Improved abutment dosimetry in segmented-field electron conformal therapy

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IMPROVED ABUTMENT DOSIMETRY IN SEGMENTED FIELD ELECTRON CONFORMAL THERAPY

A THESIS

Submitted to the Graduate Faculty of the Louisiana State University and Agricultural and Mechanical Collage in partial fulfillment of the requirements for the Degree of Master of Science in The Department of Physics & Astronomy

by

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B.S. Louisiana State University, 2001
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# Table of Contents

Acknowledgments ............................................................................................................... ii

List of Tables ....................................................................................................................... v

List of Figures ..................................................................................................................... vi

Abstract ............................................................................................................................. xii

1. Introduction ................................................................................................................ ....1
   1.1 Electron Conformal Therapy ..................................................................................1
   1.2 Bolus and Segmented-Field ECT ............................................................................1
   1.3 Hypothesis and Specific Aims ................................................................................4

2. Methods and Materials...................................................................................................7
   2.1 Aim 1: Design of Methods for Matching Penumbra ..............................................7
      2.1.1 Theoretical Penumbra Width .........................................................................8
      2.1.2 Variable-SCD Applicator ...............................................................................9
   2.2 Aim 2: Verification of Electron Beam Commissioning in TPS ...................................12
      2.2.1 Calculated Dose Distributions ......................................................................12
         2.2.1.1 Hogstrom Pencil-beam Algorithm ..................................................12
         2.2.1.2 Treatment Planning Dose Distributions ..........................................13
      2.2.2 Relative Dose Distribution Measurements ..................................................14
         2.2.2.1 Varian 21EX Radiotherapy Accelerator .........................................14
         2.2.2.2 Film Phantom ..................................................................................14
         2.2.2.3 Measurement Setup .........................................................................17
      2.2.3 Film Dosimetry ............................................................................................19
         2.2.3.1 Analysis ...........................................................................................20
         2.2.3.2 Calibration .......................................................................................21
   2.3 Aim 3: Construction of Cerrobend® Blocks with Diverging Edges ...........................27
      2.3.1 Block Fabrication with the Compu•cutter® System .......................................27
         2.3.1.1 General ............................................................................................27
         2.3.1.2 Blocks for Field Abutment ..................................................................29
      2.3.2 Assessment of Accuracy of Block Fabrication Method ...................................31
         2.3.2.1 Accuracy and Precision of Off-axis Abutment Test ............................34
         2.3.2.2 Zero and 5-cm Off-axis Abutment Test ..........................................34
         2.3.2.3 Dosimetric Impact of Errors in Block Fabrication Test .................34
   2.4 Aim 4a: Evaluate Dose Homogeneity in Simple Targets ........................................35
      2.4.1 Simple Targets .............................................................................................36
         2.4.1.1 Two-step Block Target ......................................................................36
         2.4.1.2 Wedge Target ....................................................................................37
         2.4.1.3 Pentagon Target .................................................................................37
         2.4.1.4 Inverted Well Target ............................................................................37
      2.4.2 Treatment Planning ......................................................................................39
      2.4.3 Plan Evaluation and Comparisons ....................................................................40
2.5 Aim 4b: Evaluate Dose Homogeneity in Patient PTV

2.5.1 Patient Treatment Planning

2.5.1.1 Treatment Planning Objectives

2.5.1.2 Segmented-field ECT Treatment Planning

2.5.2 Patient Case

3. Results and Discussion

3.1 Aim 1: Design of Methods for Matching Penumbra

3.2 Aim 2: Verification of Electron Beam Commissioning in TPS

3.3 Aim 3: Construction of Cerrobend® Blocks with Diverging Edges

3.4 Aim 4a: Evaluate Dose Homogeneity in Simple Targets

3.4.1 Treatment Planning Results

3.4.1.1 Two-step Block Target

3.4.1.2 Wedge Target

3.4.1.3 Pentagon Target

3.4.1.4 Inverted Well Target

3.4.1.5 Summary of Treatment Planning Results

3.4.2 Variable-SCD Plan Verification

3.5 Aim 4b: Treatment Planning Results in Patient PTV

4. Conclusions

References

Appendix A: Variable-SCD Applicator Design

Appendix B: 15x15-cm² Isodose Plots

Vita
List of Tables

1a. Polystyrene stopping power and scattering power ratios from TG-25 published data .................................................................................................................................15

1b. Stopping power and scattering power ratio lookup table used to calculate dose in a polystyrene phantom ..............................................................................................................15

1c. CT number to density table used for dose computation .................................................................................................................................15

2. Data for 10-point calibration curve. Monitor units (MU) delivered to each field is shown followed by dose (D’), analog to digital 16-bit scanner value (AD), and optical density (OD) .....................................................................................................................26

3. Summary of beam parameters used for treatment planning. All values were obtained using TPS data for a 10x10-cm$^2$ field size at 100-cm SSD with modified air gaps (5, 7.5, 11.5, 17.5, and 19.5 cm) .....................................................................................................................42

4. Summary of results of 2.5-cm off-axis abutment test .................................................................................................................................64

5. Summary of treatment planning results for irradiation of the two-step block target using the standard and variable-SCD methods .........................................................................................................69

6. Summary of treatment planning results for irradiation of the wedge target using the standard and variable-SCD methods ............................................................................................................72

7. Summary of treatment planning results for irradiation of the pentagon target using the standard and variable-SCD methods ............................................................................................................76

8. Summary of treatment planning results for irradiation of the inverted well target using the standard and variable-SCD methods ............................................................................................................79

9. Summary of treatment planning results for dose to the four simulated PTVs using the standard and variable-SCD methods ............................................................................................................80

10. Summary of treatment planning results for patient plan using the standard and variable-SCD applicators .................................................................................................................................94
List of Figures

1. (a) Bolus conformal therapy using a single 20 MeV beam. (b) Segmented-field conformal therapy [Zackrisson and Karlsson 1996] .................................................................3

2. (a) The variable-SSD method illustrating the effect of non-coincident edges as the gap between the two fields increases with depth. (b) The variable-SCD method illustrating beam edge coincidence .............................................................5

3. $P_{90-10}$ and $P_{80-20}$ illustrated using a measured 12 MeV beam profile at 3-cm depth for a 10x10-cm$^2$ field size .............................................................................7

4. Photograph of modified 15x15-cm$^2$ applicator from direction of Cerrobend® insert insertion. The insert shown is positioned upstream from the standard location and created a 11.5-cm air gap ........................................................................10

5. Side view sketch of the shelf structure of the modified applicator ......................10

6. (a) Downstream view of the insert frame with a Cerrobend® insert. (b) Aluminum insert frame .........................................................................................................................11

7. (a) Photo of the Hi-impact, white opaque polystyrene film phantom shown in open position with 10"x12" Kodak XV film loaded for dose measurement. (b) Phantom shown upright ...........................................................................................................16

8. Schematic of gantry and treatment room coordinate system ................................18

9. Measured central-axis depth dose curves for a 12x12-cm$^2$ 12 MeV beam. The measured depth dose without the C-clamp shows the influence of 0.25 mm air in the phantom along the surface of the film .........................................................................................18

10. The standard 15x15-cm$^2$ applicator (with 10x10-cm$^2$ insert) attached to the Varian 2100EX. The film phantom is in measurement position with the C-clamp in place .........................................................................................................................19

11. Example of calibration film #3, used to generate four points on the calibration curve ........................................................................................................................................22

12. Scanned film used to account for scatter and leakage dose contributions to the calibration fields .................................................................................................................................................23

13. Diagram of the dose contribution from adjacent fields to $D_2'$ ............................................24

14. Dose response curve for xv film. Background OD was subtracted from the raw value to give net optical density for this plot .................................................................................................................................25

15. Example plot of AD value vs. dose for a 10-point calibration curve .................26
16. Geometry used by the Compu•cutter® software to fabricate foam blocks for making Cerrobend® inserts .................................................................28

17. Sample plot of the scaled field shapes used to guide placement of the foam field shapes ..................................................................................30

18. (a) Field shape printout taped to the bottom plate of the 15x15-cm² mold. (b) The steel mold with foam in place. (c) The resulting inserts for a single 20 MeV field and two 12 MeV fields ........................................................................30

19. Illustration of $\Delta D$ and $\Delta x$ using off-axis profiles at 1-cm depth for two 12 MeV beams with an 11.5-cm air gap and abutted on central-axis ($x_{int} = 0$) with $\Delta x = +2$ mm .....................................................................................................................32

20. Dosimetric impact ($\Delta D$) of beam gap/overlap ($\Delta x$) at 1-cm depth for abutment of two 12 MeV beams using the variable-SCD applicator with an 11.5-cm air gap ..........................................................................................................................33

21. Illustration of the field shapes used to fabricate Cerrobend® inserts for central-axis abutment with 2-mm overlap .................................................................................................................................35

22. Illustration of the two-step block target .................................................................................................................................36

23. Illustration of the wedge target .................................................................................................................................37

24. Illustration of the pentagon target .................................................................................................................................38

25. Illustration of the inverted well target .................................................................................................................................38

26. Single transverse slice illustrating the PTV segmentation method for a patient case planned using 1 cm of bolus .................................................................................................................................43

27. Example of BEV depth map. (a) Transverse view of contours. (b) MAP contours from the beams-eye view .................................................................................................................................45

28. Beams eye view of PTVPM (green wire frame) generated by expanding the PTV (solid blue) by 1 cm in all directions .................................................................................................................................45

29. Room's eye view of treatment setup with the gantry at 110°, the table at 180°, and the SSD at 100 cm .................................................................................................................................49

30. Sagittal view of the 9, 12, 16, and 20 MeV beams projected at a depth of 2 cm in the patient .................................................................................................................................49

31. $P_{80-20}$ calculated in water as a function of air gap at 1.5-cm depth .................................................................................................................................50

32. $P_{80-20}$ calculated in water as a function of air gap at 2-cm depth .................................................................................................................................51
33. $P_{80-20}$ calculated in water as a function of air gap at 2.5-cm depth. The 6 MeV penumbra width is not connected as central-axis depth dose is less than 50% at this depth..........................................................51

34. Plot of theoretical $P_{80-20}$ (calculated in water) as a function of depth from 0 cm to $\sim R_{80}$ for all beam energies using the standard 5-cm air gap.................................52

35. Plot of theoretical $P_{80-20}$ (calculated in water) as a function of depth from 0 cm to $\sim R_{80}$ for all beam energies using the variable-SCD air gaps.............................53

36. 6 MeV central xz plane measured and calculated isodose plots for a 4x4-cm$^2$ field using the standard applicator (5-cm air gap).................................................54

37. 9 MeV central xz plane measured and calculated isodose plots for a 4x4-cm$^2$ field using (a) the standard applicator (5-cm air gap) and (b) the variable-SCD applicator (7.5-cm air gap) .................................................................56

38. 12 MeV central xz plane measured and calculated isodose plots for a 4x4-cm$^2$ field using (a) the standard applicator (5-cm air gap) and (b) the variable-SCD applicator (11.5-cm air gap) .................................................................57

39. 16 MeV central xz plane measured and calculated isodose plots for a 4x4-cm$^2$ field using (a) the standard applicator (5-cm air gap) and (b) the variable-SCD applicator (17.5-cm air gap) .................................................................59

40. 20 MeV central xz plane measured and calculated isodose plots for a 4x4-cm$^2$ field using (a) the standard applicator (5-cm air gap) and (b) the variable-SCD applicator (19.5-cm air gap) .................................................................61

41. 2.5-cm off-axis abutment tests comparing measured off-axis profiles at 1-cm depth for two 12 MeV fields abutted at 2.5 cm off-axis and a single 12 MeV field. (a) Trial 1, (b) Trial 2, (c) Trial 3 ...........................................63

42. Dosimetric impact of errors in block fabrication. Two 12 MeV beams abutted on central-axis with errors introduced in edge placement are compared with a single 12x12-cm$^2$ field. (a) $\Delta x = -4$ mm resulting in $\Delta D$ of -35.8%, (b) $\Delta x = -2$ mm resulting in $\Delta D$ of -18.3%, (c) $\Delta x = 0$ mm resulting in $\Delta D$ of +1.2%, (d) $\Delta x = +2$ mm resulting in $\Delta D$ of +14.6%, and (e) $\Delta x = +4$ mm resulting in $\Delta D$ of +34.1% .................................................................65

43. Measured dosimetric impact of errors in beam edge placement plotted with the calculated line with 8.5 %/mm slope. Results are plotted as measured $\Delta D$ vs. intended $\Delta x$ ...........................................................................66

44. Isodose plots of TPS calculated dose distributions for irradiation of the two-step block target in a polystyrene phantom for (a) the standard applicator plan and (b) the variable-SCD applicator plan .................................................68
45. Cumulative DVHs for the two-step block target showing PTV and Outside PTV ROIs .............................................................................................................69

46. Isodose plots of TPS calculated dose distributions for irradiation of the wedge target in the polystyrene phantom for (a) the standard applicator plan and (b) the variable-SCD applicator plan.................................................................71

47. Cumulative DVHs for the wedge block target showing PTV and Outside PTV ROIs .............................................................................................................72

48. Isodose plots of TPS calculated dose distributions for irradiation of the pentagon target in the polystyrene phantom for (a) the standard applicator plan and (b) the variable-SCD applicator plan ..................................................................................74

49. Cumulative DVHs for the pentagon target showing PTV and Outside PTV ROIs .............................................................................................................75

50. Isodose plots of TPS calculated dose distributions for irradiation of the inverted well target in the polystyrene phantom for (a) the standard applicator plan and (b) the variable-SCD applicator plan ..................................................................................78

51. Cumulative DVHs for the inverted well target showing PTV and Outside PTV ROIs .............................................................................................................79

52. Measured and TPS calculated dose data for irradiation of the two-step block target using the variable-SCD applicator plan. (a) Isodose plot of central xz plane dose distributions. (b) Off-axis dose profiles at 2-cm depth.............82

53. Measured and TPS calculated depth dose plots for irradiation of the two-step block target at (a) -3.4 cm off-axis, and (b) 2.6 cm off-axis.........................................................83

54. Measured and TPS calculated dose data for irradiation of the wedge target using the variable-SCD applicator plan. (a) Isodose plot of central xz plane dose distributions. (b) Off-axis dose profiles at 2-cm depth ........................................84

55. Measured and TPS calculated central axis depth dose plots for irradiation of the wedge target .............................................................................................................85

56. Measured and TPS calculated dose data for irradiation of the pentagon target using the variable-SCD applicator plan. (a) Isodose plot of central xz plane dose distributions. (b) Off-axis dose profiles at 2-cm depth ........................................86

57. Measured and TPS calculated central axis depth dose plots for irradiation of the pentagon target .............................................................................................................87

58. Measured and TPS calculated dose data for irradiation of the inverted well target using the variable-SCD applicator plan. (a) Isodose plot of central xz plane dose distributions. (b) Off-axis dose profiles at 2-cm depth .................89
59. Measured and TPS calculated depth dose plots for irradiation of the inverted well target at (a) 0 cm off-axis, and (b) -3.4 cm off-axis .......................................................... 90

60. Sagittal view of the field segmentation scheme showing locations of transverse slices 73, 94 and 110 (z = 1.5, -4.5, and -9.3 cm, respectively) ............................................. 91

61. TPS dose distributions in transverse slice #73 calculated using (a) the standard applicator, and (b) the variable-SCD applicator. Two 12 MeV beams and a 16 MeV beam were used to treat this portion of the PTV .............................................. 91

62. TPS dose distributions in transverse slice #94 calculated using (a) the standard applicator, and (b) the variable-SCD applicator. A 9 MeV beam, two 12 MeV beams, and a 20 MeV beam were used to treat this portion of the PTV ................................. 92

63. TPS dose distributions in transverse slice #110 calculated using (a) the standard applicator, and (b) the variable-SCD applicator. A 9 MeV and 12 MeV beam were used to treat this portion of the PTV .......................................................... 92

64. Cumulative DVHs for segmented-field ECT patient plan using the standard applicator and the variable-SCD applicator. Dose to the PTV, lacrimal gland, eye, cord, and optic nerve are shown........................................................................ 94

A1. 3D view of the leg and back plate design with (a) tabs housing frame alignment holes attached and (b) tabs without alignment holes attached................................. 104

A2. (a) Front view of leg and back plate design with tabs in place. Threaded holes for tab attachment are shown. (b) Rear view of leg and back plate design ...... 104

A3. Front view of leg design detailing dimensions concerning tab placement and spacing. Note: the holes in this figure were threaded for a 4-40 screw.............. 105

A4. (a) 3D view of tab design showing tab dimensions. (b) Top view of tab showing placement and dimensions of alignment holes. (c) Front view of tab showing placement and dimensions of holes (not threaded) used for leg attachment........................................................................................................... 106

A5. 3D view of frame design detailing all dimensions including peg diameter ...... 107

B1. Isodose plots for 6 MeV, 15x15-cm$^2$ field measured and calculated using the standard 5-cm air gap........................................................................................................ 108

B2. Isodose plots for 9 MeV, 15x15-cm$^2$ field measured and calculated using the standard 5-cm air gap........................................................................................................ 109

B3. Isodose plots for 9 MeV, 15x15-cm$^2$ field measured and calculated using the variable-SCD method with a 7.5-cm air gap ................................................................. 110
B4. Isodose plots for 12 MeV, 15x15-cm² field measured and calculated using the standard 5-cm air gap.................................................................111

B5. Isodose plots for 12 MeV, 15x15-cm² field measured and calculated using the variable-SCD method with a 11.5-cm air gap ........................................112

B6. Isodose plots for 16 MeV, 15x15-cm² field measured and calculated using the standard 5-cm air gap.................................................................113

B7. Isodose plots for 16 MeV, 15x15-cm² field measured and calculated using the variable-SCD method with a 17.5-cm air gap ........................................114

B8. Isodose plots for 20 MeV, 15x15-cm² field measured and calculated using the standard 5-cm air gap.................................................................115

B9. Isodose plot for 20 MeV, 15x15-cm² field measured and calculated using the variable-SCD method with a 19.5-cm air gap .................................116
Abstract

**Purpose:** Segmented-field electron conformal therapy is characterized by dose heterogeneity due to unmatched penumbra of abutted fields of differing energy. The present work investigates the potential to decrease dose heterogeneity by approximately matching beam penumbra using energy-specific source-to-collimator distances (SCDs). It was hypothesized that a clinically practical, variable-SCD method that utilizes Cerrobend® custom inserts can deliver segmented-field electron conformal therapy in the energy range of 6-20 MeV with less than ±5% variation in dose spread in the abutment regions of hypothetical planning target volumes (PTVs), i.e. constrain the PTV dose to 85%-105%.

**Methods:** A Varian 15x15-cm² electron applicator was modified to allow energy-dependent SCDs resulting in energy-dependent air gaps. Air gaps were chosen based on theoretical calculations to approximately match penumbra for 6, 9, 12, 16, and 20 MeV beams at a depth of 1.5 cm in water. Treatment plans developed for four simulated PTVs and a single patient using the variable-SCD applicator were compared to identical plans using the current constant-SCD applicator. Dose plans for the simulated PTVs using the variable-SCD applicator with electron inserts cut with diverging edges were delivered to film in a polystyrene phantom to assess feasibility.

**Results:** Treatment planning results in the four simulated PTVs showed that dose homogeneity in agreement with the hypothesis can be achieved using the variable-SCD applicator. Minimum dose was increased by an average of 4%, and maximum dose was decreased by an average of 4%. On average, the standard deviation of the dose decreased by 29%, and \( D_{90-10} \) decreased by 32%. Measured dose in the abutment regions for all
four simulated targets using the modified applicator agreed well with TPS predicted dose. For the patient PTV, the variable-SCD applicator plan predicted a 14% increase in minimum dose, a 10% decrease in maximum dose, and a 22% reduction in both the standard deviation of the dose distribution and D$_{90-10}$ as compared to the standard applicator plan.

**Conclusion:** The results of this study demonstrated that dose homogeneity in segmented-field electron conformal therapy can be substantially improved by using energy-dependent SCDs to match beam penumbra.
Chapter 1
Introduction

1.1 Electron Conformal Therapy

Electron conformal therapy (ECT) is the use of one or more electron beams (1) to contain the PTV in the 90% dose surface, (2) to deliver as homogenous a dose distribution as possible or a prescribed heterogeneous distribution to the PTV, and (3) to deliver minimal dose to underlying critical structures and normal tissues. Modulated electron therapy (MET) is the use of energy modulation and/or intensity modulation to achieve ECT. Energy MET can be delivered using custom bolus (bolus ECT) or multiple abutted beams of different energies (segmented-field ECT) (Hogstrom et al. 2003a). Intensity modulated electron therapy (IMET) can be achieved with scanned electron beams (Lief, Larsson, and Humm 1996), electron multi-leaf collimators (eMLC) (Lee, Jiang, and Ma 2000, Hogstrom et al. 2004), few-leaf collimators (Al-Yahya et al. 2005), or multiple field cutouts. IMET is presently impractical due to the lack of commercially available treatment delivery hardware (e.g. eMLC) and treatment planning software (Hogstrom et al. 2004).

1.2 Bolus and Segmented-Field ECT

Bolus ECT (Starkschall et al. 1991; Low et al. 1992) has been used since 1990 at The University of Texas MD Anderson Cancer Center to treat posterior chest wall sarcoma (Low et al. 1995), post-mastectomy chest wall (Perkins et al. 2001), and head and neck disease (Kudchadker et al. 2002, 2003), and that technology is presently being implemented into the Pinnacle treatment planning system (Philips Medical Systems, Inc, Andover, MA). Segmented-field ECT, an alternative to bolus ECT, is the use of multiple
electron fields of differing energy and weight to deliver a dose distribution that conforms the 90% dose surface to the distal surface of the PTV (Hogstrom et al. 2003a).

Zackrisson and Karlsson (1996) compared the quality of bolus ECT and segmented-field ECT treatment plans for electron irradiation of the post-mastectomy chest wall. Bolus ECT always delivers a higher skin dose due to dose buildup occurring in the bolus material; this effect can be advantageous in some cases, but a limiting factor in others. Segmented-field ECT does not inherently carry this effect; hence, in cases that require increased surface dose, the use of constant thickness bolus may be required, e.g. Superflab (Radiation Products Design, Inc, Albertville, MN).

One advantage of bolus ECT is that continuous energy modulation can be achieved via variable bolus thickness (Figure 1a). Conversely, segmented-field conformal therapy is at a disadvantage for energy modulation because most linacs offer only coarsely spaced energies (6, 9, 12, 16 and 20 MeV) corresponding to coarsely spaced $R_{90}$ depths (defined as the depth of the distal 90% dose on the central-axis depth dose curve) in unit density tissue (1.8, 2.7, 3.6, 4.8 and 6.1 cm respectively). Although there is no reason radiotherapy machines could not have finer spacing (e.g. 1 MeV), this current limitation might lessen the utility of segmented ECT. One disadvantage to the continuous energy modulation of bolus ECT is that it uses a single beam energy. The energy is selected to reach the deepest portion of the PTV, and this results in needless dose distal to shallow regions of the PTV due to increased $R_{90-10}$ compared to that for segmented-field ECT.
Figure 1: (a) Bolus conformal therapy using a single 20 MeV beam. Hot spots of 110% are due to loss of side scatter equilibrium as a consequence of the steep gradient in the bolus material. (b) Segmented-field conformal therapy. The three beams share a common source position. Hot spots of 110% are due to abutment of beams of different energy having unmatched penumbra. The reference point was used for dose normalization. [Zackrisson and Karlsson 1996]

A significant obstacle to delivering clinically acceptable segmented-field ECT is abutment dosimetry. Current clinical methods approach segmented-field ECT without modification of the beam penumbra, and this often produces dose variations (hot/cold spots) greater than ±5% from the ideal target volume dose spread (90% - 100%) (Figure 1b). These dose heterogeneities are due to the wider penumbra width of lower energy beams compared to higher energy beams, creating a hot spot (cold spot) just inside the edge of the high (low) energy field. By employing more closely matched penumbras it should be possible to reduce dose heterogeneity in the abutment region to less than ±5% in most cases.

Possible methods to modify (broaden) the penumbra are (1) introducing a scatter wedge at the beam edge (Kurup, Wang, and Glasgow 1992), (2) using a combed-tooth collimating edge (Kalend et al. 1985), and (3) increasing the air gap between collimator
and patient surface (Maor et al. 1985; Lachance, Tremblay, and Pouliot 1997). From a practical perspective, because the typical electron field is irregularly shaped, the air gap method is the most viable method. Therefore, in the present study, use of variable air gaps (i.e. energy-dependent) to improve abutment dosimetry for segmented-field ECT was investigated. Although only Cerrobend® (Lipowitz metal) inserts are used in this study, the method could potentially be implemented using a variable-source-to-collimation-distance (SCD) eMLC (Hogstrom et al. 2004).

Two methods of varying the air gap are possible. The variable source-to-surface-distance (SSD) method uses the existing collimating system (i.e. Cerrobend® inserts in the applicator and constant SCD) and extends the SSD by moving the treatment table (Figure 2a). This method, presently doable, suffers from the following deficiencies: (1) different virtual source positions (with respect to the patient) for fields of different energy result in non-coincidence of beam edges and (2) it requires repositioning of the treatment table for each energy used. The latter makes clinical implementation of the variable-SSD method impractical; therefore, the focus of this study was on the variable-SCD method. With the variable-SCD method (Figure 2b), the air gap is increased by varying the source to collimator distance (SCD), which eliminates the problem of beam non-coincidence and table repositioning, but requires use of redesigned electron applicators that can position the Cerrobend® insert at energy-specific SCDs.

1.3 Hypothesis and Specific Aims

The hypothesis of this research is that a clinically-practical, variable-SCD method that utilizes Cerrobend® custom inserts can deliver segmented-field ECT using electron beams in the energy range of 6-20 MeV with less than ±5% variation in dose spread in
abutment regions of the PTV for hypothetical PTVs, (i.e. constrain the PTV dose to 85%-105%). Four specific aims have been completed to test this hypothesis.

![Diagram](image)

Figure 2: (a) The variable-SSD method with the inner edge of the lower energy beam (E1) abutted to the outer edge of the higher energy beam (E2) at the patient surface illustrates the effect of non-coincident edges as the gap between the two fields increases with depth. (b) The variable-SCD method for the same abutment scheme illustrates beam edge coincidence as beams share a common virtual source. [Note: both schematics are in the reference frame of the patient.]

Aim 1. Design a geometry for providing approximately equal penumbra widths at a constant depth (≈1.5 cm) for different energy beams for the variable-SCD method. Design and fabricate the necessary hardware for the variable-SCD method by modifying a Varian 2100EX 15x15-cm² applicator (Varian Medical Systems, Inc, Palo Alto, CA).

Aim 2. Verify that electron beams for the Varian 2100EX 6/18 radiotherapy accelerator (6, 9, 12, 16, and 20 MeV) are commissioned in the Philips Pinnacle treatment planning system so that its electron pencil-beam algorithm correctly calculates
dose in a water or polystyrene phantom for the standard and modified (variable-SCD) collimating systems.

Aim 3. Develop a procedure for constructing Cerrobend® custom inserts with diverging edges for the modified, variable-SCD applicator, and assess the impact of fabrication inaccuracy by irradiating a film phantom with segmented-fields of the same energy.

Aim 4a. Test and compare the dose homogeneity of standard beam delivery (constant SSD and constant SCD) with that of the variable-SCD method for 1-D segmented-fields for four hypothetical PTVs and assess a possible method for QA by comparing calculated with measured dose distributions for the 1-D segmented-field treatment plans.

Aim 4b. Illustrate the clinical potential of the variable-SCD method by comparing the dose homogeneity of standard beam delivery (constant SSD and constant SCD) with that of the variable-SCD method for 2-D segmented-fields for a patient case.
Chapter 2
Methods and Materials

2.1 Aim 1: Design of Methods for Matching Penumbra

Ideal abutment requires that the shape of the dose distributions in the abutted penumbras regions be broad, matched, and have a common virtual source (which is characteristic of the variable-SCD system) (Khan et al. 1991). By using an increased air gap (decreased SCD), the penumbra shape of a higher energy beam can be broadened to approximately match that of a lower energy beam at a specified depth. To determine the magnitude of air gaps necessary to achieve acceptable abutment dosimetry, penumbra width ($P_{80-20}$ or $P_{90-10}$ illustrated in Figure 3) was calculated using Fermi-Eyges theory (Hogstrom, Mills, and Almond 1981; Hogstrom 1986; Hogstrom 1996).

Figure 3: $P_{90-10}$ and $P_{80-20}$ illustrated using a measured 12 MeV beam profile at 3-cm depth for a 10x10-cm$^2$ field size. The profile has been normalized to 100% of the central-axis dose.
2.1.1 Theoretical Penumbra Width

The energy-specific air gaps used in the implementation of the variable-SCD method were chosen based on theoretical calculations of penumbra width. The penumbra width, which depends on energy, air gap (SSD - SCD), and depth, can be calculated from the beam sigma $\sigma_x(z)$, a measure of the lateral spatial Gaussian distribution of electrons about the mean direction at a point in the x-y plane at depth $z$. Hogstrom et al. (1981) showed that the penumbra width is related to the sigma of a normally incident point (or pencil) beam by:

$$P_{90-10} = 2.56 \times \sigma_x(z)$$  \hspace{1cm} (1a)$$

$$P_{80-20} = 1.68 \times \sigma_x(z),$$  \hspace{1cm} (1b)$$

with $\sigma_x$ given by:

$$\sigma_x(z) = [(L_0 + z)^2 \sigma_{\theta_x}^2 + A_z(z)]^{1/2},$$  \hspace{1cm} (2)$$

where $L_0$ (cm) is the air gap (SSD-SCD), $z$ (cm) is depth in water, $\sigma_{\theta_x}$ (radians) is the root-mean-square (RMS) spread of the projected angular distribution of a point beam at the collimator (due to scatter in the dual scattering foil system and air), and $A_z$ is the second moment of the linear angular scattering power at depth $z$ in water (characterizing scatter in the phantom) given by:

$$A_z(z) = \frac{1}{2} \int_0^z T_{water}(z')(z - z')^2 \, dz'.$$  \hspace{1cm} (3)$$
The scattering power in water, $T_{water}$, was computed using Werner’s approximation: $T_{water}(E) = 4.525 \times E^{-1.78}$ (Werner 1982) and Harder’s relationship for electron energy at depth: $E(z) = E_0(1 – z/R_p)$ (Khan et al. 1991). Thus,

$$T_{water}(z') = 4.525 \times E_{p,0}^{-1.78} \left(1 - \frac{z'}{R_p}\right)^{-1.78}, \quad (4)$$

where $E_{p,0}$ is the most probable incident energy and $R_p$ is the practical range. Combining (3) and (4),

$$A_z(z) = \frac{1}{2} \int_0^z 4.525 \times E_{p,0}^{-1.78} \left(1 - \frac{z'}{R_p}\right)^{-1.78} (z - z')^2 \, dz'. \quad (5)$$

A computer program, which accepts the air gap ($L_0$) and initial beam energy ($E_{p,0}$) as input, was written using Microsoft Visual Studio to calculate the above quantities and output $P_{90-10}$ and $P_{80-20}$ as a function of depth.

**2.1.2 Variable-SCD Applicator**

Based on preliminary results of penumbra calculations, a 15x15-cm² electron applicator was modified to accept placement of Cerrobend® inserts at specific locations upstream from the conventional location (95-cm SCD, which gives a 5-cm air gap to isocenter). The modified applicator (Figure 4) allowed for adjustable air gaps in 2-cm increments from 7.5 cm to 19.5 cm. To minimize the potential for alignment errors, the modification was designed and machined to provide electron insert placement perpendicular to the beam axis (x-y plane) with ±0.13 mm accuracy.

The variable-SCD applicator was designed as a series of shelves on which a frame holding an electron insert can be placed (Figure 5, see detailed drawings in Appendix A).
Figure 4: Photograph of modified 15x15-cm² applicator from direction of Cerrobend® insert insertion. The insert shown is positioned upstream from the standard location, creating a 11.5-cm air gap.

Figure 5: Side view sketch of the shelf structure of the modified applicator.
Each shelf is comprised of four tabs attached to four perpendicular legs. Two of the four tabs have alignment holes that constrain the frame location via two alignment pins on the frame holding the Cerrobend® cutout. Also, the aluminum insert frame includes two setscrews to ensure consistent placement of the Cerrobend® insert in the frame (Figure 6). To obtain a rigid structure, the two back plates, each holding two legs, were milled out of a single piece of aluminum.

Figure 6: (a) Photograph of downstream side of the insert frame with a Cerrobend® insert. Setscrews push the insert snugly against the opposite corner of the frame (the allen wrench indicates the location of one set screw). (b) Photograph of aluminum insert frame.
2.2 Aim 2: Verification of Electron Beam Commissioning in TPS

The goal of Aim 2 was to assess the ability of the electron pencil-beam algorithm commissioned in the Philips Pinnacle\textsuperscript{3} (version 7.4f) treatment planning system (TPS) to accurately compute dose in a water or polystyrene phantom using the standard and variable-SCD collimating systems. Staff medical physicists at Mary Bird Perkins Cancer Center (MBPCC) commissioned the Pinnacle's implementation of the Hogstrom electron pencil beam algorithm for all beam energies (6, 9, 12, 16, and 20 MeV) on a Varian 2100EX 6/18 radiotherapy accelerator (S/N 1251) for the standard applicator geometry according to the recommendations of Hogstrom and Steadham (1996). The beam data for the variable-SCD applicator "Machine" in Pinnacle\textsuperscript{3} were assumed identical to that of the standard SCD (95 cm), with the exception of L_0 and \(\sigma_\theta\). The actual energy-specific air gaps (called "drift distance" in the TPS) were used, and a small modification was made to \(\sigma_\theta\) for each beam to account for the new SCDs (Hogstrom 1982), i.e.

\[
\sigma_\theta^\text{var} = \sigma_\theta^\text{std} \frac{SCD_{var}}{SCD_{std}}.
\] (6)

Algorithm commissioning for the standard and variable-SCD methods were verified by comparing Pinnacle\textsuperscript{3} calculations with measured dose distributions for 4x4-cm\textsuperscript{2} and 15x15-cm\textsuperscript{2} fields at 100-cm SSD. Data comparisons for 110-cm SSD were not performed because the variable-SCD method uses a 100-cm SSD exclusively.

2.2.1 Calculated Dose Distributions

2.2.1.1 Hogstrom Pencil-beam Algorithm

The Pinnacle\textsuperscript{3} calculates three dimensional electron beam dose distributions using the 3D implementation (Starkschall et al. 1991) of the Hogstrom pencil-beam algorithm
(Hogstrom, Mills and Almond, 1981). The algorithm models arbitrary field shapes using a collection of pencil beams (typically 2x2 mm² projected to isocenter) defined at the level of final collimation. The dose distribution for each pencil beam is modeled to include Fermi-Eyges multiple Coulomb scatter theory, beam divergence, and Collisional energy loss, the latter requiring a measured broad beam central-axis depth-dose curve for a rectangular field that approximates the field size to be calculated. Off-axis ratios determined from measured beam profiles are used to weight pencil beams, hence accounting for beam non-uniformity. A particularly useful feature of the Hogstrom pencil beam algorithm is its ability to account for the impact of arbitrary air gaps (SSD - SCD) on the beam penumbra without impacting its accuracy for dose effects due to the patient's irregular surface and internal heterogeneity.

2.2.1.2 Treatment Planning Dose Distributions

- Setup

Dose distributions for all field size and energy combinations were calculated in the Philips Pinnacle³ TPS. Beams were applied to the Pinnacle's generic water phantom using a 15x15-cm² applicator with the gantry at 180° (beam direction perpendicular to the phantom surface). The collimator was rotated from the default 180° to 90° with the block "rotate with collimator" option set to "no" for computation. This allows for a higher resolution in beam edge placement in the transverse plane. A dose grid of 151x151x80 voxels with a voxel size of (0.2 cm)³ was used for all calculations.

The nominal 100-cm source-to-axis distance (SAD) was used for computation. It would have been preferable to use a 90-cm virtual source to axis distance to model the actual virtual source position of the Varian 2100EX (Shiu 1994). Although the software
allows the user to specify the virtual source distance, the input value has no impact on algorithm output, i.e. 100 cm is used regardless.

To model the measurement phantom (hi-impact white polystyrene) the medium-to-water linear collisional stopping power and linear scattering power ratio lookup table was modified to map polystyrene ratios to the density of water. Ratios taken from data published in TG-25 (Table 1a) were inserted into the file NewElectronStoppingPower.db so that polystyrene ratios are used for densities of 0.98 to 1.02 g/cm$^3$ (Table 1b). TG-25 specifies a polystyrene density of 1.054 g/cm$^3$. The measured mass density of our polystyrene phantom was 1.053 ± 0.003 g/cm$^3$. The CT number of the virtual phantom was 1024, which specified a density of 1 g/cm$^3$ (Table 1c).

- **Data Analysis**

The TPS outputs the dose matrix to a binary file, from which the dose distribution in the central xz plane ($y = 0$) was extracted using in-house software. These data were then converted to the format required by the SigmaPlot® contour plot feature. Once imported into SigmaPlot®, dose distributions were normalized to 100% of the central-axis dose maximum, and 2D isodose plots and 1D dose profile plots were generated.

### 2.2.2 Relative Dose Distribution Measurements

#### 2.2.2.1 Varian 2100EX Radiotherapy Accelerator

Electron beams of 6, 9, 12, 16 and 20 MeV were provided by a Varian 2100EX radiotherapy accelerator with a nominal source-to-axis distance (SAD) of 100 cm.

#### 2.2.2.2 Film Phantom

Beam data were acquired by irradiating a hi-impact, white opaque polystyrene ($C_8H_8 + TiO_2$) film phantom (25.4 x 35.6 x 10 cm$^3$) with 10x12 in (25.4 x 30.5 cm$^2$)
Table 1a: Polystyrene stopping power and scattering power ratios from TG-25 published data.

<table>
<thead>
<tr>
<th></th>
<th>Water</th>
<th>High-impact polystyrene (white-opaque)</th>
<th>Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Linear collision stopping power (MeV/cm)</td>
<td>1.968</td>
<td>2.01</td>
<td>1.021</td>
</tr>
<tr>
<td>Linear angular scattering power (radian²/cm)</td>
<td>0.0695</td>
<td>0.0603</td>
<td>0.868</td>
</tr>
</tbody>
</table>

Table 1b: Stopping power and scattering power ratio lookup table used to calculate dose in a polystyrene phantom.

<table>
<thead>
<tr>
<th>Density (g/cm³)</th>
<th>Linear collision stopping power Ratio</th>
<th>Linear angular scattering power</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.000</td>
<td>0.001</td>
<td>0.001</td>
</tr>
<tr>
<td>0.291</td>
<td>0.311</td>
<td>0.292</td>
</tr>
<tr>
<td>0.927</td>
<td>0.933</td>
<td>0.729</td>
</tr>
<tr>
<td>0.980</td>
<td>1.021</td>
<td>0.868</td>
</tr>
<tr>
<td>1.020</td>
<td>1.021</td>
<td>0.868</td>
</tr>
<tr>
<td>1.047</td>
<td>1.051</td>
<td>1.040</td>
</tr>
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<td>1.100</td>
<td>1.098</td>
<td>1.135</td>
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<td>1.427</td>
<td>1.422</td>
<td>1.863</td>
</tr>
<tr>
<td>1.940</td>
<td>1.940</td>
<td>3.026</td>
</tr>
<tr>
<td>7.000</td>
<td>7.000</td>
<td>9.900</td>
</tr>
</tbody>
</table>

Table 1c: CT number to density table used for dose computation.

<table>
<thead>
<tr>
<th>CT Number</th>
<th>Density (g/cm³)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0.000</td>
</tr>
<tr>
<td>45</td>
<td>0.001</td>
</tr>
<tr>
<td>255</td>
<td>0.195</td>
</tr>
<tr>
<td>524</td>
<td>0.495</td>
</tr>
<tr>
<td>963</td>
<td>0.967</td>
</tr>
<tr>
<td>999</td>
<td>0.991</td>
</tr>
<tr>
<td>1024</td>
<td>1.000</td>
</tr>
<tr>
<td>1074</td>
<td>1.062</td>
</tr>
<tr>
<td>1082</td>
<td>1.071</td>
</tr>
<tr>
<td>1262</td>
<td>1.161</td>
</tr>
<tr>
<td>1862</td>
<td>1.609</td>
</tr>
<tr>
<td>15000</td>
<td>11.300</td>
</tr>
</tbody>
</table>
Kodak XV film (Figure 7). Hi-impact, white polystyrene was chosen because of its near water equivalence (Khan et al. 1991). Exact alignment (±0.1 mm) of the edge of the film

Figure 7: (a) Photo of the Hi-impact, white opaque polystyrene film phantom shown in open position with 10"x12" Kodak XV film loaded for dose measurement. (b) Phantom shown upright; note light opaque tape and plastic hinges that lie outside the radiation beam.
with the phantom surface is important to prevent electron scatter artifacts (Khan et al. 1991). For consistent and accurate film positioning (±0.1 mm reproducibility) within the phantom, alignment holes were drilled in the film using an electric three hole punch (Swingline Model 525) allowing film placement to be guided by corresponding alignment pegs in the phantom. A random error of 0.1 mm in the position of the alignment holes was determined by measuring the distance from film edge to the perimeter of the three alignment holes for six test films. To make the phantom light tight, black opaque “3M Scotch Photographic Tape 235” was applied along the surface where the beam is incident. The remaining perimeter of the phantom was covered with black felt attached to a permanent Velcro "hooks" lining (Bova 1990).

### 2.2.2.3 Measurement Setup

For film irradiation, the phantom was placed on the treatment table, which was adjusted so that the "hinged" surface was positioned at the appropriate source to surface distance (100-cm SSD) using the optical distance indicator. Alignment along the x- and y-axes was done using the light field cross hairs such that the film lay in the plane of gantry rotation (xz plane, Figure 8). Measurements were made with the gantry positioned at 180°.

For fields using the nominal 5-cm air gap, Cerrobend® blocks were positioned in the standard 15x15-cm² applicator such that the downstream surface of the block was 5 ± 0.2 cm from isocenter. For fields using the variable-SCD method, inserts were placed in the variable-SCD applicator at 7.5, 11.5, 17.5 and 19.5 cm from isocenter for 9, 12, 16, and 20 MeV beams respectively.
Figure 8: Schematic of gantry and treatment room coordinate system. The gantry is positioned at 180°. “I” indicates isocenter. The gantry rotates about the y-axis in the x-z plane.

Initially a thin air gap (~0.25 mm) between the film and the phantom along the plane of the film resulted in poor agreement between measured and calculated %DD (cf. Figure 9). This was due to scatter effects as described by Dutreix and Dutreix (1969).

Figure 9: Measured central-axis depth dose curves for a 12x12-cm\(^2\) 12-MeV beam. The measured depth dose without the C-clamp (dashed line) shows the influence of approximately 0.25 mm air in the phantom along the surface of the film compared to the measured depth dose with the C-clamp (solid line).
Subsequently, a large C-clamp was employed to compress the phantom and alleviate this issue. The final measurement setup is shown in Figure 10 with the standard $15 \times 15 \text{cm}^2$ applicator.

![Figure 10: The standard $15 \times 15 \text{cm}^2$ applicator (with $10 \times 10 \text{cm}^2$ insert) attached to the Varian 2100EX. The film phantom is in measurement position with the C-clamp in place.](image)

### 2.2.3 Film Dosimetry

All films used in this study were irradiated "edge on" with an exposure corresponding to a maximum central-axis dose of approximately 50 cGy, i.e. films were irradiated to the number of monitor units (MU) that would deliver a maximum central-axis dose in water of 50 cGy. This resulted in a maximum film optical density of approximately 1.5.
2.2.3.1 Analysis

- **Film Developing**

  All films were developed using a X-OMAT 270 RA processor. Before use, the processor was allowed to warm up to a developer temperature readout of 94.2 ± 0.1°F, if necessary. To ensure the processor temperature was stable and to eliminate any abnormal initial conditions (such as dust inside the film transport mechanism), four blank films were run through prior to processing irradiated films. When processing a set of exposed films, the temperature readout was checked after every four to five films processed. If the temperature reading deviated by more than ±0.2°F the temperature was allowed to stabilize before processing was resumed.

- **Readout**

  The 2D distribution of exposure of each developed film was read using the Vidar DosimetryPro™ Advantage 16-bit digitizer with a scanning resolution of 356 μm (corresponding to the average value of 16 89-μm pixels.) Incandescent light transmitted through the film is registered by a linear array of 89-μm CCD detector elements. The signal from each CCD element is recorded as a 16-bit (65,536 gray level) A/D value. RIT software version 4.2 was used to apply a 9x9 low pass filter to the calibration films and a 7x7 low pass filter to all other films to remove high frequency noise.

- **Data Analysis**

  The 2D distribution of exposure was transformed to dose using the calibration curve, ie. D=D(OD), discussed in the next section. Analysis of the measured 2-D dose distributions was performed using Microsoft SigmaPlot® 9.0 software. To facilitate this, a central-axis depth dose curve and multiple cross axis profiles in 5-mm increments from
a depth of 5 mm to just beyond $R_p$ were exported as text files for each film analyzed. The files were then used to create a 2-D dose matrix formatted for the SigmaPlot® contour plot feature using in-house software. In SigmaPlot®, dose distributions were normalized to 100% at $D_{max}$, as determined from the film central-axis depth-dose data, and isodose plots were created.

### 2.2.3.2 Calibration

Calibration films were exposed perpendicular to the beam. Khan et al. (1991) reported perpendicular and parallel film exposure give the same results under the correct conditions, i.e. no air gap between the film and the phantom and precise film edge placement for parallel exposure. In accordance with common practice for relative film dosimetry measurements, a single calibration curve was created for each batch of film used (box of 50 films) (Bos et al. 2002).

Calibration curves were determined from 10 calibration points, created using four calibration films and applied to the measured dose distributions using RIT software. The first calibration film was not irradiated, providing base + fog optical density level of the film. The remaining packaged films were placed at $R_{100}$ (the depth of maximum dose) for a 9 MeV beam (2 cm of Plastic Water®) and irradiated using a 10x10-cm² insert in a standard 10x10-cm² applicator, i.e. the setup for which the accelerator is calibrated to deliver 1 cGy/MU.

The RIT software was used to generate calibration curves by creating a piecewise polynomial fit to the measured data. To prevent the piecewise polynomial fit from underestimating dose (or estimating negative doses) at low exposures, the second calibration film was irradiated with 1 MU (1 cGy). Four fields were delivered to the third
and fourth films. The third film (illustrated in figure 11) received 10, 18, 26, and 34 MU, and the fourth received 42, 50, 58, and 66 MU. The films were scanned and the average scanner value for a 2x2-cm\(^2\) box in the center of each field was correlated to the delivered dose to create points on the calibration curve.

![Figure 11: Example of calibration film #3, used to generate four points on the calibration curve. This film received 10, 18, 26 and 34 MU. 42, 50, 58 and 66 MU were delivered in the same fashion to calibration film #4.](image)

Total dose to each field on the four-field films was as much as 3% greater than the direct dose delivered because of scattered and leakage dose from adjacent fields. To determine the total dose, a single region of a film was irradiated to a high dose (Figure 12) and scanner values at adjacent field locations were converted to optical density (OD) using the OD calibration step-wedge provided with the scanner.
Scattered and leakage dose per monitor unit for adjacent fields was determined by converting the OD of regions adjacent to the high dose field into dose using the slope of the linear region of the dose response curve. Because dose contributions to adjacent fields is proportional to monitor units, the total dose delivered to a field region was determined by summing the dose per monitor unit from adjacent fields. For example, the dose to field 2, $D_2'$, is given by (Figure 13):

$$D_2' = MU_2 + D_{2/3} + D_{2/4} + D_{2/5}$$

where $D_{ij}$ is the dose to field i from field j:

Figure 12: Scanned film used to account for scatter and leakage dose contributions to the calibration fields. The dark area received 100 MU under reference conditions. OD$_{ij}$ denotes optical density to field i after the irradiation of field j.
\[ D_{2/3} = (OD_{3/6} - OD_{BG}) \times S_0 \frac{MU_3}{MU_6}, \quad (8a) \]

\[ D_{2/4} = (OD_{4/6} - OD_{BG}) \times S_0 \frac{MU_4}{MU_6}, \quad (8b) \]

\[ D_{2/5} = (OD_{5/6} - OD_{BG}) \times S_0 \frac{MU_5}{MU_6}. \quad (8c) \]

\( S_0 \) is the inverse of the slope of the linear dose response region (cGy/OD), \( OD_{BG} \) is the background optical density, \( OD_{ij} \) is the OD at field \( i \) due to field \( j \), and \( MU_n \) is the monitor units delivered to field \( n \). Total dose, \( D'_2 \), to fields 3, 4, and 5 were determined in the same way.

Figure 13: Diagram of the dose contribution from adjacent fields to \( D'_2 \).
The slope of the linear region of the dose response curve (Figure 14) was determined iteratively by implementing equations 7 and 8 in a spreadsheet and plotting OD vs. dose. OD values were determined by converting AD to OD using the OD calibration step-wedge. The initial $S_0$ was obtained from the slope of a linear regression of the OD vs. dose plot for the 0-MU, 10-MU, and 18-MU fields assuming no scattered and leakage dose. Plotting the same curve with scattered and leakage dose to each field included gave a new $S_0$ value which was used to recalculate total dose to each field. This was repeated until the value of $S_0$ changed by less than 0.003%. Figure 15 shows an example of the AD (analog to digital pixel value) vs. dose curve with the piecewise polynomial applied to measured dose distributions by the RIT software; Table 2 lists the data for one calibration curve.

![Net OD vs. Dose](image)

**Figure 14:** Dose response curve for XV film. Background OD was subtracted from the raw value to give net optical density for this plot.
Figure 15: Example plot of AD value vs. dose for a ten-point calibration curve. The 1-cGy point prevents the piecewise polynomial from prematurely intersecting the ordinate. Note: the linear and spline fits were not used.

Table 2: Data for a ten-point calibration curve. Monitor units delivered to each field are shown followed by total dose (D'), 16-bit analog to digital pixel value (AD), and optical density (OD).

<table>
<thead>
<tr>
<th>MU</th>
<th>D' (cGy)</th>
<th>AD</th>
<th>OD</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0.0</td>
<td>44952</td>
<td>0.17</td>
</tr>
<tr>
<td>1</td>
<td>1.0</td>
<td>41476</td>
<td>0.20</td>
</tr>
<tr>
<td>10</td>
<td>10.21</td>
<td>18921</td>
<td>0.54</td>
</tr>
<tr>
<td>18</td>
<td>18.18</td>
<td>10080</td>
<td>0.81</td>
</tr>
<tr>
<td>26</td>
<td>26.20</td>
<td>5262</td>
<td>1.08</td>
</tr>
<tr>
<td>34</td>
<td>34.17</td>
<td>3088</td>
<td>1.33</td>
</tr>
<tr>
<td>42</td>
<td>42.49</td>
<td>1724</td>
<td>1.58</td>
</tr>
<tr>
<td>50</td>
<td>50.46</td>
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<td>58</td>
<td>58.48</td>
<td>730</td>
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</tr>
<tr>
<td>66</td>
<td>66.45</td>
<td>523</td>
<td>2.10</td>
</tr>
</tbody>
</table>

The total dose to each region of the four-field films determined using the method described above was verified by ion chamber measurements. A Slimline Microchamber (model A16) was placed under 2 cm of plastic water and scattered and leakage dose per
monitor unit from a high dose field was measured at adjacent field locations. These data were used to calculate total dose for each field. The total dose was also measured by placing the ion chamber at the location of each field and delivering the four field pattern. The results in both cases agreed with the results of the method described above within 0.3%.

### 2.3 Aim 3: Construction of Cerrobend® Blocks with Diverging Edges

Each plan developed using segmented-field ECT required that at least one Cerrobend® insert be fabricated for each beam energy used. Optimal abutment dosimetry requires that beam edges of adjacent beams exactly coincide. One requisite to this is that the Cerrobend® inserts be fabricated with diverging edges. This section describes how the electron inserts were fabricated with diverging edges and how the accuracy of fabricated edge placement was assessed.

#### 2.3.1 Block Fabrication with the Compu•cutter® System

##### 2.3.1.1 General

Inserts were fabricated by molding Cerrobend® inside a steel frame with apertures defined by foam blocks, which is the standard method used in the MBPCC radiation therapy clinic. Different from standard clinical practice for electron therapy, a computer-controlled hot wire system, Compu•cutter® system (Huestis Medical, Bristol, RI), was used to cut foam blocks with edges diverging from the virtual radiation source.

The Compu•cutter® software accepts three geometric parameters (Figure 16): (1) the source-to-tray distance (STD), which is the distance from the radiation source to the bottom of the shielding tray; (2) the source-to-film distance (SFD), which is the distance from the radiation source to the plane in which the field shape is defined
(isocenter in the present work), and (3) the SAD, which is the distance from the radiation source to isocenter. In the present work the radiation source is defined as the virtual source (Schröder-Babo 1983). A 90-cm virtual SAD was assumed for simplicity through this work (Shiu et al. 1993). Therefore, the STD was 85 cm. Because the desired field size is most easily defined at isocenter, a 90-cm SFD was used.

Figure 16: Geometry used by the Compu•cutter® software to fabricate foam blocks for making Cerrobend® inserts.

The Compu•cutter® cuts field shapes in foam blocks using a four-axis servo controlled hot nichrome wire cutting mechanism. To account for the melted material on either side of the hot wire (70° C), the "burn thickness" is measured and entered into the software. The software then shifts the cutting wire away from the field by half the "burn
thickness." In the present work a 0.4 mm diameter wire was used with a "burn thickness" of 0.97 mm.

The RF digitizer normally used to input field shapes has a manufacturer quoted accuracy of 0.5 mm and input data points are subject to operator error. Therefore, coordinates of points defining the aperture (simple square and rectangular field shapes in this study) were input manually, i.e. to simulate accuracy that could be expected if field shapes were transferred digitally from a treatment planning system.

2.3.1.2 Blocks for Field Abutment

Ideal abutment requires that beam edges coincide (Hogstrom 2003b). Due to the sharp dose falloff in the penumbral region, slight beam-edge misalignments can result in large dose inhomogeneities. For example, for a 12 MeV beam profile at a depth of 2 cm, the distance from the 45% to 55% dose point is approximately 1 mm. Thus, two 12 MeV beams delivered with the standard 5-cm air gap and abutted with overlap of 1 mm will give rise to dose increase (hot spot) of approximately 10%. Similarly, a gap of 1 mm will result in a dose decrease (cold spot) of approximately 10%.

During block fabrication a primary source of error is inaccurate foam placement in the steel mold. To reduce this positioning error, placement of the foam field shapes was guided by a computer printout of the field shape. To accomplish this, points defining field shapes at isocenter for each beam were scaled by the factor \((\text{SCD}-T_f)/\text{SAD}_{\text{vir}}\), where \(T_f\) is the foam thickness (1.4 cm). A plot of the resulting points (Figure 17) was then printed to scale and attached to the bottom plate of the steel frame (corresponding to the top of the Cerrobend insert). The example in Figure 18 shows how the printouts were
attached to the bottom plate, the bottom plate attached to the steel frame with foam aligned to the printed outline, and the completed electron inserts.

Figure 17: Sample plot of the scaled field shapes used to guide placement of the foam field shapes.

Figure 18: (a) Field shape printout taped to the bottom plate of the 15x15-cm² mold. (b) The steel mold with foam in place. The lead weight prevented movement of the foam when liquid Cerrobend® was poured into the frame. (c) The resulting inserts for a single 20 MeV field (left) and two 12 MeV fields (right). (Figure continued)
2.3.2 Assessment of Accuracy of Block Fabrication Method

The dosimetric impact of fabricating electron inserts as described in the previous section was assessed by irradiating the hi-impact polystyrene film phantom with 12x12-cm$^2$ fields consisting of two abutted 6x12-cm$^2$ fields of the same energy. For the method to be acceptable, the resulting dose distributions should be identical to within ±3% to that of a single field.

For analysis, it was necessary to determine the impact of beam misalignment $\Delta D/\Delta x$ (%/mm) using the variable-SCD applicator. $\Delta D$, defined as the maximum percent dose difference between the off-axis profile for a single square field (normalized to 100% on central-axis) and that for two abutted fields (with identical penumbra), should occur at the location of the junction point $x_{int}$ (Figure 19).

The profile for the abutted fields is normalized to agree with the off-axis profile for the single field at a dose point ($D_N$) in a flat region of the off-axis profile, i.e. away from the abutment region. $\Delta D$ is then given by:

$$\Delta D(\%) = 100\% \times \left\{ \frac{[D_{F1}(x_{int}) + D_{F2}(x_{int})]}{D_N} - 1 \right\} \quad (9)$$
where $D_{F1}(x)$ and $D_{F2}(x)$ are the doses due to fields 1 and 2 at off-axis position $x$, and $x_{int}$ is the off-axis profile intersection point. $D_N$ is expected dose, i.e. the sum of the photon transmission dose for a fully blocked field and the dose for an open field; therefore, $D_N$ is greater than 100%. For ideal abutment ($\Delta D = 0$) of two fields, the percent dose of each field at the junction point should be $D_N/2$.

![Diagram showing overlap and off-axis distance](image)

Figure 19: Illustration of $\Delta D$ and $\Delta x$ using off-axis profiles at 1-cm depth for two 12 MeV beams with an 11.5-cm air gap and abutted on central-axis ($x_{int} = 0$) with $\Delta x = +2$ mm. Note: $\Delta x$ is measured as the distance between $D_N/2$ dose points for each field, which is 51.3% in this example due to the photon dose contribution of $\sim 2.6\%$ for a 12 MeV beam at this depth.

Beam gap or overlap is defined by $\Delta x$ (Figure 19), with $\Delta x < 0$ indicating a field gap and $\Delta x > 0$ indicating a field overlap. In order that $\Delta x = 0$ corresponds to $\Delta D = 0$, $\Delta x$ is defined as the off-axis distance between the $D_N/2$ dose points along the abutment edge of each field:
\[ \Delta x = \frac{x_{D_N}^1}{2} - \frac{x_{D_N}^2}{2} \]  

(10)

Figure 20 plots \( \Delta D \) vs. \( \Delta x \) for a 10x10-cm\(^2\), 12-MeV beam at 1-cm depth (dotted line). In the range \(-5 \, \text{mm} < x < 5 \, \text{mm}\), a linear regression showed the slope \( (\Delta D/\Delta x) \) to be 8.5% mm\(^{-1}\), which can be used to assess the magnitude of \( \Delta x \) from \( \Delta D \). In the present study, the air gap for each beam energy was chosen to achieve penumbra matching, thus \( \Delta D/\Delta x \) for all energies was approximately equal at 1-cm depth, i.e. 8.0, 8.7, 8.5, 7.8, and 8.8% mm\(^{-1}\) for 6, 9, 12, 16, and 20 MeV respectively. Consequently, the impact of error in fabrication and placement of block edges should be independent of beam energy. Therefore, the study of this impact was restricted to 12 MeV in the present work.

![Figure 20: Dosimetric impact (\( \Delta D \)) of beam gap/overlap (\( \Delta x \)) at 1-cm depth for abutment of two 12 MeV beams using the variable-SCD applicator with an 11.5-cm air gap. The dotted line represents the calculated relationship derived from the lateral falloff of a film-measured profile. The solid line represents a linear regression of the data for -5 < \( \Delta x \) < 5, yielding a slope of 8.48 %D/mm.](image-url)
2.3.2.1 Accuracy and Precision of Off-axis Abutment Test

The precision of abutment was ascertained at 2.5-cm off-axis ($x_{int} = 2.5$ cm) by irradiating 3 films with 3 independently fabricated rectangular pairs (8.5x12 cm$^2$ and 3.5x12 cm$^2$). $\Delta D$ was measured from the film profiles and $\Delta x$ was calculated from $\Delta x = \Delta D(\Delta D/\Delta x)^{-1}$. While the sample standard deviation represents the precision of a single measurement, the mean dose error gives the accuracy at $x_{int} = 2.5$ cm.

2.3.2.2 Zero and 5-cm Off-axis Abutment Test

The impact of the location of and off-axis abutment on the accuracy of abutted edges was ascertained by placing field edges at 0-cm and $+5$-cm off-axis. Hence, two insert pairs were fabricated (8.5x12 cm$^2$ paired with 3.5x12 cm$^2$, and 6x12 cm$^2$ paired with 6x12 cm$^2$). A single film was irradiated for each pair of inserts. The precision of the data measured at 0-cm and $+5$-cm off-axis was assumed to be equal to that at $+2.5$-cm off-axis.

2.3.2.3 Dosimetric Impact of Errors in Block Fabrication Test

The expected effect of errors in block cutting was validated by measuring the error resulting from purposely displaced field edges for central-axis abutment ($x_{int} = 0$). Four separate block pairs (6x12 cm$^2$ each) were fabricated with “errors” resulting in $\Delta x$ values of $\pm 2$ mm and $\pm 4$ mm by displacing each central-axis aperture edge by half the desired $\Delta x$. For example, $\Delta x$ of $+2$ mm was achieved by offsetting the central-axis edge of each insert by 1 mm toward the other field (Figure 21), resulting in hot spots. Conversely, offsetting the edges by 1 mm in the other direction resulted in $\Delta x$ of $-2$ mm and a cold spot.
2.4 Aim 4a: Evaluate Dose Homogeneity in Simple Targets

The potential for the variable-SCD method to improve dose homogeneity was assessed using the Philips Pinnacle³ treatment planning system by studying simple 1D segmented-field plans for irradiation of four hypothetical planning target volumes (PTVs). Each PTV was defined by a single slice (0.25-cm slice thickness) region of interest (ROI) in a virtual polystyrene phantom (50 x 50 x 50 cm³) to evaluate the method in the absence of common patient features (irregular surface, bone, air cavities, etc). A second single-slice ROI called "Outside Target" was defined for each plan to aid in analysis of dose outside the PTV. In all cases, the "Outside Target" ROI extended from the surface to 7 cm in depth (maximum depth of PTVs ranged from 4 to 5.5 cm) and ±8 cm off-axis (all PTVs spread from -5 to 5 cm) with the PTV excluded.
Two trials were created for each hypothetical target, the first using the variable-SCD method and the second using the standard method. For comparison, all planning parameters (beam energy, field shapes and beam weight) were developed for the variable-SCD trial and copied exactly to the standard trial. The ability to deliver each dose distribution calculated by Pinnacle³ using the variable-SCD method was evaluated by delivering the plan to the film phantom and measuring the dose distribution in the central xz plane. The measurement and data analysis methods used in this section were identical to that used for verification of electron beam commissioning described in aim 2.

2.4.1 Simple Targets

2.4.1.1 Two-step Block Target

The two-step block target represented in Figure 22 was chosen to assess the potential of the variable-SCD method to reduce dose heterogeneity when treating with two beam energies abutted near central-axis. This target might approximate a typical volume targeted for postmastectomy irradiation, i.e. 4-cm depth simulating the internal mammary chain lymph nodes and 2.5-cm depth simulating the chest wall thickness.

Figure 22: Illustration of the two-step block target. The dashed line represents the phantom surface.
2.4.1.2 Wedge Target

The wedge target represented in Figure 23 was chosen to assess the potential of the variable-SCD method to reduce dose heterogeneity when treating with three adjacent beam energies abutted off-axis.

![Wedge Target Diagram](image)

Figure 23: Illustration of the wedge target. The dashed line represents the phantom surface.

2.4.1.3 Pentagon Target

The pentagon target represented in Figure 24 was chosen to assess the potential of the variable-SCD method to reduce dose heterogeneity when treating with three beams of different energy abutted off-axis while conforming the 90% dose contour to a sharp gradient in target depth.

2.4.1.4 Inverted Well Target

The inverted well target represented in Figure 25 was chosen to assess the potential of the variable-SCD method to reduce dose heterogeneity when treating with three fields (two beam energies) abutted off-axis while minimizing dose in the central region beyond the distal target surface. This target might approximate a PTV around the
spinal cord, e.g. for treatment of the paraspinal muscles for mesenchymal chondrosarcoma.

Figure 24: Illustration of the pentagon target. The dashed line represents the phantom surface.

Figure 25: Illustration of the inverted well target. The dashed line represents the phantom surface.
2.4.2 Treatment Planning

The 1D energy segmentation plans for treatment of the simple PTVs using the variable-SCD method were designed with the following goals:

1. Conform the 90% isodose contour to the PTV.
2. Minimize dose heterogeneity in the target, i.e. maximum dose < 105% and minimum dose > 85%

To accomplish the first goal, beam energies required to treat a particular portion of the target volume were selected based on the $R_{90}$ depths of a 10x10-cm$^2$ field size. Blocks were then created for each beam in order of highest energy to lowest. To account for the fact that the beam edge locates the 50% dose point at the $R_{100}$ depth, each field edge is extended beyond its target volume by a lateral margin of at least $0.5*P_{90-10} @ R_{100}$, with higher energies taking territorial precedence over lower energies.

After defining all beam edges and calculating dose for each beam, the following beam weight procedures were followed with the goal of minimizing dose heterogeneity:

1. A single beam is weighted 100%, i.e. all other beams are weighted to 0%.
2. A normalization point is chosen for that beam such that the maximum dose is 100%.
3. Monitor units are assigned to the beam so that dose to the normalization point is approximately 50 cGy.
4. Steps 1 to 3 are repeated until a normalization point is assigned to each beam.
5. All beams are weighted by the monitor units determined in steps 1-4 giving a composite dose distribution.
6. The absolute dose at each normalization point from step 2 is checked. If the dose to all normalization points is greater than 51 cGy (102%), monitor units are reduced for all beams uniformly (scaled) such that the dose to at least one normalization point is approximately 50 cGy.

7. The composite dose distribution is normalized to 100% at a selected normalization point with an absolute dose of approximately 50 cGy.

8. Monitor units are adjusted for an individual beam if (1) dose to the target volume is >105% and the modification does not result in diminished target coverage or (2) target coverage can be improved without violating the heterogeneity goals.

2.4.3 Plan Evaluation and Comparisons

Dose distributions in the PTV and the "outside PTV" ROIs for the standard applicator and variable-SCD applicator plans were compared quantitatively based on the maximum and minimum dose, dose spread (maximum - minimum), mean dose, and σ (standard deviation) of the dose distribution. Additionally, $D_{90-10}$ was used to compare dose homogeneity in the PTVs. $D_{90-10}$ is defined as the difference between the dose received by 90% of the PTV volume and the dose received by 10% of the PTV volume, which was determined using cumulative dose volume histograms (DVHs).

2.5 Aim 4B: Evaluate Dose Homogeneity in Patient PTV

The potential for the variable-SCD method to improve dose homogeneity in a heterogeneous target was assessed by studying a single patient case. A plan was designed such that the 90% isodose surface covers the PTV with dose homogeneity comparable to treatment with a single square field.
2.5.1 Patient Treatment Planning

2.5.1.1 Treatment Planning Objectives

The goals of the patient treatment plan developed using the variable-SCD method were as follows:

1. Conform the 90% isodose contour to the PTV.
2. Minimize dose to all critical structures.
3. Minimize dose heterogeneity in the target, especially in the abutment region of adjacent beams.

2.5.1.2 Segmented-field ECT Treatment Planning

Prior to beginning the segmented-field ECT treatment planning methods outlined in this section, initial setup parameters common to all fields (i.e. gantry angle, couch position, and SSD) were selected. These parameters are ideally specified such that (1) the patient or bolus surface is located at isocenter (100-cm SSD), (2) beam central-axis intersects approximately the center of the PTV, and (3) the distance from the surface to the maximum PTV depth along the direction of beam central-axis is minimized.

To create an acceptable treatment plan using segmented-field electron conformal therapy, it was necessary to determine the minimum beam energy needed to treat the maximum depth of each portion of the PTV. Segmented fields were then defined with appropriate margins to insure good coverage. A field segmentation and energy modulation method developed by Rogers (2005) was modified for the present aim.

- PTV Segmentation

PTV segmentation is done to aid the planner in choosing the appropriate beam energies to treat subsets of the PTV. PTV segmentation was based on the distal $R_{90}$
values of fields having side scatter equilibrium for the electron beam energies available 
with the 2100EX 6/18 radiotherapy accelerator. Table 3 lists $R_{90}$ values for a 10x10-cm$^2$ 
field size for each beam energy.

Table 3: Summary of beam parameters used for treatment planning. All 
values were obtained using TPS data for a 10x10-cm$^2$ field size at 100-cm 
SSD with modified air gaps (5, 7.5, 11.5, 17.5, and 19.5 cm).

<table>
<thead>
<tr>
<th>Treatment planning beam data reference for 10x10-cm$^2$ field</th>
</tr>
</thead>
<tbody>
<tr>
<td>Distance (cm)</td>
</tr>
<tr>
<td>$R_{100}$</td>
</tr>
<tr>
<td>$R_{90}$</td>
</tr>
<tr>
<td>$(P_{90-10})/2$ at $R_{100}$</td>
</tr>
</tbody>
</table>

The following steps outline the PTV segmentation procedure used to create volume 
contours (notated as the energy followed by "V").

1. The skin surface (or bolus surface if present) is contoured for all slices 
containing the PTV. The resulting contour is defined "Surface" (Figure 26b).
2. The surface contour is contracted for each beam energy to be used by its 
corresponding $R_{90}$ value (Table 3). The resulting contours are defined 20T, 16T, 
12T, 9T, and 6T ("T" for temporary) (cf. Figure 26c).
3. Contour 6V is created as a copy of the PTV with the 6T volume excluded.
4. Contour 9V is created as a copy of the PTV with the 6V and 9T volumes 
excluded.
5. Contour 12V is created as a copy of the PTV with the 6V, 9V, and 12T 
volumes excluded.
6. Contour 16V is created as a copy of the PTV with the 6V, 9V, 12V and 16T volumes excluded.

7. Contour 20V is created as a copy of the PTV with the 6V, 9V, 12V, 16V and 20T volumes excluded.

(Note: Steps 3-7 are accomplished in Pinnacle\(^3\) by expanding the PTV contour by 0-cm with the contours to be excluded set as limiting structures.)

The resulting 6V, 9V, 12V, 16V, and 20V contours (Figure 26d) contain the PTV volume that can be treated by that energy. These contours can be displayed one at a time in the beams-eye view (BEV) window.

![Figure 26: Single transverse slice illustrating the PTV segmentation method for a patient case planned using 1 cm of bolus. The white dotted line indicates beam direction. Note: All contours are shown in color wash except the PTV which is shown in outline. (a) Lavender PTV contour, (b) orange surface contour, (c) contours 6T (green), 9T (blue), 12T (red), and 16T (yellow), (d) contours 6V (green), 9V (blue), 12V (red), and 16V (yellow).](image-url)
• **BEV Depth Map**

The contours created by PTV segmentation were used to create additional contours that can be viewed simultaneously in the BEV window as a depth map (Figure 27). The following steps detail how this was done:

1. Expand the highest energy "V" contour that contains the PTV by $R_{90} + 0.2$ cm toward the skin and parallel to the beam central-axis with no limiting structures. If the beam direction is not parallel to one of the orthogonal axes, vector components can be used to specify the direction of expansion. The resulting contour should extend beyond the skin surface slightly and is defined "E MAP" where E is the beam energy.

2. Expand the next lowest energy "V" contour by $R_{90} + 0.2$ cm toward the skin and parallel to the beam central-axis with the previous energy MAP contour as a limiting structure. The resulting contour should extend beyond the skin surface slightly but should not extend beyond the previous energy's MAP contour.

3. Repeat step 2 for the remaining energies, each time using all previously created MAP contours as limiting structures.

• **Segmented-field Energy Modulation**

The volume contours described in the PTV segmentation portion of this aim are used to create energy-segmented fields. It is also possible to manually draw the desired fields using the BEV depth map.

To aid in applying appropriate beam margins it is useful to create a "PTV plus margin" contour (PTVPM). The PTVPM (Figure 28) extends the PTV in all directions.
perpendicular to the beam by the desired margin (usually 1 - 1.5 cm) so that the planner can allow for dose falloff around the perimeter of the target volume (Rogers 2005).

Figure 27: Example of BEV depth map. (a) Transverse view of contours 6 MAP (green), 9 MAP (blue), 12 MAP (red), and 16 MAP (yellow). (b) MAP contours from the beams-eye view.

Figure 28: Beams eye view of PTVPM (green wire frame) generated by expanding the PTV (solid blue) by 1 cm in all directions.
The following steps detail the method used to create segmented energy fields using the volume contours:

1. The first beam is added to the beam list and given the desired setup geometry, i.e. the planner selects the beam direction and the applicator to be used based on their best judgment. This beam is set to the highest energy to be used.

2. A block is added in the beam modifier list and the blocking mode is set to "expose" the appropriate "V" contour with a minimum margin of \((P_{90-10})/2\) at \(R_{100}\) for that beam energy (cf. Table 3). The block mode is then changed from auto-block to manual allowing the planner to modify the block shape. At this point the block may be expanded to a) include small patches that may be nearby but separate from the larger exposed portions or b) expand the block to expose the PTVPM if the distance from the block edge to the PTVPM edge is on the order of 1 cm (this eliminates unnecessary decreases in energy in these areas, allowing the natural dose falloff to follow the tapering in depth of the PTV).

3. The next highest energy beam is created using the TPS "copy beam" function, which generates an identical copy of the previous beam with its blocks. The block used to expose the target for the previous beam is now set to "block" (this ensures field edges coincide) and the energy is reduced to the next highest energy.

4. Steps 2 and 3 are repeated until all beams have been defined.

Often blocks are created with two or more separated apertures or with block regions that are not connected to the outer perimeter of the block. In these cases the TPS will not calculate dose. Procedures for dealing with these circumstances are described below:
1) If a beam has two or more separate exposure regions that are 2x2 cm$^2$ or larger, the beam is copied and each exposure region is assigned to a separate beam.

2) If a separate exposure region is smaller than 2x2 cm$^2$, it is connected to the nearest exposure region using the brush function in the BEV. When this method is used, the connector region is added when the beam is initially created. This allows the "beam copy" segmentation method to account for the connector region, avoiding overlaps.

3) If one or more block regions are not connected with the perimeter of the block, the beam is copied and the block is divided between the two beams such that the common edge intersects the disconnected area(s). This is done after the "beam copy" segmentation method is complete.

**Segmented-field Modification**

When applied to a patient case, the methods described above can still result in poor target coverage. This is most commonly due to the presence of a curved or otherwise irregular patient surface. In the presence of a sloped surface dose tends to extend beyond the beam edge due to disproportionate scatter into the region. To account for this, beam edges along the perimeter of the target are adjusted to reduce the field size. The presence of thick regions of bone in the target can also result in poor coverage as decreased penetration in bone has not been taken into account. In these cases, beam energy is increased if possible.
2.5.2 Patient Case

A segmented-field treatment plan was developed for a patient case to compare the standard and variable-SCD applicator methods. The case was chosen with assistance from a radiation oncologist to reflect a typical PTV that could be treated with electron conformal therapy, e.g. custom bolus or segmented-field electron therapy.

The patient selected for this study had recurrent squamous cell carcinoma of the left ear and had undergone a total auriculectomy with parotidectomy and upper neck dissection before receiving post-operative radiation therapy. A dose of 50 Gy was delivered in 25 fractions to the ear area and neck using 4 MV photons, and to the left scalp using 9 MeV electrons. The radiation oncologist contoured the PTV covering 62 3-mm CT slices. The PTV extends from the skin surface to a minimum depth of 1 cm and a maximum depth of 3.2 cm. The left lacrimal gland, left eye, left optic nerve and the spinal cord were also contoured by the radiation oncologist as dose limiting structures.

Because the PTV extended to the skin surface, a 1-cm bolus contour was drawn on the patient surface. Using the "density override" function, the contoured volume was given a density of 1 g cm$^{-3}$. The procedures described in the previous section were carried out with the bolus surface at isocenter, the gantry at 90°, and the couch at 180°, resulting in the volume contours and depth map in Figures 26 and 27 respectively. To reduce plan complexity (i.e. the number of fields used), the 6V contour was combined with the 9V contour. To reduce the dose to the eye and lacrimal gland, the gantry angle was increased to 110° (Figure 29) and a 1-cm lead eye shield was contoured over the eye in 7 transverse slices (2.1 cm) and extended over 5 slices (1.5 cm) in both the superior and inferior directions.
The 9, 12, 16, and 20 MeV beams in Figure 30 were defined using the segmented-field energy modulation procedures described in the previous section. A beam weight of 19.87% (212 MU) was assigned to the 9 MeV beam and the two 12 MeV beams, 20.71% (221 MU) to the 16 MeV beam, and 19.68% (210 MU) to the 20 MeV beam.

Figure 29: Room's-eye view of treatment setup with the gantry at 110°, the table at 180°, and the SSD at 100 cm.

Figure 30: Sagittal view of the 9, 12, 16, and 20 MeV beams projected at a depth of 2 cm in the patient.
3.1 Aim 1: Design of Methods for Matching Penumbra

The air gaps to be used in the implementation of the variable-SCD method were selected to reduce dose heterogeneity for any combination of beam energies based on inspection of penumbra width calculations (in water) as a function of air gap. Figures 31, 32, and 33 plot the results of these calculations for all energies at 1.5 cm, 2 cm, and 2.5-cm depth, respectively. The points connected by the dotted lines represent the air gaps selected for each energy. The air gaps were selected from the available insert tray positions (7.5-19.5 cm in 2 cm increments) so that $P_{80-20}$ penumbra widths were approximately equal for all beam energies at a depth of 1.5 cm in water for 100-cm SSD.

Figure 31: $P_{80-20}$ calculated in water as a function of air gap at 1.5-cm depth.
Figure 32: $P_{80-20}$ calculated in water as a function of air gap at 2-cm depth.

Figure 33: $P_{80-20}$ calculated in water as a function of air gap at 2.5-cm depth. The 6 MeV penumbra width is not connected as central-axis depth dose is less than 50% at this depth.
The lowest energy, 6 MeV, has the broadest penumbra and was therefore assigned the standard 5-cm air gap. The 20 MeV beam has the sharpest penumbra and was assigned the largest air gap available using the variable-SCD applicator (19.5 cm). To approximately match the 6, 9, and 12 MeV at 1.5 cm, air gaps of 7.5 cm and 11.5 cm were assigned to 9 MeV and 12 MeV respectively. A 17.5-cm air gap was assigned to 16 MeV so that the 12, 16, and 20 MeV penumbras are approximately matched at 2 and 2.5 cm while maintaining decent abutment dosimetry between 9 MeV and 16 MeV. Comparing penumbra width plotted as a function of depth using the standard air gap (Figure 34) and the selected variable-SCD air gaps (cf. Figure 35), one sees that the latter achieves improved penumbra agreement, particularly from 1 to 2-cm depth.

Figure 34: Plot of theoretical $P_{80-20}$ (calculated in water) as a function of depth from 0 cm to ~R$_{80}$ (maximum therapeutic range) for all beam energies using the standard 5-cm air gap.
3.2 Aim 2: Verification of Electron Beam Commissioning in TPS

The ability of the commissioned electron pencil beam algorithm to accurately predict dose to a phantom was assessed by comparing calculated and measured 2D dose distributions for 4x4-cm² and 15x15-cm² fields at 100-cm SSD. Previously commissioned beam data were verified by doing the comparison for the standard applicator. Commissioned beam data with modified parameters (energy-specific air gaps and modified $\sigma_{\theta_i}$) for the variable-SCD applicator were verified similarly.

Comparisons of central-axis depth dose and off-axis profiles were performed using central xz-plane isodose plots normalized to 100% of the central-axis maximum dose. Because the variable-SCD method requires an air gap of 5 cm be used for the 6 MeV beam, dose distribution comparisons were performed for 6 MeV using the standard 15x15-cm² applicator. Because segmented-field ECT will often necessitate the use of
small fields and rarely the use of large open fields, measured and calculated dose
distributions for 6, 9, 12, 16, and 20 MeV are presented in the text for a 4x4-cm$^2$ field
size. The results of comparisons for 15x15-cm$^2$ fields are presented in Appendix B.

3.2.1 6 MeV

Figure 36 shows the comparison of measured and calculated dose distributions for
a 6-MeV, 4x4-cm$^2$ field using the standard air gap. Good agreement (< 1-mm distance to
agreement (DTA)) is seen in field size (defined by the 50% isodose contour) as well as
central-axis depth dose for depths greater than 0.5 cm. In the region of therapeutic dose,
the 80% isodose contours agree within 1-mm DTA, while the measured data for dose
points greater than 90% are less than calculated dose by as much as 5% (1.5-cm depth
and -1-cm off-axis). For dose less than 5%, the calculated dose underestimates the
measured dose by as much as 5% between 1 to 2-cm depth, and overestimates the
measured dose by as much a 5% for depth greater than 2 cm (i.e. corners of the dose
falloff region). Results for the 15x15-cm$^2$ field size (cf. Figure B1) are similar to the
4x4-cm$^2$ field size results.

![Figure 36: 6 MeV central xz plane measured (solid line) and calculated
dashed line) isodose plots for a 4x4-cm$^2$ field using the standard
applicator (5-cm air gap).](image)

54
3.2.2 9 MeV

Figure 37 shows isodose plots of measured and calculated dose distributions overlaid for a 9 MeV, 4x4-cm² field using the standard applicator (5-cm air gap) and the variable-SCD applicator (7.5-cm air gap). In both cases, agreement is within 1-mm DTA on central-axis depth dose, as well as for off-axis dose > 30%.

The influence of the increased air gap in the penumbra region is evidenced by the distance from the measured 90% to 10% isodose lines at 1.5-cm depth increasing from 1.3 cm for the standard applicator dose distribution to 1.6 cm for the variable-SCD applicator dose distribution.

Results for the 15x15-cm² field size (cf. Figure B2 and B3) are similar to the 4x4-cm² field size results. However, a notable difference for both applicators is seen between ±5-cm and ±7.5-cm off-axis from 1.5-cm to 2-cm depth, where calculated dose overestimates measured dose by as much as 5%. This is likely due to sub-optimal off-axis ratios being used for calculation.

3.2.3 12 MeV

Figure 38 shows isodose plots of measured and calculated dose distributions overlaid for a 12 MeV, 4x4-cm² field using the standard applicator (5-cm air gap) and the variable-SCD applicator (11.5-cm air gap). In both cases, agreement is within 1-mm on central-axis depth dose beyond 3.5 cm, as well as off-axis dose for dose ≥ 30%. From 1 cm to 3.5 cm in depth, and ±2 cm off-axis (a low dose gradient region), agreement in percent dose is within 2%.
Figure 37: 9 MeV central xz plane measured (solid line) and calculated (dashed line) isodose plots for a 4x4-cm² field using (a) the standard applicator (5-cm air gap) and (b) the variable-SCD applicator (7.5-cm air gap). (figure continued)
Figure 38: 12 MeV central xz plane measured (solid line) and calculated (dashed line) isodose plots for a 4x4-cm$^2$ field using (a) the standard applicator (5-cm air gap) and (b) the variable-SCD applicator (11.5-cm air gap).
The influence of the increased air gap is evidenced by the distance from the measured 90% to 10% isodose lines at 2-cm depth increasing from 1.2 cm for the standard applicator to 1.7 cm for the variable-SCD applicator.

Results for the 15x15-cm² field size (cf. Figure B4 and B5) are similar to the 4x4-cm² field size results. However, a notable difference for the standard applicators is seen between ±5-cm and ±7.5-cm off-axis from 1.5-cm to 3.5-cm depth, where calculated dose overestimates measured dose by as much as 5%. Interestingly, this discrepancy is not seen in the variable-SCD applicator results for the 15x15-cm² field size.

3.2.4 16 MeV

Figure 39 shows isodose plots of measured and calculated dose distributions overlaid for a 16 MeV, 4x4-cm² field using the standard applicator (5-cm air gap) and the variable-SCD applicator (17.5-cm air gap). Agreement is within 1-mm DTA on central-axis depth dose in the standard applicator distribution for depths from 3.5 cm to 8 cm. Similar agreement is seen in the variable-SCD applicator distribution, with the exception of 3.5 to 5 cm in depth, where DTA is as much as 2 mm. In both cases, from 1.5 cm to 3.5 cm in depth, and within ±2 cm off-axis (a low dose gradient region), agreement is within 2% and agreement in lateral dose falloff for doses ≥ 30% is within 1 mm.

The large disparity seen in the 5% isodose contour of the standard applicator distribution is likely due to low energy electrons generated in the block by photons in the beam. As a result of the difference in distance between the Cerrobend block and the phantom surface for standard and variable-SCD applicators, low energy electrons escaping the distal surface of the block have a less pronounced effect in the latter case.
Figure 39: 16 MeV central xz plane measured (solid line) and calculated (dashed line) isodose plots for a 4x4-cm$^2$ field using (a) the standard applicator (5-cm air gap) and (b) the variable-SCD applicator (17.5-cm air gap).
The influence of the 12.5 cm increase in air gap is evidenced by the distance from the measured 90% to 10% isodose lines at 2-cm depth increasing from 1.1 cm for the standard applicator distribution to 1.8 cm for the variable-SCD applicator distribution.

Results for the 15x15-cm\(^2\) field size (cf. Figure B6 and B7) are generally similar to the 4x4-cm\(^2\) field size results. As in the case of the 12 MeV beam results, a notable difference for the standard applicators is seen between ±5-cm and ±7.5-cm off-axis from 1.5-cm to 3.5-cm depth, where calculated dose overestimates measured dose by as much as 5%. Again, this is likely due to sub-optimal off-axis ratios being used for calculation. Additionally, for the variable-SCD applicator results for the 15x15-cm\(^2\) field size, a significant difference is seen in depth dose from 1-cm to 2-cm, where calculated dose overestimated measured dose by as much as 5%. This is likely due to less electrons scatter from the collimator reaching the phantom surface as a result of the increased air gap.

### 3.2.5 20 MeV

Figure 40 shows isodose plots of measured and calculated dose distributions overlaid for a 20 MeV, 4x4-cm\(^2\) field using the standard applicator (5-cm air gap) and the variable-SCD applicator (19.5-cm air gap). In both cases, central-axis depth dose DTA is within 2 mm from 5.5 cm to 8 cm depth, but increases to as much as 3.5 mm from 8 cm to 10 cm depth. In contrast, good agreement in lateral dose falloff from 1.5 cm to 4 cm depth for doses between 30% and 80% indicates good field size agreement. The source of the CAX depth dose discrepancy is not clear, but it should be noted that measured distributions for larger field sizes do not exhibit this incongruity (cf. Figures B8 and B9).
Figure 40: 20 MeV central xz plane measured (solid line) and calculated (dashed line) isodose plots for a 4x4-cm$^2$ field using (a) the standard applicator (5-cm air gap) and (b) the variable-SCD applicator (19.5-cm air gap).
As in the case of the 16 MeV distributions, the large disparity seen in the 5% isodose contour of the standard applicator distribution is likely due to low energy electrons generated in the block by photons in the beam. In this case, however, the effect is more pronounced as more electrons with sufficient energy to escape the block are generated using the 20 MeV beam.

The influence of the 14.5 cm increase in air gap is evidenced by the increase in off-axis distance between the measured 90% and 10% isodose lines from 1.1 cm for the standard applicator to 1.6 cm for the variable-SCD applicator at 2-cm depth.

Results for the 15x15-cm$^2$ field size (cf. Figure B8 and B9) are similar to the 4x4-cm$^2$ field size results with the exception of the differences in depth dose noted above. Also, as in the case of the 16 MeV beam results, a notable difference for the standard applicators is seen between ±5-cm and ±7.5-cm off-axis from 1.5-cm to 3-cm depth, where calculated dose overestimates measured dose by as much as 5%.

3.3 Aim 3: Construction of Cerrobend® Blocks with Diverging Edges

The accuracy of the beam collimation method (i.e. construction of Cerrobend inserts using the Compu●cutter® system and insert placement) was evaluated for the variable-SCD applicator by comparing off-axis dose profiles for a single square field to that of two equally-weighted, abutted fields for each of the three tests in this section. Results for a 12 MeV beam (SCD = 78.5 cm) are shown.

3.3.1 Accuracy and Precision of Off-axis Abutment Test

Figures 41 (a-c) compare the off-axis profile at 1-cm depth of a single 12 MeV field (12x12-cm$^2$) with three abutment trials consisting of two 12 MeV fields (3.5x12-cm$^2$ and 8.5x12-cm$^2$) abutted 2.5 cm off-axis. In trial 1 (Figure 41a) a dose deficiency of
Figure 41: 2.5-cm off-axis abutment tests comparing measured off-axis profiles at 1-cm depth for two 12 MeV fields (3.5x12 cm$^2$ and 8.5x12 cm$^2$) abutted at 2.5 cm off-axis (solid line) and a single 12 MeV field (12x12-cm$^2$) (dashed line). The arrows indicate the abutment region. (a) Trial 1, (b) Trial 2, (c) Trial 3.
-1.46% is seen at 2.5-cm off-axis corresponding to $\Delta x = -0.17$ mm. In trial 2 (Figure 41b) a dose deficiency of -1.8% is seen at 2.5-cm off-axis, corresponding to $\Delta x = -0.21$ mm. In trial 3 (Figure 41c) a dose deficiency of -5.5% at 2.5-cm off-axis is seen corresponding to $\Delta x = -0.65$ mm.

The results of the three trials, summarized in table 4, show that there was a sample mean dose error ($\overline{\Delta D}$) of -2.9 ± 1.3% and a sample standard deviation ($\sigma_{\Delta D}$) of 2.2%. Based on calculated $\Delta D/\Delta x$ of 8.5%/mm, this $\overline{\Delta D}$ implies an accuracy (systematic error) in edge fabrication and alignment of $\overline{\Delta x} = -0.30 \pm 0.15$ mm. Additionally, $\sigma_{\Delta D} = 2.2\%$ implies a precision (random error) in fabrication and alignment of $\sigma_{\Delta x} = 0.27$ mm.

Table 4: Summary of results of 2.5-cm off-axis abutment test.

<table>
<thead>
<tr>
<th># meas</th>
<th>$\overline{\Delta D} \pm \sigma_{\overline{\Delta D}}$ (%)</th>
<th>$\sigma_{\Delta D}$ (%)</th>
<th>$\overline{\Delta x} \pm \sigma_{\overline{\Delta x}}$ (mm)</th>
<th>$\sigma_{\Delta x}$ (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
<td>-2.9 ± 1.3</td>
<td>2.2</td>
<td>-0.30 ± 0.15</td>
<td>0.27</td>
</tr>
</tbody>
</table>

3.3.2 Zero and 5-cm Off-axis Abutment Test

The abutment tests made at $x_{\text{int}} = +2.5$ cm were repeated using a single film measurement for $x_{\text{int}} = 0$ cm (central-axis) and for $x_{\text{int}} = 5$ cm. The measured values for $\Delta D$ were +1.2 and -4.2 respectively, which gave values for $\Delta x$ of 0.14 mm and -0.49 mm. Hence, abutment accuracy at 0, +2.5, and +5.0 cm off-axis were $+0.14 \pm 0.27$ mm, $-0.34 \pm 0.15$ mm, and $-0.49 \pm 0.27$ mm, respectively. These results indicate a slight correlation of $\Delta x$ with off-axis location ($\Delta x \approx -0.01 \times \text{off-axis location}$).

3.3.3 Test of Dosimetric Impact of Errors in Block Fabrication

The impact of error in block edge fabrication and alignment is illustrated in Figure 42, which plots off-axis profiles measured for the intentional block abutment
Figure 42: Dosimetric impact of errors in block fabrication. Two 12 MeV beams (6x12 cm\(^2\) and 6x12 cm\(^2\)) abutted on central-axis \((x_{int} = 0)\) with errors introduced in edge placement (solid line) are compared with a single 12x12-cm\(^2\) field (dashed line). (a) \(\Delta x = -4\) mm resulting in \(\Delta D\) of -35.8\%, (b) \(\Delta x = -2\) mm resulting in \(\Delta D\) of -18.3\%, (c) \(\Delta x = 0\) mm resulting in \(\Delta D\) of +1.2\%, (d) \(\Delta x = +2\) mm resulting in \(\Delta D\) of +14.6\%, and (e) \(\Delta x = +4\) mm resulting in \(\Delta D\) of +34.1\%. 
errors of $\Delta x = -4, -2, 0, +1, \text{ and } +4 \text{ mm}$. Resulting values of $\Delta D$ were -35.8%, -18.3%, +1.2%, +14.6%, and +34.1%, respectively.

Assuming an uncertainty of 2.2% (determined in previous data), Figure 43 plots $\Delta D$ versus $\Delta x$. These data are consistent ($\chi^2 = 0.64$ and $p = 0.6$) with the theoretical line, which corresponds to 8.5%/mm, determined for the 12 MeV beam.

Based on these results, random error in block edge fabrication and alignment ($\sigma_{\Delta x} = 0.27 \text{ mm}$) should be less than 0.54 mm (2$\sigma$). This results in a 95% confidence interval for $\Delta D$ of 4.6% (0.54 mm x 8.5%/mm).

![Figure 43: Measured dosimetric impact of errors in beam edge placement plotted with the calculated line with 8.5%/mm slope. Results are plotted as measured $\Delta D$ vs. intended $\Delta x$. Error bars represent $\sigma_{\Delta D} = 2.2\%$ determined in the 2.5-cm off-axis abutment test.](image)

### 3.4 Aim 4a: Evaluate Dose Homogeneity in Simple Targets

This section presents the results of studying the utilization of the variable-SCD applicator for segmented-field conformal therapy for four idealized PTVs. The first set
of results for each PTV compares a treatment plan using the variable-SCD applicator to that using the standard applicator. The second set of results compares dose measurement of the variable-SCD plan delivered to a phantom to the calculated dose distribution for each PTV.

3.4.1 Treatment Planning Results

3.4.1.1 Two-step Block Target

This target was divided into two regions, region 1 (negative off-axis position) having a distal target depth of 2.5 cm and region 2 (positive off-axis position) having a distal target depth of 4 cm. A 9 MeV beam ($R_{90} = 2.6$ cm) was selected to irradiate region 1 and a 16 MeV beam ($R_{90} = 5.1$ cm) to irradiate region 2. Beam edges were defined to account for lateral falloff by placing the edges of the 16 MeV beam 1 cm beyond the inner boundary of region 2 and 1.3 cm beyond the outer boundary. The 9 MeV beam was then defined with its inner edge at the same position as the 16 MeV beam's inner edge and the outer edge 1.3 cm beyond the outer boundary of region 1. The 16 MeV beam normalization point (+1.6 cm off-axis, 3 cm depth) was chosen as the reference dose point for the composite distribution, allowing an increase from 48 to 49 MU for the 9 MeV beam to pull the 90% isodose surface toward the surface, resulting in better PTV coverage. A beam weight of 49 MU (49.5%) was assigned to the 9 MeV beam and a weight of 50 MU (50.5%) to the 16 MeV beam.

Treatment planning dose distributions using the standard and variable-SCD applicators are presented as isodose plots in Figures 44a and 44b, respectively. The DVHs for both trials are overlaid in Figure 45, and the DVH statistics are summarized in Table 5.
Figure 44: Isodose plots of TPS calculated dose distributions for irradiation of the two-step block target in a polystyrene phantom for (a) the standard applicator plan and (b) the variable-SCD applicator plan.
Figure 45: Cumulative DVHs for the two-step block target showing PTV and Outside PTV ROIs. The standard trial is represented by dotted lines and the variable-SCD trial by solid lines.

Table 5: Summary of treatment planning results for irradiation of the two-step block target using the standard and variable-SCD methods.

<table>
<thead>
<tr>
<th>Trial</th>
<th>ROI</th>
<th>Mean Dose (%)</th>
<th>σ (%)</th>
<th>Min (%)</th>
<th>Max (%)</th>
<th>D_{90-10} (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standard</td>
<td>PTV</td>
<td>97.6</td>
<td>4.2</td>
<td>78.0</td>
<td>108.5</td>
<td>10.3</td>
</tr>
<tr>
<td>Variable-SCD</td>
<td>PTV</td>
<td>97.2</td>
<td>3.2</td>
<td>85.8</td>
<td>102.8</td>
<td>7.7</td>
</tr>
<tr>
<td>Standard</td>
<td>Outside PTV</td>
<td>44.0</td>
<td>33.9</td>
<td>3.8</td>
<td>108.3</td>
<td></td>
</tr>
<tr>
<td>Variable-SCD</td>
<td>Outside PTV</td>
<td>43.9</td>
<td>32.6</td>
<td>3.8</td>
<td>102.8</td>
<td></td>
</tr>
</tbody>
</table>

The variable-SCD applicator plan in Figure 44b provides a visibly more uniform dose distribution than the standard applicator plan in Figure 44a. This is reflected in the variable-SCD applicator plan's significantly steeper DVH, 24% reduction in σ, and 25% reduction in D_{90-10} for the PTV. The unmatched penumbra in the standard applicator plan results in a cold spot in the PTV as low as 78% and a hot spot as great as 109%. There
remains in the variable-SCD applicator plan (1) a small cold spot of 86% (0.7 cm depth, -0.8 cm off-axis) due to the 16 MeV penumbra being greater than the 9 MeV penumbra at depths less than 1.5 cm (cf. Figure 35), and (2) a small hot spot of 103% (2 cm depth, -3.6 cm off-axis) due to 49 rather than 48 MU having been assigned to the 9 MeV beam in order to contain the proximal target surface in the 90% isodose contour.

Due to the location of the abutment edge, the hot spot in the standard applicator plan that results from unmatched penumbra is located partially outside the PTV. Therefore, the penumbra matching in the variable-SCD plan results in a 5% reduction in dose spread outside the target and a 4% reduction in $\sigma$. As expected however, the difference in mean doses outside the target is less than 0.5%.

### 3.4.1.2 Wedge Target

This target was divided into three regions, region 1 (negative off-axis position) having a distal target depth range of 5.5 cm to 4.6 cm, region 2 (about central-axis) having a distal target depth range of 4.6 cm to 3.7 cm, and region 3 (positive off-axis position) having a distal target depth range of 3.7 cm to 2.7 cm. A 20 MeV beam ($R_{90} = 5.9$ cm) was selected to treat region 1, a 16 MeV beam ($R_{90} = 5.1$ cm) to treat region 2, and a 12 MeV beam ($R_{90} = 3.9$ cm) to treat region 3. The 20 MeV beam was defined with the inner edge at -1.5 cm and the outer edge placed 2.5 cm beyond the outer boundary of region 1 at -7.5 cm to obtain good target coverage. Beam edges for the 16 MeV beam (region 2) were defined at $\pm 1.5$ cm off-axis. The 12 MeV beam was then defined with the inner edge at $+1.5$ cm to match the 16 MeV field, and the outer edge placed 1.5 cm beyond the outer boundary of region 3. The 12 MeV beam normalization point ($+4$ cm off-axis, 3 cm depth) was chosen as the reference point for the composite
Figure 46: Isodose plots of TPS calculated dose distributions for irradiation of the wedge target in the polystyrene phantom for (a) the standard applicator plan and (b) the variable-SCD applicator plan.
Figure 47: Cumulative DVHs for the wedge block target showing PTV and Outside PTV ROIs. The standard trial is represented by dotted lines and the variable-SCD trial by solid lines.

Table 6: Summary of treatment planning results for irradiation of the wedge target using the standard and variable-SCD methods.

<table>
<thead>
<tr>
<th>Trial</th>
<th>ROI</th>
<th>Mean Dose (%)</th>
<th>σ (%)</th>
<th>Min (%)</th>
<th>Max (%)</th>
<th>D_{90-10} (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standard</td>
<td>PTV</td>
<td>97.6</td>
<td>2.8</td>
<td>88.9</td>
<td>104.2</td>
<td>7.4</td>
</tr>
<tr>
<td>Variable-SCD</td>
<td>PTV</td>
<td>97.6</td>
<td>2.4</td>
<td>89.2</td>
<td>102.6</td>
<td>6.1</td>
</tr>
<tr>
<td>Standard</td>
<td>Outside PTV</td>
<td>56.1</td>
<td>32.0</td>
<td>5.7</td>
<td>100.6</td>
<td></td>
</tr>
<tr>
<td>Variable-SCD</td>
<td>Outside PTV</td>
<td>56.2</td>
<td>31.4</td>
<td>5.7</td>
<td>101.2</td>
<td></td>
</tr>
</tbody>
</table>

dose distribution. A beam weight of 48 MU (34%) was assigned to the 20 MeV beam, 46 MU (32.6%) to the 16 MeV beam, and 47MU (33.3%) to the 12 MeV beam.

Treatment planning dose distributions using the standard and variable-SCD applicators are presented in Figures 46a and 46b respectively. The DVH for both trials are overlaid in Figure 47 and the DVH statistics are summarized in Table 6. Because the
abutted fields have adjacent energies, the variable-SCD plan in Figure 46b provides only a moderate improvement in dose homogeneity over the standard applicator plan in Figure 46a. The penumbra matching of the variable-SCD applicator provided a 1.5% reduction in maximum dose to the PTV as compared to the standard applicator plan, and less than a 1% increase in minimum dose. This is reflected in a 14% reduction in $\sigma$ and a 18% reduction in $D_{90-10}$ for the PTV DVH. Also, less than 2% reduction is seen in the $\sigma$ of the dose outside the PTV and less than 1% reduction in dose spread.

3.4.1.3 Pentagon Target

This target was divided into three regions, region 1 (negative off-axis position) and region 3 (positive off-axis position) having minimum distal target depths of 2.5 cm, and region 2 (about central-axis) having a maximum distal target depth of 5.5 cm. A 12 MeV beam ($R_{90} = 3.9$ cm) was selected to treat regions 1 and 2, and a 20 MeV beam ($R_{90} = 5.9$ cm) to treat region 2. Beam edges were defined for the center 20 MeV beam (region 2) at $\pm3.5$ cm off-axis. The 12 MeV beams for regions 1 and 3 were then defined with inner edges at $\pm3.5$ cm to match the 20 MeV field, and the outer edges placed 1 cm beyond the outer boundaries of the target at $\pm6$ cm off-axis. The central 20 MeV beam normalization point (0 cm off-axis, 2.2 cm depth) was chosen as the reference point for the composite dose distribution. A beam weight of 49 MU (49.5%) was assigned to the 20 MeV beam and 50 MU (50.5%) to the 12 MeV beams.

The two 12 MeV fields for this case were defined using a single block. Because Pinnacle$^3$ does not calculate dose for non-contiguous block definitions, a 3-mm wide "connector region" was included in the block definition (located well outside the transverse plan containing the PTV).
Treatment planning dose distributions using the standard and variable-SCD applicators are presented in Figure 48a and 48b, respectively. The DVHs for both trials are overlaid in Figure 49, and the DVH statistics are summarized in Table 7.

The variable-SCD applicator plan in Figure 48b provides a significantly more uniform dose distribution than the standard applicator plan in Figure 48a. For the PTV, a 39% reduction in $\sigma$, 34% reduction in $D_{90-10}$, and a 32% reduction in dose spread are seen, all of which are reflected in the sharper target DVH of the variable-SCD applicator plan.

Figure 48: Isodose plots of TPS calculated dose distributions for irradiation of the pentagon target in the polystyrene phantom for (a) the standard applicator plan and (b) the variable-SCD applicator plan. (figure continued)
Figure 49: Cumulative DVHs for the pentagon target showing PTV and Outside PTV ROIs. The standard trial is represented by dotted lines and the variable-SCD trial by solid lines.
Table 7: Summary of treatment planning results for irradiation of the pentagon target using the standard and variable-SCD methods.

<table>
<thead>
<tr>
<th>Trial</th>
<th>ROI</th>
<th>Mean Dose (%)</th>
<th>σ (%)</th>
<th>Min (%)</th>
<th>Max (%)</th>
<th>D&lt;sub&gt;90-10&lt;/sub&gt; (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standard</td>
<td>PTV</td>
<td>99.2</td>
<td>3.6</td>
<td>88.4</td>
<td>109.6</td>
<td>8.8</td>
</tr>
<tr>
<td>Variable-SCD</td>
<td>PTV</td>
<td>99.0</td>
<td>2.2</td>
<td>90.8</td>
<td>105.2</td>
<td>5.8</td>
</tr>
<tr>
<td>Standard</td>
<td>Outside PTV</td>
<td>45.7</td>
<td>32.9</td>
<td>4.4</td>
<td>107.4</td>
<td></td>
</tr>
<tr>
<td>Variable-SCD</td>
<td>Outside PTV</td>
<td>45.1</td>
<td>31.9</td>
<td>4.4</td>
<td>104.2</td>
<td></td>
</tr>
</tbody>
</table>

The unmatched penumbra in the standard applicator plan resulted in a hot spot as great as 110% in the PTV, which was reduced to 105% in the variable-SCD applicator plan. This limited reduction in magnitude is due to the 12 MeV beam penumbra becoming broader than that of the 20 MeV beam at depths greater than 1.5 cm (cf. Figure 35). As in the case of the two-step block target, the hot spot in the standard plan overlaps the PTV and Outside PTV ROIs resulting in a hot spot of 107% outside the target. Therefore, a 3% reduction in dose spread and σ outside the target is provided by the penumbra matching of the variable-SCD applicator. As expected, the reduction in mean dose outside the target is less than 2%.

3.4.1.4 Inverted Well Target

This target was divided into three regions, regions 1 and 3 having a maximum distal target depth of 5.5 cm and region 2 having a maximum distal target depth of 3.5 cm. A 20 MeV beam (R<sub>90</sub> = 5.9 cm) was selected to treat regions 1 and 3, and a 12 MeV beam (R<sub>90</sub> = 3.9 cm) to treat region 2. The 20 MeV beams (defined using two separate blocks) for regions 1 and 3 were defined with inner edges at ±2 cm respectively and the outer edges placed 2 cm beyond the outer boundaries of regions 1 and 3 at ±7 cm off-axis. Beam edges were defined for the center 12 MeV beam (region 2) at ±2 cm off-axis.
to match the 20 MeV beams. The central 12 MeV beam normalization point (0 cm off-axis, 2.7 cm depth) was chosen as the reference point for the composite dose distribution. Beam weights were assigned with 47 MU (33.8%) to each of the two 20 MeV beams and 45 MU (32.4%) to the 12 MeV beam.

Treatment planning dose distributions using the standard and variable-SCD applicators are presented in Figure 50a and 50b, respectively. The DVHs for both trials are overlaid in Figure 51 and the DVH statistics are summarized in Table 8.

The variable-SCD applicator plan in Figure 50b provided a visibly more uniform dose distribution than the standard applicator plan in Figure 50a. This is reflected in the variable-SCD applicator’s significantly steeper DVH and 35% reduction in σ, 43% reduction in D_{90-10}, and 31% reduction in dose spread. The unmatched penumbra in the standard applicator plan resulted in a cold spot in the PTV as low as 85% and a hot spot as great as 108%. There remained in the variable-SCD applicator plan (1) a small cold spot of 88% due to the 20 MeV penumbra being greater than the 12 MeV penumbra at depths less than 1.5 cm (cf. Figure 35), and (2) a small hot spot of 104% due to the 12 MeV beam penumbra being greater than that of the 20 MeV beam at depths greater than 1.5 cm.

The small 103% hot spot outside the target due to unmatched penumbra in the standard applicator plan was reduced to 100% in the variable-SCD applicator plan, and a slight increase in the minimum dose outside the target was seen. This resulted in the dose spread outside the target being reduced by 4% as well as a 6% reduction in σ. As expected, a negligible change was seen in mean dose outside the target.
Figure 50: Isodose plots of TPS calculated dose distributions for irradiation of the inverted well target in the polystyrene phantom for (a) the standard applicator plan and (b) the variable-SCD applicator plan.
Figure 51: Cumulative DVHs for the inverted well target showing PTV and Outside PTV ROIs. The standard trial is represented by dotted lines and the variable-SCD trial by solid lines.

Table 8: Summary of treatment planning results for irradiation of the inverted well target using the standard and variable-SCD methods.

<table>
<thead>
<tr>
<th>Inverted well target</th>
<th>Trial</th>
<th>ROI</th>
<th>Mean Dose (%)</th>
<th>σ (%)</th>
<th>Min (%)</th>
<th>Max (%)</th>
<th>D_{90-10} (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Standard</td>
<td>PTV</td>
<td>97.4</td>
<td>4.6</td>
<td>84.8</td>
<td>108.2</td>
<td>13.8</td>
</tr>
<tr>
<td></td>
<td>Variable-SCD</td>
<td>PTV</td>
<td>97.4</td>
<td>3.0</td>
<td>87.6</td>
<td>103.8</td>
<td>7.9</td>
</tr>
<tr>
<td></td>
<td>Standard</td>
<td>Outside PTV</td>
<td>61.4</td>
<td>26.4</td>
<td>8.6</td>
<td>103.0</td>
<td></td>
</tr>
<tr>
<td></td>
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<td>61.3</td>
<td>24.8</td>
<td>9.6</td>
<td>100.0</td>
<td></td>
</tr>
</tbody>
</table>

3.4.1.5 Summary of Treatment Planning Results

The results of treatment planning PTV dose statistics for the four simulated targets using the standard applicator and the variable-SCD applicator are summarized in Table 9.
Table 9: Summary of treatment planning results for dose to the four simulated PTVs using the standard and variable-SCD methods.

<table>
<thead>
<tr>
<th>Target</th>
<th>Trial</th>
<th>Mean Dose (%)</th>
<th>Min (%)</th>
<th>Max (%)</th>
<th>Max-Min (%)</th>
<th>$\sigma$ (%)</th>
<th>$D_{90-10}$ (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Std</td>
<td>97.6</td>
<td>78.0</td>
<td>108.5</td>
<td>30.5</td>
<td>4.2</td>
<td>10.3</td>
<td></td>
</tr>
<tr>
<td>Mod</td>
<td>97.2</td>
<td>85.8</td>
<td>102.8</td>
<td>17.0</td>
<td>3.2</td>
<td>7.7</td>
<td></td>
</tr>
<tr>
<td>Std</td>
<td>97.6</td>
<td>88.7</td>
<td>104.2</td>
<td>15.5</td>
<td>2.8</td>
<td>7.4</td>
<td></td>
</tr>
<tr>
<td>Mod</td>
<td>97.6</td>
<td>89.1</td>
<td>103.0</td>
<td>13.9</td>
<td>2.4</td>
<td>6.1</td>
<td></td>
</tr>
<tr>
<td>Std</td>
<td>99.2</td>
<td>88.4</td>
<td>109.6</td>
<td>21.2</td>
<td>3.6</td>
<td>8.8</td>
<td></td>
</tr>
<tr>
<td>Mod</td>
<td>99.0</td>
<td>90.8</td>
<td>105.2</td>
<td>14.4</td>
<td>2.2</td>
<td>5.8</td>
<td></td>
</tr>
<tr>
<td>Std</td>
<td>97.4</td>
<td>84.8</td>
<td>108.2</td>
<td>23.4</td>
<td>4.6</td>
<td>13.8</td>
<td></td>
</tr>
<tr>
<td>Mod</td>
<td>97.4</td>
<td>87.6</td>
<td>103.8</td>
<td>16.2</td>
<td>3.0</td>
<td>7.9</td>
<td></td>
</tr>
<tr>
<td>Average</td>
<td>↓ 0.2%</td>
<td>↑ 4%</td>
<td>↓ 4%</td>
<td>↓ 32%</td>
<td>↓ 29%</td>
<td>↓ 32%</td>
<td></td>
</tr>
</tbody>
</table>

The mean dose decreased by 0.2% on average, indicating the total dose in the PTV was not impacted significantly using the variable-SCD applicator. In all four cases, dose homogeneity in the PTV using the variable-SCD applicator was in agreement with the hypothesis, i.e. minimum PTV dose was $\geq 85\%$ and maximum dose was $\leq 105\%$.

Minimum dose was increased by 4% on average, and maximum dose was decreased by 4% on average. An average reduction of 32% in dose spread (maximum-minimum), 29% in the $\sigma$ of the PTV dose distribution, and 32% in $D_{90-10}$ indicate that treatment plans calculated using the variable-SCD applicator provide a substantial improvement in dose homogeneity over plans using the standard applicator.

3.4.2 Variable-SCD Plan Verification

The results of delivering the variable-SCD applicator plans, described in the previous section, are presented using isodose plots and off-axis profiles at 2 cm depth. The monitor units assigned to each beam in the treatment planning phase were delivered...
to the polystyrene film phantom, and the composite measured dose distribution for each plan was normalized to 100% at the reference point defined in the treatment plan. The measured dose distributions were compared with the TPS calculations to assess the ability to deliver the planned dose distributions.

3.4.2.1 Two-step Block Target

The isodose plots and off-axis profiles at a depth of 2 cm for measured and calculated dose distributions for irradiation of the two-step block target using the variable-SCD applicator plan are overlaid in Figure 52a and 52b, respectively. Depth dose curves at -3.4 cm and 2.6 cm off-axis are shown in Figure 53a and 53b, respectively.

Field size agreement is seen comparing measured and calculated lateral dose falloffs at 2 cm depth in Figure 52b. For depths greater than 2 cm, the 90% isodose region in Figure 52a agree within 2%, with the exception of a small region of 3% difference from -3 to -5.5 cm off-axis and 2 cm depth (cf. Figure 52b and 53a), where calculated dose overestimates measured dose. This is likely due to beam output for the 9 MeV beam using the variable-SCD applicator being less than the measured standard applicator output factors used in the treatment planning system. For depths less than 2 cm, the 90% isodose contours agree within 3%, with the exception of a region within ± 0.5 cm of the abutment edge, where the measured dose is seen to be more uniform than calculated. This resulted in the absence of the 86% cold spot predicted by the treatment plan.

Differences as much as 5% due to the pencil-beam algorithm's overestimation of dose less than 20% in the falloff regions outside of the 9 MeV beam edges are seen from 3 to 4 cm depth and -7.5 to -9 cm off-axis in Figure 52a. Similar discrepancies are seen
Figure 52: Measured (solid red) and TPS calculated (dashed black) dose data for irradiation of the two-step block target using the variable-SCD applicator plan. (a) Isodose plot of central xz plane dose distributions. Note: The solid circle indicates the point at which the measured and calculated dose distributions are normalized to 100%. (b) Off-axis dose profiles at 2-cm depth.
outside the 16 MeV beam edges from 6 to 8 cm depth from 8 to 10 cm off-axis from -3 to -5 cm off-axis. The influence of the latter discrepancies on depth dose in the region of the 9 MeV beam is evident in Figure 53a from 6 to 8 cm depth.

3.4.2.2 Wedge Target

The isodose plots and off-axis profiles at 2 cm depth for measured and calculated dose distributions for irradiation of the wedge target using the variable-SCD applicator plan are overlaid in Figure 54a and 54b, respectively. The central-axis depth dose curves are shown in Figure 55.

Agreement in lateral dose falloff in Figure 54b indicates field size agreement. Profile agreement at this depth is within 2% generally, with the exception of a region within ± 0.5 cm of the abutment edge of the 16 and 20 MeV beams, where calculation exceeds measurement by as much as 5%. In Figure 54a, the measured dose is generally 2% less than calculated in the region of the PTV covered by the 16 MeV beam. This
Figure 54: Measured (solid red) and TPS calculated (dashed black) dose data for irradiation of the wedge target using the variable-SCD applicator plan. (a) Isodose plot of central xz plane dose distributions. Note: The solid circle indicates the point at which the measured and calculated dose distributions are normalized to 100%. (b) Off-axis dose profiles at 2-cm depth.
effect may be due to 16 MeV beam output for this field size (3x12 cm^2) using the variable-SCD applicator being less than the standard applicator output factors used in the TPS calculation. More clinically significant, however, is the 6% measured dose deficiency (1 cm DTA) at the 90% isodose in the region covered by the 16 MeV beam (cf. Figure 55). This discrepancy is likely due to a combination of inaccurate beam output and the dose algorithm's overestimation of the low dose contribution of the 12 MeV beam between 4 cm and 6 cm depth (cf. Figure 38).

3.4.2.3 Pentagon Target

The isodose plots and off-axis profiles at 2 cm depth for measured and calculated dose distributions for irradiation of the pentagon target using the variable-SCD applicator plan are overlaid in Figure 56a and 56b, respectively. The central-axis depth dose curves are shown in Figure 57.
Figure 56: Measured (solid red) and TPS calculated (dashed black) dose data for irradiation of the pentagon target using the variable-SCD applicator plan. (a) Isodose plot of central xz plane dose distributions. Note: The solid circle indicates the point at which the measured and calculated dose distributions are normalized to 100%. (b) Off-axis dose profiles at 2-cm depth.
Figure 57: Measured (solid red) and TPS calculated (dashed black) central axis depth dose plots for irradiation of the pentagon target.

Agreement in lateral dose falloff (< 1 mm DTA) in Figure 56b indicates field size agreement. The same plot shows agreement in the region of the 20 MeV beam to be within 2%, while the measured dose in the regions of the two 12 MeV beams is less than calculated by as much as 5%. This difference in percent dose is seen between 2 cm and 4 cm depth in the PTV, and is likely due to output for the 12 MeV beams using the variable-SCD applicator being less than the standard applicator output factors used in the TPS calculations. Further, between ±3 cm off-axis for 4-6 cm depths, measured dose is less than calculated dose by as much as 4%. Discrepancies in this region are most significant on central axis (Figure 57), and are clearly due in part to the dose algorithm's overestimation of low dose (< 20%) contributions for the 12 MeV beams (cf. Figure 38). From 80% to 10% however, central-axis depth dose agreement is within 2 mm DTA.
The central axis depth dose curve in Figure 57 also exhibits a difference of as much as 5% from 0.5 to 2 cm depth, where calculated dose exceeds measured. This difference is consistent with the commission verification results (cf. Figure B9), and is likely due to depth dose curves measured using the standard applicator being used to calculate the variable-SCD applicator dose distribution.

3.4.2.4 Inverted Well Target

The isodose plots and off-axis profiles at 2 cm depth for measured and calculated dose distributions for irradiation of the inverted well target using the variable-SCD applicator plan are overlaid in Figure 58a and 58b, respectively. Depth dose curves at 0 cm and -3.4 cm off-axis are shown in Figure 59a and 59b, respectively.

Agreement in lateral dose falloff (< 1 mm DTA) in Figure 58b indicates field size agreement. At this depth, measured and calculated dose generally agree within 1%, with the exception of dose to points at ±1 cm from central-axis, where agreement is within 2%. Similarly, agreement in the PTV for depths greater than 2 cm is within 2%. The measured hot spot (105.7%) at -2.8 cm off-axis and 3.5 cm depth is approximately 1.7% greater than the TPS predicted at that point. This may be due to a slight beam overlap (probably less than 0.3 mm). The symmetric discrepancy in the 90% and 95% isodose contours from 3 to 4 cm off-axis and from -3 to -4 cm off-axis is again due to the dose algorithm's overestimation of the low dose contribution of the 12 MeV beam. Similar effects due to the 20 MeV beam are seen in the 30% and 10% isodose contours between ±2 cm off-axis. The influence of these discrepancies on central axis depth dose is seen in Figure 59a from 7 to 11 cm depth where calculated dose exceeds measured dose by as much as 8%. Also, due to measured depth dose curves for the standard applicator being
Figure 58: Measured (solid red) and TPS calculated (dashed black) dose data for irradiation of the inverted well target using the variable-SCD applicator plan. (a) Isodose plot of central xz plane dose distributions. Note: The solid circle indicates the point at which the measured and calculated dose distributions are normalized to 100%. (b) Off-axis dose profiles at 2-cm depth.
used to calculate the variable-SCD applicator dose distribution, calculated dose exceeds measured dose by as much as 6% on central axis (Figure 59a) from 0.5 to 1 cm depth, and by as much as 8% at -3.4 cm off-axis (Figure 59b) from 0.5 to 2 cm depth.

Figure 59: Measured (solid red) and TPS calculated (dashed black) depth dose plots for irradiation of the inverted well target at (a) 0 cm off-axis, and (b) -3.4 cm off-axis.

3.5 Aim 4b: Treatment Planning Results in Patient PTV

The results of segmentation-field treatment plans developed for the patient case (squamous cell carcinoma of the left ear) using the standard and variable-SCD applicators are presented in this section. Figure 60 shows the field segmentation scheme and locations of CT slices presented in this section. Dose distributions (normalized such that 5000 cGy = 90%) calculated using both applicators are shown for CT slices 73, 94, and 110 in Figures 61, 62, and 63, respectively.
Figure 60: Sagittal view of the field segmentation scheme showing locations of transverse slices 73, 94 and 110 (z = 1.5, -4.5, and -9.3 cm, respectively).

Figure 61: TPS dose distributions in transverse slice #73 calculated using (a) the standard applicator, and (b) the variable-SCD applicator. Two 12 MeV beams and a 16 MeV beam were used to treat this portion of the PTV. The PTV is contoured in lavender and the cord in yellow.
Figure 62: TPS dose distributions in transverse slice #94 calculated using (a) the standard applicator, and (b) the variable-SCD applicator. A 9 MeV beam, two 12 MeV beams, and a 20 MeV beam were used to treat this portion of the PTV.

Figure 63: TPS dose distributions in transverse slice #110 calculated using (a) the standard applicator, and (b) the variable-SCD applicator. A 9 MeV and 12 MeV beam were used to treat this portion of the PTV. The eye is contoured in skyblue and the lacrimal gland in green. The white strip over the eye is the lead shield.
In slice 73 (Figure 61), a hot spot of 111% using the standard applicator was reduced to 108% using the variable-SCD applicator. In slice 94 (Figure 62), a more dramatic reduction in maximum dose is seen (114% to 108%) due to 12 MeV and 20 MeV being non-adjacent energies. Conversely, the modest reduction in the maximum PTV dose (103% to 102%) in slice 110 (Figure 63) is attributed to 9 MeV and 12 MeV being adjacent beam energies. Also, a cold spot of 85% in slice 94 for the standard applicator plan (due to a slight beam gap along the anterior abutment edge of the 12 MeV and 20 MeV beams) is increased to 90% in the variable-SCD applicator plan.

The DVHs for both trials are overlaid in Figure 64, and the DVH statistics are summarized in Table 10. The dose to the critical structures was within acceptable limits for both plans. 95% of the lacrimal gland received less 35 Gy and the cord, eye, and optic nerve doses were well under tolerance dose (45 Gy, 50 Gy, and 50 Gy, respectively). Dose homogeneity in the PTV was improved significantly in the variable-SCD applicator plan. This is reflected by the $\sigma$ of the PTV being reduced by 22%, the minimum dose being increased by 14%, the maximum dose being decreased by 10%, and the $D_{90-10}$ dose spread decreasing by 22%. Compared to the treatment planning results for simple PTVs, these results show a more significant improvement in the maximum PTV dose (4% reduction compared to 10% reduction) and minimum PTV dose (4% increase compared to 14% increase), while the improvements in $\sigma$ and $D_{90-10}$ are less significant (30% reductions compared to 22% reductions).

The 66% cold spot in the standard applicator plan occurs at the junction of the 12 MeV and 20 MeV beams. This cold spot was eliminated in the variable-SCD applicator plan. The remaining 75% cold spot in the variable-SCD applicator plan occurs where the
PTV nearly intersects the lacrimal gland and is a result of the efforts to minimize dose to the lacrimal gland and eye, i.e. the dose at the location of the cold spot could be increased at the expense of increased dose to the critical structures.

Figure 64: Cumulative DVHs for segmented-field ECT patient plan using the standard applicator (dashed lines) and the variable-SCD applicator (solid lines). Dose to the PTV (lavender), lacrimal gland (green), eye (skyblue), cord (yellow) and optic nerve (red) are shown.

Table 10: Summary of treatment planning results for patient plan using the standard and variable-SCD applicators.

<table>
<thead>
<tr>
<th>Trial</th>
<th>ROI</th>
<th>Mean Dose (%)</th>
<th>Min (%)</th>
<th>Max (%)</th>
<th>Max-Min (%)</th>
<th>σ (%)</th>
<th>D_{90-10} (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standard</td>
<td>PTV</td>
<td>98.6</td>
<td>66.2</td>
<td>124.3</td>
<td>58.1</td>
<td>5.4</td>
<td>13.9</td>
</tr>
<tr>
<td>Variable-SCD</td>
<td>PTV</td>
<td>98.0</td>
<td>75.3</td>
<td>111.5</td>
<td>36.2</td>
<td>4.2</td>
<td>10.9</td>
</tr>
<tr>
<td>Percent Difference</td>
<td>↓ 0.6%</td>
<td>↑ 13.7%</td>
<td>↓ 10.3%</td>
<td>↓ 37.7%</td>
<td>↓ 22.2%</td>
<td>↓ 21.6%</td>
<td></td>
</tr>
</tbody>
</table>
The limitations of the planning tools made segmented-field optimization prohibitively time consuming. As such, a limited number of variations on this plan were investigated. For example, the 112% hot spot occurs in the region of the 20 MeV field. A 20 MeV beam was selected to treat the deepest portion of the PTV because a 16 MeV beam would not provide adequate penetration for the field size used (approximately 3x3-cm²). However, one could increase the penetration of the 16 MeV beam by increasing the field size, which may result in a less conformal dose distribution, but would decrease the magnitude of the hot spot in the PTV.
The results of this research support the hypothesis that a clinically practical method that utilizes Cerrobend® custom inserts can deliver segmented-field electron conformal therapy in the energy range of 6-20 MeV with less than ±5% variation in dose spread in abutment regions of hypothetical PTVs, i.e. constrain the PTV dose to 85%-105%.

Four specific aims were completed to test the hypothesis. In Aim 1, air gaps were selected (5, 7.5, 11.5, 17.5, and 19.5 cm for 6, 9, 12, 16, and 20 MeV beams respectively) for implementation of the variable-SCD method and a standard 15x15-cm² applicator was modified to allow collimator placement at the selected locations.

In Aim 2, the ability of the TPS to accurately calculate dose in a phantom using the standard and variable-SCD applicators was assessed. The results showed that the electron pencil-beam algorithm commissioned for the variable-SCD applicator using the standard applicator commissioning data predicts measured relative dose distributions with the same degree of accuracy as it does for the standard applicator at 100 cm SSD.

In Aim 3, a method of fabricating Cerrobend® inserts with diverging edges to define precise field shapes was developed and the accuracy of beam collimation for abutted fields using the variable-SCD applicator was evaluated. The results show that field edges can be abutted with ±0.3 mm accuracy. This resulted in ±2.2% dose variation in the abutment region at a depth of 1 cm for a 12 MeV beam, which should be typical of all beam energies using the variable-SCD applicator due to the penumbra widths matching.
In Aim 4a, segmented-field plans for the irradiation of four hypothetical PTVs were developed using the standard and variable-SCD applicators. The ability to deliver the variable-SCD plans was assessed by irradiating a film phantom. Treatment plans for the four simulated PTVs showed dose homogeneity in agreement with the hypothesis can be achieved using the variable-SCD applicator. On average, the minimum dose increased 4%, maximum dose was decreased 4%, the σ of the dose decreased by 29%, and $D_{90-10}$ decreased by 32%. Mean dose was decreased by an average of 0.2%, indicating that the improvement in dose homogeneity did not impact the total dose to the PTV. In cases 1, 3, and 4, unacceptable maximum and minimum dose in the PTV for the standard plan was reduced to acceptable levels (85%-105%) in the variable-SCD plan.

The results of film phantom measurements showed the variable-SCD applicator plans can be delivered with the predicted improvement in dose homogeneity in the abutment regions. However, significant disagreement between calculated and measured dose distributions in central-axis depth dose (as much as 1 cm DTA in the 90% isodose contours) was observed in case 2. This is likely due to (1) the limitations of the dose algorithm in low dose regions (<20%) and (2) output factors measured using the standard applicator being used for the variable-SCD treatment plans. Also, the 2% tolerance on clinical accelerator output may have contributed to the discrepancies observed. Therefore, it should be possible to significantly improve calculated and measured dose distribution agreement by (1) using a more accurate dose algorithm (e.g. the pencil-beam redefinition algorithm or Monte Carlo), and (2) fully commissioning the variable-SCD applicator.
In Aim 4b, the potential to improve dose homogeneity for a patient PTV was assessed by developing a treatment plan for a single patient case. Comparison of identical plans using the standard and variable-SCD applicators with four beam energies show that dose homogeneity can be substantially improved using the variable-SCD applicator. A 22% reduction was seen in the standard deviation and $D_{90-10}$ of the PTV dose distribution. The minimum dose was increased from 66% to 75% (a 14% increase) and the maximum dose was decreased from 124% to 112% (a 10% decrease). The magnitude of the remaining hot could be reduced at the expense of decreased conformality by substituting a 16 MeV beam in the place of the 20 MeV beam and increasing the field size. The remaining cold spot could be increased in dose at the expense of increased dose to critical structures. Other possible sources of dose heterogeneity in both plans include (1) the definition of pencil beam locations for each field, i.e. field dimensions were not multiples of the 2x2 mm$^2$ pencil beams, (2) field gaps and overlaps due to the limited treatment planning field-segmentation tools, and (3) irregular patient surface and variations in tissue density.

- **Future Studies**

Future work that would build upon this research should be focused on (1) optimization of the variable-SCD method, (2) treatment planning, and (3) verification of patient plans. The air gaps used in this research were chosen with the goal of optimizing abutment dosimetry for any combination of beam energies. Investigation of optimal air gaps for subsets of energy combinations would likely yield different results, possibly further improving dose homogeneity. For example, one might developed a set of air gaps
for the 6 MeV, 9 MeV, and 12 MeV beams to be used for shallow PTVs and another set of air gaps for deeper PTVs using the 12 MeV, 16 MeV, and 20 MeV beams.

In regards to treatment planning, two tools could be developed to aid in beam segmentation. Firstly, a beams-eye view "depth map" is essential to allowing the user to choose appropriate beam energies and field shapes to treat the target volume. While the method presented in this work is effective, an automated method would be preferable for clinical implementation. Secondly, a TPS feature to automatically adjust block definitions such that gaps and overlaps are disallowed when a single block definition is modified would greatly reduce planning time.

The ability to deliver patient treatment plans using multiple energy-segmented fields requires further study. The results of plan verification for simple hypothetical PTVs showed substantial discrepancies in depth dose which was attributed to the limitations of the pencil-beam dose algorithm as well as the output factors measured using the standard applicator being used for the variable-SCD applicator plans. Because of increased complexity in patient plans, these discrepancies could possibly be increased. Therefore, a study of plan verification for patient plans calculated using the standard pencil-beam algorithm, the pencil-beam redefinition algorithm, and Monte Carlo methods would be prudent and should coincide with an investigation into the influence of the increased air gaps on beam output.

Assessment of the potential for the variable-SCD method to deliver dosimetry comparable to bolus electron conformal therapy in patient PTVs also requires further study. Ideally, these studies would be carried out using patient cases treated using custom bolus or by developing custom bolus and variable-SCD plans for patient cases.
- **Technology Required for Clinical Implementation**

  The variable-SCD method results in broadened penumbra for all beam edges, regardless of whether the edges is abutted to another field or is an outer edge. This results in increased dose to normal tissue; however, the method could be improved by defining a single outer-field-edge collimator at the standard location (e.g. 5-cm air gap) in addition to the abutment-field-edge collimators at the variable SCDs. This would allow for sharp lateral dose falloff, sparing normal tissue around the perimeter of the target. While this is currently practical with the variable-SCD applicator, the treatment planning system does not allow for such a setup.

  For the method to be implemented clinically on a routine basis, there are several technologies that need to be commercially available. Specifically, they include: (1) variable-SCD applicators for the treatment machine, including allowing the outer edge of all fields to be defined by a common insert located at 95-cm SCD, the currently standard insert location; (2) improved treatment planning tools for creating the BEV depth map and accomplishing segmentation of the treatment field into optimal subfields for each beam energy; and (3) dose calculations that are more accurate than the PBA and that can model the multiple collimation levels.
References


101


Appendix A
Variable-SCD Applicator Design

Detailed drawings of the components of the variable-SCD applicator are presented in this appendix. All dimensions are in centimeters.

Figure A1: 3D view of the leg and back plate design with (a) tabs housing frame alignment holes attached and (b) tabs without alignment holes attached.

Figure A2: (a) Front view of leg and back plate design with tabs in place. Threaded holes for tab attachment are shown. (b) Rear view of leg and back plate design.
Figure A3: Front view of leg design detailing dimensions concerning tab placement and spacing. Note: the holes in this figure were threaded for a 4-40 screw.
Figure A4: (a) 3D view of tab design showing tab dimensions. (b) Top view of tab showing placement and dimensions of alignment holes. (c) Front view of tab showing placement and dimensions of holes (not threaded) used for leg attachment.
Figure A5: 3D view of frame design detailing all dimensions including peg diameter.
Appendix B
15x15-cm² Isodose Plots

Isodose plots of measured and TPS calculated dose distributions for 15x15-cm² fields using the standard and variable-SCD applicators are presented in this appendix.

Figure B1: Isodose plots for 6 MeV, 15x15-cm² field measured (solid line) and calculated (dashed line) using the standard 5-cm air gap.
Figure B2: Isodose plots for 9 MeV, 15x15-cm² field measured (solid line) and calculated (dashed line) using the standard 5-cm air gap.
Figure B3: Isodose plots for 9 MeV, 15x15-cm² field measured (solid line) and calculated (dashed line) using the variable-SCD method with a 7.5-cm air gap.
Figure B4: Isodose plots for 12 MeV, 15x15-cm$^2$ field measured (solid line) and calculated (dashed line) using the standard 5-cm air gap.
Figure B5: Isodose plots for 12 MeV, 15x15-cm$^2$ field measured (solid line) and calculated (dashed line) using the variable-SCD method with a 11.5-cm air gap.
Figure B6: Isodose plots for 16 MeV, 15x15-cm\(^2\) field measured (solid line) and calculated (dashed line) using the standard 5-cm air gap.
Figure B7: Isodose plots for 16 MeV, 15x15-cm$^2$ field measured (solid line) and calculated (dashed line) using the variable-SCD method with a 17.5-cm air gap.
Figure B8: Isodose plots for 20 MeV, 15x15-cm$^2$ field measured (solid line) and calculated (dashed line) using the standard 5-cm air gap.
Figure B9: Isodose plot for 20 MeV, 15x15-cm$^2$ field measured (solid line) and calculated (dashed line) using the variable-SCD method with a 19.5-cm air gap.
Vita

John Richert was born in Lake Charles, Louisiana, in 1977. From that year until his high school graduation, he lived in Sulphur, Louisiana, with his parents, brother, and two sisters. He began his undergraduate education in 1995 at Louisiana State University. After two and a half years of coursework, he traveled to go to San Francisco, California. Eight months later, he returned to Louisiana and resumed his undergraduate education at LSU. Upon receiving his Bachelor of Science degree in physics with a concentration in computer science in 2000, he moved to New York City. There he lived for a year and three months. Again, he returned to his home town in Louisiana, this time highly motivated to pursue a stable career. To this end, he applied to the Medical Physics and Health Physics graduate program at LSU. To his delight, he was accepted and found the course work fascinating. He currently plans to begin a career as a clinical medical physicist.