</tr>
<tr>
<td>Sagittal</td>
<td>0.38</td>
<td>0.29</td>
</tr>
</tbody>
</table>
3.2.1.3 Parotid Mixed Beam Therapy Measurements

The CAX depth-dose curves for the 3 measurements in the sagittal plane and the 3 RGF measurements in the CAX planes are compared in Figures 3.18 and 3.20, respectively. Results show a maximum percent dose difference of ~3.2% in the high dose region for the CAX and 2.0% in the high dose region for the sagittal planes. The maximum DTA difference in the high gradient region was approximately 1.9mm in the CAX plane and 1.3mm in the sagittal plane.

The average measured dose distribution for the sagittal and CAX planes are shown in Figures 3.17 and 3.19, respectively. All parotid mixed beam isodose measurements were normalized to 300 cGy. For all parotid ECT measurement planes, the percent standard error of the mean is typically less than 1% in the HDR and LDR, while the standard error of the mean DTA is typically less than 0.35mm in the HGR. These results indicate very high precision of the measured data.

The percent standard error of the mean and standard deviation of the standard error of the mean for the HDR, HGR, and LDR was calculated for all six measurement planes (Table 3.3). Values for the HDR and LDR were normalized to 300 cGy.

Mean standard error values in the HDR were similar for the CAX and sagittal planes, while the other planes exhibited a much lower standard error. The standard error in the mean for the HGR and LDR is about twice as large in the superior 5.0cm plane then the other measurement planes. This difference is likely attributed to a slight difference (<0.5mm) in the registration process. In Figure 3.17 there is a region on the left side of the distribution between 30% and 50% isodose lines that has standard error of the mean DTA exceeding 1.5mm. The gradient of the dose in this region levels out, which could produce large differences in distance to agreement between measurements while still maintaining small differences in dose.
Figure 3.17: Parotid measured mixed beam sagittal isodose.

Figure 3.18: Parotid mixed beam sagittal depth dose of three film measurements (Red line in Figure 3.17).

Figure 3.19: Parotid measured mixed beam CAX isodose plot.
Figure 3.20: Parotid mixed beam CAX depth dose of three measurements (Red line in Figure 3.19).

Figure 3.21: Parotid mixed beam depth dose along the intersection between the CAX and sagittal measurement planes.

Table 3.3: Parotid mixed beam measurement statistics. Mean percent standard error and standard deviation of the mean percent standard error in the three dose regions of each measurement plane.

<table>
<thead>
<tr>
<th>Measurement Plane</th>
<th>High Dose Region (HDR)</th>
<th>High Gradient Region (HGR)</th>
<th>Low Dose Region (LDR)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$\sigma_{err}$ (%)</td>
<td>$\sigma_{err}$ (%)</td>
<td>$\sigma_{err}DTA$ (mm)</td>
</tr>
<tr>
<td>Central Axis (CAX)</td>
<td>0.63</td>
<td>0.16</td>
<td>0.39</td>
</tr>
<tr>
<td>Superior 2.5 cm</td>
<td>0.60</td>
<td>0.13</td>
<td>0.25</td>
</tr>
<tr>
<td>Superior 5.0 cm</td>
<td>0.59</td>
<td>0.16</td>
<td>0.39</td>
</tr>
<tr>
<td>Inferior 2.5 cm</td>
<td>0.77</td>
<td>0.18</td>
<td>0.32</td>
</tr>
<tr>
<td>Inferior 5.0 cm</td>
<td>0.71</td>
<td>0.14</td>
<td>0.29</td>
</tr>
<tr>
<td>Sagittal</td>
<td>0.78</td>
<td>0.21</td>
<td>0.31</td>
</tr>
</tbody>
</table>
Differences between the sagittal and CAX transverse depth dose curves in the HDR are as great as 4.0% relative to 300 cGy, exceeding the uncertainty in both measurements. Possible reasons for the larger difference are similar to those discussed previously in section 3.2.1.1. The sagittal depth dose curve is greater than the transverse in the low dose region for all three measured treatment modalities. In all three measurement modalities, the sagittal and CAX depth dose curves are not significantly different, as the standard error bars overlap along the majority of the depth dose profiles.

3.2.2 Chest Wall Case

3.2.2.1 Chest Wall Electron Conformal Therapy Measurements

The CAX depth-dose curves for the 3 RGF measurements in the sagittal plane and the 3 RGF measurements in the CAX planes are compared in Figures 3.23 and 3.25, respectively. Results show a maximum percent dose difference of ~1.6% in the high dose region for the CAX plane and 4.9% for the sagittal plane. The maximum DTA difference in the high gradient region was approximately 1.2mm in both the CAX and sagittal planes.

The average measured dose distribution for the sagittal and CAX planes are shown in Figures 3.22 and 3.24, respectively. All chest wall ECT isodose measurements were normalized to 43.5 cGy. For all chest wall ECT measurement planes, the percent standard error of the mean is typically less than 1.25% in the HDR and LDR, while the standard error of the mean DTA is typically less than 0.4mm in the HGR. These results indicate very high precision of the measured data.

The standard error in the HDR was at least 25% larger in the sagittal plane than the other measurement planes. One of the film measurements was ~2-3% different than the other film measurements in the sagittal plane, resulting in a larger percent standard error.
Figure 3.22: Chest Wall measured ECT sagittal isodose.

Figure 3.23: Chest Wall ECT sagittal depth dose of three film measurements (Red line in Figure 3.22).

Figure 3.24: Chest Wall measured ECT CAX isodose.
Figure 3.25: Chest Wall ECT CAX depth dose of three film measurements (Red line in Figure 3.24).

Figure 3.26: Chest wall ECT comparing depth dose along the intersection between the CAX and sagittal measurement planes.

Error bars for the sagittal and CAX transverse depth dose curves are spaced every 5mm along the depth dose and represent the standard error of the mean dose for a 5mm$^2$ region surrounding the dose point with the error bar. The error bars of the CAX transverse and sagittal depth dose curves overlap except in the first 1cm from the phantom surface.
Table 3.4: Chest wall ECT measurement statistics. The mean percent standard error and standard deviation of the mean standard in the three dose regions of each measurement plane.

<table>
<thead>
<tr>
<th>Measurement Plane</th>
<th>High Dose Region (HDR)</th>
<th>High Gradient Region (HGR)</th>
<th>Low Dose Region (LDR)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$\sigma_{%\text{err}}$ (%)</td>
<td>$\sigma_{%\text{err}}$ (%)</td>
<td>$\sigma_{%\text{err}}$ (%)</td>
</tr>
<tr>
<td>Central Axis (CAX)</td>
<td>0.66 0.19</td>
<td>0.33 0.07</td>
<td>0.14 0.09</td>
</tr>
<tr>
<td>Superior 2.0 cm</td>
<td>0.86 0.24</td>
<td>0.24 0.05</td>
<td>0.16 0.09</td>
</tr>
<tr>
<td>Superior 6.0 cm</td>
<td>0.82 0.21</td>
<td>0.25 0.08</td>
<td>0.16 0.07</td>
</tr>
<tr>
<td>Inferior 2.0 cm</td>
<td>0.82 0.18</td>
<td>0.36 0.05</td>
<td>0.18 0.05</td>
</tr>
<tr>
<td>Inferior 6.0 cm</td>
<td>0.79 0.2</td>
<td>0.18 0.04</td>
<td>0.15 0.07</td>
</tr>
<tr>
<td>Sagittal</td>
<td>1.01 0.25</td>
<td>0.26 0.04</td>
<td>0.15 0.07</td>
</tr>
</tbody>
</table>

3.2.2.2 Chest Wall Intensity Modulated X-ray Therapy Measurements

The CAX depth-dose curves for the 2 RGF measurements in the sagittal plane and the 3 RGF measurements in the CAX plane are compared in Figures 3.28 and 3.30, respectively. Results show a maximum percent dose difference of ~3.4% in the CAX plane and 3.6% in the sagittal plane.

The average measured dose distribution for the sagittal and CAX planes are shown in Figures 3.27 and 3.29, respectively. All chest wall IMXT isodose measurements were normalized to 250 cGy. For all chest wall IMXT measurement planes, the percent standard error of the mean is typically less than 1.5%, indicating very high precision of the measured data.

Mean standard error and standard deviation values are slightly higher for the Inferior 6.0cm, Superior 2.0cm, and sagittal measurement planes (Table 3.5). The increased variation tends to occur in regions with a high dose gradient, indicating measurement alignment or registration differences may have contributed to the larger standard error. The standard error in the sagittal plane is higher due to only using two measurements to calculate the uncertainty.

Precision of each transverse measured data set was further evaluated by comparing the depth dose curves corresponding to the intersection line between the transverse and sagittal planes. The depth dose along the intersection between the sagittal and CAX measurements is shown in Figure 3.31 and correlates to the -5mm position in both Figures 3.27 and 3.29.
Figure 3.27: Chest Wall measured IMXT sagittal isodose.

Figure 3.28: Chest Wall IMXT sagittal depth dose of two film measurements (Red line in Figure 3.27).

Figure 3.29: Chest Wall measured IMXT CAX isodose.
Figure 3.30: Chest Wall IMXT CAX depth dose of three film measurements (Red line in Figure 3.29).

Figure 3.31: Chest wall IMXT comparing depth dose along the intersection between the CAX and sagittal measurement planes.

The sagittal and CAX transverse depth dose curves differ by a significant amount between 4cm and 6cm depth, with the CAX dose ~6% lower than the sagittal depth dose curve. This difference exceeds the uncertainty in both measurement sets.

Table 3.5: Chest wall IMXT measurement statistics. The mean percent standard error and standard deviation of the mean percent standard error for each measurement plane.

<table>
<thead>
<tr>
<th>Measurement Plane</th>
<th>$\sigma_{err}$ (%)</th>
<th>$\sigma_{err}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Central Axis (CAX)</td>
<td>0.55</td>
<td>0.36</td>
</tr>
<tr>
<td>Superior 2.0 cm</td>
<td>0.88</td>
<td>0.60</td>
</tr>
<tr>
<td>Superior 6.0 cm</td>
<td>0.39</td>
<td>0.32</td>
</tr>
<tr>
<td>Inferior 2.0 cm</td>
<td>0.63</td>
<td>0.46</td>
</tr>
<tr>
<td>Inferior 6.0cm</td>
<td>0.84</td>
<td>0.56</td>
</tr>
<tr>
<td>Sagittal</td>
<td>0.78</td>
<td>0.78</td>
</tr>
</tbody>
</table>
3.2.2.3 Chest Wall Mixed-Beam Therapy Measurements

The CAX depth-dose curves for the 3 RGF measurements in the sagittal plane and the 2 RGF measurements in the CAX planes are compared in Figures 3.31 and 3.33, respectively. Results show a maximum percent dose difference of ~2.9% in the high dose region for the CAX and 1.2% in the high dose region for the sagittal planes. The maximum DTA difference in the high gradient region was approximately 0.5mm in the CAX plane and 0.5mm in the sagittal plane.

The average measured dose distribution for the sagittal and CAX planes are shown in Figures 3.32 and 3.34, respectively. All chest wall mixed beam isodose measurements were normalized to 300 cGy. For all chest wall ECT measurement planes, the percent standard error of the mean is typically less than 1.7% in the HDR and LDR, while the standard error of the mean DTA is typically less than 0.5mm in the HGR. These results indicate very high precision of the measured data.

The standard error in the HDR was at least 40% smaller in the sagittal and Inferior 2.0cm planes than the other measurement planes. The standard error of the distance to agreement in the HGR was at least 50% higher for the transverse Superior 6.0cm. This measurement plane only had two measurements, which possibly contributes to the larger discrepancy in the DTA standard error. The transverse Superior 6.0cm and Superior 2.0cm measurement planes had a larger standard error in the LDR.

Precision of the each transverse measured data set was further evaluated by comparing the depth dose curves corresponding to the intersection line between the transverse and sagittal planes. The depth dose along the intersection between the sagittal and CAX measurements is shown in Figure 3.36 and correlates to the -5mm position in both Figures 3.32 and 3.34.

The agreement between the transverse CAX and sagittal measured depth dose curves was within the standard error, with the only discrepancies occurring in the first 5mm from the phantom surface and a small deviation around 6cm depth.
Figure 3.32: Chest wall measured mixed beam sagittal isodose.

Figure 3.33: Chest wall mixed beam sagittal depth dose of three film measurements (Red line in Figure 3.32).

Figure 3.34: Chest wall mixed beam CAX isodose.
Figure 3.35: Chest wall measured mixed beam CAX depth dose of two film measurements (Red line in Figure 3.34).

Figure 3.36: Chest wall mixed beam comparing depth dose along the intersection between the CAX and sagittal measurement planes.

Table 3.6: Chest wall mixed beam measurement statistics. The mean percent standard error and standard deviation of the mean standard error in the three dose regions of each measurement plane.

<table>
<thead>
<tr>
<th>Measurement Plane</th>
<th>High Dose Region (HDR)</th>
<th>High Gradient Region (HGR)</th>
<th>Low Dose Region (LDR)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$\bar{\sigma}_{err}$ (%)</td>
<td>$\bar{\sigma}_{err}$ (%)</td>
<td>$\bar{\sigma}_{errDTA}$ (mm)</td>
</tr>
<tr>
<td>Central Axis (CAX)</td>
<td>1.03</td>
<td>0.24</td>
<td>0.41</td>
</tr>
<tr>
<td>Superior 2.0 cm</td>
<td>1.09</td>
<td>0.26</td>
<td>0.43</td>
</tr>
<tr>
<td>Superior 6.0 cm</td>
<td>0.88</td>
<td>0.25</td>
<td>0.63</td>
</tr>
<tr>
<td>Inferior 2.0 cm</td>
<td>0.47</td>
<td>0.10</td>
<td>0.23</td>
</tr>
<tr>
<td>Inferior 6.0 cm</td>
<td>0.91</td>
<td>0.19</td>
<td>0.16</td>
</tr>
<tr>
<td>Sagittal</td>
<td>0.38</td>
<td>0.12</td>
<td>0.32</td>
</tr>
</tbody>
</table>
3.2.3 Observations Pertaining to All Measurement Planes

Large bumps at depths greater than 9cm in the depth dose profiles represent a registration mark and not dosimetric data. While the standard error values in measurement isodose figures are all less than 1.5%, it is important to note that these represent an average of 400 individual dose points. An individual dose point can have a standard error larger than these given values.

The standard error in the high dose region for all measurement planes was much larger than the standard error in the low dose region. Since both regions are normalized to the same value, the percent standard error of the mean would be expected to correlate to the relative amount of dose within the region. Thus the absolute percent standard error is expected to be larger for the high dose region and lower for the low dose region.

3.3 Aim 3: Evaluate the Accuracy of the PBA and PBRA by Comparing Calculated Dose Distributions to the Set of Parotid and Chest Wall Measurements

Pinnacle TPS dose calculation data sets for the ECT, IMXT, and mixed beam treatment plans and in-house PBRA dose calculation data sets for ECT and mixed beam treatment plans were compared with the measured data in Section 3.1. The IMXT component for both the PBA and PBRA mixed beam plans was identical. A criterion of ±4% dose difference was applied to the HDR and LDR of the ECT and mixed beam plans and ±2mm distance to agreement was applied to the HGR. Both criteria were applied to all dose points in the IMXT plan, with a dose point only required to pass a single criterion. This section only provides data for the central axis and sagittal planes, with the comparisons for all five transverse planes provided in Appendix B. For all isodose plots, dashed lines represent measured data and solid lines represent calculated data. Dose points located in the HDR and LDR that are colored red or blue represent points that fail the ±4% dose difference criterion, while dose points in the HGR colored green fail the ±2mm DTA criterion. To differentiate between the exact and approximate bolus shape used to create a specific isodose distribution, a solid blue bolus contour was included in the CAX and sagittal isodose plots, while a blue bolus outline was included in all off-axis transverse slices.
3.3.1 Parotid Case

3.3.1.1 Electron Conformal Therapy

Central axis and sagittal isodose curves comparing the Pinnacle TPS PBA calculated dose set to the averaged measured data set are shown in Figures 3.37 and 3.43, respectively. The red line in the CAX and sagittal isodose plots represent the depth profiles in Figures 3.39 and 3.44, while the green line represents the lateral profiles in Figures 3.39 and 3.45. Both the measured and calculated dose was normalized to 44.5cGy. The calculated PBRA dose distributions for the same CAX and sagittal planes immediately follow the corresponding PBA calculated plots.

Figure 3.37: Parotid ECT PBA CAX isodose plot comparing measured and calculated dose distributions.

Figure 3.38: Parotid ECT PBRA CAX isodose plot comparing measured and calculated dose.
Figure 3.39: Parotid ECT PBA CAX depth dose

Figure 3.40: Parotid ECT PBRA CAX depth dose.
Figure 3.41: Parotid ECT PBA CAX lateral profile.

Figure 3.42: Parotid ECT PBRA CAX lateral profile.
Figure 3.43: Parotid ECT PBA sagittal isodose plot comparing measured and calculated dose distributions.

Figure 3.44: Parotid ECT PBRA sagittal isodose plot comparing measured and calculated dose distributions.
Figure 3.45: Parotid ECT PBA sagittal depth dose.

Figure 3.46: Parotid ECT PBRA sagittal depth dose.
The measured dose is consistently less than the PBA and PBRA calculated dose within the first several millimeters of the film edge, often exceeding the -4% criterion. One possible reason for this difference is the inhomogeneous Cherenkov radiation produced by the phantom that resulted in a lower dose region of the film. While the applied Cherenkov mask corrected for much of the dose difference, it is reasonable that the mask did not account for the entire Cherenkov effect. A number of dose points in the deep penumbra between 20% and 10% isodose fail the 2mm criteria. This is an expected result of
the PBA and is a consequence of the algorithm’s inability to account for the loss of electron fluence and scattering through large angles at increasing depth. The instances where the PBRA fails the DTA criteria in the lateral penumbra are likely due to the algorithm’s resolution to calculate the edges of the electron insert. In the transverse direction, the PBRA calculates the position of the block to 1mm resolution at 100cm SSD, producing a larger variability for the extended SSD of the bolus ECT setup. A difference in the position of the beam edge of ~1mm at the surface may produce differences of more than 2mm in the lateral penumbra at depths near the range of the electrons. A number of dose points also fail the 2mm criteria between 80% and 60%, with the calculation underestimating the dose. Both the PBA and PBRA failed in the same regions, indicating the error may be due to alignment of the phantom during the measurement. The PBA also overestimated the dose along the central axis in the 35% - 10% region of the inferior 2.5cm and inferior 5.0cm measurement planes, likely not accounting for the loss of electron fluence.

In the HGR of both the CAX and sagittal depth profiles the measured and calculated dose differ by no more than 0.5mm. The measured dose differs from the calculated dose by less than 1% in the LDR of the depth profile for both the sagittal and CAX planes. The lateral profiles have similar accuracy in the HGR, however the PBA and PBRA calculated dose underestimates the measured dose by 2%-3% in the LDR of the sagittal plane. Within the HDR of the lateral profiles of both the sagittal and CAX planes the film measurements differ from the calculation by up to 3.5%.

Dose points within the HDR and LDR were binned into 1% increments and represented as a histogram in Figure 3.49 (PBA) and Figure 3.50 (PBRA), with all dose points within ±4% passing the applied criteria. Dose points in the HGR were binned into 1mm increments and represented as a histogram in Figure 3.51a (PBA) and Figure 3.51b (PBRA) with all dose points less than 2mm passing the applied criteria. The percent of dose points passing each criterion and the percent of all dose points passing either criterion is given in Table 3.7.
Table 3.7: Parotid ECT calculated statistics. Percent of dose points passing the ±4% dose criteria in the HDR and LDR, the ±2mm DTA criteria in the HGR, the applied criteria for the HDR, HGR, and LDR, and percent dose points either the ±4% dose or ±2mm criteria. The given percentages are the sum of all six measurement planes.

<table>
<thead>
<tr>
<th></th>
<th>HDR &amp; LDR ±4%</th>
<th>HGR ±2mm DTA</th>
<th>HDR, LDR, and HGR combined</th>
<th>±4% or ±2mm</th>
</tr>
</thead>
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<tr>
<td>PBA % Pass</td>
<td>98.8</td>
<td>93.4</td>
<td>96.6</td>
<td>98.9</td>
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<tr>
<td>PBRA % Pass</td>
<td>99.0</td>
<td>97.7</td>
<td>98.5</td>
<td>98.9</td>
</tr>
</tbody>
</table>

Figure 3.49: Parotid ECT PBA histograms for % dose difference in the HDR and LDR of all 6 planes.

Figure 3.50: Parotid ECT PBRA % dose difference histogram in the HDR and LDR of all 6 planes.
The number of dose points passing the ±4% dose criteria in the LDR and HDR combined with the dose points passing the ±2mm DTA criteria in HGR results in 96.6% of all dose points passing the regionally applied criteria for the PBA and 98.5% for the PBRA. The percent of dose points passing either criterion applied to every dose point results in a 98.9% pass rate for both the PBA and PBRA.

3.3.1.2 Intensity Modulate X-Ray Therapy

Central axis and sagittal isodose curves comparing the Pinnacle TPS calculated dose set to the averaged measured data set are shown in Figure 3.52 and 3.55, respectively. The red line in the CAX and sagittal isodose plots represent the depth profiles in Figures 3.53 and 3.56, while the green line represents the cross profiles in Figures 3.54 and 3.57. Green shaded regions in the isodose plots represent dose points failing both the ±4% and ±2mm criteria.
Figure 3.53: Parotid IMXT CAX depth dose.

Figure 3.54: Parotid IMXT CAX lateral profile.
Figure 3.55: Parotid IMXT sagittal isodose plot comparing measured and calculated dose distributions. Green regions represent dose points failing both applied criteria of ±4% and ±2mm.

Figure 3.56: Parotid IMXT sagittal depth dose.
The edges of the film measurement exhibit dose points failing both the ±4% dose and ±2mm DTA criteria. These dose points are within the first 2mm of the film edge and have two primary causes. First, the Pinnacle TPS sets the dose in air to zero and begins calculating dose at the phantom edge. When using a dose grid with 2.5mm resolution, Pinnacle interpolates from zero up to the dose at the first dose point. This can result in the calculated dose underestimating the measured dose within the first 2mm of the phantom. Second, there are regions of the film that have significantly high measured dose in the first 2mm-3mm from the film edge, as seen in the upper right side of the sagittal plane in Figure 3.55. The measured data also exhibits several isolated hotspots within the PTV that exceed the ±4% criteria. Many of these were within the regions that had percent standard error values exceeding 2.5%, which likely contributed to some of the calculated data failing the applied criteria.

In addition, the measured dose exceeds the calculated dose in the low dose region by up to 4% as noted in Figure 3.56 and 3.57. In low dose regions, the ratio of low energy scattered photons to high energy photons from the primary beam increases. As the ratio of low to high energy photons increases, radiographic over responds by 3-4% as more of the dose is delivered via the photoelectric effect (Pai et al. 2007). The percent of dose points passing either criteria for each plane is given in Table 3.8.
Table 3.8: Parotid IMXT calculated statistics. Percent of dose points passing either the ±4% dose or ±2mm DTA criteria for each plane.

<table>
<thead>
<tr>
<th>Measurement Plane</th>
<th>% Dose points Passing Either ±4% Dose or ±2mm DTA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Central Axis</td>
<td>96.6</td>
</tr>
<tr>
<td>Superior 2.5cm</td>
<td>91.4</td>
</tr>
<tr>
<td>Superior 5.0cm</td>
<td>91.2</td>
</tr>
<tr>
<td>Inferior 2.5cm</td>
<td>93.0</td>
</tr>
<tr>
<td>Inferior 5.0cm</td>
<td>93.6</td>
</tr>
<tr>
<td>Sagittal</td>
<td>95.9</td>
</tr>
<tr>
<td>Total</td>
<td>93.7</td>
</tr>
</tbody>
</table>

All transverse planes except for the CAX exhibit lower criteria pass rates while the sagittal and CAX planes have pass rates of 95.9% and 96.6%, respectively.

3.3.1.3 Mixed Beam Therapy

Central axis and sagittal isodose plots comparing the Pinnacle TPS PBA calculated dose set to the averaged measured data set are shown in Figure 3.58 and 3.64, respectively. The red line in the CAX and sagittal isodose plots represent the depth profiles in Figures 3.60 and 3.66, while the green line represents the cross profiles in Figures 3.62 and 3.68. Central axis and sagittal isodose plots comparing the PBRA calculated data to the measured data sets follow the corresponding PBA plots.

Figure 3.58: Parotid mixed beam PBA CAX isodose plot comparing measured and calculated dose distributions.
Figure 3.59: Parotid mixed beam PBRA CAX isodose plot comparing measured and calculated dose distributions.

Figure 3.60: Parotid mixed beam PBA CAX depth dose.

Figure 3.61: Parotid mixed beam PBRA CAX depth dose.
Figure 3.62: Parotid mixed beam PBA CAX lateral profile.

Figure 3.63: Parotid mixed beam PBRA CAX lateral profile.
Figure 3.64: Parotid mixed beam PBA sagittal isodose plot comparing measured and calculated dose distributions.

Figure 3.65: Parotid mixed beam PBRA sagittal isodose plot comparing measured and calculated dose distributions.
Figure 3.66: Parotid mixed beam PBA sagittal depth dose.

Figure 3.67: Parotid mixed beam PBRA sagittal depth dose.
Figure 3.68: Parotid mixed beam PBA sagittal lateral profile.

Figure 3.69: Parotid mixed beam PBRA sagittal lateral profile.

The measured dose consistently deviates from the PBA calculated dose within the first several millimeters of the film edge, often exceeding the 4% criterion, as discussed in section 3.3.1.1. Regions of dose points on the left and right sides of the transverse and sagittal planes between 80% and 50% also fail the 2mm criteria, with the both the PBA and PBRA calculation overestimating the dose. This region has a much shallower dose gradient then the rest of the HGR, resulting in small dose differences.
creating large differences in distance to agreement. In addition, these regions exhibit large standard errors, ranging from 0.6mm to 1.6mm, in the HGR of all transverse measurement planes.

The PBRA mixed beam plan had 85.3% of dose points failing the 2mm criterion for the HGR. This is predominantly a result of the IMXT component of the mixed beam plan being created by optimizing over the parotid PBA ECT dose distribution. The PBA ECT dose distribution consistently underestimated the measured dose in the HGR, while the PBRA provided an accurate dose distribution in the HGR. The resulting IMXT plan delivered the necessary dose for the PBA mixed beam to accurately compare to the measured data, while delivering too much dose for the PBRA mixed beam plan, causing the PBRA to over-estimate the measured data in the HGR.

In the HGR of both the CAX and sagittal depth dose plots, the measured and PBA calculated dose differ by no more than 1.2mm. The measured dose exceeds the PBA and PBRA calculated dose by as much as 4.5% in the LDR of the depth and lateral profile for the sagittal planes, as discussed in section 3.3.1.1. The left side of the lateral profile (Figure 3.54) for the CAX plane differs by as much as 1.3mm. Within the HDR of the sagittal and CAX lateral profiles the film measurements differ from the PBA and PBRA calculated data by up to 2.8%.

Dose points within the HDR and LDR were binned into 1% increments and represented as a histogram in Figure 3.71 (PBA) and Figure 3.72 (PBRA) with all dose points within ±4% passing the applied criteria. Dose points in the HGR were binned into 1mm increments and represented as a histogram in Figure 3.70a (PBA) and Figure 3.70b (PBRA) with all dose points less than 2mm passing the applied criteria. The percent of dose points passing each criteria is given in Table 3.9.

| Table 3.9: Parotid mixed beam calculated statistics. The given percentages are the sum of all six planes. |
|-----------------|-----------------|-----------------|-----------------|
|                | HDR & LDR ±4%   | HGR ±2mm DTA    | HDR, LDR, and HGR combined ±4% or ±2mm |
| PBA % Pass     | 97.9%           | 89.2%           | 95.0%           | 97.5           |
| PBRA % Pass    | 97.9%           | 85.3%           | 93.8%           | 97.2           |

The number of dose points passing the ±4% dose criteria in the LDR and HDR combined with the dose points passing the ±2mm DTA criteria in HGR results in 95.0% of all dose points for the PBA and
93.8% for the PBRA passing the regionally applied criteria. The percent of dose points passing either criterion applied to every dose point results in a 97.5% pass rate for the PBA and 97.2% for the PBRA.

Figure 3.70: Parotid Mixed Beam DTA Histogram for PBA(a) and PBRA(b) in the HGR of all 6 measurement planes.

Figure 3.71: Parotid mixed beam PBA histograms for % dose difference in the HDR and LDR of all 6 measurement planes.

Figure 3.72: Parotid Mixed Beam PBRA histogram for % dose difference in the HDR and LDR of all 6 measurement planes.
3.3.2 Chest Wall Case

3.3.2.1 Electron Conformal Therapy

Central axis and sagittal isodose curves comparing the Pinnacle TPS PBA calculated dose set to the averaged measured data set are shown in Figures 3.73 and 3.79, respectively. The red line in the CAX and sagittal isodose plots represent the depth profiles in Figures 3.75 and 3.81, while the green line represents the lateral profiles in Figures 3.77 and 3.83. Both the measured and calculated dose distributions are normalized to 43.5cGy. The calculated PBRA dose distributions for the same CAX and sagittal planes immediately follow the corresponding PBA calculated plots.

![Image](image1.png)

Figure 3.73: Chest Wall ECT PBA CAX isodose comparing measured and calculated dose distributions.

![Image](image2.png)

Figure 3.74: Chest Wall ECT PBRA CAX isodose comparing measured and calculated dose distributions.
The PBA calculated dose distribution overestimates the measured dose distribution for the CAX plane in the deep lateral parts of the HDR, while underestimated the measured dose in the HGR, as seen in Figure 3.73. The PBRA conforms better in the HDR while failing in the lateral HGR, seen in Figure 3.74.

Figure 3.75: Chest Wall ECT PBA CAX depth dose.

Figure 3.76: Chest Wall ECT PBRA CAX depth dose.
The PBA has a maximum percent dose difference in the HDR of the CAX depth dose curve of ~3% and a maximum DTA in the HGR of 1.6mm. The PBRA has better average conformity in all parts of the depth dose curve, with a maximum percent dose difference of ~2.5% and a maximum DTA of ~1.4mm.

Both the PBA and PBRA are over 6% higher than the measured data along the left edge of the lateral CAX profile, indicating an error in the film measurement at that location. Both calculations differ by up to ~2.5% across the lateral profile, with no significant difference between the two algorithms.
The PBA underestimates the measured data in some regions of the sagittal HGR of Figure 3.79 by more than 2mm and by more than 4% in the right lateral HDR and LDR. By comparison, the PBRA conforms better to the measured data, slightly overestimating the measured data in the HGR of Figure 3.80 by ~1mm.
Both the PBA and PBRA have a maximum percent dose difference in the HDR of the sagittal depth dose curve of ~2.9%. The PBRA has better average conformity in the HGR of the depth dose curve, with a maximum DTA of ~1.1mm compared to the PBA maximum DTA of 1.4mm.
The PBRA differs by up to ~1.8% across the HDR of the sagittal lateral profile in Figure 3.84, while the PBA underestimates the measured data by up to ~3.1%, as seen in Figure 3.83. The PBRA has better agreement in the HGR while both algorithms are lower than the measured dose in the LDR along the left lateral edge. The film may be over-responding to the low energy photons in this region.
The chest wall PBA ECT underestimates the measured dose along the central axis for all measurement planes. In each of the transverse planes, it consistently overestimates the dose in the lateral portions of the HDR. This is likely caused by the PBA’s inability to account for large angle scatter and loss of electron fluence. The chest wall PBRA consistently fails the 2mm criteria in the lateral penumbra for all transverse measurement planes. As discussed in section 3.3.1.1, this is due to the 1mm resolution that defines the insert edge used in the PBRA dose calculation. It is possible that a higher percentage of dose points would pass the criteria if a finer insert resolution were used.

Dose points within the HDR and LDR were binned into 1% increments and represented as a histogram in Figure 3.86 (PBA) and Figure 3.87 (PBRA) with all dose points within ±4% passing the applied criteria. Dose points in the HGR were binned into 1mm increments and represented as a histogram in Figure 3.85a (PBA) and Figure 3.85b (PBRA) with all dose points less than 2mm passing the applied criteria. The percent of dose points passing each criterion is given in Table 3.10.

![Histograms](image_url)

Figure 3.85: Chest wall ECT (a)PBA and (b) PBRA distance to agreement histograms in the HGR for all 6 measurement planes.

<table>
<thead>
<tr>
<th></th>
<th>HDR &amp; LDR ±4%</th>
<th>HGR ±2mm DTA</th>
<th>HDR, LDR, and HGR combined</th>
<th>±4% or ±2mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>PBA % Pass</td>
<td>96.2</td>
<td>94.5</td>
<td>95.6</td>
<td>97.3</td>
</tr>
<tr>
<td>PBRA % Pass</td>
<td>98.9</td>
<td>95.7</td>
<td>97.7</td>
<td>98.7</td>
</tr>
</tbody>
</table>
Compared to the PBA, the PBRA had 2.1% more dose points passing the criteria in the HDR, HGR, and LDR for all six measurement planes while 1.4% more dose points passed either criteria applied to the all six measurement planes. The PBRA notably overestimated the dose in the lateral penumbra region for each plane.

**3.3.2.2 Intensity Modulate X-Ray Therapy**

Central axis and sagittal isodose curves comparing the Pinnacle TPS IMXT calculated dose set to the averaged measured data set are shown in Figure 3.88 and 3.91, respectively. The red line in the CAX
and sagittal isodose plots represent the depth profiles in Figures 3.89 and 3.92, while the green line represents the cross profiles in Figures 3.90 and 3.93. Green shaded regions in the isodose plots represent dose points failing both the ±4% and ±2mm criteria.

Figure 3.88: Chest Wall IMXT CAX isodose comparing measured and calculated dose distributions. Green regions represent dose points failing both applied criteria of ±4% and ±2mm.

Figure 3.89: Chest Wall IMXT CAX depth profile.
The Pinnacle v8.0 TPS calculated IMXT dose distribution differed by up to 5.4% along the depth dose profile (Figure 3.89) of the CAX plane and up to 4.8% along the lateral profile (Figure 3.90). Dose points appeared to fail both applied criteria along specific IMXT beam paths, indicating possible over exposure due to rounding the delivered monitor units to the nearest whole number.
The Pinnacle v8.0 TPS calculated IMXT dose distribution differed by up to 4.3% along the depth dose profile (Figure 3.92) of the sagittal plane and up to 6.2% along the lateral profile (Figure 3.93). The measured data was higher in the lateral low dose regions, indicating a possible over-response of the film to scattered low energy photons.

Figure 3.92: Chest Wall IMXT sagittal depth profile.

Figure 3.93: Chest Wall IMXT sagittal lateral profile
### Table 3.11: Chest Wall IMXT calculated statistics.

<table>
<thead>
<tr>
<th>Measurement Plane</th>
<th>% Dose points Passing Either ±4% Dose or ±2mm DTA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Central Axis</td>
<td>92.3</td>
</tr>
<tr>
<td>Superior 2.0cm</td>
<td>96.6</td>
</tr>
<tr>
<td>Superior 6.0cm</td>
<td>95.8</td>
</tr>
<tr>
<td>Inferior 2.0cm</td>
<td>97.3</td>
</tr>
<tr>
<td>Inferior 6.0cm</td>
<td>96.6</td>
</tr>
<tr>
<td>Sagittal</td>
<td>96.0</td>
</tr>
<tr>
<td>Total</td>
<td>95.8</td>
</tr>
</tbody>
</table>

The percent of dose points in each measurement plane passing both criteria are given in Table 3.11. All four off-axis transverse planes and the sagittal plane had similar pass rates, while the CAX plane had ~4% fewer dose points passing both criteria. The total number of dose points passing the applied criteria for all 6 measurement planes was 95.8%, which is ~2.1% higher than the parotid IMXT treatment plan. Some of the measurement regions where dose points failed the criteria had percent standard error exceeding 2%. For the chest wall IMXT plan, the first 2mm of dose points from the phantom surface were eliminated to avoid any interpolation issues produced by the Pinnacle calculation algorithm.

#### 3.3.2.3 Mixed Beam Therapy

The PBA underestimates the measured data in some regions of the CAX HGR of Figure 3.94 by more than 2mm and by more than 4% in the left lateral LDR. By comparison, the PBRA conforms better to the measured data, overestimating the measured data in the right lateral HGR and LDR of Figure 3.95. This is possibly a blocking issue as discussed in section 3.3.2.1 and seen in Figure 3.74.

Both the PBA and PBRA accurately calculated the dose in the HDR of the CAX depth dose profile, differing by no more than 2.7%. The PBRA closely conformed to the measured data in the HGR (Figure 3.97), differing by a maximum of 1.2mm, while the PBA differed from the measured dose by a maximum of 2.9mm (Figure 3.96). The PBRA underestimated the dose in the LDR by ~1.6%.
Figure 3.94: Chest Wall mixed beam PBA CAX isodose comparing measured and calculated dose distributions.

Figure 3.95: Chest Wall mixed beam PBRA CAX Isodose comparing measured and calculated dose distributions.

Figure 3.96: Chest Wall mixed beam PBA CAX depth dose.
Figure 3.97: Chest Wall mixed beam PBRA CAX depth profile.

Figure 3.98: Chest Wall mixed beam PBA CAX lateral profile

Figure 3.99: Chest Wall mixed beam PBRA CAX Lateral Profile.
Figure 3.100: Chest Wall mixed beam PBA sagittal isodose comparing measured and calculated dose distributions.

The PBA fails the 4% criteria in a few regions of the sagittal HDR (Figure 3.100); however, the HGR and the LDR accurately conform to the measured data. By comparison, the PBRA better predicts the measured dose in the HDR, while overestimating the measured data in the right and left lateral HGR and LDR of Figure 3.101. This is possibly a blocking issue as discussed in section 3.3.2.1 and seen in Figure 3.80.
Both the PBA and PBRA underestimate the measured dose in the first centimeter of the sagittal depth dose curve. The PBA also differs by a maximum of 1.9mm in the HGR, as seen in Figure 3.102. The PBRA closely conforms to the measured data in the HDR and HGR; however, the calculated PBRA dose was 3.2% lower than the measured dose in the LDR, seen in Figure 3.103.
Figure 3.104: Chest Wall mixed beam PBA sagittal lateral profile

Figure 3.105: Chest Wall mixed beam PBRA sagittal lateral profile

The PBRA calculated dose differs from the measured dose in the HGR of the sagittal lateral profile by up to 1.8mm, seen in Figure 3.105. The PBA accurately calculated the lateral dose profile, being within 0.8mm in the HGR and 1.5% in the HDR.
Figure 3.106: Chest Wall mixed beam PBA % dose difference histogram in the HDR and LDR of all 6 measurement planes.

Figure 3.107: Chest wall mixed beam PBRA % dose difference histogram in the HDR and LDR of all 6 measurement planes.

Figure 3.108: Chest wall mixed beam PBA (a) and PBRA (b) distance to agreement histogram in the HGR of all 6 measurement planes.
Table 3.12: Chest Wall mixed beam calculated statistics.

<table>
<thead>
<tr>
<th></th>
<th>HDR &amp; LDR ±4%</th>
<th>HGR ±2mm DTA</th>
<th>HDR, LDR, and HGR combined</th>
<th>±4% or ±2mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>PBA % Pass</td>
<td>94.7%</td>
<td>95.1%</td>
<td>94.8%</td>
<td>95.7%</td>
</tr>
<tr>
<td>PBRA % Pass</td>
<td>97.4%</td>
<td>96.1%</td>
<td>97.0%</td>
<td>98.0%</td>
</tr>
</tbody>
</table>

Dose points within the HDR and LDR were binned into 1% increments and represented as a histogram in Figure 3.106 (PBA) and Figure 3.107 (PBRA) with all dose points within ±4% passing the applied criteria. Dose points in the HGR were binned into 1mm increments and represented as a histogram in Figure 3.108a (PBA) and Figure 3.108b (PBRA) with all dose points less than 2mm passing the applied criteria. The percent of dose points passing each criteria is given in Table 3.12.

Compared to the PBA, the PBRA had 2.2% more dose points passing the criteria in the HDR, HGR, and LDR for all six measurement planes while 2.3% more dose points passed either criteria applied to the all six measurement planes. The PBRA notably overestimated the dose in the lateral penumbra region for each plane. This is due to the 1mm resolution that defines the insert edge used in the PBRA dose calculation. It is possible that a higher percentage of dose points would pass the criteria if a finer insert resolution were used.

The chest wall mixed beam PBA plan underestimated the dose in the lateral regions of the HDR. This is directly caused to the chest wall ECT PBA dose distribution. It is notable that the PBRA has better agreement for the chest wall mixed plan, whereas the PBA has better agreement for the parotid mixed beam plan. As discussed in section 3.3.1.3, the IMXT plan was created by optimizing the dose distribution over the parotid ECT PBA dose, resulting in the PBRA mixed beam plan overestimating the dose in the HGR. The fraction of ECT:IMXT used in the chest wall plan (2.08:1) was greater than the fraction used for the parotid plan (1.73:1). The smaller component of IMXT used in the chest wall mixed beam PBRA resulted in a more accurate agreement with the measured data.
3.3.3 Summary

Table 3.13 summarizes the results for the chest ECT, IMXT, and mixed beam plans. It was expected that for the ECT plans, the PBRA would produce a slightly more accurate dose distribution in a homogenous phantom material than the PBA, a hypothesis confirmed by the data. For the parotid mixed beam plan, the PBA was more accurate than the PBRA; however, this was likely a result of the IMXT component being optimized for the PBA dose distribution. The PBRA dose distribution was more accurate for the chest wall mixed beam plan, likely due to the smaller fraction of IMXT used.

Table 3.13: Summary of calculated vs. measured results.

<table>
<thead>
<tr>
<th>Plan and Delivery Modality</th>
<th>Percent of Dose points Passing Regional (HDR, LDR, HGR) Criteria</th>
<th>Percent of Dose points Passing Either ±4% OR ±2mm</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Parotid:</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>ECT PBA</td>
<td>96.6</td>
<td>98.9</td>
</tr>
<tr>
<td>ECT PBRA</td>
<td>98.5</td>
<td>98.9</td>
</tr>
<tr>
<td>IMXT</td>
<td>N/A</td>
<td>93.7</td>
</tr>
<tr>
<td>Mixed Beam PBA</td>
<td>95.0</td>
<td>97.5</td>
</tr>
<tr>
<td>Mixed Beam PBRA</td>
<td>93.8</td>
<td>97.2</td>
</tr>
<tr>
<td><strong>Chest Wall:</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>ECT PBA</td>
<td>95.6</td>
<td>97.3</td>
</tr>
<tr>
<td>ECT PBRA</td>
<td>97.7</td>
<td>98.7</td>
</tr>
<tr>
<td>IMXT</td>
<td>N/A</td>
<td>95.8</td>
</tr>
<tr>
<td>Mixed Beam PBA</td>
<td>94.8</td>
<td>95.7</td>
</tr>
<tr>
<td>Mixed Beam PBRA</td>
<td>97.0</td>
<td>98.0</td>
</tr>
</tbody>
</table>
Chapter 4: Conclusions

4.1 Summary of Results

Hypothesis: At least 98% of dose points for the Bolus ECT and Bolus ECT + IMXT measured dose distribution will be within ±4% of the calculated dose in the low gradient regions and ±2mm distance to agreement in the high gradient regions of both the PBA and PBRA calculated dose distributions for a head and neck and post-mastectomy patient-like PTV modeled in a cylindrical HIWO polystyrene phantom.

Three specific aims were completed to test the hypotheses. In Aim 1, a parotid and a post-mastectomy chest wall bolus ECT and IMXT mixed beam treatment plan were constructed in a HIWO polystyrene film phantom.

In Aim 2, two to four radiographic film measurements were taken in each of the six planes (five transverse and one sagittal) for the bolus ECT, IMXT, and bolus ECT + IMXT components of both the parotid and chest wall treatment plans. Results showed accurate measured data sets, with the percent standard error of the measurements typically less than 1.0% in the low gradient dose regions and less than 0.4mm in the high gradient regions. The data showed consistency along the intersection of the sagittal plane with the transverse planes.

In Aim 3, the calculated IMXT, bolus ECT, and mixed beam bolus ECT + IMXT dose distributions were compared with the measured data sets. The bolus ECT dose distributions were calculated using the Pinnacle PBA and an in-house PBRA. Accuracy of the ECT and mixed beam dose distributions were evaluated with a ±4% criterion applied to the HDR and LDR and a ±2mm criterion applied to all dose points in the HGR, while the accuracy of the IMXT dose distribution was evaluated with all dose points passing either criterion. Also, results passing either criteria showed:

- The parotid ECT (PBA and PBRA) plans had more than 98% of dose points pass either the ±4% dose difference or ±2mm DTA criteria.
• The parotid mixed beam (PBA and PBRA) plans had more than 97% of dose points pass either the ±4% dose difference or ±2mm DTA criteria.

• The chest wall ECT and mixed beam PBRA plans had more than 98% of dose points pass either the ±4% dose difference or ±2mm DTA criteria.

• The chest wall ECT PBA plan had more than 97% of dose points pass either the ±4% dose difference or ±2mm DTA criteria.

• The chest wall mixed beam PBA plan had 95.7% of dose points pass either the ±4% dose difference or ±2mm DTA criteria.

4.2 Conclusions

1. Accuracy of the PBA and accuracy of the PBRA are comparable for the polystyrene phantom for the parotid and chest wall.

2. From a clinical perspective, both the PBA and PBRA are sufficiently accurate for calculating bolus ECT and mixed beam dose distributions in a homogenous phantom, i.e. either the PBA or PBRA can be used to calculated dose in homogenous sites.

3. Mixed beam therapy can be delivered in a polystyrene phantom with an agreement to the planned (calculated) dose distribution that is clinically acceptable.

4.3 Future Work

1. Further analysis of data in this thesis should include:
   • Modify bolus ECT and IMXT measurements to account for output discrepancies.
   • Internal consistency check: Combine measured bolus ECT dose distribution with IMXT dose distribution to evaluate whether mixed beam measurements are achieved.
   • Separate dose histograms to show separate distribution of dose points in the HDR and LDR.
   • Investigate differences in the tail region of the mixed beam PBA and PBRA calculated dose distributions.
These corrections and analysis will slightly modify the results; however, the effect of the changes are expected to be minimal and have little, if any, impact on the conclusions.

2. Suggestions for future supplementary work.
   - Repeat Specific Aim 2 and Specific Aim 3 for a heterogeneous phantom with bolus.
   - Evaluate the accuracy of the PBA dose calculation in a heterogeneous patient by comparing the Pinnacle TPS calculated dose distributions to PBRA calculated dose distributions.
References


Appendix A: Parotid and Chest Wall ECT, IMXT, and Mixed Beam Transverse Measurement Planes

Figure A.1: Parotid ECT Superior 5cm Isodose.

Figure A.2: Parotid ECT superior 5cm depth dose of film measurements located at -5mm in Figure A.1.

Figure A.3: Parotid ECT comparing depth dose of the intersection between the superior 5cm and sagittal measurement planes.
Figure A.4: Parotid ECT superior 2.5cm isodose.

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Figure A.12: Parotid ECT comparing depth dose of the intersection between the inferior 2.5cm and sagittal measurement planes.
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Figure A.15: Parotid ECT comparing depth dose of the intersection between the inferior 5.0cm and sagittal measurement planes.
Figure A.16: Parotid IMXT superior 5cm isodose plot.

Figure A.17: Parotid IMXT superior 5.0cm depth dose of film measurements located at -5mm in Figure A.16.

Figure A.18: Parotid IMXT comparing depth dose of the intersection between the superior 5.0cm and sagittal measurement planes.
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Figure A.21: Parotid IMXT comparing depth dose of the intersection between the superior 2.5cm and sagittal measurement planes.
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Figure 23: Parotid IMXT CAX depth dose of film measurements located at -5mm in Figure A.22.

Figure 24: Parotid IMXT comparing depth dose along the intersection between the CAX and sagittal measurement planes.
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Figure A.63: Chest wall IMXT comparing depth dose along the intersection between the superior 6.0cm and sagittal measurement planes.
Figure A.64: Chest wall IMXT superior 2.0cm isodose plot.

Figure A.65: Chest wall IMXT superior 2.0cm depth dose of film measurements at -5mm in Figure A.64.

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Figure A.72: Chest wall IMXT comparing depth dose along the intersection between the inferior 2.0cm and sagittal measurement planes.
Figure A.73: Chest wall IMXT inferior 6.0cm isodose plot.

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Figure A.75: Chest wall IMXT comparing depth dose along the intersection between the inferior 6.0cm and sagittal measurement planes.
Figure A.76: Chest wall MB superior 6.0cm isodose plot.

Figure A.77: Chest wall MB superior 6.0cm depth dose of film measurements at -5mm in Figure A.76.

Figure A.78: Chest wall MB comparing depth dose along the intersection between the superior 6.0cm and sagittal measurement planes.
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Figure A.85: Chest wall MB inferior 2.0cm isodose plot.

Figure A.86: Chest wall MB inferior 2.0cm depth dose of film measurements at -5mm in Figure A.85.

Figure A.87: Chest wall MB comparing depth dose along the intersection between the inferior 2.0cm and sagittal measurement planes.
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Figure B.2: Parotid ECT superior 5.0 cm depth dose comparing PBA calculated to measured data.
Figure B.3: Parotid ECT superior 5.0cm lateral profile comparing PBA calculated to measured data.

Figure B.4: Parotid ECT superior 2.5cm isodose plot comparing the PBA calculated to the measured data.

Figure B.5: Parotid ECT superior 2.5cm depth dose comparing PBA calculated to measured data.

Figure B.6: Parotid ECT superior 2.5cm lateral profile comparing PBA calculated to measured data.
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Figure B.9: Parotid ECT CAX lateral profile comparing PBA calculated to measured data.
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Figure B.11: Parotid ECT inferior 2.5cm depth dose comparing PBA calculated to measured data.

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Figure B.15: Parotid ECT inferior 5.0cm lateral profile comparing PBA calculated to measured data.
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Figure B.18: Parotid ECT superior 5.0cm lateral profile comparing PBRA calculated to measured data.
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Figure B.129: Chest wall mixed beam CAX lateral profile comparing PBA calculated to measured data.
Figure B.130: Chest wall mixed beam inferior 2.0 cm isodose comparing the calculated PBA to measured data.

Figure B.131: Chest wall mixed beam inferior 2.0 cm depth dose comparing calculated PBA to measured data.

Figure B.132: Chest wall mixed beam inferior 2.0 cm lateral profile comparing PBA calculated to measured data.
Figure B.133: Chest wall mixed beam inferior 6.0cm isodose comparing the calculated PBA to measured data.

Figure B.134: Chest wall mixed beam inferior 6.0 cm depth dose comparing calculated PBA to measured data.

Figure B.135: Chest wall mixed beam inferior 6.0cm lateral profile comparing PBA calculated to measured data.
Figure B.136: Chest wall mixed beam superior 6.0cm isodose comparing the calculated PBRA to measured data.

Figure B.137: Chest wall mixed beam superior 6.0 cm depth dose comparing calculated PBRA to measured data.

Figure B.138: Chest wall mixed beam superior 6.0cm lateral profile comparing PBRA calculated to measured data.
Figure B.139: Chest wall mixed beam superior 2.0cm isodose comparing the calculated PBRA to measured data.

Figure B.140: Chest wall mixed beam superior 2.0 cm depth dose comparing calculated PBRA to measured data.

Figure B.141: Chest wall mixed beam superior 2.0cm lateral profile comparing PBRA calculated to measured data.
Figure B.142: Chest wall mixed beam CAX isodose comparing the calculated PBRA to measured data.

Figure B.143: Chest wall mixed beam CAX depth dose comparing calculated PBRA to measured data.

Figure B.144: Chest wall mixed beam CAX lateral profile comparing PBRA calculated to measured data.
Figure B.145: Chest wall mixed beam inferior 2.0cm isodose comparing the calculated PBRA to measured data.

Figure B.146: Chest wall mixed beam inferior 2.0 cm depth dose comparing calculated PBRA to measured data.

Figure B.147: Chest wall mixed beam inferior 2.0cm lateral profile comparing PBRA calculated to measured data.
Figure B.148: Chest wall mixed beam inferior 6.0 cm isodose comparing the calculated PBRA to measured data.

Figure B.149: Chest wall mixed beam inferior 6.0 cm depth dose comparing calculated PBRA to measured data.

Figure B.150: Chest wall mixed beam inferior 6.0 cm lateral profile comparing PBRA calculated to measured data.
Vita

James Kavanaugh was born in Geelong, Australia, in November of 1985. Much of his early childhood was spent in the city of Torquay, swimming at some of the great beaches of the southern continent. At the age of seven he moved with his parents and younger brother to the historic family farm in Carlton, Minnesota, where he lived until graduating from Carlton High School at the age of eighteen. Growing up on an active farm, James learned many skills and values that had been passed down by the previous three generations of Kavanaugh family members who had worked the same soil and it was these family values to which he credits much of his success.

In the fall of 2004, James began his college education at the University of Saint Thomas, located in the beautiful city of Saint Paul, Minnesota. It was here that he found both his passion for physics and his fiancée, Katherine Tschida. After graduating from St. Thomas in 2008, James moved to Baton Rouge, Louisiana, to begin study in the Medical Physics program at Louisiana State University.

James plans on completing his Master of Science degree during the summer of 2011. After graduation, James will continue his medical physics education through the Medical Physics Residency program at Washington University in St. Louis.