Health Science Campus

FINAL APPROVAL OF THESIS
Master of Science in Biomedical Sciences

Dosimetric Verification of the ADAC Pinnacles Pencil Beam Algorithm for Clinical Electrons in Presence of Cerrobend Blocking

Submitted by:
Philip Chan

In partial fulfillment of the requirements for the degree of Master of Science in Biomedical Sciences

Examination Committee

Major Advisor: E. Ishmael Parsai, Ph.D.

Academic Advisory Committee: Michael Dennis, Ph.D.
John Feldmeier, D.O.

Senior Associate Dean
College of Graduate Studies
Michael S. Bisesi, Ph.D.

Date of Defense: December 12, 2007
Dosimetric Verification of the ADAC Pinnacle\(^3\) Pencil Beam Algorithm

For Clinical Electrons In Presence of Cerrobend Blocking

Philip Chan

University of Toledo

Health Science Campus

2008
Acknowledgements

I wish to express my greatest appreciation to my advisors, Dr. Ishmael Parsai, Dr. John Feldmeier, and Dr. Michael Dennis for their guiding voices. Their knowledge and quality of work are the standard I will forever strive to achieve. I wish to thank David Pearson for his knowledge on performing radiation measurements and Chetan Patel for his experience with beam modeling. Also I want to thank Dan Schifter for providing some of the measurements used in this work.
<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>Introduction</td>
<td>4</td>
</tr>
<tr>
<td>Literature</td>
<td>6</td>
</tr>
<tr>
<td>Materials and Methods</td>
<td>19</td>
</tr>
<tr>
<td>Results</td>
<td>33</td>
</tr>
<tr>
<td>Discussion</td>
<td>64</td>
</tr>
<tr>
<td>Conclusion</td>
<td>69</td>
</tr>
<tr>
<td>Bibliography</td>
<td>71</td>
</tr>
<tr>
<td>Abstract</td>
<td>74</td>
</tr>
</tbody>
</table>
Introduction

With the advent of megavoltage machines, electron beam therapy has become an important component in radiation therapy. Early accelerators such as the betatron delivered low energy electron beams and thus were used only for treating superficial tumors. As available beam energies increased, greater depths could be treated using the favorable characteristics unique to electron beams.

Electron therapy treatments are commonly performed with a single unopposed field. A custom block is used to shape the electron radiation field to the shape of the tumor. Then, the field is aimed perpendicular to the skin surface. Furthermore, treatments are simple as there are only two choices of importance: beam energy and the use of bolus.

One of the most important characteristics of the electron beam is the homogeneous dose delivered from the skin surface to a given depth that is proportional to beam energy. This region is called the therapeutic range, and at further depths, there is a very sharp falloff in radiation dose, a positive aspect of electrons that is not provided by photon beams. This allows electron beams to be aimed directly at critical structures without contributing significant dose to such structures.

The depth of treatment, often the therapeutic range $R_{90}$, can be controlled by choosing the proper beam energy. This depth in centimeters is approximated by dividing the beam energy in MeV by three. Conversely, the depth of treatment can be multiplied by three to determine the minimum energy in MeV necessary to treat the entire tumor to at least ninety percent of maximum dose. The use of bolus provides another method for moving the depth of treatment as current accelerators provide a selection of beam
energies—every centimeter depth from approximately two to seven. Bolus may be used in situations to protect critical structures at depth or to improve skin surface dose.

Electron dosimetry can be based solely on reference dose distributions measured in a water phantom. This dose calculation often does not include the effects of heterogeneity, irregular skin surface, or blocking on the dose distribution. Hand calculations for instance are only monitor unit calculations based on reference measurements with tabulated correction factors. This provides no information about the tumor volume coverage shown with computed isodose distributions.

One of the most commonly implemented computer algorithms for dose calculation has been the Hogstrom electron pencil beam algorithm. It is the basis for many commercial and in-house treatment planning systems, and it is the implemented in ADAC Pinnacle³ (ADAC Laboratories, Milpitas, CA). The goals of this project are to commission planning electrons for clinical use and to verify the planning system for common clinical settings. All energies will be commissioned, but a subset of commonly used energies 6, 9, and 12 MeV will be examined further.

The objectives are first to gather needed beam data for developing a beam model. Second, the beams are modeled in the planning system using a combination of measured data and modeling parameters. Next, the dose calculations are commissioned through comparison with absolute dose measurements performed in water. Lastly, relative dose distributions of various blocked fields not included as beam data are measured for comparison. This final test is used to verify the predictive accuracy of the algorithm for clinical electron therapy.
Literature

Almond et al. (1999) described a method for dose calibration of electron beams. This method includes determining correction factors for measuring absolute dose in water as well as the protocol for standardizing dose output for linear accelerators.

Blomquist et al. (1996) designed a method for testing a commercial treatment planning system for electron beams. Open field depth dose curves verify accurate entry of beam data into the planning system. Corresponding profiles at two depths can then be used to verify the accuracy of modeling of in-air scatter variance along with the accuracy of dose calculation along the profile. Extended treatment distance tests further verify the in-air scattering value and the algorithm’s virtual SSD and inverse-square correction. For evaluation of central axis depth dose data, Blomquist recommends calculating percent difference for points from the surface to $R_{80}$, and then using distance of deviation for the higher dose gradient in the fall-off region to 10% dose. For profiles, the author recommends similar analysis for the clinically relevant central region above 80%, and then the penumbra region laterally to 10%. The author also recommends 2% and 2 mm as the tolerances of acceptance. Normalization for comparison of depth dose curves is done at 0.4 cm depth instead of the depth of dose maximum. The planning system tested showed problems at the farther end with obliquely incident beam. However, it properly predicted the therapeutic range decreased while the practical range increased, and it accurately predicted the higher maximum dose.

Cheng et al. (1996) performed a verification of the MDAH (M. D. Anderson Hospital) pencil beam model to verify accuracy of input beam data and to determine situations where the algorithm is accurate. Electron fluence readily decreases with depth
because of straggling but the Hogstrom algorithm generalizes this process only with depth dose curves. Tests were performed following ECWG (Electron Collaborative Working Group) recommendations. For central axis depth dose beyond the buildup the acceptable percentage difference is 2%. Profiles were given 4% in the low dose gradient region and 4 mm in the penumbra. For the sloped surface test, the tolerance was given as 5% and 5 mm. Cheng described the possible shortcomings of the Hogstrom algorithm to explain the calculated inaccuracies at depth in the penumbra region. The algorithm does not model straggling and simply assumes electron energy at depth is calculated following Harder’s equation. Electrons may in fact scatter at higher angles and have much lower energies for its depth. The algorithm assumes constant fluence across the field at any depth and the only difference is decreasing energy with depth. Fluence in fact should decrease as well because electrons may be absorbed with longer effective path lengths. The result is the calculated low isodose lines may spread more than actual as there is an overestimation of the electron fluence near the practical range. Cheng stressed the beam obliquity experiment is good for testing the excess scatter dose predicted by the algorithm. Because of the obliquity, the overestimation near the practical range along one pencil beam will affect “shallower” depths further along the field towards the distal edge with an accumulation of overestimation. Using the ECWG data set, Cheng was able to get 98% accuracy with an acceptance limit of 5% or 5 mm. The algorithm was accurate for open fields, extended treatment distances, blocked fields, and oblique beam incidence. Inaccuracies were well defined and expected in certain experimental tests.

Deasy et al. (1996) described the three-dimensional (3D) implementation of the Hogstrom algorithm in the Pinnacle planning system. The basis of the algorithm is the
Fermi-Eyges theory which assumed that incident monoenergetic charged particles undergo very many small angle scattering events (due to collisions with atomic nuclei Coulomb fields and also with atomic electron Coulomb fields), and all scattering events are assumed to deflect through angles much smaller than the mean angle of total deflection at that depth. The effective scattering in a medium is a function of energy determined with Harder’s equation for probable energy at an effective depth. The lateral distribution is considered Gaussian. All pencil beams for a field follow the same Gaussian shape, and thus path lengths of scattered electrons are ignored. The physics is not modeled because the algorithm approximates energy loss.

\[ E_p = E_{p,0} \left( 1 - \frac{z_{\text{eff}}}{R_p} \right) \]  

The FMCS value in electron beam modeling is a multiplicative factor applied to the scattering powers included in the original M.D. Anderson implementation. As stated previously the basis of this algorithm, the scattering powers are based on the Fermi-Eyges theory of multiple Coulomb scattering. It provides depth correction for the scattering powers. These powers are significantly smaller than actual measured values but applying the appropriate FMCS value allows for proper modeling of scatter. The factor controls the lateral spread of dose distributions at depth in medium. This is contrasted with the sigma-theta-x which controls in-air scatter and dose shape at shallow depths. Because Pinnacle uses similar, low scattering powers as the original algorithm, the FMCS modeling parameter is recommended to be above 1.0.

The virtual source position is used only as a correction for the increased deflection of electrons with extended treatment distances. The virtual SSD is typically
evaluated to be less than 100 cm thus making the beam appear more divergent. This provides a small correction for in-air scattering. The other similar factor is sigma-theta-x which models the increase in penumbra width with treatment distance caused by the increased in-air scattering.

Fraas et al. (1998) provided recommendations for performing quality assurance on treatment planning systems. Recommendations include commissioning or verification measurements of common clinical treatment settings with electron therapy. For electron algorithms, recommendations include characterizing an algorithm’s deficiencies by verifying depth dose calculation with field blocking and extended treatment distances as well as analyzing any energy dependence.

Hogstrom (2006) provided a survey of electron beam therapy. The survey includes standards for characterizing electron beams for clinical use, specifically methods for dose measurement and calibration. Also included is a brief history of the development of electron pencil beam algorithms.

Hogstrom et al. (1981) provided the first computer algorithm for calculating electron dose distributions that uses measured beam data as input. The algorithm calculates dose with a summation of the contribution by each Gaussian pencil beam in a broad, collimated beam to some calculation point. It is two-dimensional (2D) as it includes the effects of heterogeneity and beam obliquity although on a single computed tomography slice made infinite in the lateral extent. The algorithm described accepts measured data and manipulates the data according to the beam setup. Energy loss through scatter is inherently included with percent depth dose in water. The angular spread of electrons in air and in water is included with the penumbra region of profile
data in both media. Photon contamination is included as a uniform component derived from a depth greater than the practical range. Published data of linear angular scattering power per CT number are used for heterogeneity correction. Percent depth dose can be calculated for rectangular fields using a square root method to combine dose curves from two different square fields instead of using an equivalent square method. The algorithm integrates over the area of the collimator aperture which is typically the block insert, to calculate the dose to a single point at $X, Y, Z$ from pencil beams at $X', Y'$. Each pencil beam has a divergence from the beam source along with the root mean square spread given by angles $\sigma_{thx}$ and $\sigma_{thy}$. These RMS values are considered to be equal so only $\sigma_{thx}$ is considered. The Fermi-Eyges theory states that for each pencil beam the $\sigma_{thx}$ or “angular spread at any point in space is Gaussian”. Hogstrom recommends using film at varying source-to-film distances to measure the penumbra width for the calculation of $\sigma_{thx}$.

The dose to a point is calculated by the equation below:

$$
D(X, Y, Z) = \int_{\text{collimator}} S(X', Y') d(X' - X, Y' - Y, Z) dX' dY' \quad (2)
$$

The dose factor $d(X, Y, Z)$ in the equation combines a calculated percent depth dose $g(Z)$ multiplied by an off-axis term $f(X, Y, Z)$ to derive dose to a calculation from an arbitrary pencil beam.

$$
d(X, Y, Z) = f(X, Y, Z) g(Z) \quad (3)
$$

The other term $S(X', Y')$ represents the strength of a particular pencil beam. This allows for the characterization of beam inconsistencies such as beam asymmetry. The $g(Z)$ term represents the percent depth dose of the particular pencil beam.
The off-axis term is characterized by the “lateral flux distribution due to thick-target multiple Coulomb scattering (MCS)”. This in turn is based on $\sigma^2_{MCS}$ or the root mean square of the lateral distribution.

$$f(X,Y,Z) = \frac{1}{2\pi\sigma^2_{MCS}} e^{-\frac{X^2+Y^2}{2\sigma^2_{MCS}}}$$ (4)

$$\sigma^2_{MCS} = \frac{1}{2} \int_{-L_0}^{L_0} (Z-Z')^2 \frac{d\sigma_{MCS}}{dZ'} dZ'$$ (5)

The factor $\sigma^2_{MCS}$ is the linear angular scattering power of water based on data from ICRU 21 for linear scattering power. This factor is energy dependent following Harder’s equation for electron energy and depth, and requires fluence conversion factors for heterogeneity correction. It can be seen that the apparent effect of scatter increases dramatically with distance from the collimator.

The factor $g(Z)$ is the calculated percent dose at depth. For heterogeneity correction the effective depth must be evaluated. The algorithm again uses ICRU 21 values for linear collision stopping power of different materials. Effective depth is calculated normally as the sum of linear stopping powers corrected relative to water and multiplied by the thickness of the slab. A further correction for extended SSD allows the algorithm to inherently perform accurate calculations with irregular surfaces. $g_0$ is the percent depth dose in water.

$$Z_{eff}(Z) = \int_{-L_0}^{L_0} \frac{(dE/dZ')}{(dE/dZ)_{H_2O}} dZ'$$ (6)

$$g(Z) = g_0 \left[ \frac{SSD + Z_{eff}}{SSD + Z} \right]^2$$ (7)
The original equation $D(X, Y, Z)$ can be used to calculate the percent depth dose for an arbitrary field by determining $g(Z)$ using $D_0(X, Y, Z)$ which again is the dose to a point in water as calculated by output factors and Task Group No. 51 (TG-51) calibration. The final equation is presented concisely with the ending terms utilizing a measured percent depth dose curve in water and corrected for field size and treatment distance. It is important to note the algorithm ignores the applicator size and follows only the unblocked field size.

$$D(X, Y, Z) = \left( \frac{1}{2\pi\sigma^2} \right) \left( \int_{\text{collimator}} S(X', Y') e^{-\frac{(X-X')^2 + (Y-Y')^2}{2\sigma^2}} \, dx' \, dy' \right) \times D_0(0,0,Z_{eff}) \left( \text{erf} \left( \frac{WX_\theta Z}{\sqrt{2}\sigma_{\theta}} \right) \left( \frac{SSD + Z_{eff}}{SSD + Z} \right)^2 \right)$$

The term $\sigma^2$ is calculated based on $\sigma_{\text{MCS}}^2$ and the drift distance from the collimator or cutout to the treatment surface. This is how the algorithm represents increased beam penumbra with extended treatment distances.

$$\sigma^2 = \sigma_{\text{MCS}}^2 + (Z + L_0)^2 \sigma_{\text{th}}^2$$

$WX_\theta Z$ is the square field size dimension of the measured depth dose curve being used for the calculation. Then for rectangular fields, the square root method can be applied to combine depth dose values from measured depth dose curves from two square field sizes $WX$ and $WY$.

$$D_{(0,0,Z)}^{WX, WY} = \left( D_{(0,0,Z)}^{WX, WX} D_{(0,0,Z)}^{WY, WY} \right)^{\frac{1}{2}}$$

Bremsstrahlung or photon contamination is measured at depths greater than the practical range of electrons. For shallow depths the photon dose is accounted for by the inverse-square correction. This constant value is simply summed during dose calculation within the collimated field.
Measured data are manipulated by the algorithm for different fields or for other settings. Percent depth dose curves calculated by Hogstrom for small field sizes were based on depth dose curves measured with larger field sizes. Good results were demonstrated by calculating the depth dose curve for 6x6 from measured 10x10 cm² data. The 4x4 showed significant differences which Hogstrom attributed to electron-electron scattering not being accounted for by the algorithm. His solution was simply measuring smaller field sizes so the algorithm could manipulate these depth dose curves. In general, low dose regions of 10% isodose were different from measurements. This was probably due to the algorithm having ignored the presence of large angle electron scattering which would result in increased path length but diminished depth of penetration. The measured data showed 10% wider isodose curves than calculated. Additionally with such low doses, the photon contribution became significant and therefore small differences in dose were in fact much significant in percentage errors.

Hogstrom et al. (1984) provided dosimetric analysis of a 2D implementation of the pencil beam algorithm at M.D. Anderson Cancer Center. Water phantom studies showed the algorithm accurately calculated isodose distributions with extended treatment distances and surface irregularities without the need for additional beam data as input. The penumbra width was predicted with different air gap and surface slope with the algorithm’s use of scattering powers. Also this was the first version of the algorithm were off-axis ratios could be applied to points on beam profiles to account for beam non-uniformity. Low dose regions were inaccurate because the photon contamination model is a simple summation and the model for multiple Coulomb scattering includes only small angle scatter electrons and not large angle electron-electron scatter. Extended
surface distances were tested for open applicators and showed accurate prediction of increased scatter in air particularly with low energy electrons. For the oblique case, the algorithm predicted the high dose region and the widening of the depth dose curve, namely the decrease in therapeutic range and increase in practical range.

Khan (2003) described the factors involved in the Hogstrom pencil beam algorithm. The rate of energy loss for electrons in a medium is determined by the medium stopping power. Stopping power is the sum of collisional losses with other electrons and radiative losses from bremsstrahlung. The value is given as energy loss per unit path length. Collisional losses increase with lower energy and with higher electron density which is often lower atomic number. Radiative losses are increased with energy and atomic number. Electrons will encounter multiple scattering from Coulomb interactions with electrons and nuclei. An acceptable method is to approximate the angular and spatial spread with a Gaussian distribution. The value is given as mean square scattering angle per unit path length.

Khan et al. (1991) described important beam characteristics relevant to proper beam data collection. Increasing field size will increase the therapeutic range and decrease the surface dose relatively. The reason is the increased lateral scatter that will reach the central axis at depth, resulting in a relatively higher dose at depth than at the surface. With higher electron energies, the contribution of scatter becomes significant at greater depths. The greater energy means scatter will diffuse over a longer path length and lateral scatter from relatively larger field sizes will significantly increase central axis depth dose and shift the maximum and the therapeutic range downstream. Smaller fields will oppositely diminish lateral scatter equilibrium at shallower depths. Increasing beam
energy results in higher surface dose because there is decreased multiple Coulomb scattering and thus, less of a relative build-up. The dose gradient is much longer because of the longer path length. Even with extended treatment distances, the penumbra at the surface is minimal because of decreased scattering in air. With low energy beams where the isodose of the therapeutic range bulges out from increased multiple Coulomb scattering in phantom, the same range for a high energy beam narrows at depth because of the slow diffusion over a greater path length. Measured output factors will vary depending on energy. Because high energy beams are less affected in air, the scatter onto the surface from the collimation system does not change significantly with field size. For low energy beams, in-air scattering is strongly dependent on field size. Small fields with low energies will diminish lateral scatter equilibrium resulting in an upstream shift of the depth dose thereby decreasing the apparent output factor.

McNutt and Tome (2002) derived a method for applying a correction factor to a pencil-beam algorithm for monitor unit calculation. This method is used in the Pinnacle implementation of the Hogstrom algorithm. The method requires extensive user input of output factors per energy, cone, blocking, and source-skin distance for a multiplicative correction factor for the raw dose per monitor unit calculated by the algorithm. The calculation takes into account the different effects of scatter with applicator sizes and the differing output at extended source-to-skin distance. The Hogstrom pencil beam algorithm inherently includes field size and extended treatment distance dependence for central axis depth dose prediction. McNutt and Tome describe a way to use the pencil beam calculation process and apply it to monitor unit calculation.
The pencil beam algorithm is able to predict central axis relative depth dose but does not include the effects on absolute dose by scatter. Scattering effects include collimator settings and treatment distance. The purpose of measuring combinations of applicator, field size and treatment distance is to determine what the pencil beam algorithm should calculate for output factors. Using measured and calculated factors, Pinnacle stores correction factors to convert from the raw pencil beam calculation $OF_p$ to the measured output factor $OF_{rel}$ for monitor unit calculation.

$$OF_c(cone,W,SSD,D_{max}) = \frac{OF_{rel}(cone,W,SSD,D_{max})}{OF_p(cone,W,SSD,D_{max})} \quad (11)$$

$$D_{MU}(r) = OF_c A_{cal} D_{pb}(r) \quad (12)$$

$A_{cal}$ is the $OF_c$ for the reference field as 10x10 at 100 SSD measured at $d_{max}$. The calculation algorithm attempts to find a rectangle of best fit for an irregular field by selecting the smallest area rectangle circumscribing the irregular field. Then it calculates the output correction factor using the square root method for two square field size output correction factors.

$$OF_c(X,Y) = \sqrt{OF_c(X,X)OF_c(Y,Y)} \quad (13)$$

Linear interpolation is employed for intermediate field sizes and treatment distances. This is possible with extended treatment distance because the Hogstrom algorithm uses an inverse-square correction for relative depth dose curves. Clinical fields were shown to be accurate to within 1-2%. Further irregular fields were accurate to within 3-4% as long as their shape could be approximated with a rectangular field shape.

McShan et al. (1994) performed verification measurements to verify their implementation of the Hogstrom algorithm. McShan measured depth dose curves with
diode detectors along with five or more profiles per axis. Interpolation of such data provided isodose distributions parallel to beam central axis. Then, film measurements were performed in solid water phantoms. Because of difficulties in precise water scans, all experiments for a specific testing condition were renormalized so that data could be self-consistent. McShan found having a proper grid size for dose calculation within their treatment planning system was important for reducing differences between measured and calculated isodose. Accuracy in general was very accurate. Although large errors occurred in the high gradient regions of the penumbra, it was considered acceptable as the distances of difference was less than 2 mm. For the modeling procedure, it was recommended that the virtual source distance and Gaussian spread be used as modeling parameters rather than as measured values. This would allow for better prediction of beam behavior with different clinical settings.

Muller-Runkel and Cho (1997) tested the CMS planning system FOCUS, specifically the electron pencil beam algorithm with experiments designed to include most clinical settings. This included open fields at standard and extended distance, changes in percent depth dose with blocking, fields with beam obliquity, and centered rectangular blocks. The recommendation for $\sigma_{\theta_x}$ is that although derived the value directly, the variance did not yield calculated profiles that provided accurate fit to measured beam data. It along with $FMCS$ could be selected solely for model fitting and not necessarily be based on measurement. One experiment tested small fields with and without 2x2 and 4x4 cm$^2$ input beam data for the model. The FOCUS algorithm underestimated the upstream shift with increased blocking without the small field beam data. In general for measured depth dose curves, the therapeutic range shifted upstream...
with increased blocking, but the difference gradually decreases until an almost common practical range with an unblocked cone. Large fields were accurate with all energies but medium to high energy beams required small field data to show the shallower depth of maximum dose along with a slower and more gradual dose falloff because of the gradual diffusion of higher energy electrons over a greater range.

Shiu (1991) provided recommendations for developing an electron model with a treatment planning system that used the Hogstrom pencil beam algorithm. Important modeling parameters are the drift distance, sigma-theta-x, FMCS, and off-axis weighting factors. An included summary provides a step-by-step example for developing each electron model.

Shiu et al. (1992) described a data set to be measured for verification of an electron pencil beam algorithm for accuracy of dose calculation in common clinical settings and for validation of the algorithm. The experimental design was to choose a low and high energy to eliminate any possible energy dependence for accuracy. Water scans were performed along the major axes in the cross and in-plane directions. In general, the purpose of the experiments is to test the accuracy with various energies, field sizes, blocking, treatment distance, and irregular or sloped surfaces.

Xing et al. (2005) compared a Monte Carlo based algorithm to a pencil beam algorithm for electrons in two planning systems. It was found that for homogeneous phantoms, both systems accurately predicted isodose distributions for extended treatment distances and for obliquely incident beams.
Materials and Methods

Materials

The accelerator used in this project is the Elekta SL-25 (Stockholm, Sweden). For water scanning, a tank from Scanditronix-Wellhofer (Louvain-la-Neuve, Belgium) provides motorized three-dimensional movement. The associated control and analysis software is OmniPro-Accept and is provided with the tank. A small chamber of 0.125 cm$^3$ volume allows for accurate measurements of dose gradients and beam penumbra. A smaller MEDTEC tank (Kalona, Iowa) with only manual depth adjustment is used for absolute dose measurements with a larger chamber of 0.6 cm$^3$ volume. RIT113 software (Radiological Imaging Technology, Colorado Springs, CO) is used for exporting data from dose distributions computed by Pinnacle.

TG-25

The recommendation in AAPM (American Association of Physicists in Medicine) TG-25 regarding the most probable energy $E_{p,0}$ is that it should be calculated using the practical range $R_p$ evaluated from measured percent depth dose data. The practical range is determined at the depth where the tangent at the inflection point of the falloff portion of the curve intersects the bremsstrahlung background. Broad beam geometry with a narrow detector is used for this determination. The protocol recommends applicators larger than 10x10 cm$^2$ be used for measuring the depth dose curve, and Pinnacle in fact recommends using the largest cone. Beam divergence must be corrected for with a multiplicative factor based on the effective or virtual SSD and applied to the percent depth dose curve at all sampled points. However with beam energies below 20 MeV, the decreased accuracy is not relevant to clinical settings and the nominal SSD of 100 cm
may be used. Percent depth dose curves measured in a water phantom demonstrate the field size dependence. Depth dose curves must be corrected for depth dependence. For smaller field sizes typically made with cutout inserts for some set applicator size, the depth dose curves will move shallower as compared to an open applicator. To ensure accuracy, such small field sizes must be measured for modeling and to verify computed depth dose curves. Such small field depth dose curves help facilitate the computed depth dose curves provided by the Hogstrom pencil beam algorithm, again the basis of the Pinnacle algorithm. These curves help with the calculation of rectangular fields following the square-root approximation from depth dose curves from square fields. This is particularly important as irregular fields are also approximated by Pinnacle via a rectangular fit.

The TG-25 protocol recommends measuring output factors for a range of clinically relevant fields in their associated applicators. The jaws, multileaf collimator, and the applicator system all provide dramatic differences in lateral electron scattering, and therefore greatly affects beam characteristics such as surface dose, penumbra width, and photon contamination. Extensive output factor measurements at $d_{\text{max}}$ are recommended to account for such changes in lateral scatter equilibrium. The protocol recommends all relevant combinations of applicators and inserts as long as the multileaf collimator positions are constant per applicator. The protocol includes corrections for extended treatment distances and for oblique beam incidence. These corrections are not necessary for the Hogstrom-based electron algorithm in so far as measuring output factors for combinations of applicator, block cutouts, and treatment SSD.

**Machine and Beam Data**
The Pinnacle physics manual provides a complete list of necessary beam data for modeling in the planning system. The data used for each beam energy model are summarized here. Applicator sizes are defined at SAD in Pinnacle. The applicator sizes for the Elekta SL-25 are defined instead at the block insert by the source collimator distance which is nominally 94.7 cm. The drift distance is 5.3 cm from the block cutout insert to machine SAD of 100 cm. For example, the 10x10 cm\(^2\) cone is approximately 10.5 cm at SAD. Another complication is the accuracy permitted for field sizes. Typically field sizes are defined to the nearest millimeter so a full field width of 10.5 cm cannot be divided into equivalent left and right field widths. Therefore, full field widths need to be approximated with the nearest even millimeter. The decision on the actual size can be based on measured beam profiles.

Virtual SSD is determined in air with cross-beam profiles measured at source to chamber distances from 100 to 120 cm. The full width half maximums, which by definition determine field size, are plotted against the chamber distance, and the intercept of the linear fit curve through the data determines the virtual SSD. In other words the field size is back-projected to a common point. This value is not particularly sensitive or critical to the model. The other value that can be determined with in-air measurements is sigma-theta-x (\(\sigma_{0x}\)). This value is critical to the model as it controls the angular scattering of electrons in air and thus affects the entrance of a beam on a surface. This is of particular importance for extended treatment distances or for sloped surfaces. The widths of the 80%-20% penumbra from the same in-air profile data are plotted versus isocenter to chamber distance. The slope of the linear fit curve is then multiplied by 0.595 to determine \(\sigma_{0x}\). Alternatively, relative dose profile measurements in water were used to
determine this value as it affects the size of the penumbra region in water as well as in air. Its effect was seen in the beam modeling which was done based on the profiles measured in a water phantom.

Transmission factors through the Cerrobind cutout material were measured at $R_p + 2$ to determine the photon transmission in the blocked regions of a clinical electron field. Any dose readings at such a depth or region are deemed to be photon contamination. In addition to dose profiles in water described next, these values allow Pinnacle to add in the contribution of photon contamination inside and outside of the clinical field. Output factors are measured at $d_{max}$ and normalized to the reference field, which is an open 10x10 cm$^2$ beam at 100 cm SSD. Factors must be determined for all the typical clinical settings with variable applicator, SSD, and blocking percentage. The input values of these for Pinnacle must combine all these factors and must be normalized to the reference field. It is recommended that each applicator and block insert be measured at all distances to more accurately account for the differing angular scatter of electrons.

An alternative approximation that was used was to measure each output factor individually and to derive other output factors based on these relationships, for example measuring extended SSD factors with the 6x6 cone and applying such factors to all other block cutouts within that cone. Such factors were available with the existing dosimetry system. Relative output factors for Pinnacle were calculated by first combining extended SSD factors with the applicator factors. The blocking factors were then applied to each combination of SSD and applicator. Percent depth dose and orthogonal profiles were measured in a water phantom at 100 cm SSD for field sizes of 2, 3, 4, 5, 6, 8, 10, 14, 20,
and 25 cm using block cutouts if needed with the appropriate applicator. These measurements are necessary for calculating relative isodose curves within a phantom. The tail region of the electron depth dose curve which occurs after the practical range gives the relative contribution by the photon contamination. Additionally, photon contamination is included with a cross-plane profile at depth $R_p + 2$ determined from the percent depth dose curve with the largest applicator 25x25 cm².

Pinnacle requires that the scan depth $R_p + 2$ for photon contamination be the same for all field sizes for the energy. It is recommended by Khan in TG-25 that $R_p$ be determined using an applicator of at least 10x10 cm² at 100 cm SSD in a water phantom. Other profiles are not required but are generally recommended to account for any beam asymmetries resulting from one or two scattering foils and for evaluating that beam scatter is properly modeled. A complete set was taken for these beam models. Orthogonal profiles at approximate depths of $R_{90}$, $R_{70}$, and $R_{50}$ are used during modeling to check that the scatter within the water phantom is calculated accurately through the range of the beam. The effect of sigma-theta-x can be seen here. At greater depths the penumbra regions become increasingly wider, specifically when going from profiles at $R_{90}$ to $R_{50}$. A larger sigma-theta-x increases the rate of this spread so adjusting its value should basically fit calculated profiles to the measured cross and in-plane profiles. Additional orthogonal profiles taken around $d_{\text{max}}$ or at a depth less than $R_{90}$ are used to model asymmetries and “horns” within each beam defined by a different field size. Off-axis ratios are applied as a multiplicative factor to the computed profile to match to the measured profile. These factors typically affect a change of a few percent and at most several percent, usually occurring at the horns or in the “shoulders” of the computed
profile. These off-axis ratios are applied to fine tune the computed profile fit, and any large deviations require adjusting the sigma-theta-x. Off-axis ratios are calculated as the computed value at some offset from the central axis divided by the measured value at the same offset. Ratios are defined at evenly spread points through the profile typically in the region above 90% of the central axis value.

**TG-51**

The recommendations of this AAPM protocol pertain not only to ensuring output constancy of electron beams but also to performing absolute point dose measurements with a Farmer-type ionization chamber in a water phantom. The relevant equation for calculating dose given a charge reading from a dosimeter is given in the protocol along with measurements for various correction factors.

\[ D_w^Q = M_{i,r}^Q k'_{R_{50}} k_{\text{ecal}} N_{D,w}^{\text{Co}} \text{(Gy)} \]  

(14)

\[ M = P_{\text{ion}} P_{\text{dp}} P_{\text{elec}} P_{\text{pol}} M_{\text{raw}} \text{(C)} \]  

(15)

The charge reading is corrected for ion recombination, temperature and pressure, electrometer inaccuracies, and polarity effects. For beam quality and the conversion of the dose-to-water calibration factor per beam quality, correction factors pertain to the depth of \( R_{50} \) to adjust for any change in absorbed dose in water with the electron beam versus calibration done with a Cobalt-60 source. Ion recombination results from chamber inefficiency when collecting charge. The efficiency is gauged by measuring the same beam of radiation with two different bias voltages applied by the electrometer.

\[ P_{\text{ion}}(V_H) = \frac{1-(V_H/V_L)^2}{M_{\text{raw}}^H/M_{\text{raw}}^L - (V_H/V_L)^2} \]  

(16)
Temperature of the water phantom and pressure of the chamber air correct for conditions different from the standard calibration conditions.

\[
P_{TP} = \frac{273.2 + T}{273.2 + 22} \times \frac{101.33}{P}
\]  

(17)

The calibrated electrometer used for such measurements has a correction factor at unity. The correction for the polarity effect takes into account different readings when measuring with a positive or negative voltage. As stated in TG-25 as well, readings taken with each opposite bias should be averaged. With this protocol, the average is applied as a correction factor to the raw charge reading.

\[
P_{pol} = \left( \frac{M_{raw}^+ - M_{raw}^-}{2M_{raw}} \right)
\]  

(18)

The effective point of measurement with cylindrical ionization chambers following this protocol is at the central axis. Although the effective point of measurement is upstream or closer to the source, measurements for some arbitrary depth are performed with the central axis of the chamber at that depth. The protocol states with this formalism the gradient effects are included “explicitly by the term \( P_{gr}^0 \) for electrons”.

This protocol implicitly compensates for the effective point of measurement by performing an upstream shift of a depth ionization curve before determining \( I_{50} \) to calculate \( R_{50} \) and \( d_{ref} \) using a given equation. An alternative given is to convert a depth ionization curve to depth dose curve following TG-25 to explicitly determine \( R_{50} \).

\[
R_{50} = 1.029I_{50} - 0.06 \text{ (cm)} \quad \text{(for } 2 \leq I_{50} \leq 10 \text{ cm})
\]  

(19)

\[
R_{50} = 1.059I_{50} - 0.37 \text{ (cm)} \quad \text{(for } I_{50} > 10 \text{ cm})
\]  

(20)

\[
d_{ref} = 0.6R_{50} - 0.1 \text{ (cm)}
\]  

(21)
The remaining factors are based on the values of $R_{50}$ and $d_{ref}$. An additional measurement at a deeper depth for cylindrical chambers, shifted by half the chamber radius, determines $P^Q_{gr}$ which explicitly scales the reading from the effective point of measurement upstream of the central axis to the desired point of measurement at the central axis. Beam quality factor $k'_{R90}$ again pertains to $R_{50}$. $kecal$ values are listed in tables by cylindrical chamber. The chamber used was similar to PTW N30001.

\[
k'_{R90} = 0.9905 + 0.0710e^{-\frac{R_{90}}{3.67}}
\]

\[
P^Q_{gr} = \frac{M_{raw}(d_{ref} + 0.5r_{cay})}{M_{raw}(d_{ref})}
\]

With these factors, machine output should be calibrated to a standard which is typically 1 cGy per 1 MU at $d_{max}$. Simply, these factors allow for accurate calculation of dose to the point of measurement, and such factors can be applied to any other depths of measurement to obtain absolute dose to a point for comparison with the point dose calculations of the treatment planning system.

**Scanditronix-Wellhofer**

All data were taken with the Scanditronix-Wellhofer 3D water scanning system. The associated OmniPro-Accept software provided all the relevant beam data such as $d_{max}$, $R_{50}$, or penumbra width. The software was used to smooth all curves, and all profiles were centered according to the each half maximum which defines the field size of a beam. Additionally, depth ionization curves were converted to dose curves with the option of TG-25 or TG-51. The software was configured to follow the given TG when calculating beam characteristics.

**Cerrobend**
Cerrobind is a low-melting point alloy used for making custom blocks or conformal field shapes. This metal is used for both photon and electron field blocking although the thicknesses for electron beams are much less compared to for photon beams. The melting point is 70 °C, and it consists of 13.3% tin, 50% bismuth, 26.7% lead, and 10% cadmium. Cutouts or inserts are made by cutting the open field shape in foam, centering and placing the foam in an applicator template, and then pouring the molten Cerrobind around the placed foam into the shaped template. Pouring must be done slowly to avoid trapping air within the insert. Cooling must occur in a well ventilated area because of hazardous fumes.

**Beam Modeling**

Pinnacle requires field sizes evaluated at SAD of 100 cm. Therefore field sizes, such as 2x2, at the applicator must be scaled for the 5.3 cm drift distance to machine SAD. Field sizes can be adjusted by viewing the fit of a calculated profile at FWHM to its measured profile. In some situations where 1-2 mm precision is necessary, profiles need only be approximately parallel to each other in the penumbra region at the field edge. Percent depth dose curves are affected by the incident or most probable energy $E_{p,0}$ for an electron beam. It is recommended that this be derived from the $R_p$ with the equation, known as the Markus relation, provided in the manual. Likewise, the OmniPro-Accept software also gave these values based on PDD curves per energy. Adjusting the most probable energy may be necessary to ensure the depth dose curves fit well with all field sizes. This accuracy must be ensured before continuing with beam modeling. As stated earlier, the fit of the computed depth profiles to the measured profiles are affected mostly by the angular scattering variance defined by sigma-theta-x. Larger value
increases the penumbra region and reduces the flat region. An accurate sigma-theta-x allows the computed profiles for all depths to closely match the measured profiles. This must be true before continuing to the fine tuning process.

Fine tuning is performed with off-axis ratios, dividing the computed value by the measured value at an offset from central axis. This allows more accurate modeling of beam irregularities resulting from scattering foil design and machine collimation scatter. Large adjustments can adversely affect the associated depth dose curve for a field size. For example, giving a beam rounder “shoulders” would increase the penumbra region. This in turn would make the depth dose curve shallower. Oppositely making a beam more flat would improve percent depth dose at greater depths and push $d_{\text{max}}$ deeper, essentially reducing the penumbra width. Such ratios must be determined at approximately the depth of maximum dose or half $R_{90}$ depth. Pinnacle applies such factors to its models at half $R_{90}$ depth.

**TG-53**

The recommendations of this protocol pertain to commissioning and verification of treatment planning systems. The focus of this project is the electron pencil beam algorithm, specifically accuracy in dose calculation and isodose distribution with open fields and with block inserts. All regions of the beam were implicitly checked with percent depth dose curves, orthogonal profiles perpendicular to the central axis, and absolute dose calibration. Emphasis was placed on the absolute dose calculated to some normalization point and on the penumbra region.

Annual quality assurance (QA) was performed on the electron beams so certain measurements were used to check accuracy of the beam model. Depth dose curves for
open fields must show no significant change in beam energy, and profiles must show the beam to be within specifications for flatness and symmetry following TG-25 and TG-40. Output factors for applicators normalized to the 10x10 applicator were also checked. Extended SSD factors were measured as well at 105 and 110 cm for all open applicators and energies. The absolute dose calculated to a point was compared to dose measured following TG-51. The choice of depth was approximately $R_{80}$ which is close to the therapeutic depth $R_{90}$ but in the steep region of the depth dose curve. This allowed for the measurement to be more sensitive in detecting deviations in central axis dose calculation at depths providing a higher dose gradient. Additional point dose measurements were performed separately at extended surface distance for comparison with a Pinnacle treatment plan using a theoretical water phantom. Acceptable error is 2% and 2 mm.

Further verification was done to test the Pinnacle electron beam algorithm for accuracy with clinical field sizes with the three most commonly used applicators—6x6, 10x10, and 14x14. Following two sets of Electron Contract Working Group test cases on algorithm verification, measurements were performed with a low 6 MeV and high 12 MeV energy beam. Note that open block and small field beam data were already measured for the beam modeling. A range of field sizes from 25% to 75% blocked were evaluated at standard distance and at 105 cm SSD. A specific instance of oblique beam incidence with blocking was measured as well. Another specific instance was comparing dose for rectangular fields. The goal was to test the accuracy of the model to adjust the PDD and penumbra region according to the change in scatter resulting from blocking and treatment distance. Depth dose curves and profiles were measured for all these settings.

Commissioning
Electron beam commissioning can be performed by measuring absolute point doses for comparison with the absolute dose calculation by the treatment planning system. The acceptable error is 2% and 2 mm because the sharp dose gradient of electron beams will yield large errors even with good setup accuracy. Specifically, measuring even a millimeter upstream or downstream of the desired depth may result in a large percentage error. Such setup error of a few millimeters is clinically irrelevant and can be ignored.

Point measurements are taken for all energies and all applicators, at standard and extended source-to-surface distances. Although depth dose curves may shift upstream or downstream with field size, one depth of $R_{80}$ approximately is used per energy for all field and distance settings. The calibration procedure described in TG-51 must be performed first to correct for beam fluctuations on the day of commissioning as well as for accurately converting charge readings to absolute dose. The correction values included are as follows: ion recombination, temperature and pressure, gradient effects, beam quality factor, and beam output. These are the multiplicative factors applied to the charge reading in addition to the ADCL dose conversion factor.

**Test Scans**

The selection of energies and applicator sizes were for verification of clinical electrons. This was not necessarily only to verify the Pinnacle implementation of the Hogstrom algorithm but also to verify the model for the most oft used beams and cones. These were the 6, 9, and 12 MeV electron beams with 6x6, 10x10, and 14x14 cm$^2$ cones. Water phantom was used because it would be most accurate and convenient for measuring depth dose curves and testing the algorithm’s predictive accuracy. With
relative measurements, the use of a cylindrical chamber was appropriate even with the lowest energy 6 MeV beam.

Open applicators with the selected energies were measured. This test demonstrates the algorithm properly manipulates the beam data to predict accurate depth dose curves and profiles. In fact the calculation should be perfect because exactly such data were given to develop the beam model. Small block inserts were used with each selected applicator. Electron treatments are typically of this size—small and regularly shaped. A range of square fields 25% to 75% Cerrobend blocked was used. This would check the algorithm’s ability to predict the change in depth dose curve as the field size decreases within each applicator. The small field sizes entered as beam data were all for the 6x6 applicator, so additional testing was done with small field sizes within the larger 10x10 and 14x14 applicators. The small field sizes with the 6x6 applicator were tested to verify the algorithm similarly as the applicators with open cutouts. Another common field size shape is a rectangular cutout. This is particularly important as the Pinnacle algorithm bases its calculation of irregular fields on a rectangular field of best fit surrounding the irregular field. This ensures the square root method for combining two square field percent depth dose curves to predict the rectangular field depth dose curve is accurate.

In combination with all these field sizes at 100 cm SSD, the same depth dose curve and profile data were measured at extended SSD. This would test the algorithm’s ability to predict the effects of the scattering differences with the extended treatment distance on the depth dose curve and the change in penumbra width.

Relative Comparison
In all test cases, comparisons were made between measured and computed relative percent depth dose curves and profiles. The Pinnacle software allows for output of planar doses. The planar doses provide absolute dose information which must be normalized for relative comparison with measured data that are inherently relative. For calculation of a planar dose that encompasses the percent depth dose along with profiles in the cross-plane, an extra beam is placed with its beam central axis orthogonal to both the depth dose and the cross-plane profile. In other words this extra beam must have its central axis in the in-plane direction of the gantry. Once the planar doses are exported for each electron beam setting being tested, the RIT software can be used to extract depth and cross-plane profiles. The data are exported in ASCII format. As stated earlier, because the planar doses provide absolute dose, the depth dose values are normalized to its maximum, and the profiles are normalized to its central axis value. This allows for comparison with the relative water scans done with the Wellhofer scanning system. For the central axis depth dose, the acceptable limits are 2% and 2 mm in the therapeutic range after the depth of maximum dose. In the buildup region and in the penumbra region, the acceptable limits are 5% and 5 mm. For profiles, similar limits of 5% and 5 mm are allowed for the penumbra region, respectively. The flat region must follow specifications for flatness and symmetry.
Results

In-air Measurement

The data obtained with in-air measurements were performed with the 25x25 cm² applicator. The back-projection of the average FWHM versus source to chamber distance was obtained and subtracted from one hundred to give the virtual source location of each beam. Table I shows the determination of virtual source distances.

Table I. Virtual Source

<table>
<thead>
<tr>
<th>SDD (cm)</th>
<th>6 MeV</th>
<th>9 MeV</th>
<th>12 MeV</th>
<th>15 MeV</th>
<th>18 MeV</th>
<th>22 MeV</th>
</tr>
</thead>
<tbody>
<tr>
<td>105</td>
<td>27.89</td>
<td>27.83</td>
<td>27.83</td>
<td>27.82</td>
<td>27.89</td>
<td>27.85</td>
</tr>
<tr>
<td>110</td>
<td>29.24</td>
<td>29.17</td>
<td>29.17</td>
<td>29.16</td>
<td>29.26</td>
<td>29.2</td>
</tr>
<tr>
<td>115</td>
<td>30.63</td>
<td>30.55</td>
<td>30.54</td>
<td>30.54</td>
<td>30.64</td>
<td>30.58</td>
</tr>
<tr>
<td>120</td>
<td>32.03</td>
<td>31.92</td>
<td>31.92</td>
<td>31.9</td>
<td>32.04</td>
<td>31.97</td>
</tr>
<tr>
<td>V-SSD</td>
<td>95.68</td>
<td>96.66</td>
<td>96.73</td>
<td>97.10</td>
<td>95.48</td>
<td>96.22</td>
</tr>
</tbody>
</table>

Average penumbra widths were recorded from the Wellhofer software for calculation of the angular scattering variance. During the modeling process, these data could not be used, as sigma-theta-x had to be tuned for modeling of the beam profiles. The values used in modeling are given in Table II with the measured values.
Table II. Sigma-theta-x

<table>
<thead>
<tr>
<th>IDD (cm)</th>
<th>6 MeV</th>
<th>9 MeV</th>
<th>12 MeV</th>
<th>15 MeV</th>
<th>18 MeV</th>
<th>22 MeV</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0.78</td>
<td>0.63</td>
<td>0.54</td>
<td>0.5</td>
<td>0.45</td>
<td>0.43</td>
</tr>
<tr>
<td>5</td>
<td>1.31</td>
<td>0.81</td>
<td>0.71</td>
<td>0.61</td>
<td>0.56</td>
<td></td>
</tr>
<tr>
<td>10</td>
<td>1.89</td>
<td>1.43</td>
<td>1.1</td>
<td>0.96</td>
<td>0.81</td>
<td>0.71</td>
</tr>
<tr>
<td>15</td>
<td>2.47</td>
<td>1.85</td>
<td>1.44</td>
<td>1.23</td>
<td>1.02</td>
<td>0.9</td>
</tr>
<tr>
<td>20</td>
<td>3.06</td>
<td>2.3</td>
<td>1.78</td>
<td>1.52</td>
<td>1.28</td>
<td>1.09</td>
</tr>
</tbody>
</table>

In-air $\sigma_{\theta x}$

| Model $\sigma_{\theta x}$ | 0.068 | 0.050 | 0.037 | 0.030 | 0.025 | 0.020 |
| Model $\sigma_{\theta x}$ | 0.090 | 0.085 | 0.075 | 0.060 | 0.055 | 0.045 |

Photon Contamination Measurement

The measurements were performed in a water phantom using the FC-65P ionization chamber also used for all absolute dose measurements and dose calibration procedures. Each $R_{p}+2$ depth used for measurement corresponded to the practical range of the 10x10 percent depth dose curve input for the model of each beam. The measured transmission data are shown in Table III with the given measurement settings.

Table III. Cutout Transmission

<table>
<thead>
<tr>
<th>MeV</th>
<th>MU</th>
<th>Depth (cm)</th>
<th>Open (nC)</th>
<th>Block (nC)</th>
<th>Transmission</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>200</td>
<td>5.35</td>
<td>0.3244</td>
<td>0.3415</td>
<td>0.1981</td>
</tr>
<tr>
<td>9</td>
<td>200</td>
<td>6.62</td>
<td>0.3938</td>
<td>0.3942</td>
<td>0.2560</td>
</tr>
<tr>
<td>12</td>
<td>200</td>
<td>8.04</td>
<td>0.8201</td>
<td>0.8211</td>
<td>0.5261</td>
</tr>
<tr>
<td>15</td>
<td>200</td>
<td>9.39</td>
<td>1.4190</td>
<td>1.4180</td>
<td>0.9057</td>
</tr>
<tr>
<td>18</td>
<td>200</td>
<td>10.85</td>
<td>1.3220</td>
<td>1.3220</td>
<td>0.8800</td>
</tr>
<tr>
<td>22</td>
<td>200</td>
<td>13.36</td>
<td>1.9770</td>
<td>1.9630</td>
<td>1.2570</td>
</tr>
</tbody>
</table>

Beam Output Correction Factors

As stated earlier, Pinnacle requires output factors for combinations of applicator, treatment distance, and blocking, all normalized to the output of the reference field of 10x10 at 100 cm SSD with the open insert. Data from the present dosimetry book were
used to determine all necessary output factors. Additionally, such factors are checked annually for accuracy, and a commissioning process would also verify these factors.

The 6 MeV data are shown here separately for each treatment situation followed by the combined factors input into the planning system. Pinnacle further requires the depth at which the beam delivers a known dose, typically at $d_{\text{max}}$ where the output is 1 cGy per 1 MU. These factors were verified with commissioning measurements. Tables IV, V, and VI show beam data included in the dosimetry book used for monitor unit calculation. The combined data for 6 MeV electron beam shown in Table VII are provided to the planning system.

Table IV. Applicator Factors

<table>
<thead>
<tr>
<th></th>
<th>6x6</th>
<th>10x10</th>
<th>14x14</th>
<th>20x20</th>
<th>25x25</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.916</td>
<td>1.000</td>
<td>1.003</td>
<td>1.008</td>
<td>1.022</td>
<td></td>
</tr>
</tbody>
</table>

Table V. SSD Factors

<table>
<thead>
<tr>
<th>SSD (cm)</th>
<th>6x6</th>
<th>10x10</th>
<th>14x14</th>
<th>20x20</th>
<th>25x25</th>
</tr>
</thead>
<tbody>
<tr>
<td>100</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
</tr>
<tr>
<td>105</td>
<td>0.864</td>
<td>0.882</td>
<td>0.901</td>
<td>0.907</td>
<td>0.893</td>
</tr>
<tr>
<td>110</td>
<td>0.752</td>
<td>0.786</td>
<td>0.807</td>
<td>0.824</td>
<td>0.820</td>
</tr>
<tr>
<td>115</td>
<td>0.668</td>
<td>0.710</td>
<td>0.739</td>
<td>0.752</td>
<td>0.725</td>
</tr>
<tr>
<td>120</td>
<td>0.585</td>
<td>0.641</td>
<td>0.674</td>
<td>0.680</td>
<td>0.663</td>
</tr>
</tbody>
</table>

Table VI. Blocking Factors

<table>
<thead>
<tr>
<th>Blocking</th>
<th>6x6</th>
<th>10x10</th>
<th>14x14</th>
<th>20x20</th>
<th>25x25</th>
</tr>
</thead>
<tbody>
<tr>
<td>25%</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
</tr>
<tr>
<td>50%</td>
<td>0.983</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
</tr>
<tr>
<td>75%</td>
<td>0.877</td>
<td>0.998</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
</tr>
<tr>
<td>90%</td>
<td>0.723</td>
<td>0.865</td>
<td>0.971</td>
<td>1.000</td>
<td>1.000</td>
</tr>
</tbody>
</table>
Table VII. 6 MeV Output Correction Factors

<table>
<thead>
<tr>
<th>SSD (cm)</th>
<th>blocking</th>
<th>6x6</th>
<th>10x10</th>
<th>14x14</th>
<th>20x20</th>
<th>25x25</th>
</tr>
</thead>
<tbody>
<tr>
<td>100</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>25%</td>
<td>0.916</td>
<td>1.000</td>
<td>1.003</td>
<td>1.008</td>
<td>1.022</td>
<td></td>
</tr>
<tr>
<td>50%</td>
<td>0.900</td>
<td>1.000</td>
<td>1.003</td>
<td>1.008</td>
<td>1.022</td>
<td></td>
</tr>
<tr>
<td>75%</td>
<td>0.803</td>
<td>0.998</td>
<td>1.003</td>
<td>1.008</td>
<td>1.022</td>
<td></td>
</tr>
<tr>
<td>90%</td>
<td>0.662</td>
<td>0.865</td>
<td>0.974</td>
<td>1.008</td>
<td>1.022</td>
<td></td>
</tr>
<tr>
<td>105</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>25%</td>
<td>0.791</td>
<td>0.882</td>
<td>0.903</td>
<td>0.915</td>
<td>0.913</td>
<td></td>
</tr>
<tr>
<td>50%</td>
<td>0.778</td>
<td>0.882</td>
<td>0.903</td>
<td>0.915</td>
<td>0.913</td>
<td></td>
</tr>
<tr>
<td>75%</td>
<td>0.694</td>
<td>0.880</td>
<td>0.903</td>
<td>0.915</td>
<td>0.913</td>
<td></td>
</tr>
<tr>
<td>90%</td>
<td>0.572</td>
<td>0.763</td>
<td>0.877</td>
<td>0.915</td>
<td>0.913</td>
<td></td>
</tr>
<tr>
<td>110</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>25%</td>
<td>0.688</td>
<td>0.786</td>
<td>0.810</td>
<td>0.830</td>
<td>0.838</td>
<td></td>
</tr>
<tr>
<td>50%</td>
<td>0.677</td>
<td>0.786</td>
<td>0.810</td>
<td>0.830</td>
<td>0.838</td>
<td></td>
</tr>
<tr>
<td>75%</td>
<td>0.604</td>
<td>0.785</td>
<td>0.810</td>
<td>0.830</td>
<td>0.838</td>
<td></td>
</tr>
<tr>
<td>90%</td>
<td>0.498</td>
<td>0.680</td>
<td>0.786</td>
<td>0.830</td>
<td>0.838</td>
<td></td>
</tr>
<tr>
<td>115</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>25%</td>
<td>0.612</td>
<td>0.710</td>
<td>0.741</td>
<td>0.758</td>
<td>0.740</td>
<td></td>
</tr>
<tr>
<td>50%</td>
<td>0.601</td>
<td>0.710</td>
<td>0.741</td>
<td>0.758</td>
<td>0.740</td>
<td></td>
</tr>
<tr>
<td>75%</td>
<td>0.537</td>
<td>0.709</td>
<td>0.741</td>
<td>0.758</td>
<td>0.740</td>
<td></td>
</tr>
<tr>
<td>90%</td>
<td>0.442</td>
<td>0.615</td>
<td>0.720</td>
<td>0.758</td>
<td>0.740</td>
<td></td>
</tr>
<tr>
<td>120</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>25%</td>
<td>0.536</td>
<td>0.641</td>
<td>0.676</td>
<td>0.685</td>
<td>0.678</td>
<td></td>
</tr>
<tr>
<td>50%</td>
<td>0.527</td>
<td>0.641</td>
<td>0.676</td>
<td>0.685</td>
<td>0.678</td>
<td></td>
</tr>
<tr>
<td>75%</td>
<td>0.470</td>
<td>0.640</td>
<td>0.676</td>
<td>0.685</td>
<td>0.678</td>
<td></td>
</tr>
<tr>
<td>90%</td>
<td>0.388</td>
<td>0.554</td>
<td>0.657</td>
<td>0.685</td>
<td>0.678</td>
<td></td>
</tr>
</tbody>
</table>

Beam Modeling

Beam data input to the planning system were checked for consistency. Data sets measured on separate occasions were compared. Although annual QA of the Elekta SL-25 showed no significant change in electron beams, comparing the data sets provided confidence the input data set were accurate. Percent depth dose curves for the 10x10 applicator were measured after machine service August 2007. Annual QA for all applicators was performed August 2006. Percent depth dose curves for all applicators and all cutouts as recommended by Pinnacle were measured August 2007. Then the original data set with all recommended depth curves and profiles were measured September 2003.
For profile data, the original beam data were used as they were within specification for flatness and symmetry. Additional comparison was done of the shoulder region to ensure there was no change in scatter in the penumbra region from the collimation system. For percent depth dose data, there were beam changes demonstrated in the data from September 2003, although all relevant data after this date show consistency. Data for all energies for applicators of at least 10x10 are consistent from the 2006 annual. For the rest of the fields less than 10x10 cm², the original data were used as the depth dose curves measured in August 2007 were essentially identical.

The change of only the larger field sizes is likely due to fluctuations in flatness and symmetry from each water scan performed in 2003 and then 2007. The characteristic change is the 2003 data have higher surface dose and greater practical range. “Horns” present in the outer regions of a beam will contribute greater scatter to the central axis at depth to increase range, and for low energy beams there will be more surface dose from scatter as well. As these regions of asymmetry are blocked with smaller field sizes, the beam becomes relatively more flat. This would be the reason the small field depth dose curves have remaining unchanged. Any changes are blocked out and unnoticed at the central axis of such cutouts.

**Commissioning**

The FC-65P ionization chamber was used for all commissioning measurements. Dose calibration following TG-51 was performed to determine the relevant correction factors. These correction factors shown in Table VIII were determined live at commissioning.
Table VIII. TG-51 Electron Beam Dosimetry

<table>
<thead>
<tr>
<th>MeV</th>
<th>R50</th>
<th>k(ecal)</th>
<th>k’R50</th>
<th>Pgr</th>
<th>Pion</th>
<th>cGy/MU</th>
<th>CF*</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>2.68</td>
<td>0.897</td>
<td>1.025</td>
<td>0.9832</td>
<td>1.007</td>
<td>1.002</td>
<td>0.9085</td>
</tr>
<tr>
<td>9</td>
<td>3.61</td>
<td>0.897</td>
<td>1.017</td>
<td>0.9937</td>
<td>1.006</td>
<td>1.010</td>
<td>0.9029</td>
</tr>
<tr>
<td>12</td>
<td>4.84</td>
<td>0.897</td>
<td>1.009</td>
<td>0.9942</td>
<td>1.005</td>
<td>1.012</td>
<td>0.8936</td>
</tr>
<tr>
<td>15</td>
<td>5.91</td>
<td>0.897</td>
<td>1.005</td>
<td>0.9928</td>
<td>1.006</td>
<td>1.007</td>
<td>0.8941</td>
</tr>
<tr>
<td>18</td>
<td>7.15</td>
<td>0.897</td>
<td>1.001</td>
<td>0.9928</td>
<td>1.007</td>
<td>1.015</td>
<td>0.8844</td>
</tr>
<tr>
<td>22</td>
<td>8.87</td>
<td>0.897</td>
<td>0.997</td>
<td>0.995</td>
<td>1.005</td>
<td>1.014</td>
<td>0.8819</td>
</tr>
</tbody>
</table>

* CF = kQ*Pion/(cGy/MU)

The ADCL factor for the ionization chamber is 4.885 cGy/nC, and the $C_{ip}$ were recorded at the time of each set of measurements. The measurements were taken within 2 mm of the approximate $R_{50}$, and all measurements were within 2% of the calculated values of the planning system. The standard and extended distance measurements for the 10x10 applicator are shown in Table IX and X, respectively. Other applicators showed similar dosimetric accuracy with both treatment distances. This is consistent with the algorithm’s use of extensive output factors and depth dose curves.

Table IX. Commissioning – 10x10 cone, 100 cm SSD

<table>
<thead>
<tr>
<th>MeV</th>
<th>Depth</th>
<th>Average</th>
<th>Ctp</th>
<th>CF</th>
<th>Nc</th>
<th>Measured</th>
<th>ADAC</th>
<th>% diff</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>2.1</td>
<td>18.205</td>
<td>1.019</td>
<td>0.908</td>
<td>4.885</td>
<td>82.307</td>
<td>81.0</td>
<td>1.588</td>
</tr>
<tr>
<td>9</td>
<td>3.1</td>
<td>16.550</td>
<td>1.019</td>
<td>0.903</td>
<td>4.885</td>
<td>74.365</td>
<td>74.5</td>
<td>-0.181</td>
</tr>
<tr>
<td>12</td>
<td>4.2</td>
<td>17.570</td>
<td>1.019</td>
<td>0.894</td>
<td>4.885</td>
<td>78.134</td>
<td>78.3</td>
<td>-0.212</td>
</tr>
<tr>
<td>15</td>
<td>5.0</td>
<td>17.875</td>
<td>1.019</td>
<td>0.894</td>
<td>4.885</td>
<td>79.536</td>
<td>79.9</td>
<td>-0.458</td>
</tr>
<tr>
<td>18</td>
<td>6.0</td>
<td>18.605</td>
<td>1.019</td>
<td>0.884</td>
<td>4.885</td>
<td>81.886</td>
<td>83.3</td>
<td>-1.727</td>
</tr>
<tr>
<td>22</td>
<td>7.1</td>
<td>18.720</td>
<td>1.019</td>
<td>0.882</td>
<td>4.885</td>
<td>82.162</td>
<td>83.8</td>
<td>-1.994</td>
</tr>
</tbody>
</table>
Table X. Commissioning – 10x10 cone, 105 cm SSD

<table>
<thead>
<tr>
<th>MeV</th>
<th>Depth</th>
<th>Average</th>
<th>Ctp</th>
<th>CF</th>
<th>Nc</th>
<th>Measured</th>
<th>ADAC</th>
<th>% diff</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>2.1</td>
<td>15.865</td>
<td>1.019</td>
<td>0.908</td>
<td>4.885</td>
<td>71.728</td>
<td>71.2</td>
<td>0.736</td>
</tr>
<tr>
<td>9</td>
<td>3.1</td>
<td>14.585</td>
<td>1.019</td>
<td>0.903</td>
<td>4.885</td>
<td>65.536</td>
<td>66.6</td>
<td>-1.624</td>
</tr>
<tr>
<td>12</td>
<td>4.2</td>
<td>15.610</td>
<td>1.019</td>
<td>0.894</td>
<td>4.885</td>
<td>69.418</td>
<td>70.7</td>
<td>-1.847</td>
</tr>
<tr>
<td>15</td>
<td>5.0</td>
<td>16.300</td>
<td>1.019</td>
<td>0.894</td>
<td>4.885</td>
<td>72.528</td>
<td>71.8</td>
<td>1.003</td>
</tr>
<tr>
<td>18</td>
<td>6.0</td>
<td>16.705</td>
<td>1.019</td>
<td>0.884</td>
<td>4.885</td>
<td>73.523</td>
<td>74.6</td>
<td>-1.464</td>
</tr>
<tr>
<td>22</td>
<td>7.1</td>
<td>16.790</td>
<td>1.019</td>
<td>0.882</td>
<td>4.885</td>
<td>73.691</td>
<td>74.1</td>
<td>-0.555</td>
</tr>
</tbody>
</table>

Clinical Test Cases

The percent depth dose comparisons are presented for the 6 and 12 MeV blocked fields at standard and extended treatment distances. Percent difference is presented as well for the entire data set. There is also the shift tolerance of 2 and 5 mm in the respective regions of the dose curves checked visually on each plot.

The results of the depth dose curve comparison are as expected. In all cases, the Pinnacle algorithm overestimates the low doses near the practical range. The prediction of the central axis curve with Cerrobend blocked fields is accurate in all cases at the standard treatment distances with a tolerance of 2% and 2 mm, shown in Figures 1-10. This is true even in the buildup region and the steep dose gradient region.
Figure 1. 6 MeV depth dose – 3x3 cutout, 6x6 cone, 100 SSD

![Graph of 6 MeV depth dose with Pinnacle, Wellhofer, and percent difference markers.]

---

Figure 2. 12 MeV depth dose – 3x3 cutout, 6x6 cone, 100 SSD

![Graph of 12 MeV depth dose with Pinnacle, Wellhofer, and percent difference markers.]

---

---

---
Figure 3. 6 MeV depth dose – 5x5 cutout, 10x10 cone, 100 SSD

---

Figure 4. 12 MeV depth dose – 5x5 cutout, 10x10 cone, 100 SSD

---
Figure 5. 6 MeV depth dose – 7x7 cutout, 14x14 cone, 100 SSD

![Graph showing 6 MeV depth dose for Pinnacle and Wellhofer with percent difference markers.]

---

Figure 6. 12 MeV depth dose – 7x7 cutout, 14x14 cone, 100 SSD

![Graph showing 12 MeV depth dose for Pinnacle and Wellhofer with percent difference markers.]

---
Figure 7. 6 MeV depth dose – 10x10 cutout, 14x14 cone, 100 SSD

Figure 8. 12 MeV depth dose – 10x10 cutout, 14x14 cone, 100 SSD
Figure 9. 6 MeV depth dose – 12x12 cutout, 14x14 cone, 100 SSD

Figure 10. 12 MeV depth dose – 12x12 cutout, 14x14 cone, 100 SSD
With extended treatment distances, the algorithm falters with the small field sizes although there is no demonstrable difference between the lower and higher energy beams shown in Figures 11 and 12. This is consistent with the inverse square correction used. With the small fields, the algorithm overestimates the upstream shift by its modeling of the increased scatter with air gap. Again this is expected and the results are within tolerance of 5% and 5 mm in the buildup and dose gradient regions. In the therapeutic range after the depth of maximum dose the prediction is generally accurate to within 2% and 2 mm. These cutouts are shown in Figures 13-20.

Figure 11. 6 MeV depth dose – 3x3 cutout, 6x6 cone, 105 SSD
Figure 12. 12 MeV depth dose – 3x3 cutout, 6x6 cone, 105 SSD

Figure 13. 6 MeV depth dose – 5x5 cutout, 10x10 cone, 105 SSD
Figure 14. 12 MeV depth dose – 5x5 cutout, 10x10 cone, 105 SSD

---

Figure 15. 6 MeV depth dose – 7x7 cutout, 14x14 cone, 105 SSD
Figure 16. 12 MeV depth dose – 7x7 cutout, 14x14 cone, 105 SSD

--- Pinnacle
---- Wellhofer
×××× Percent difference

Figure 17. 6 MeV depth dose – 10x10 cutout, 14x14 cone, 105 SSD

--- Pinnacle
---- Wellhofer
×××× Percent difference
Figure 18. 12 MeV depth dose – 10x10 cutout, 14x14 cone, 105 SSD

Figure 19. 6 MeV depth dose – 12x12 cutout, 14x14 cone, 105 SSD
Figure 20. 12 MeV depth dose – 12x12 cutout, 14x14 cone, 105 SSD

The rectangular field calculation is very accurate with the standard distance as shown in Figures 21 and 22, and although the 3x3 results for extended distance are poor, the results shown in Figures 23 and 24 for the 12x3 are accurate. It is possible the accuracy of the algorithm with the large 12x12 field at extended distance compensates for the inaccuracy of small field prediction. Also possible is the lateral scatter is not as adversely affected in the rectangular field and thus the sigma-theta-x approximation is acceptable.
Figure 21. 6 MeV depth dose – 12x3 cutout, 14x14 cone, 100 SSD

---

Pinnacle
---

Wellhofer

×××× Percent difference

---

Figure 22. 12 MeV depth dose – 12x3 cutout, 14x14 cone, 100 SSD

---

Pinnacle
---

Wellhofer

×××× Percent difference
Figure 23. 6 MeV depth dose – 12x3 cutout, 14x14 cone, 105 SSD

--- Pinnacle
--- Wellhofer
×××× Percent difference

Figure 24. 12 MeV depth dose – 12x3 cutout, 14x14 cone, 105 SSD

--- Pinnacle
--- Wellhofer
×××× Percent difference
The profile comparisons are presented for the 6 and 12 MeV blocked fields at standard and extended treatment distances. Percent difference is presented as well for the field width. The shift tolerance is 5 mm in the penumbra region checked visually on each plot.

The central region of these profiles at the depth of maximum dose are not detailed as this region need only be within specification of flatness and symmetry with the largest 25x25 cone. As stated earlier in the comparison of annual water scans with the original data scans, the flatness and symmetry of the beams were adjusted numerous times. Differences can be seen in the profile comparisons presented for commissioning.

The important region is the penumbra width affected by the scatter model. The calculations of blocked fields at standard treatment distances are accurate and well within the tolerance of 5% and 5 mm, shown in Figures 25-34.

Figure 25. 6 MeV profile – 5x5 cutout, 10x10 cone, 100 SSD
Figure 26. 12 MeV profile – 5x5 cutout, 10x10 cone, 100 SSD

Figure 27. 6 MeV profile – 7x7 cutout, 14x14 cone, 100 SSD
Figure 28. 12 MeV profile – 7x7 cutout, 14x14 cone, 100 SSD

Figure 29. 6 MeV profile – 10x10 cutout, 14x14 cone, 100 SSD
Figure 30. 12 MeV profile – 10x10 cutout, 14x14 cone, 100 SSD

Figure 31. 6 MeV profile – cross-plane, 12x3 cutout, 14x14 cone, 100 SSD
Figure 32. 12 MeV profile – cross-plane, 12x3 cutout, 14x14 cone, 100 SSD

Figure 33. 6 MeV profile – in-plane, 12x3 cutout, 14x14 cone, 100 SSD
At extended distance though, the planning system overestimates the in-air scatter and penumbra size as shown in Figures 35-44. This is consistent with the upstream shift of the depth dose curve at extended distance as shown earlier. The results are still within 5% and 5 mm in the penumbra region from 80% to 20% of central axis ionization.
Figure 35. 6 MeV profile – 3x3 cutout, 6x6 cone, 105 SSD

Figure 36. 12 MeV profile – 3x3 cutout, 6x6 cone, 105 SSD
Figure 37. 6 MeV profile – 5x5 cutout, 10x10 cone, 105 SSD

![Graph 6 MeV profile]

--- Pinnacle
--- Wellhofer
×××× Percent difference

Figure 38. 12 MeV profile – 5x5 cutout, 10x10 cone, 105 SSD

![Graph 12 MeV profile]

--- Pinnacle
--- Wellhofer
×××× Percent difference
Figure 39. 6 MeV profile – 12x12 cutout, 14x14 cone, 105 SSD

Pinnacle
--- Wellhofer
×××× Percent difference

Figure 40. 12 MeV profile – 12x12 cutout, 14x14 cone, 105 SSD

Pinnacle
--- Wellhofer
×××× Percent difference
Figure 41. 6 MeV profile – cross-plane, 12x3 cutout, 14x14 cone, 105 SSD

Figure 42. 12 MeV profile – cross-plane, 12x3 cutout, 14x14 cone, 105 SSD
Figure 43. 6 MeV profile – in-plane, 12x3 cutout, 14x14 cone, 105 SSD

Figure 44. 12 MeV profile – in-plane, 12x3 cutout, 14x14 cone, 105 SSD
Discussion

The sigma-theta-x values measured could not be used because the computed profiles were significantly different from the measured profiles. This “problem” has been encountered before and as discussed, the value can be used solely for modeling. It can be noted the modeling values are greater than those measured. One possible reason is the excessive size of the scanning field chamber used.

Many output factors must be entered manually, and as a result there may be errors in typing values into the system. This can be evaluated during use with secondary calculation checks which is required for dosimetry work. The planning electrons are based on the present reference dosimetry system so there is already a system for monitor unit calculation, even without commercial second-check software.

Furthermore the factors are checked during the commissioning process. The main purpose of the commissioning should be to ensure all beam data were properly entered into the system. Particularly in this case where data entered to the algorithm are directly used, accuracy is essential. Point dose measurements at \( R_{80} \) ensure the percent depth dose data are accurate in position in case the scanning chamber setup was inaccurate. If the chamber was not properly zeroed at the water surface during water scans, then the entire depth dose curve would be shifted either upstream or downstream. In conjunction with such relative depth dose data, the output factors are applied by the algorithm at \( d_{\text{max}} \) to calculate dose delivered per monitor unit at \( R_{80} \).

The whole modeling process was very tedious. An extensive amount of values needed to be entered into the model, specifically off-axis ratios for shaping the shoulder regions in addition to beam output factors for combinations of applicator, treatment
distance, and blocking. The commissioning process was somewhat excessive for the sole purpose of verifying a subset of output factors. A simpler method of commissioning would be to compare depth dose curves with annual data and then verify the output of the beam as one cGy per MU at $d_{max}$. If the depth dose curves match and the absolute dose at one point is verified, then the planning electrons can be considered equivalent to the clinical beams.

As for the beam data itself, it is important to recognize if depth dose and profile data are acceptable for use. The TG-25 among other publications gives descriptions of beam characteristics. This was used as a guideline for understanding upstream shifts of dose curves or fluctuations of output factors. The main focus here is what is practical and acceptable for clinical use. One difficulty in maintaining consistent data is adjustments of electron beam steering to correct flatness and symmetry. As seen in comparing old data to current data, large field depth dose curves have changed—the surface and falloff regions. The reason is that electron scatter provides significant dose contribution in electron beam therapy with this type of accelerator. Changes in in-air scatter determine the fluence and most probable energy at the patient surface. Because the beam has changed on entrance, the scatter in medium can be affected resulting in shifts or a completely different dose curve shape. There has been reasonable consistency with recent water scans but the beams have changed since the original beam data was measured in 2003. With the comparison to annual data though, there are certain profiles where there are large differences. Again, it should be accepted that as long as the shoulders are reasonably modeled, and the penumbra widths are accurate at various depths, the sigma-theta-x values and off-axis ratios have provided sufficient modeling control. With depth
dose curves, the main problem is shifting of the entire data set up or downstream. Clinical tolerances of a few millimeters are acceptable.

The profile data were accepted for use as long as the flatness and symmetry and shoulder regions were consistent. Then the calculated profiles had to have penumbra regions that were parallel to those of the measured profiles. Depth dose curves require a tighter tolerance and thus were re-measured and selected based on the physics of electron beams. The data had to be self-consistent so that for example the upstream shift of dose curves with reduced field size had to be accurate even though the entire data set may have been shifted due to inaccurate placement of the chamber at zero depth during scanning.

For modeling, there are several common approaches which purposely do not model all the nuances of the beam data. Asymmetric scans can be halved or averaged and made symmetric. Instead of measuring orthogonal profiles, one symmetric data set can be applied to both $X$ and $Y$ axes. The idea is that the beam may at any time be adjusted to improve flatness and symmetry. Also, it can be difficult to measure a complete set that is both accurate and self-consistent. It can be seen in the commissioning comparison with annual data; some profiles measured at annuals are rather asymmetric. Then the precision of the zero depth may change during the course of scanning.

The scatter model and the use of inverse square correction provide reasonable results in most clinically relevant cases. The results for blocked fields at standard distance were well within tolerance. Of particular importance is the rectangular field because of its use for approximating irregular-shaped fields. The expectation is that regardless of the applicator used, the Cerrobend blocking does an effective job of
blocking a larger beam size into a common smaller field size. For example calculating a 5x5 field size within the 10x10 applicator makes use of the input beam data for the 5x5 field within the 6x6 applicator. Naturally such a small field size would never be treated in a 10x10 applicator aside from the rectangular or irregular-shaped cutouts.

For extended treatment distances, the scatter model has been shown to be less accurate. Even though the sigma-theta-x values provide good matching with the standard distance, the same values cannot properly predict the penumbra region of the profiles at extended distance. The penumbra width is much wider and the scatter exaggerated. This would imply the sigma-theta-x values are too high. One possible solution would be to allow for different sigma-theta-x values per treatment distance. With the CMS planning system for instance, unique FMCS values can be assigned to each field size. The Pinnacle system allows for only one per energy. Such an addition though would complicate the modeling process as it would require additional water scans at extended distances. Again, tolerance of 5% and 5 mm is clinically acceptable for the penumbra region. Although, TG-25 recommends that extended treatment distances should be avoided.

The central axis depth dose is accurate at extended distance except with the small field sizes. This is a known problem with the inverse square correction and clinically, a small 3x3 field would never be treated at extended distance. Although the results are poor, it was shown the rectangular field depth dose could still be accurate. The more common fields at extended distance would be very long rectangular fields set diagonal across an applicator with one dimension being sufficiently large. As shown with the larger blocks, the depth dose curves are well within tolerance.
In the tail region, there is some overestimation of low doses. As with in-air scatter, path lengths and energy straggling are completely ignored. The result is the modeled beams have higher energy at depth following Harder’s equation along with electron fluence at depth that is unchanged from the fluence at the surface.
Conclusion

The goals of this project were to commission planning electrons in the Pinnacle treatment planning system and to further verify the pencil beam algorithm for blocked fields of clinical relevance. The objectives accomplished were measuring required beam data, developing a beam- and parameter-based electron model, commissioning the accuracy of dose calculation, and comparing measured and predicted dose distributions for unique Cerrobend blocked fields.

The pencil beam algorithm is accurate at the standard treatment distance for blocked fields. This encompasses most electron treatments. The algorithm is equally accurate along the central axis with extended treatment distances except in the non-critical case of very small fields. This is important because in most cases the patient surface is irregular. It is essential the algorithm be accurate with varying source-surface distance. The penumbra region of a beam is controlled by the Gaussian distribution used to approximate dose contribution of each pencil beam. The inaccuracies were demonstrated with the profile comparisons at extended distances. The only solution with the present system would be to model data taken at extended distances and creating separate beams with different sigma-theta-x constants.

The inaccuracies have been characterized and clinically, errors are within tolerance. The main advantage of this algorithm is that it provides a complete dose calculation to patient CT data. Dose to the entire target volume can be gotten. Another convenience is the shifting of the depth dose and changes in surface dose with Cerrobend blocking. In a dosimetry system with only reference beam data, such changes in beam characteristics would typically be ignored. Visualization of dose coverage allows for a
better decision on what beam energy to use and if bolus is necessary for boosting skin surface dose.
Bibliography


electrons employing a two-dimensional heterogeneity correction. Int J Radiat Oncol Biol Phys 10, 561-569.


planning systems using either a pencil beam or a Monte Carlo algorithm. Int J Radiat Oncol Biol Phys 63, 622-633.
Abstract

The purpose of this study was to commission planning electrons in the ADAC Pinnacle\textsuperscript{3} treatment planning system and to verify its accuracy with clinical Cerrobend blocked fields. Pinnacle utilizes the Hogstrom electron pencil beam algorithm to perform its electron dose calculation. Complete set of measurement beam data in water phantom was first acquired and verified for consistency. The data was then entered into the planning system to develop a measurement and parameter-based model per beam. After the system was commissioned for clinical use, additional verification measurements were performed with unique field blocking. Comparisons between measured and computed dose from Pinnacle demonstrated the calculation was accurate and within clinical tolerance for all field blocking at standard treatment distance. For extended distance, relative depth dose prediction was poor only in the case with very high percentage field blocking. Relative cross-beam dose results were also poorer but were still acceptable for clinical use.