PHILIPS

Pinnacle^3

Physics

Reference Guide

Release 16.4 (English)

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Medical Device Directive
Pinnacle³ Radiation Therapy Planning System is CE Marked to the Medical Device Directive 93/42/EEC.

CE 2797

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Device Description

The Pinnacle³® Radiation Therapy Planning (RTP) software is composed of several modules including the core Pinnacle³ functionality, Syntegra™, P³IMRT®, and AcQSim³™. The Pinnacle³ RTP software runs on a Solaris UNIX (or UNIX compliant) computer, which allows qualified medical personnel to enter patient data into the system, use that data to construct a plan for radiation therapy, and evaluate the plan. Optionally, the qualified medical personnel may output the plan in an electronic or printed form for use by other systems in the delivery of treatment to a patient.

Pinnacle³ includes networking capabilities to provide connectivity to other Pinnacle³, Syntegra, AcQSim³, or P³IMRT workstations, input devices and output devices, as well as access to the Pinnacle³ database from any Pinnacle³ workstation available on the network. The system can be run from a single workstation but has network capability to other Pinnacle³ workstations and to both input and output devices via local area network (LAN) or wide area network (WAN). The Pinnacle³ RTP System also has an enterprise configuration for larger sites requiring a centralized data center environment. This system consists of one or multiple rack mountable servers on which Pinnacle³ sessions run which enables display on any network accessible computer. The expected service lifetime of the software is five (5) years.

To enable use of the software, a license key must be issued by the installer after software installation has been completed. If you have any software issues, please contact your service representative: Customer Support (1-800-722-9377 for US and Canada), your local Philips Customer Service provider, or your local Distributor. Beginning with Pinnacle³ 16.4, you will need to provide your service representative with the equipment ID assigned to your software kit. Your equipment ID can be found in the **About** window and is used to direct your call to the appropriate service personnel.

Intended Use

Pinnacle³ Radiation Therapy Planning System is a software package intended to provide planning support for the treatment of disease processes, utilizing photon, proton, electron and brachytherapy techniques.

Indications for Use

Pinnacle³ Radiation Therapy Planning System is a software package intended to provide planning support for the treatment of disease processes. Pinnacle³ Radiation Therapy Planning System incorporates a number of fully integrated subsystems, including Pinnacle³ Proton, which supports proton therapy planning. The full Pinnacle³ Radiation Therapy Planning System software package provides planning support for the treatment of disease processes, utilizing photon, proton, electron and brachytherapy techniques.

Pinnacle³ Radiation Therapy Planning System assists the clinician in formulating a treatment plan that maximizes the dose to the treatment volume while minimizing the dose to the surrounding normal tissues. The system is capable of operating in both the forward planning and inverse planning modes. Plans generated using this system are used in the determination of the course of a patient's radiation treatment. They are to be evaluated, modified and implemented by qualified medical personnel.

Contraindications

There are no known contraindications regarding the use of the Pinnacle³ RTP system.

Intended Audience

This manual is written for qualified medical personnel trained as users of Pinnacle³ RTP systems. You should make sure that you have thoroughly read and completely understand the manuals and release notes that are delivered with the software. Keep this manual and all other manuals delivered with the software near your Pinnacle³ system and review them periodically. The initial installation procedure will be performed by a Field Service Engineer. If you suspect that your system has an error, discontinue its use and contact Customer Support or your local distributor.

Minimum Requirements

This section provides the minimum requirements concerning hardware and software related to running the current product release as intended.

Professional

- Solaris 11 version 11.3 or later with Oracle X6-2 *(Note for existing sites: 11.2 or later with Oracle X5-2)*
- 64GB RAM (based on number of concurrent users)
  - Additional RAM requires upgrade
  - Upgradeable to 384GB RAM
- Rack Management: 1U each server

Network

- 10,000/1,000/100 Mbps Base-T Ethernet
- 1,000/100 Mbps Base-T Ethernet for remote management (ILOM)
- 4/8/16GBs HBA ports supported for communication between Smart Enterprise Cluster servers and the SAN

Third-Party Software

- OVDC(VCC) is used to access Pinnacle^3
- Third-party software installations are not supported on any Pinnacle^3 system server without proper validation.

Smart Enterprise (Cluster Server Nodes)

- Solaris 11 version 11.3 or later with Oracle X6-2 *(Note for existing sites: 11.2 or later with Oracle X5-2)*
- Solaris Cluster required on cluster nodes
- 64GB RAM (based on number of concurrent users)
  - Additional RAM requires upgrade
  - Upgradeable to 384GB RAM
- Rack Management: 1U each cluster server

Smart Enterprise (Application Servers)

- Solaris 11 version 11.3 or later with Oracle X6-2 *(Note for existing sites: 11.2 or later with Oracle X5-2)*
- 64GB RAM (based on number of concurrent users)
  - Additional RAM requires upgrade
  - Upgradeable to 384GB RAM
- Rack Management: 1U each application server

Notes

Equipment specifications are subject to alteration without notice. All changes will be in compliance with regulations governing manufacture of medical equipment.

Use of Symbols in Labeling

Philips Healthcare complies with International Standards and FDA requirements for the use of symbols in labeling. An online glossary of symbols used by Philips Healthcare is available at http://www.symbols.philips.com.

General Device Warnings

Do not load non-system software onto the computer used by this system without the direct authorization of Philips Medical Systems. Feature performance and safety may be compromised.

To assure proper treatment, it is critical that a qualified medical person review and verify all system treatment plan parameters using an independent verification method prior to treating patients using the plan.

Report the occurrence of a serious incident in relation to this device to Philips Customer Support and the competent authority of the Member State in which the user and/or patient is established.

We recommend that you review TG40, TG53, and other pertinent radiation therapy treatment standards and incorporate those methods into your clinical practice to ensure that your use of the system results in the most accurate treatment plans. TG40, TG53, and other reports are available publicly at the American Association of Physicists in Medicine (AAPM) website.

- **Comprehensive QA for radiation oncology: Report of AAPM Radiation Therapy Committee Task Group 40. *Medical Physics* 21(4), 1994.**
- **American Association of Physicists in Medicine Radiation Therapy Committee Task Group 53: Quality assurance for clinical radiotherapy treatment planning. *Medical Physics* 25(10), 1998.**

The following clinical practices are recommended to verify the accuracy of each treatment plan:

- **An independent calculation of the monitor units for each beam of a plan and treatment time for each brachytherapy plan.**
- **Acquisition and review of portal images or review of multi-leaf collimator (MLC) leaf positions after import to the treatment system.**
- **A chart check prior to the plan being delivered or during the first week of treatment.**
- **Independent review of the treatment plan prior to the delivery.**
- **Cross-functional review of the plan in a weekly chart round.**
- **Manual verification of record and verify settings after transfer to the treatment machine.**
- **Verification of the SSD and field shape during patient setup.**

These reviews should be performed for a new plan or when a change is made to any component of a plan.

U.S.A. law: CAUTION: Federal law restricts this device to sale by or on the order of a physician.

This software product is a Medical Device.

Only qualified medical personnel should operate the system. New personnel should receive training prior to unsupervised operation of the system. For more information, contact Customer Support or your local distributor.

Serious injury to patients can result due to the misapplication of this product. Make sure that you thoroughly understand all the user instructions prior to using this device.

Pinnacle³ treatment plans may include the statement NOT FOR CLINICAL USE. Based on the machine or isotope data and the treatment plan, the software determined that the plan cannot be delivered clinically. Do not treat patients with plans that are not for clinical use.

Pinnacle³ includes sample data. This information is for reference purposes only. Do not treat patients with plans based on sample machines or other sample data.

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1 Introduction

This guide describes the physics algorithms used in the Pinnacle³ Treatment Planning System as well as beam and source data you need to collect. The guide includes descriptions of physics requirements and geometry definitions to help you collect the following types of beam and source data:

- Photon beams
- Stereotactic radiosurgery
- Electron beams
- Brachytherapy sources

In addition, this guide includes reference material that supplements the information in the *Pinnacle³ Physics Instructions for Use*.

The *Additional Reference Information* chapter includes a list of water phantom interfaces you can use with Pinnacle³, along with file handling procedures. More information will be supplied by your sales consultant and submitted on your site survey.

If you have questions about the information in this guide, contact Customer Support at 1-800-722-9377 (US and Canada) or your local distributor for assistance.

2 Physical Machine Characteristics

The information in this chapter supplements the multi-leaf collimator (MLC) information found in the *Photon, Stereotactic, and Electron Machine Definition* chapter in the *Pinnacle³ Physics Instructions for Use*.

Leaf/jaw overlap

When a control point requires that one or more leaf pairs be closed, and these leaf pairs are not completely covered by the jaws, the software pushes the leaf pair junctions behind one of the jaws. This prevents the transmission through the minimum dynamic leaf gap from contributing to the incident energy fluence.

The software uses the **Maximum leaf/jaw overlap** and **Minimum leaf/jaw overlap** fields found in the **MLC Editor** window to determine how far to push the leaves beneath the jaw.

> **NOTE**
> Closed MLC leaf pairs are pushed behind the jaws for machines in which interdigitation is allowed and MLC does not replace jaws. In the **MLC Editor** window, the **Allow opposing adjacent leaves to overlap** option must be set to **Yes**.
>
> For machines that do not allow interdigitation but use a flagpole, closed MLC leaf pairs are pushed behind the jaws to the maximum leaf jaw overlap.

Maximum leaf/jaw overlap

This option lets you specify the largest allowable distance beneath the jaws that any given leaf can extend. The software automatically attempts to push the leaf to the maximum leaf/jaw overlap. However, in cases where the field width is too large to accommodate this distance, the actual leaf/jaw overlap may be smaller than the maximum leaf/jaw overlap.

Minimum leaf/jaw overlap

This option lets you specify the smallest allowable distance beneath the jaws that any given leaf can extend. If the machine cannot meet this requirement due to the size of the field width, the software splits the beam.

The beams will be split if the following equation is satisfied:

$$FS > MTD - LG - LJO$$

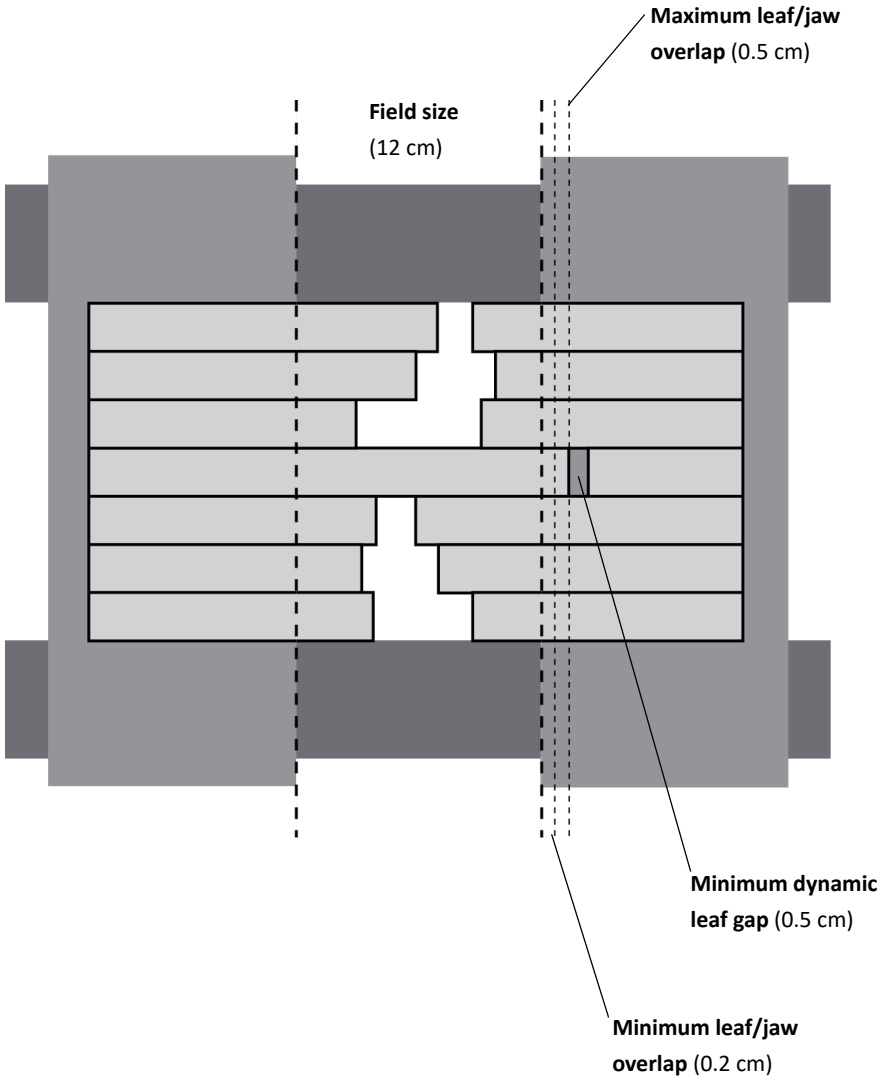
where *FS* is the field size along the leaf-travel direction, *MTD* is the maximum tip difference allowed by the machine, *LG* is the minimum dynamic leaf gap, and *LJO* is the minimum leaf/jaw overlap.

In some cases, the beams may be required to split even if the field size is smaller than the maximum tip difference overall. For example, given a field size of 14 cm, a maximum tip difference

of 14.5 cm, a minimum dynamic leaf gap of 0.05 cm, and a minimum leaf/jaw overlap of 0.5 cm, the beams would be required to split because the field size is larger than the maximum tip difference minus the minimum dynamic leaf gap and the minimum leaf/jaw overlap.

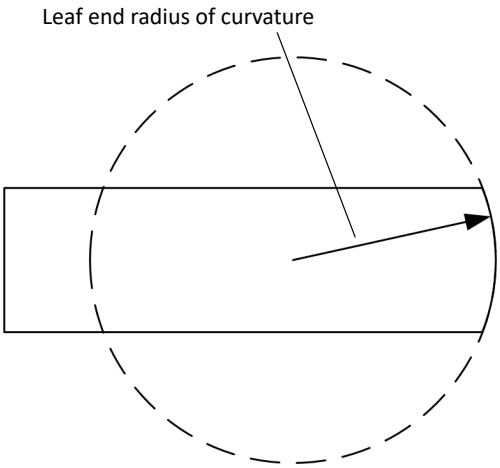
Leaf/jaw overlap example

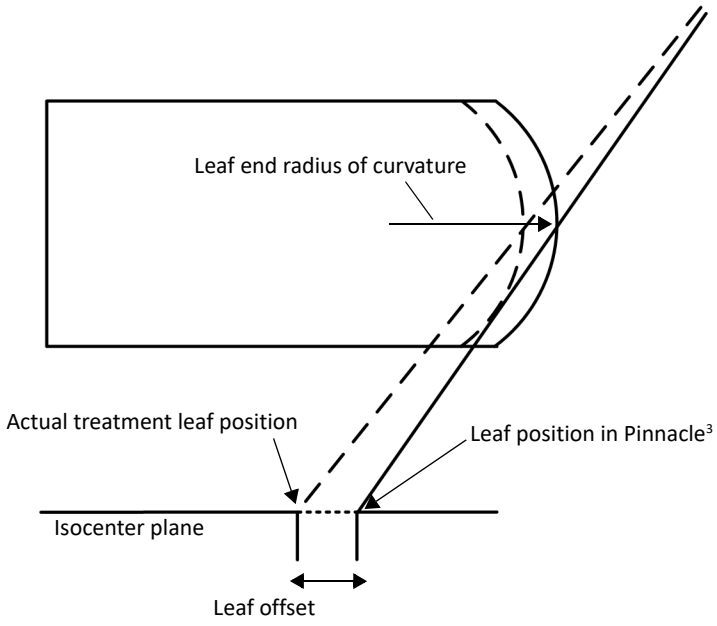
The example below illustrates a field size of 12 cm on a machine with a maximum tip difference of 14.5 cm. The **Minimum leaf/jaw overlap** is set to 0.2 cm, and the **Maximum leaf/jaw overlap** is set to 0.5 cm. In this example, the maximum tip difference allowed by the machine is large enough to accommodate the maximum leaf/jaw overlap.



Rounded leaf ends

The rounded leaf end is modeled as a circle segment that extends between the top and bottom of the MLC leaf (defined by MLC leaf thickness). The rounded leaf end model approximates the actual shape for the leaves that do not have a perfectly circular profile. The radiation through the leaf tip is attenuated by the thickness of the leaf traversed by the beam at each point in the tip.





The radius of curvature for the MLC leaves in published literature is about 8 cm for a Varian MLC[1], and this is the default in Pinnacle[3]. The radius of curvature for Elekta MLC leaves should be

approximately 12.2 cm for Beam Modulator machines and 15 cm for the other Elekta machines, according to the manufacturer. The Siemens MLC does not have rounded leaf ends, so set **MLC Has Rounded Leaves** to **No** in the **MLC Editor** window.

Due to the approximation, radii of 4-20 cm are reasonable if they generate the best fit to the measured data. To determine the optimal radius, vary the radius between 4 cm and 20 cm then compare the models with the measure profiles. Decreasing the radius widens the penumbra in the direction parallel to the leaf motion. Increasing the radius sharpens the penumbra.

The leaf position displayed, stored, and exported by Pinnacle^3 is the leaf position listed on the readout of the accelerator. It is the position of the tip of the leaf projected to the isocenter plane. The actual leaf position used in dose computation is the leaf position in Pinnacle^3 plus the leaf offset.

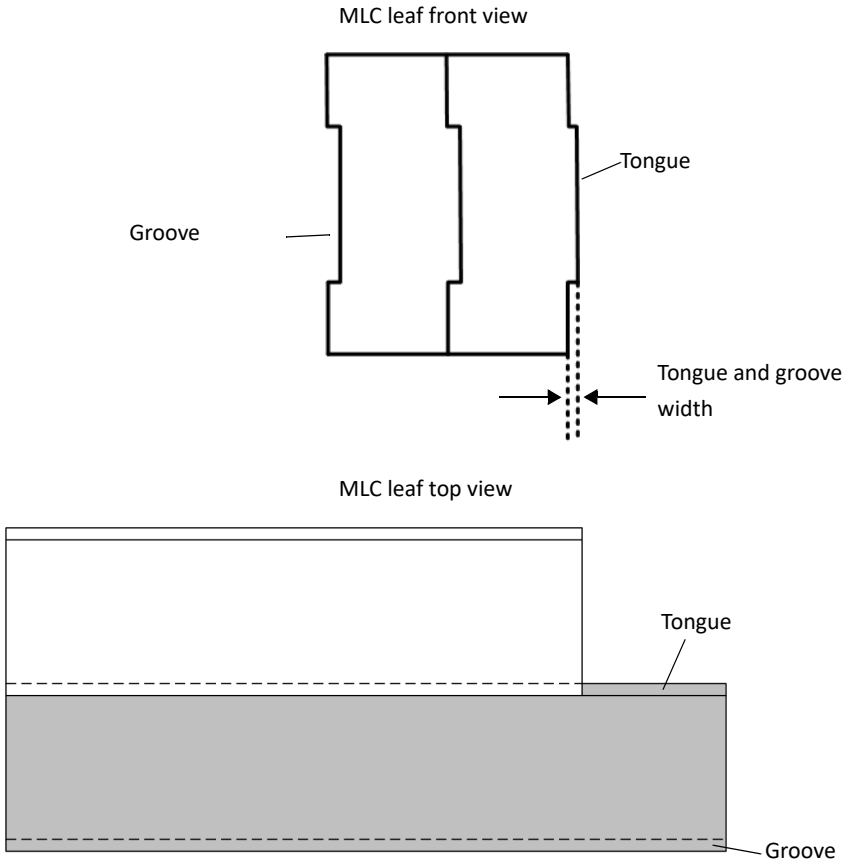
> **NOTE**
> The leaf offset is negative when the actual treatment leaf position is greater than the leaf position in Pinnacle^3. The leaf offset is positive when the actual treatment leaf position is less than the leaf position in Pinnacle^3.

Use the **Leaf Offset Calibration** table in the **Rounded Leaf End Specification** window to correct for the difference between the actual leaf positions and the leaf positions in Pinnacle^3. The table shifts the penumbra based on the leaf positions but does not affect the penumbra shape. For more information about the offset table, see Graves, et al.[2]

Tongue and groove

The tongue and groove are the parts of a leaf that overlap with its adjacent leaves. Each leaf has a tongue on one side and a groove on the other. The tongue and groove limit the interleaf leakage by attenuating the radiation by at least half of the MLC thickness. The tongue and groove width is the amount of overlap between the tongue of a leaf and the groove of its adjacent leaf. Set the tongue and groove width (in centimeters) on the **Leaves** tab in the **MLC Editor** window.

To model the tongue and groove effect, vary the widths from 0.005 to 0.200 cm and compare the models to the measured profiles. Increasing the width widens the penumbra in the direction perpendicular to the leaf motion. Decreasing the width sharpens the penumbra.



Interleaf leakage transmission

The interleaf leakage transmission, in conjunction with the tongue and groove characteristics, models the intersection of two adjacent MLC leaves. At the intersection, there is a small area where leakage occurs and the total thickness through the tongue and groove is less than the total thickness of the center of the MLC leaves. The **Additional interleaf leakage transmission** field on the **Leaves** tab of the **MLC Editor** window lets you specify the additional interleaf leakage transmission that should be added to the MLC transmission in the region where two adjacent leaves intersect. The range is 0–0.1 (10%). Published interleaf leakage transmission values are about 1%[1].

The additional interleaf leakage transmission can be determined by comparing computed profiles at maximum phantom resolution for small MLC fields with measured data perpendicular to the leaf motion in the physics tool. Use 1 cm x 1 cm and 2 cm x 2 cm MLC fields for best results. At a phantom resolution of 0.2 cm or less, enough detail is available to approximate the amplitude of interleaf transmission spikes.

References

LoSasso T., C. Chui, and C. Ling. 1998. Physical and Dosimetric Aspects of a Multileaf Collimation System Used in the Dynamic Mode for Implementing Intensity Modulated Radiotherapy. *Medical Physics* 25 (10):1919-1927.

Graves M., A. Thompson, M. Martel, D. McShan, and B. Fraass. 2001. Calibration and Quality Assurance for Rounded Leaf-end MLC Systems. *Medical Physics* 28(11) 2227-2233.

3 Photon Beam Physics & Measured Data Requirements

This section describes the photon beam physics dose calculation and the photon measured data requirements.

Convolution/superposition dose calculation

The convolution algorithm employed in the Pinnacle^3 planning system is based upon the work of Mackie (1985, 1988, 1990) and Papanikolaou (1993). Rather than correcting measured dose distributions, the algorithm computes dose distributions from first principles and, therefore, can account for the effects of beam modifiers, the surface of the patient, and tissue heterogeneities on the dose distribution.

The implemented algorithm has the following characteristics:

- Modeling the incident energy fluence as it exits the accelerator head.
- Head scatter is modeled using a Gaussian-based flattening filter scatter model (Ahnesjo, 1994).
- Primary transport for a beam is accomplished by projecting the incident energy fluence through the density representation of a patient to compute a TERMA (Total Energy Released per unit Mass) volume using polyenergetic rays, with water equivalent depth hardening and off-axis softening.
- Beam modifiers are handled with the following effects: primary attenuation, primary hardening, and primary “boosting” to account for scatter from the modifier. Wedges and compensators receive all three effects; blocks only attenuate the beam; bolus is treated as part of the patient.
- A three-dimensional superposition of the TERMA with a polyenergetic energy deposition kernel is used to compute dose. A ray-tracing technique is used during the superposition to incorporate the effects of heterogeneities on lateral scatter.
  - Polyenergetic kernel construction is based on the energy spectrum at the central axis and is determined during the modeling process.
  - To account for heterogeneities, the kernels are density-scaled during superposition. Superposition is performed using “collapsed cones.” The collapsed cones refer to the modeling of a cone in space using a single ray corresponding to the central axis of the cone. However, Pinnacle^3 does not use analytical functions to fit to the kernel (as described by Ahnesjo). Kernels are not tilted. Variable kernel sampling in zenith angle is also performed to reduce artifacts caused by non-tilted kernels, and to more uniformly distribute the energy amongst the rays.

    NOTE
    The current implementation of the photon dose algorithm uses an equivalent square field size for calculation of the OFc contribution to the output factor determination. The

algorithm does not explicitly distinguish between two rectangular fields, for example, 5 cm x 20 cm and 20 cm x 5 cm. The differences in head scatter between the two example field sizes would instead be accounted for with the Flattening Filter Scatter Source model.

An overview of the model

The Pinnacle^3 photon beam model parameters characterize the radiation exiting the head of the linear accelerator. The starting point for photon modeling is a uniform plane describing the incident energy fluence. Then Pinnacle^3 adjusts the fluence model to account for the flattening filter, the accelerator head, and beam modifiers such as blocks, wedges, and compensators.

- The “horns” in the beam produced by the flattening filter are modeled either by removing an inverted cone from the distribution or with an arbitrary profile that is radially symmetric about the central axis of the beam.
- Off-focus scatter produced in the accelerator head is modeled by defining a 2D Gaussian function as a scatter source and adjusting the incident energy fluence based on the portion of the scatter source visible from each point in the incident energy fluence plane.
- The geometric penumbra is modeled by convolving the fluence array with a focal spot blurring function.
- During planning, the shape of the field produced by blocks or multi-leaf collimators is cut out of the fluence array leaving behind the corresponding transmission through the shape-defining entity.
- Beam modifiers such as wedges and compensators are included in the fluence array by attenuating the energy fluence using the corresponding thickness of the modifier. For static wedges and compensators, a radiological depth array is also stored which allows for proper modeling of the beam hardening due to the presence of the beam modifiers during the projection of the incident fluence array.

Pinnacle^3 handles each of these aspects of the model using a parameter or set of parameters which it iteratively adjusts during the modeling process so that the dose computed by the model matches the dose generated from your machine.

During the modeling process, different regions of the measured depth doses and dose profiles are used to adjust the parameters which characterize the beam.

Measured data requirements for photon beams

The Pinnacle[3] photon dose algorithm is model-based rather than measurement-based. This allows more accurate dose computation in non-measurement configuration/geometry. Therefore, the measured data is used to characterize the beam attributes rather than to create extensive lookup tables of dose values.

During the beam modeling process, the measured beam data is used to compare how well the computed dose for a given measurement geometry matches the measured data for your machine. Once the model parameters have been adjusted to match the characteristics of your machine, the measured data is no longer used.

NOTE
When acquiring beam data needed for modeling, keep the data in different files or on separate disks. This avoids confusion when all the open field profiles are separated from the wedged field profiles for a specific energy, especially if you are entering a machine with multiple energies. You also may want to keep the depth dose profiles separate from the cross-plane and in-plane profiles.

WARNING
The accuracy of the treatment planning dose calculation depends on the quality of the data entered in the physics tool. If the quality of this data is poor, or if the computed dose is not verified with measured data, the dose calculation may be inaccurate. Please read and follow these instructions carefully.

Measured data summary

For each energy on each machine, you will need to measure the following types of data. The specific measurement requirements are covered in the following sections. All data must be measured in a water phantom or a water equivalent phantom.

- Depth dose curves from which the energy spectrum and electron contamination are determined.
- Dose profiles for the determination of incident energy fluence inside the field.
- Dose profiles extending outside the field for the determination of scatter dose and jaw transmission from the machine head components.
- Various profiles with the MLC aperture to verify dose accuracy for MLC fields.
- The calibration output factor.
- Relative output factors.
- The tray transmission factor and block and tray transmission factor.

In addition, the following data must be measured for each wedge on each machine at each energy: depth doses and dose profiles measured with the wedge in place, and wedged field relative output factors.

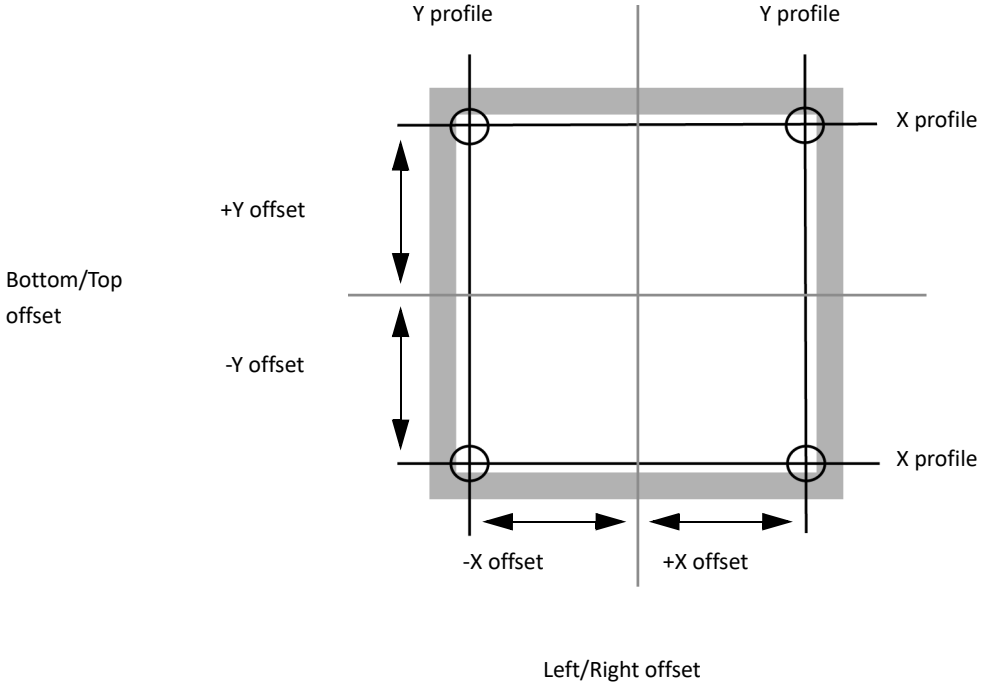
NOTE
If you have equipment that has the capacity, you can measure profiles and output factors for field sizes smaller than those recommended in the following sections. You

may need to adjust the dose grids appropriately to fully utilize this data. In general, you should measure output factors down to the smallest field sizes that you plan to use clinically.

Optional profiles for the corners of a field

If you want to evaluate your model in the corners of the field, we recommend creating additional X and Y profiles for the largest field in the measurement geometry list. To do this, create four profiles, one for each “side” of the field. The intersection of the profiles will show you the dose in the corners of the field. To enter the offset from the central axis, use the Bottom/Top Offset for X profiles and the Left/Right Offset for Y profiles.

NOTE
Some scanning software automatically computes profiles for the corners of the field.



Data collection considerations for Elekta and Siemens machines

This section contains the general procedures you need to follow to collect the jaw and MLC transmission data for Elekta and Siemens machines. The remainder of this chapter contains detailed information about all of the data you need to collect.

Elekta

In order for Pinnacle^3 to model the MLC transmission accurately (for non-Beam Modulator machines), the MLC must be positioned at least 1 cm beyond the backup jaw. The normal operation mode of an Elekta machine (with the MLC leaves aligned with the backup jaw) makes it difficult to measure only the jaw transmission or only the MLC transmission.

Open field data should be collected in the same mode in which the machine will be used. The MLC-defined fields can be measured in Elekta Service Mode.

After you have imported the profiles into Pinnacle^3, be sure to set the **MLC tracks jaws** option to **MLC tracks XY jaws for open fields** in the **MLC Editor** window.

Once you have completed automodeling, adjust the MLC transmission and the X and Y jaw transmissions as necessary, and then compute dose and validate your model. The transmission values you enter should approximate the small-field transmission values that are appropriate for your machine. The MLC transmission value is the most critical.

If you do not feel comfortable measuring the MLC-defined fields in Elekta Service Mode, you can approximate the transmission values in the physics tool. Enter the correct MLC and jaw transmission values prior to starting automodeling. Once you have completed automodeling, adjust the MLC transmission and the X and Y jaw transmissions as necessary, and then compute dose and validate your model.

Siemens

It is difficult to measure individual transmission values for the jaws and MLC on the Siemens machines for which the MLC replaces the jaw because of the way the jaws and MLC work together.

Open field data should be collected in the same mode in which the machine will be used.

You do not need to measure MLC-defined profiles as described in this chapter because the open field data allows automodeling of the MLC transmission.

Once you have completed automodeling, adjust the MLC transmission and the Y jaw transmission as necessary, and then compute dose and validate your model. The transmission values you enter should approximate the small-field transmission values that are appropriate for your machine. The MLC transmission value is the most critical.

Open field depth dose measurements for energy spectrum and electron contamination determination

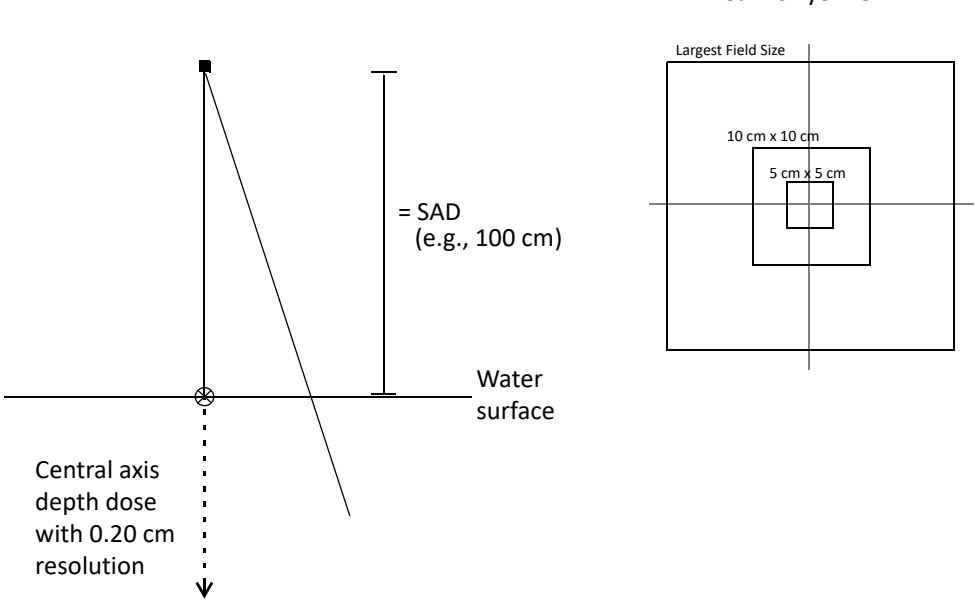
For each energy on each machine, central axis depth dose curves must be measured using open fields. In the beam modeling process, the depth dose curves are used to determine the energy spectrum. The setup information for the depth dose measurements is provided below.

NOTE
The measurement resolution (0.20 cm) is a recommended value. Data measured using higher resolutions may be used, but the calculation time will increase. The use of lower resolution data (increments of 0.30 cm or greater) can result in poor models and inaccurate clinical results.

Measurement Type	Central axis depth dose
Field Sizes – Open field (non-fixed jaw machines)	• 5 cm x 5 cm • 10 cm x 10 cm • 20 cm x 20 cm • 30 cm x 30 cm • 20 cm x 5 cm • 5 cm x 20 cm • 40 cm x 40 cm or the largest possible field size
Field Sizes – MLC field (optional) (non-fixed jaw machines)	Jaw sizes: • 20 cm x 20 cm; MLC fields: • 2 cm x 2 cm • 3 cm x 3 cm • 5 cm x 5 cm • 10 cm x 10 cm • 15 cm x 15 cm
Field Sizes – MLC field (fixed jaw machines)	• 2.4 cm x 2.4 cm • 3.2 cm x 3.2 cm • 4.8 cm x 4.8 cm • 5.6 cm x 5.6 cm • 8.0 cm x 8.0 cm • 10.4 cm x 10.4 cm • 15.2 cm x 15.2 cm • the largest possible field size
Measurement Limits	Minimum 0 to 25 cm depth
Resolution	0.20 cm increments
Setup	SSD can be defined by the user. Isocenter at the water surface or at a depth of 10 cm is recommended.
Modifiers	None (open field)

NOTE
IMRT can involve the use of very small, heavily blocked fields (2 cm x 2 cm or smaller). Be sure to measure field sizes comparable to those you expect to treat. Also, if you

expect to use very small field sizes defined by the jaws, as opposed to the MLC leaves, you should also collect 2 cm x 2 cm and 3 cm x 3 cm PDDs and profiles.



Measuring dose profiles for incident energy fluence inside field and outside field

These profiles will be used in adjusting the inside of field fluence and the outside of field transmission. The measured profiles should extend as far outside the field boundary as possible.

> **NOTE**
> The measurement resolution (0.20 cm) is a recommended value. Data measured using higher resolutions may be used, but the calculation time will increase. The use of lower resolution data (increments of 0.30 cm or greater) can result in poor models and inaccurate clinical results.

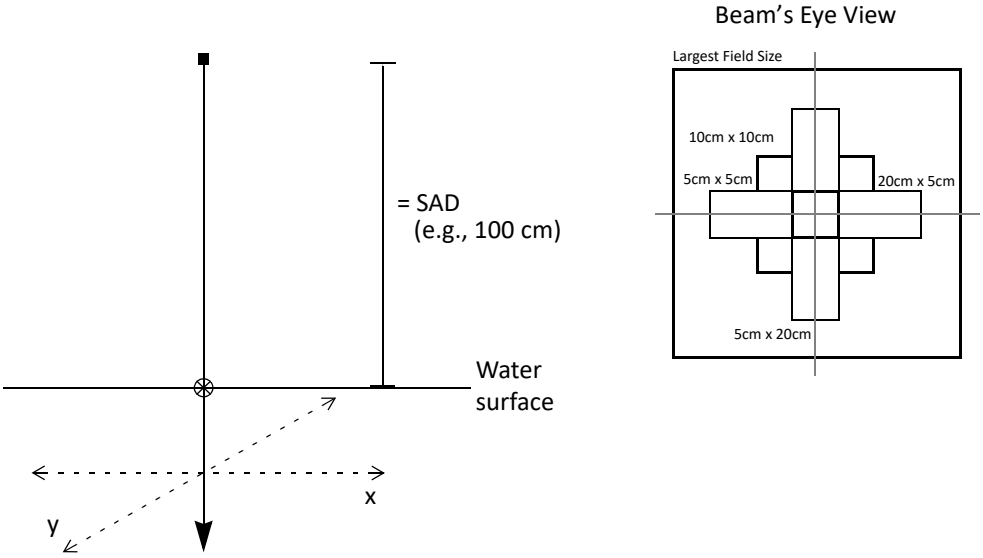
Measurement Type	Orthogonal (x and y) dose profiles through central axis.
Field Sizes* – Open field (non-fixed jaw machines)	• 5 cm x 5 cm • 10 cm x 10 cm • 20 cm x 20 cm • 30 cm x 30 cm • 20 cm x 5 cm • 5 cm x 20 cm • 40 cm x 40 cm or the largest possible field size
Field Sizes* – MLC field** † (non-fixed jaw machines)	Jaw sizes: • 20 cm x 20 cm MLC fields: • 2 cm x 2 cm • 3 cm x 3 cm • 5 cm x 5 cm • 10 cm x 10 cm • 15 cm x 15 cm
Field Sizes* – MLC field** † (fixed jaw machines)	• 2.4 cm x 2.4 cm • 3.2 cm x 3.2 cm • 4.8 cm x 4.8 cm • 5.6 cm x 5.6 cm • 8.0 cm x 8.0 cm • 10.4 cm x 10.4 cm • 15.2 cm x 15.2 cm • maximum field size
Depths	• d_{max} • 5 cm • 10 cm • 20 cm
Measurement Limits	Measurements should extend as far outside the field boundary as possible (minimum of 2 cm).
Resolution	0.20 cm or higher.
Setup	SSD can be defined by the user. Isocenter at the water surface or at a depth of 10 cm is recommended.
Modifiers	None (open field)

NOTE
IMRT can involve the use of very small, heavily blocked fields (2 cm x 2 cm or smaller). Be sure to measure field sizes comparable to those you expect to treat. Also, if you expect to use very small field sizes defined by the jaws, as opposed to the MLC leaves, you should also collect 2 cm x 2 cm and 3 cm x 3 cm PDDs and profiles.

NOTE
To model more accurately the outer corners of large fields, you may also measure one or two additional X and Y profiles for the largest field size. For these profiles, use a large offset from the central axis. For more information, see *Optional profiles for the corners of a field*.

NOTE
If you measure MLC field sizes other than those recommended above and you want to use those measurements for automodeling, make sure the MLC leaves extend past the jaws by at least 1 cm on one side or the measurements will not be used in automodeling.

NOTE
MLC scans should avoid interleaf leakage and abutted leaf ends. The software allows you to offset scans and to position closed leaf ends away from the CAX to avoid these geometries. See the *Importing and Entering Measured Beam Data* chapter in the *Pinnacle³ Physics Instructions for Use* for information.



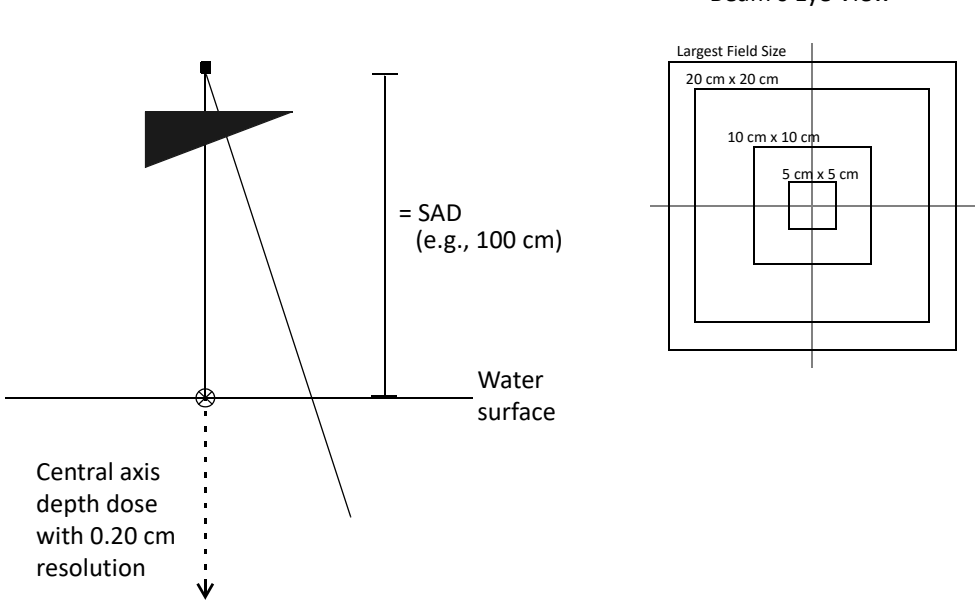
Wedged field depth dose measurements for adjusting the wedged field spectrum

These measurements are used in modeling the wedged field spectrum. If you will be using dynamic or motorized wedges, see *Requirements for dynamic wedges* and *Requirements for motorized wedges* for additional requirements to those listed below.

NOTE
The measurement resolution (0.20 cm) is a recommended value. Data measured using higher resolutions may be used, but the calculation time will increase. The use of lower resolution data (increments of 0.30 cm or greater) can result in poor models and inaccurate clinical results.

Measurement Type	Central axis depth dose
Field Sizes (non-fixed jaw machines)	• 5 cm x 5 cm • 10 cm x 10 cm • 20 cm x 20 cm • largest field size possible
Field Sizes – MLC field (fixed jaw machines)	• 2.4 cm x 2.4 cm • 3.2 cm x 3.2 cm • 4.8 cm x 4.8 cm • 5.6 cm x 5.6 cm • 8.0 cm x 8.0 cm • 10.4 cm x 10.4 cm • 15.2 cm x 15.2 cm • maximum field size
Measurement Limits	Minimum 0 to 25 cm depth
Resolution	0.20 cm increments
Setup	SSD can be defined by the user. Isocenter at the water surface or at a depth of 10 cm is recommended.
Modifiers	The wedge to be modeled.

NOTE
Be sure to measure field sizes comparable to those you expect to treat. Also, if you expect to use very small field sizes defined by the jaws, as opposed to the MLC leaves, you should also collect 2 cm x 2 cm and 3 cm x 3 cm PDDs and profiles.



Wedged field measurements for adjusting the wedge-specific model parameters

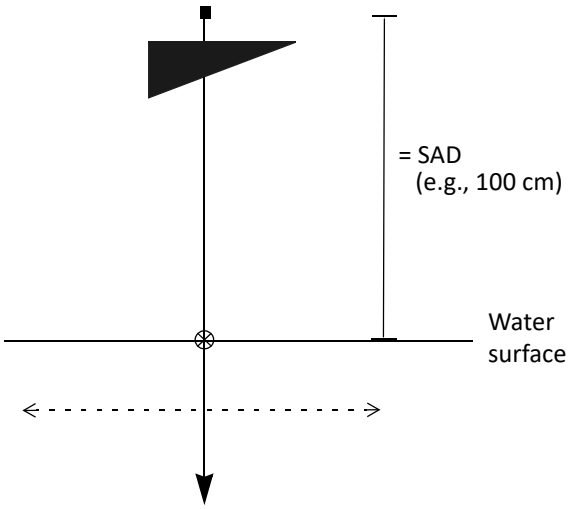
These profiles will be used in modeling the wedge-specific model parameters. If you will be using dynamic or motorized wedges, see *Requirements for dynamic wedges* and *Requirements for motorized wedges* for additional requirements to those listed below.

NOTE
The measurement resolution (0.20 cm) is a recommended value. Data measured using higher resolutions may be used, but the calculation time will increase. The use of lower resolution data (increments of 0.30 cm or greater) can result in poor models and inaccurate clinical results.

Measurement Type	Dose profiles through the central axis in the wedged direction and the non-wedged direction.
Field Sizes (non-fixed jaw machines)	In the wedged direction: • 5 cm x 5 cm • 10 cm x 10 cm • 20 cm x 20 cm • maximum field size
Field Sizes – MLC field (fixed jaw machines)	In the wedged direction: • 2.4 cm x 2.4 cm • 3.2 cm x 3.2 cm • 4.8 cm x 4.8 cm • 5.6 cm x 5.6 cm • 8.0 cm x 8.0 cm • 10.4 cm x 10.4 cm • 15.2 cm x 15.2 cm • maximum field size
Depths	• d_{max} • 5 cm • 10 cm • 20 cm (In the non-wedged direction, you only need to measure one depth for each field size to verify the model's validity.)
Measurement Limits	Measurements should extend well outside the field.
Resolution	0.20 cm
Setup	SSD can be defined by the user. Isocenter at the water surface or at a depth of 10 cm is recommended.
Modifiers	The wedge to be modeled.

NOTE
Be sure to measure field sizes comparable to those you expect to treat. Also, if you expect to use very small field sizes defined by the jaws, as opposed to the MLC leaves, you should also collect 2 cm x 2 cm and 3 cm x 3 cm PDDs and profiles.

NOTE
To model more accurately the outer corners of large fields, you may also measure one or two additional X and Y profiles for the largest field size. For these profiles, use a large offset from the central axis. For more information, see *Optional profiles for the corners of a field*.



Output factor measurements

To calibrate your machine, you measure the ionization at a specified reference depth, convert the measurement to dose at the same reference depth, and then divide it by the proper central axis percentage depth dose. The result determines the dose for a standard field size (usually 10 cm x 10 cm) and standard SSD (usually 100 cm, and equal to SAD). For fixed jaw machines, the standard field size is defined by the user and corresponds to an MLC-defined field instead of a jaw-defined field.

As the field size changes, so too does radiation output. To measure this change, divide the output at a specific depth (usually 10 cm) on the central axis for each field size by the output at the same depth on the central axis for the standard field. These ratios are called *output factors*.

For each energy on each machine, you need to measure a calibration output factor and relative output factors. In addition, you must measure wedged field output factors for each wedge that will be used with that energy.

Buildup and D_{max} doses

We strongly recommend that photon output factors be taken at 10 cm depth for all energies. This is due to the uncertainty of absolute dose computation and measurement errors at d_{max}.

Two effects contribute to dose uncertainty at d_{max} and in the buildup region:

- Contaminant electrons, which scatter off the flattening filter and collimators, enter the patient and boost the dose in the buildup region. In Pinnacle^3, an exponential function is used to model electron contamination and this electron dose is added to the photon dose. Because it is

difficult to measure the dose at d_{max}, the AAPM TG-51 protocol recommends a calibration depth of 10 cm.

- Inaccuracy and sampling problems can occur when discrete grids are used to model continuous functions. One of the inputs to the convolution/superposition model is the TERMA distribution, or total energy released per unit mass. To compute the dose, the dose spread array, or kernel, is convolved with the TERMA distribution. The TERMA must be known at all positions within the patient.

In Pinnacle^3, the TERMA distribution is computed by ray tracing through the patient model from a 2D incident energy fluence array. Pinnacle^3 automatically covers the entire patient model with a grid and computes the TERMA at each point in this grid. The resolution (voxel size) of this grid is generally set to match the resolution of the dose grid. Thus, if a 4 mm dose grid is selected, the TERMA grid will also have 4 mm resolution. When setting up the dose grid, an arbitrary resolution for the TERMA grid can also be selected on the **Dose Grid Specifications** window.

The TERMA distribution is very smooth, and this resolution is sufficient in most cases. However, 4 mm resolution may be too coarse when buildup dose or dose at d_{max} is critical. Consider a 4 MV beam with d_{max} at 1.2 cm depth. In a 4 mm TERMA grid, there are only 2 or 3 voxels between the surface and d_{max}. Clearly, this is a very coarse resolution if dose at d_{max} is critical.

If buildup dose or dose at d_{max} is critical, you can increase the TERMA grid resolution by specifying a higher resolution in the **Dose Grid Specifications** window of Pinnacle^3.

To minimize the dose uncertainties described above, please follow these guidelines:

- Do not measure output factors at d_{max}. Use a depth of 10 cm.
- When modeling electron contamination, use a fluence grid resolution of 0.20 cm and reduce the phantom size to 15 cm x 15 cm. Automodeling scripts will automatically adjust the phantom size and resolution.
- In planning mode, if prescribing to d_{max} for a single beam, or when dose in the buildup region is clinically relevant, reduce the dose grid voxel size to 3 mm or lower. The TERMA grid resolution will be set to match the dose grid resolution.

Calibration point output factor measurement

The calibration point referred to here is the point at which absolute dose is measured for Pinnacle^3. In general, this not the same as the machine calibration point.

The output factor calibration point must be measured at a depth beyond the electron contamination region. A measurement depth of 10 cm is strongly recommended.

WARNING
Measuring output factors at d_{max} (or anywhere within the electron contamination region for the energy) may result in incorrect monitor unit calculations.

The calibration output factor measurement and the relative output factor measurements (described in the following section) must be measured using the same depth.

NOTE
Currently, the software requires a 10 cm x 10 cm field size for calibration output factor measurements. For fixed jaw machines, the field size is defined by the user and corresponds to an MLC-defined field instead of a jaw-defined field.

Dose/MU at Calibration Point (cGy/MU):	user defined
Field size (non-fixed jaw machines):	10 cm x 10 cm
Field size (fixed jaw machines):	user defined (minimum 4 cm x 4 cm)
Calibration Point Depth (cm):	(10 cm is strongly recommended)
Source to calibration point distance (cm):	user defined

Open field relative output factor measurements

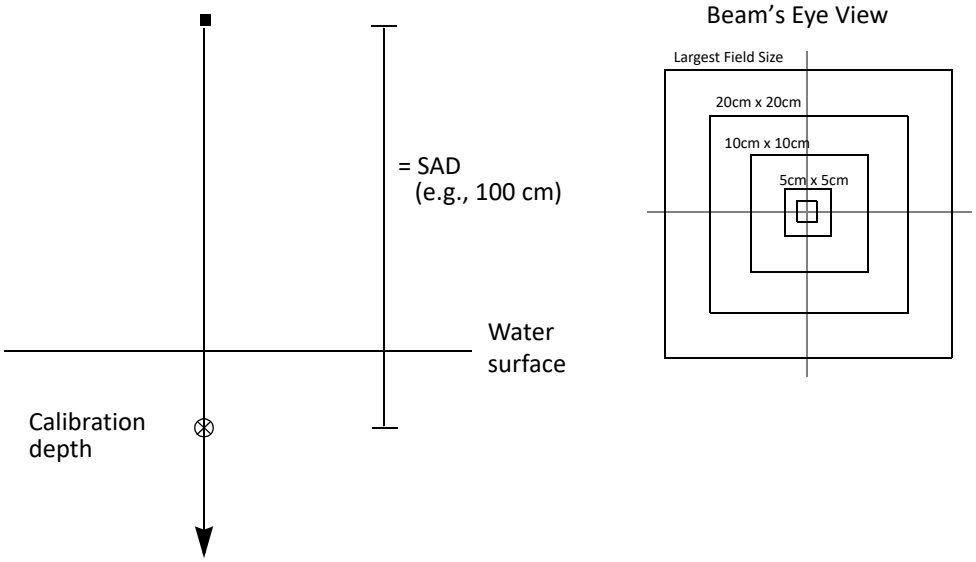
The relative output factors are measured for a standard geometry. This data is used along with the calibration output factor to determine the “true” output factor. The “true” output factor is the energy fluence per monitor unit as a function of field size. The true output factor is used to scale the magnitude of the incident fluence.

The relative output factor for an open field is calculated using the equation

$$OF = \frac{Dose_{FS}}{Dose_{CFS}}$$

where $Dose_{FS}$ is the measured dose for the given field size and $Dose_{CFS}$ is the measured dose for the calibration field.

Measurement Type	Point dose measurements
Field Sizes (non-fixed jaw machines)	• 2 cm x 2 cm • 5 cm x 5 cm • 10 cm x 10 cm • 20 cm x 20 cm • 30 cm x 30 cm • 40 cm x 40 cm • additional small field sizes if possible • maximum field size
Field sizes (fixed jaw machines):	• 2.4 cm x 2.4 cm • 3.2 cm x 3.2 cm • 4.8 cm x 4.8 cm • 5.6 cm x 5.6 cm • 8.0 cm x 8.0 cm • 10.4 cm x 10.4 cm • 15.2 cm x 15.2 cm • maximum field size
Depth	Same as depth used in calibration output factor measurement (10 cm is strongly recommended).
Normalization	Normalize relative to the calibration output factor.
Setup	Same as that used for the absolute calibration.
Modifiers	None (open field)



Wedged field output factor measurements

For each wedge used with a given energy, relative output factor measurements must be made for a range of field sizes.

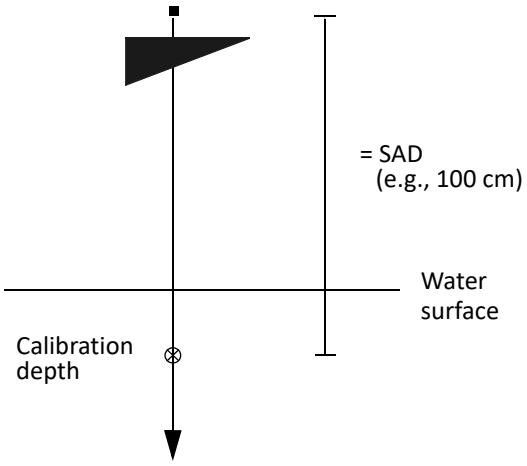
The relative output factor for a wedged field is calculated using the equation

$$OF = \frac{Dose_{WFS}}{Dose_{CFS}}$$

where $Dose_{WFS}$ is the measured dose for the given wedged field size and $Dose_{CFS}$ is the measured dose for the open calibration field. Note that the ratio is *not* the wedge transmission factor because $Dose_{CFS}$ is the fixed open calibration field dose.

WARNING
The wedge transmission factor cannot be substituted for the wedged field output factors. Using the wedge transmission factor instead of wedged field output factors will result in incorrect monitor unit calculations.

Measurement Type	Point dose measurements
Field Sizes (non-fixed jaw machines)	• 2 cm x 2 cm • 5 cm x 5 cm • 10 cm x 10 cm • 20 cm x 20 cm • 30 cm x 30 cm • largest field size possible
Field sizes (fixed jaw machines):	• 2.4 cm x 2.4 cm • 3.2 cm x 3.2 cm • 4.8 cm x 4.8 cm • 5.6 cm x 5.6 cm • 8 cm x 8 cm • 10.4 cm x 10.4 cm • 15.2 cm x 15.2 cm • largest field size possible
Depth	Same as depth used in calibration output factor measurement (10 cm is strongly recommended).
Normalization	Normalize relative to the open field calibration output factor.
Setup	Same as that used for the absolute calibration.
Modifiers	Wedge



Tray factor measurements

To avoid electron contamination, you should measure tray factors at a depth of at least 5 cm, and possibly 10 cm.

Requirements for dynamic wedges

If you will be using dynamic wedges, follow the guidelines below.

- Measure the same dose profiles as recommended earlier in this chapter for physical wedges, but take the measurements at different wedge angles. At a minimum, you should take measurements at the maximum wedge angle and an intermediate angle, such as 30 degrees.
- Measure output factors for the same field sizes as recommended for the open field earlier in this chapter. Note that the larger field sizes will be asymmetric for dynamic wedges.
- Measure output factors for dynamic wedges at the smallest and largest wedge angles, and at a few intermediate angles such as 15, 30, and 45 degrees. The system will interpolate to intermediate wedge angles for the OFc values.
- Check as many wedge angles and jaw settings as possible to verify absolute dosimetry and ensure confidence in the algorithm.

Requirements for motorized wedges

If you will be using motorized wedges, follow the guidelines below.

- Measure the same dose profiles and depth dose curves that are recommended for fixed wedges. You are only required to take measurements for the maximum motorized wedge angle.
- Measure output factors for the same field sizes that are recommended for the open field. You are only required to measure output factors for the maximum motorized wedge angle.
- Check the dose profiles and depth dose curves for as many synthesized wedge angles as possible to verify relative dosimetry. This is done in the Pinnacle^3 planning software, not in the Photon Physics tool.

- Check as many synthesized wedge angles and jaw settings as possible to verify absolute dosimetry and ensure confidence in the algorithm.

CAUTION
You should validate your beam models, particularly the output factors for motorized wedges, by comparing Pinnacle³ calculations with measurements. Use fields that are similar to those you are likely to treat, and use both square and rectangular fields. Ideally, this validation should be done before you commission the machine.

Dose computation with dynamic wedges

The *Pinnacle³ Physics Instructions for Use* describes how to set up dynamic wedges in the physics tool. This section describes the method the software uses to compute 3D dose distributions from dynamic wedges produced by Siemens and Varian linear accelerators.

In the software, relative dose modeling of dynamic wedges involves a few basic steps:

1 The open field for the given accelerator is accurately modeled.

2 A planar transmission filter is produced that models the primary wedge-shaped profile of the wedge. The production of the filter is vendor-specific.

3 The wedge-shaped profile is modified to incorporate secondary factors, which include jaw transmission and head scatter effects. This modification is *not* vendor-specific. You can create a wedge-specific model, if necessary.

4 The final dose computation is performed by first multiplying the open field incident energy fluence image by the planar transmission image prior to the TERMA computation in the phantom patient. This process, along with the convolution/superposition dose computation, completes the modeling for the relative dose from a dynamic wedge.

Certain vendor-specific aspects of the software produce a planar transmission array in the wedged direction given the desired wedge angle, the jaw settings of the “toe” and “heel” ends of the wedge, and the position information in the plane of the isocenter. Each vendor, whether Varian or Siemens, produces this transmission array using its own parameters. These vendor-specific parameters are discussed in the following sections.

Varian transmission array

To produce its transmission array, the Varian Enhanced Dynamic Wedge (EDW) uses a “Golden” Segmented Treatment Table (STT), a transmission table that produces a 60-degree wedge angle for a specific energy. This table is fixed relative to the central axis of the beam.

To calculate wedge profiles for different angles associated with the same energy, the Varian EDW employs a “ratio of tangents,” which is a weighting factor used to determine the open field and the 60-degree wedged field components of the transmission. The dose profile for a given wedge angle can be viewed as a weighted sum of the open field and the 60-degree wedge field. The desired wedge angle uniquely specifies the relative weights of these fields.

The STT is Varian's name for the 1D array produced for an arbitrary wedge angle using the Golden STT and the weighting factor derived from the ratio of tangents.

In Varian's implementation, the signal detected in the monitor unit chamber for a dynamically wedged beam at the end of the collimator motion is equal to the signal for the open field. The final transmission array is therefore normalized and the "effective wedge factor" is incorporated into the transmission image.

Siemens transmission array

The Calibration Constant and the Linear Attenuation Coefficient defined by Siemens are parameters of the analytical function used to produce the transmission array.

In the Siemens "Virtual Wedge" implementation, the signal detected in the monitor chamber for an open field is equal to the signal detected up to the point where the moving jaw crosses the central axis (or *would* cross the central axis, if the jaw is not in the beam due to collimator asymmetry). In this implementation, the "effective wedge factor" is unity and is built into the transmission from the Siemens equations and does not require normalization.

The Siemens dynamic wedge uses an analytical function to determine the 1D transmission array. This transmission array (unlike Varian's SST) is not fixed relative to the central axis of the beam, but instead has its fixed point at the geometric center of the field in the wedged direction.

Vendor-independent effects

The vendor-specific primary transmission array is modified by the following secondary effects, which are independent of the vendor.

Jaw transmission effect

This effect produces a slightly lower gradient in the dose profile due to the transmission of photons through the moving jaw. When the beam is on, there is some photon transmission through the jaws that is not accounted for in the vendor-specific implementations.

Head scatter effect

This effect also reduces the gradient in the dose profile due to the decreased accelerator output at the smaller field sizes during jaw motion. When the field is small, the head scatter contribution is smaller, hence the "toe" end of the wedge will receive slightly less dose than is accounted for in the vendor-specific implementations.

The jaw transmission and head scatter effects are both relatively small; consequently, they are modeled by a single transmission factor that is specific to each dynamic wedge and energy. The modulation of the primary transmission image by this transmission factor is shown below:

$$T_{final}(x,y) = T(x,y) + (\text{Transmission Factor})(T_{max} - T(x,y))$$

where $T(x,y)$ is the vendor-specific transmission filter, and T_{max} is the maximum transmission value present in the entire filter. And (T_{max} - $T(x,y)$) reflects the duration that the moving jaw is covering the point (x,y) in the incident fluence (or transmission) plane.

The value of the transmission factor is adjusted manually to provide the best match to the measured dose profiles for each dynamic wedge and energy. This value should be close to the "true" jaw transmission of the particular linear accelerator.

Absolute dose calculation

For the calculation of absolute dose from a dynamic wedge, note that the "effective wedge factor" associated with the vendor-specific delivery device is already included in the transmission filter. The output factor (OFc) computations are thus correcting for any differences between actual measured output factors and the vendor-stated output factors.

The OFc's that are computed in the physics tool should therefore be similar to the open field OFc's, and should reflect the smaller differences between the "effective wedge factors" and the measured "wedge factors."

The software's monitor unit calculation uses the output factors for a subset of wedge angles. The OFc values for a given wedge angle are linearly interpolated. If the wedge angle specified for a given treatment is smaller than the smallest measured wedge angle, the reported OFc will be an interpolation between the OFc of the open field and the OFc of the smallest measured wedged field. The software will not compute absolute dose for wedge angles higher than the largest wedge angle for which output factors have been computed.

For more information on computing output factors, see the *How the software uses output factors* section.

Photon model parameters

The *Photon Beam Physics & Physics Utilities* chapter in the *Pinnacle³ Physics Instructions for Use* describes the photon beam modeling process. Before beginning to model beams in the **Photon Model Editor** window, become familiar with the photon model parameters and how changes to those parameters affect the model. The following sections describe the photon model parameters in detail.

The energy spectrum

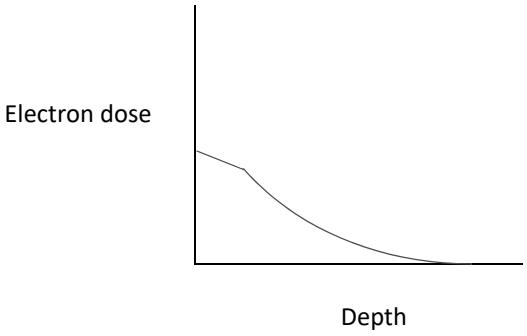
The first step in modeling is to determine the energy spectrum for the beam. The energy spectrum defines the relative number of photons of each energy which are exiting the head of the linear accelerator. You use the measured depth dose profiles to model the spectrum. In the **Photon Model Editor** window, click the **Depth Dose** tab to access the spectrum parameter options.

The software contains a number of published and interpolated spectra for use as the starting point to model your beam. Generally, you should start the modeling process with a spectrum of the same energy as your beam if one is available. If the exact energy is not available, select the spectrum with the nominal energy closest to but greater than your beam's nominal energy.

When comparing the measured and computed depth dose profiles, you will increase or decrease the relative number of photons for different energy bins depending upon the region of the depth dose which needs to be adjusted. The shape of the depth dose curves is most dependent upon the relative number of mid-range and high energy photons. The specified energy bin must correspond to the available kernel and attenuation coefficient energies.

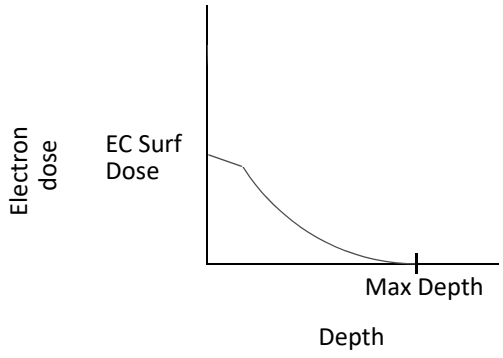
Electron contamination parameters

The software accounts for the electron contamination in a beam by adding electron dose to the photon dose. It models the electron dose as a modified exponential curve. At shallow depths the dose is linear to prevent the electron dose from becoming too high at the surface. The shape of this curve along the central axis depth dose is illustrated below.

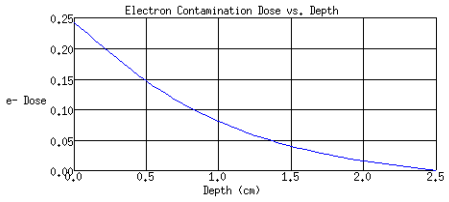


The electron contamination parameters control the amount of electron dose that is added, the maximum depth where the electrons have an effect, the rate of electron dose fall-off with respect to depth, and the change in electron contamination with field size. In the **Photon Model Editor** window, click the **Buildup** tab and click **On** next to the **Electron Contamination** option to turn on the electron contamination parameters.

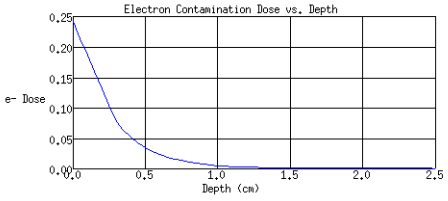
The **Max Depth** parameter determines the maximum depth receiving electron dose. The **EC Surface Dose** parameter determines the magnitude of the electron surface dose.



The **Depth Coefficient** determines the rate of the electron dose fall-off with respect to depth. Increasing the **Depth Coefficient** increases the rate of fall-off. For example, the figures below show a depth coefficient of 1 and 4.



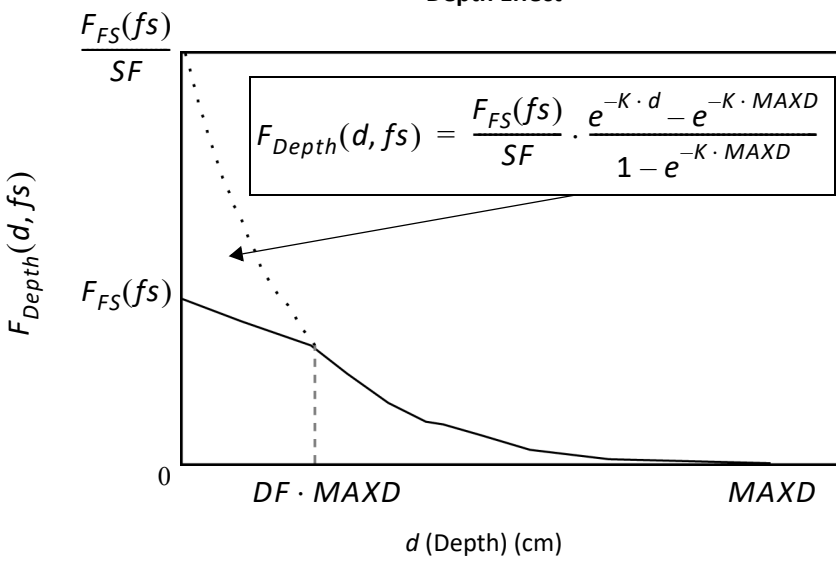
Depth Coefficient = 1



Depth Coefficient = 4

The **DF** (Depth Fraction) and **SF** (Scale Fraction) parameters determine the linear region of the curve which replaces the exponential increase in dose near the surface. The Depth Fraction determines the depth at which the linear region of the electron contamination ends and the exponential fall off begins. The Scale Fraction is the fraction by which the initial exponential electron contamination curve is adjusted to obtain the linear start point for the curve.

Depth Effect



The **Off-Axis Coefficient Electron Contamination** parameter accounts for the decrease in the effects of the electron contamination as you move off-axis. It allows you to shape the electron contamination dose magnitude.

The software models this effect using a Gaussian curve (with radial symmetry). When the Off-Axis Coefficient is set to 0, the Gaussian curve is flat. As the value increases, the Gaussian curve becomes more narrow, effectively reducing electron contamination off the central axis.

The **EC Field Size Dependence** parameters (C_1, C_2, C_3) model the change in electron contamination as a function of field size. These changes are generally linear for low energies with increasing curvature at higher energies. In the software, this effect is modeled using the equation:

$$F_{FS}(fs) = ECD_{10 \times 10} + C_1 \cdot (fs - 10) + C_2(e^{-C_3 \cdot 10} - e^{-C_3 \cdot fs})$$

where

F_{FS} = field size factor

fs = field size

ECD_{10x10} = Electron Contamination Surface Dose for a 10 x 10 field

C_1, C_2, C_3 = Model parameters fitted to match the measured data

Constraints: $C_2 > 0$

WARNING
C_1, C_2, and C_3 are used to produce a function of field size. They should be the same for each model when the models have been created for each field size.

If you have photon models that are in the same wedge group but have different field sizes, you cannot assign different values for C_1, C_2, and C_3. It causes problems in the interpolation over field size for the different field-size specific models and will prevent commissioning of the machine.

In Field model parameters

The flattening filter attenuation and off-axis softening are the two main effects seen in the field. To access these parameters, click the **In Field** tab in the **Photon Model Editor** window.

The flattening filter attenuation has two effects in the beam:

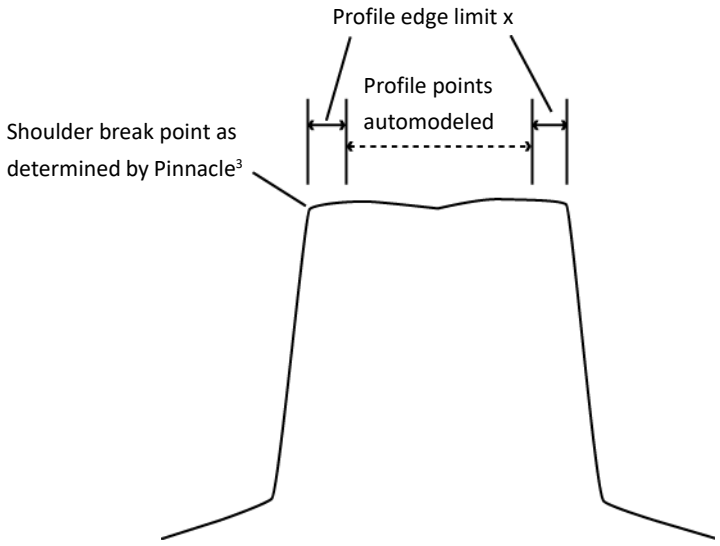
- it changes the photon fluence as a function of off-axis distance. As you get farther from the central axis, you see an increase in the number of photons per unit area. You can model this effect using either cone radius or arbitrary profile.
- it changes the distribution of the photon energies as a function of off-axis distance. The beam contains relatively more high-energy (hard) photons near the central axis and relatively more low-energy (soft) photons far from the central axis. You can model this effect with the **Spectral Off-Axis Softening** parameter.

Arbitrary Profile model

When you select **Arbitrary Profile** from the **Modeled As** option list, the **Arbitrary Profile Editor** button and the **Limit Profile Edge for Auto-Modeling by** field appear on the **In Field** tab.

The **Limit Profile Edge for Auto-Modeling by** field lets you set the boundary in the high dose region to remove data points from the penumbra. Some profiles do not have well-defined transitions from

the high dose region to the penumbra, and the automodeler could include data from the penumbra region of smaller profiles in the high dose region.



Click the **Arbitrary Profile Editor** button to open the **Edit Fluence Filter Profile** window. (This window also appears if you change the **Modeled As** option from **Cone** to **Arbitrary Profile**.) A default arbitrary fluence profile is created with 50 points. The profile is generated out to the squared sum of the largest measurement geometry and increases at 0.6% per cm.

If the arbitrary fluence table is not as wide as the incident energy fluence, the software extrapolates the table by extending the last value of the table beyond the maximum position of the table itself to the edges of the incident fluence.

Use the E_TuneAllInSections sequence to automodel the profile, or use the **Ins After**, **Ins Before**, **Delete Point**, and **Delete All Points** buttons to edit the profile. Arbitrary Profiles can contain from 2 to 200 points.

> **NOTE**
> For each point in the profile, the radius must be 0 cm–50 cm, and the incident fluence value must be 0.5–1.5, or you will not be able to compute dose.

Cone model

When you select **Cone** from the **Modeled As** option list, the **Fluence Increase/cm** and **Cone radius** fields appear on the **In Field** tab. See the *Photon Beam Physics & Physics Utilities* chapter in the *Pinnacle³ Physics Instructions for Use* for information about the cone model.

The Spectral Off-Axis Softening parameter

The off-axis softening parameter reduces the spectrum weights as a function of off-axis angle in this fashion:

$$W'_i = W_i \times \left(\frac{1}{1 + \left(\frac{E_i}{E_{max}} \right)} \right)^{OffAxisSofteningParam \times \theta}$$

where

W_i = the spectral weight for bin "i" which has an effective energy E_i

θ = the off-axis angle

E_{max} = maximum energy in the beam spectrum

The off-axis angle is calculated in the following way:

$$\theta = \text{atan}\left(\frac{OffAxisDist}{SAD} \right)$$

where *OffAxisDist* is the distance from the central axis in the isocentric plane that is orthogonal to the beam.

Wedge/Compensator Scatter Factor

For wedged fields, you may also need to adjust the Wedge/Compensator scatter factor. The convolution algorithm models beam modifiers using three processes: attenuation; hardening of the beam in the primary TERMA calculation; and primary TERMA boosting to account for scatter from the modifier. The algorithm models the third effect with the Modifier Scatter Factor (MSF).

The TERMA is increased under the modifier using the equation:

$$\Psi'(\vec{r}) = \Psi(\vec{r})(1 + MSF \times Len(\vec{R}, mod) \times MSC(mod, SSD))$$

where

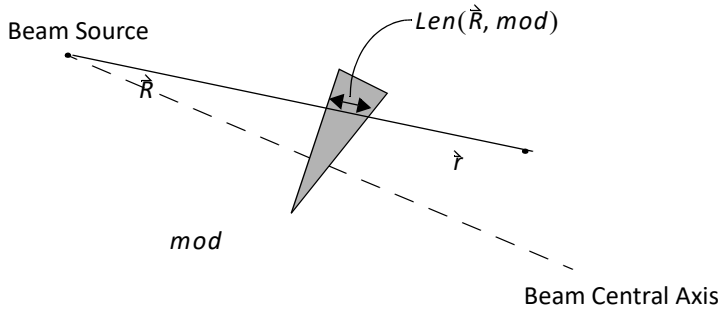
MSF = the Wedge/Compensator Scatter Factor which you must set

R = the line connecting the beam source to the point *r*.

mod = the beam modifier

Len (R, mod) = the length of the line *R* through the beam modifier

MSC = the modifier scatter capacity. The modifier scatter capacity is a function of the modifier's ability to interact with the beam and the modifier's ability to scatter photons into the phantom. For compensators this function has a value very close to zero.



Typical values for the wedge scatter factor are: 0.5 for Varian machines and 0.2 for Elekta machines with an internal wedge.

Out of Field model parameters

The penumbra and tails of the profiles are modeled with the Out of Field model parameters. To access the out of field photon model parameters, click the **Out of Field** tab in the **Photon Model Editor** window.

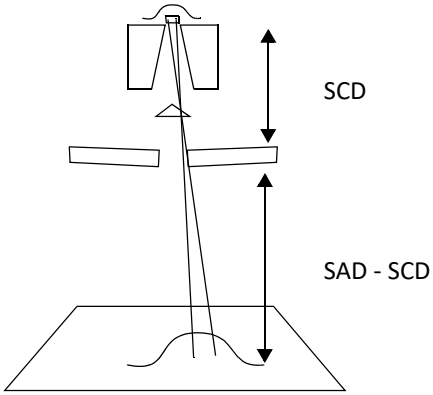
> **NOTE**
> The MLC transmission field only appears if the machine has an MLC.

Effective Source Size parameters

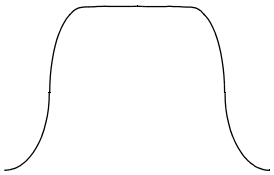
The Effective Source Size parameters model the penumbra of a beam by blurring the incident fluence model. The shape of the blurring kernel is modeled as a Gaussian with the FWHM (or Full Width at Half Maximum—describes a gaussian distribution's width) equal to the projected effective source size.

The effective source size is scaled by (SAD - SCD)/SCD, where SCD is the source to collimator (jaw) distance and SAD is the source to axis distance. Note that each dimension is handled separately.

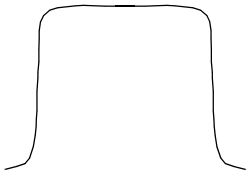
This Gaussian blurring kernel is then convolved with the incident fluence to determine the incident fluence distribution.



Increasing the effective source size makes the shoulders and base of the X and Y profiles rounder. Decreasing the effective source size makes the shoulders and base of the X and Y profiles more square.



Use a larger effective source size to make the shoulders and base of the profiles more round.



Use a smaller effective source size to make the shoulders and base of the profiles more square.

Flattening Filter Scatter Source parameters

In addition to changing the shape of the primary radiation, the flattening filter acts as a source of secondary scatter radiation. The Flattening Filter Scatter Source parameters control the model of the photon scatter from the flattening filter. This scatter is most visible in the tails of the cross-axis scans, but the software model takes this contribution into account across the entire beam. The software models this photon scatter source as a Gaussian curve.

The Gaussian Height parameter defines the fraction of the central axis energy fluence that is due to the flattening filter scatter when the collimator jaws are wide open.

The Gaussian Width parameter specifies the width (in centimeters) of the Gaussian curve used to model the flattening filter scatter source. The width is defined at the source to flattening filter distance, which is entered in the **Misc** tab of the **Machine Editor** window. The software achieves the modification to the incident fluence plane by back projecting through the jaws and performing an integration based on the amount of the Gaussian curve visible from each point in the incident fluence plane.

Transmission Factors

The **XY Jaw Transmission Equal** option allows you to select if your machine has single or multiple jaw transmission factors. The setting is global for the machine, even though you select it for one of the models and can change the transmission factors per model. If you select **Yes**, you can enter one jaw transmission factor for both pairs of jaws. If you select **No**, you can enter one transmission factor for the top/bottom pair and another for the right/left pair. If you set **MLC Replaces Jaws** to **Yes** in the **MLC Editor**, you can only set a single jaw transmission factor.

The jaw transmission factor (0.001-1.00) specifies the fraction of the energy fluence transmitted through the jaws. This models the actual jaw transmission.

The MLC transmission factor (0.001-0.200) specifies the fraction of the energy fluence transmitted through the MLC leaves.

> **NOTE**
> The MLC transmission field only appears if the machine has an MLC. The MLC transmission factor cannot be edited during planning.

Automodeling sequences

The *Photon Beam Physics & Physics Utilities* chapter in the *Pinnacle³ Physics Instructions for Use* describes the software's automodeling tool, which can optimize photon model parameters for you. When using the automodeling tool, you must choose among a set of automodeling sequences that optimize specific parameters of the model. This section describes each of the automodeling sequences in detail.

A number of automodeling optimization sequences are available. These are some of the common uses:

- Several sequences allow you to tune the entire model. For example, run the E_TuneAllInSections sequence to tune all parameters for all open field sizes using a single model. The Tolerance Factor is reset to 1.0 at the beginning of each modeling state.
- You can also perform the automodeling in steps. Run the FineTuneECAndSpectrum sequence to fine-tune the spectrum and electron contamination. Run the FineTuneCrossBeam sequence to fine-tune the cross-beam shape parameters.

All the sequences are listed below and explained in this section.

Sequences to use for all field sizes (open and/or wedge)

- E_TuneAllInSections.OptSequence
- FineTuneCrossBeam.OptSequence
- FineTuneSpectrum.OptSequence
- FineTuneECAndSpectrum.OptSequence
- FineTuneECInParallel.OptSequence
- FineTuneECInSections.OptSequence
- FineTuneECSequentially.OptSequence

Sequences to use for models containing a wedge

- FineTuneModelForOpenAndWedgedFields.OptSequence
- FineTuneModifierScatter.OptSequence

Sequence to use for wedge-only models

- FineTuneAllForWedge.OptSequence

Sequences to use for field-size specific data (open or wedge)

- FineTuneModelForFieldSize.OptSequence
- FineTuneCrossBeam.OptSequence
- FineTuneSpectrum.OptSequence

E_TuneAllInSections

This sequence currently produces the best results for tuning an entire model. The sequence tunes the electron contamination parameters in conjunction with the spectrum while tuning the entire model. It also separates the X and Y focal spot size tuning and optimizes the left/right jaw transmission, top/bottom jaw transmission, MLC transmission, and arbitrary fluence profiles. The left/right jaw transmission optimization only uses X profiles, while the top/bottom jaw transmission only uses Y profiles. The MLC transmission is optimized after the jaw transmissions, and it only uses profiles that include the MLC.

> **NOTE**
> The MLC parameters **Rounded Leaf Tip Radius**, **Tongue and groove width** and **Additional interleaf leakage transmission** are not included in this automodeling sequence, but their effects are included in the dose computation.
>
> When jaw transmissions are optimized, only the profiles in the direction of the jaws are used. When MLC transmissions are optimized, only the profiles in the direction of the MLC are used.

FineTuneCrossBeam

Use this sequence to fine-tune the cross-beam shape parameters. You can also use this sequence to fine-tune a model copied from the model library to your machine.

This sequence does *not* tune the Off-Axis Softening parameter nor the Inc Fluence Cone Angle and Radius for small fields. These parameters should only be tuned using large fields. Because of the relationship between cross-beam low dose and cross-beam high dose parameters, you should not apply this sequence to a model that does not include large and small fields.

If a single model exists for all fields, this sequence does *not* use the wedged fields for optimization. However, if the model is for wedged fields, the software uses only the corresponding wedged fields for optimization.

FineTuneSpectrum

Use this sequence to fine-tune the beam spectrum. You can use this sequence to fine-tune a model copied from the model library to your machine.

If a single model exists for all fields, this sequence does *not* use the wedged fields for optimization. However, if the model is for wedged fields, the software uses only the corresponding wedged fields for optimization.

FineTuneECAndSpectrum

Use this sequence to fine-tune the beam spectrum in the presence of electron contamination, which obtains a best fit in the buildup region of the depth dose curves while maintaining agreement at deeper depths. The success of this sequence depends on an accurate fit of the electron contamination for the tuning of the spectrum.

If a single model exists for all fields, this sequence does *not* use the wedged fields for optimization. However, if the model is for wedged fields, the software uses only the corresponding wedged fields for optimization.

FineTuneECInParallel

Use this sequence to fine-tune the electron contamination. You should use this sequence only after the beam spectrum has been tuned.

Most of the electron contamination parameters are tuned simultaneously for all fields corresponding to the model being tuned. The Depth parameter is *not* tuned. The Off Axis Coefficient is tuned as a separate phase from the other parameters.

If a single model exists for all fields, this sequence does *not* use the wedged fields for optimization. However, if the model is for wedged fields, the software uses only the corresponding wedged fields for optimization.

FineTuneECInSections

Use this sequence to fine-tune the electron contamination *after* the beam spectrum has been tuned. Most of the electron contamination parameters are tuned in sections for all fields corresponding to the model being tuned. The Depth parameter is *not* tuned. The Depth Fraction and the Scale Fraction are tuned separately. The Off Axis Coefficient is tuned as a separate phase from the other parameters.

If a single model exists for all fields, this sequence does *not* use the wedged fields for optimization. However, if the model is for wedged fields, the software uses only the corresponding wedged fields for optimization.

FineTuneECSequentially

Use this sequence to fine-tune the electron contamination. You should use this sequence only after the beam spectrum has been tuned. The electron contamination parameters are tuned

sequentially. The optimization starts with medium size fields and then progresses to all fields. This results in electron contamination parameters that are biased toward the medium size fields. The Depth parameter is *not* tuned. The Off Axis Coefficient is tuned as a separate phase from the other parameters.

If a single model exists for all fields, this sequence does *not* use the wedged fields for optimization. However, if the model is for wedged fields, the software uses only the corresponding wedged fields for optimization.

FineTuneModelForOpenAndWedgedFields

Use this sequence to fine-tune all aspects of the model both for open fields and wedged fields. The assumption is that you have a single model for all fields (wedged and open), and that you have performed open field tuning.

FineTuneModifierScatter

Use this sequence to fine-tune all aspects of the model for the wedged fields to which it corresponds. The assumption is that you have copied this model from the model library or from the open field model, and that you need to fine-tune it to fit the wedged fields.

FineTuneAllForWedge

Use this sequence to fine-tune all aspects of the model for the wedged fields to which it corresponds, including left/right jaw transmission, top/bottom jaw transmission, MLC transmission, and arbitrary fluence. The assumption is that you have copied this model from the model library or from the open field model, and that you need to fine-tune it to fit the wedged fields.

> **NOTE**
> The MLC parameters **Rounded Leaf Tip Radius** and **Tongue and groove width** are not included in this automodeling sequence, but their effects are included in the dose computation.

FineTuneModelForFieldSize

Use this sequence to fine-tune a model for the currently selected field size. The assumption is that you have copied this model from the model library, or preferably from the open field model, and that you need to fine-tune it.

This sequence does not tune the cross-beam parameters, as these must be tuned with consideration for large fields. This sequence also does not tune the modifier scatter parameter.

WARNING
C_1, C_2, and C_3 are used to produce a function of field size. They should be the same for each model when the models have been created for each field size.

If you have photon models that are in the same wedge group but have different field sizes, you cannot assign different values for C_1, C_2, and C_3. It causes problems in the interpolation over field size for the different field-size specific models and will prevent commissioning of the machine.

Photon beam output factor computation

The *Photon Beam Physics & Physics Utilities* chapter in the *Pinnacle³ Physics Instructions for Use* describes how to compute the output factors for photon beams. Before computing output factors, you might want to read the following sections that explain how the software uses output factors and handles head scatter.

How the software uses output factors

To produce the correct dose, the software multiplies the standard incident energy fluence image by a correction factor (OF_c) before performing the convolution. Consequently, the resulting dose distribution represents cGy/MU. Next, the software uses the weighting of the beam and the prescription to determine the final MU. Therefore, the incident energy fluence image scaled by OF_c represents the true energy fluence/MU exiting the accelerator.

To determine the OF_c, the software first calculates the absolute calibration for the 10 cm x 10 cm reference field (or for fixed jaw machines, the field defined by the user). It computes dose to the calibration point and compares the dose to the user-supplied calibrated output. Then the software corrects the ratio of measured and computed data using the incident energy fluence. This absolute calibration factor is used for every field size when creating the incident energy fluence image.

The next step is to determine the field-size dependent correction factors. (The OF_cs are relative to the 10 cm x 10 cm reference, or for fixed jaw machines, the field defined by the user.) First, the software computes OF_p, which is the ratio of the computed dose for the specified field over the reference field using the convolution/superposition dose model with only the 10 cm x 10 cm absolute calibration applied, or for fixed jaw machines, the field defined by the user. With the measured OF and the computed OF_p, we can deduce that $OF_c = OF/OF_p$. This computation is made for every measured relative output factor and is tabulated per field size and plotted in the Photon Output Factor Computation window. In Planning mode the OF_c is interpolated via equivalent squares. For non-fixed jaw machines, the equivalent square is computed using the jaw-defined field, and for fixed jaw machines it is computed using the minimum circumscribing rectangle surrounding the MLC-defined field.

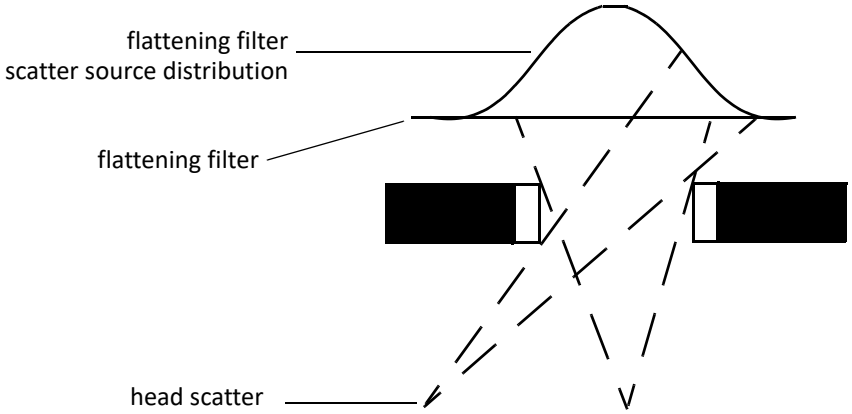
The final incident energy fluence for each beam is therefore the product of the fluence from the standard head model, the absolute calibration factor, and the field-size dependent OF_c.

CAUTION
When modeling, you want to achieve uniformity of the OF_c among the various field sizes. You can adjust the Gaussian flattening filter scatter source to obtain this uniformity. For more details on the Gaussian function, see the next section.

Head scatter and output factors

The absorbed dose from a photon beam is proportional to the energy fluence reaching the patient. Fluence is composed of both primary and scattered photons. The scattered component, principally originating from the flattening filter and primary collimator, is commonly called “head scatter,” “extra-focal radiation,” or “collimator scatter.” For accurate dose calculations, it is important to model the head scatter component of the dose correctly.

The software’s automodeling tool calculates the head scatter emanating from an scatter source located at the level of the flattening filter. This source distribution is modeled by a 2D Gaussian function. The software determines the amount of head-scattered radiation reaching any point in the incident fluence plane by integrating over the region of the source visible from the calculation point. For machines where the MLC replaces one of the jaws or beams where the MLC is on, the visible region of the source is defined by the MLC. If the MLC does not replace the jaw, the visible region of the source can be either the MLC or the jaw—whichever device produces the smallest visible region.



NOTE
The primary fluence plus the head scatter representation (the Gaussian integral) is the final incident fluence. The Gaussian flattening filter scatter source distribution is defined at the source-to-flattening-filter distance. The cone radius is defined at the isocenter plane.

The output factors (OF_c) computed in the current version of the software represent the head scatter that is not already included in the photon model. If all head scatter effects have been incorporated into the model, OF_c should be 1.00 for all field sizes. In practice, there may be a slight increase in computed OF_c values with increasing field size, although it is not unreasonable to see small decreases if the Gaussian scatter source slightly overestimates the amount of scatter generated by the flattening filter.

Because the software’s computed OF_c values reflect only the head scatter effects not included in the new model, they differ from the traditional Collimator Scatter Factors (Sc).

Why the software's scatter factors cannot be measured

Some users have questioned why their measured scatter factors (Sc) differ from the software's calculations of OF_c. The reason is simple: their definitions are different. A user begins by defining Sc as a measured quantity, whereas the software starts with the assumption that OF_c will be determined so as to ensure consistent results in determining absolute dose.

The OF_c reported by the software is calculated as shown:

$$OF_c = (OF)/(OF_p)$$

where *OF* is the user-measured relative output factor and OF_p is the calculated output factor, which includes the head scatter model in the incident fluence and phantom scatter from the superposition.

The convolution dose algorithm properly models phantom scatter by design. The superposition of the dose-spread kernel and the TERMA distribution intrinsically model the scattered radiation generated in the patient volume. OF_p can be determined directly from the integrated dose.

In practice, we set the scaling of the incident fluence (which includes the Gaussian head scatter model) to be constant, then extract a relative OF_p (for field "i" relative to field "ref") by computing the dose with the superposition:

$$OF_p(i) = \frac{D(i)_{OF_c = 1}}{D(ref)}$$

We now consider that the convolution operation is linear, meaning that scaling the number of photons in the incident fluence is the same as scaling the dose:

$$D_{OF_c(i)}(i) = D_{OF_c = 1}(i) \times OF_c(i)$$

$D_{OFc(i)}(i)$ is the final dose calculation when the proper OF_c is used for the given field size, and is synonymous with *D(i)*.

Using the previous equations in this section, and realizing that by definition

$$OF_c(ref) = OF_p(ref) = 1$$

we can verify that this approach is coherent and consistent.

$$OF(i) = \frac{D(i)}{D(ref)} = \frac{D(i)_{OF_c = 1}(i) \times OF_c(i)}{D(ref)_{OF_c = 1}(i)} = \frac{OF_p(i) \times D(ref)_{OF_c = 1}(i) \times OF_c(i)}{D(ref)_{OF_c = 1}(i)} = OF_p(i) \times OF_c(i)$$

This equation matches the first equation in this section.

References and recommended reading

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Battista, J.J. and M.B. Sharpe. 1992. True three-dimensional dose computations for megavoltage x-ray therapy: A role for the superposition principle. *Australasian Physical & Engineering Sciences in Medicine* 15(4):159-178.

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Mackie, T.R., P.J. Reckwerdt, M.A. Gehring, T.W. Holmes, S.S. Kubsad, B.R. Thomadsen, C.L. Sanders, B.R. Paliwal, and T.J. Kinsella. 1990. Clinical implementation of the convolution/superposition method. *Proceedings of the 10th International Conference on the Use of Computers in Radiation Therapy.* Council of Science and Technology, Bombay, India, 1990.

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Van Dyk, J., R.B. Barnett, J.E. Cygler, and P.C. Shragge. 1993. Commissioning and quality assurance of treatment planning computers. *International Journal of Radiation Oncology, Biology and Physics* 26(2):261-273.

Wong, J.W. and J.A. Purdy. 1990. On methods of inhomogeneity corrections for photon transport. *Medical Physics* 17(5):807-814.

4 Electron Physics & Data Requirements

WARNING
The accuracy of the treatment planning dose calculation depends on the quality of the data entered in the physics tool. If the quality of this data is poor, or if the computed dose is not verified with measured data, the dose calculation may be inaccurate.

Required electron physics data

The following measured data is required for electron beam modeling.

Physical machine measurements

You need to measure the drift distance (in centimeters) between the collimator and the surface of the phantom for the standard setup (nominal) SSD.

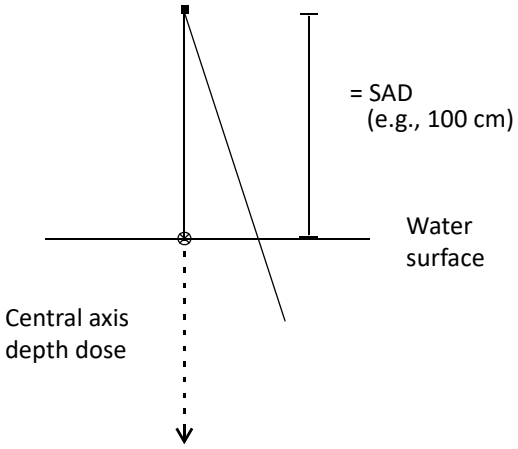
Depth dose measurements

Depth dose measurements must be taken for each energy at several square field sizes as recommended in the *Depth dose measurements for dose normalization* section. The depth dose measurements are used to determine the practical range of electrons (R_p) and are also used for depth normalization of the dose.

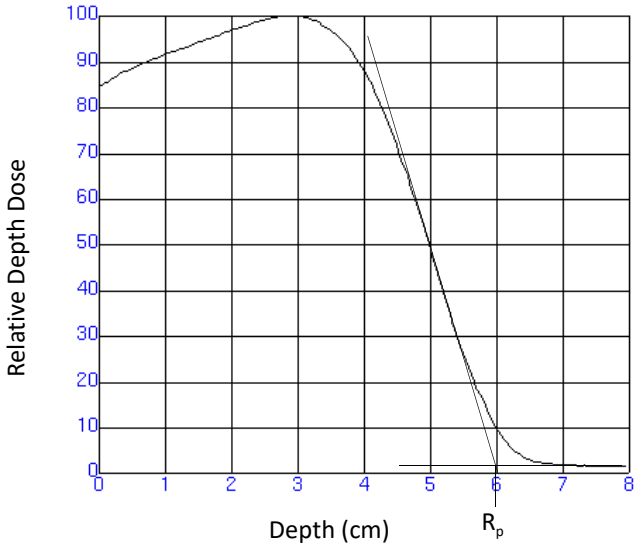
Depth dose measurement for determining R_p

Because many of the dose profile measurements are measured at depths related to the practical range of electrons (R_p), you should determine R_p for the energy before measuring any other data. You *must* determine R_p for the selected energy before measuring the cross-axis dose profiles.

Required Data for Each Energy	Description
Measurement Type	Open-field central axis depth dose
Cone Sizes	Largest available cone size
Setup	Water surface at isocenter. SSD equal to SAD.



As defined by Khan, “the practical range is the depth of the point where the tangent to the descending linear portion of the curve (at the point of inflection) intersects the extrapolated background,” as shown in the illustration below.



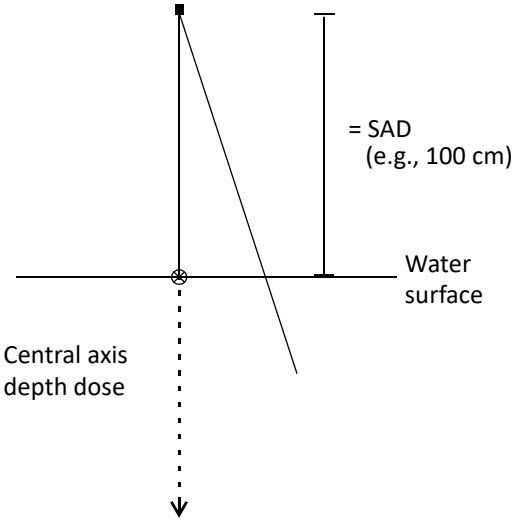
The practical range of electrons should be determined using the depth dose for the largest cone size measured for the energy. For a detailed description of the energy determination process, refer to the AAPM TG25 report on clinical electron-beam dosimetry.

Depth dose measurements for dose normalization

For each energy, the central-axis depth dose should be measured for a range of square field sizes. Each field size should be made as a cutout. This data is used to normalize the dose at varying depths.

The field sizes are independent of the cones used, but for each particular field size we recommend using the cone that you would use clinically.

Required Data for Each Energy	Description
Measurement Type	Open-field central axis depth dose
Square Field Sizes	2, 3, 4, 5, 6, 8, 10, 15, 20, 25, or open cone field size (in cm)
Measurement Limits	Measure to a depth of R_p + 5 cm
Setup	Water surface at isocenter. SSD equal to SAD.

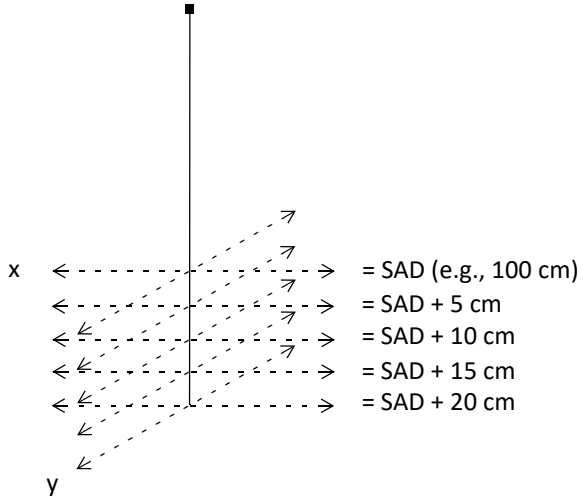


In-air cross-axis dose profile measurements

Cross-axis dose profiles must be measured in air for the determination of the virtual source distance and sigma-theta-x. These profiles do not need to be imported into the Electron Physics tool.

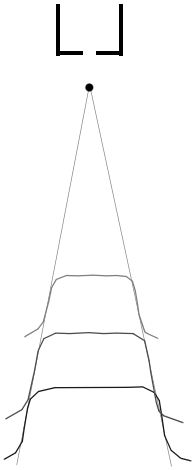
For the largest field size (20 cm x 20 cm or the largest field size closest to 20 cm x 20 cm), measure either X or Y cross-axis profiles in air using high spatial resolution. Profiles should be obtained at equally spaced distances beyond the nominal SSD (i.e., for a 100 cm SAD machine, measure profiles at 100 cm, 105 cm, 110 cm, 115 cm and 120 cm source to probe distances).

Required Data	Description
Measurement Type	Open-field orthogonal (horizontal or vertical) dose profiles in air through the central axis.
Cone Sizes	20 cm x 20 cm or the largest field size.
Source to chamber distances	Equally spaced distances beyond the nominal SSD (i.e., for a 100 cm SAD machine, measure at 100 cm, 105 cm, 110 cm, 115 cm, and 120 cm source to probe distances)
Resolution	1 mm spacing
Measurement Limits	Measurements should extend as far outside the field boundary as possible



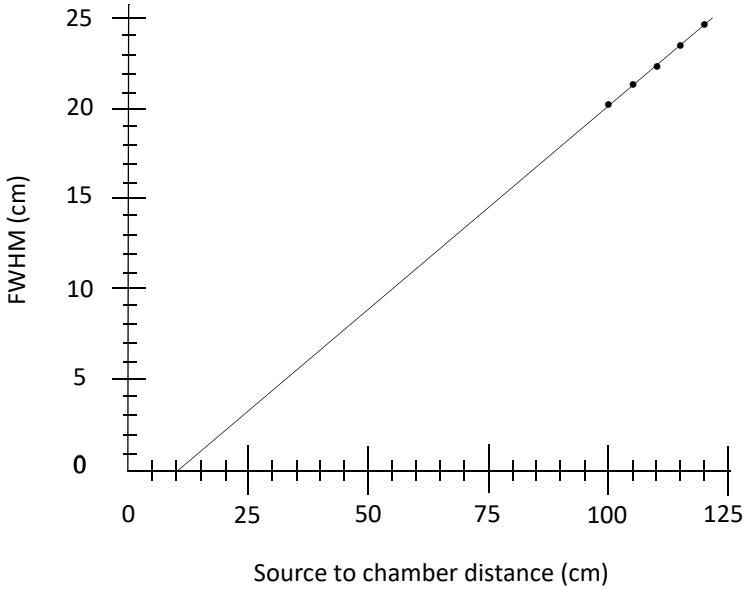
Determining the virtual source to surface distance

Unlike x-ray beams, electron beams do not emanate from a physical source in the accelerator. Instead the electron beam is spread by the scattering foils into a broad beam that appears to diverge from a point. This point is referred to as the “virtual source.” The position of the virtual source can be found by back projection of the 50% width of the beam profiles obtained at different source to chamber distances.



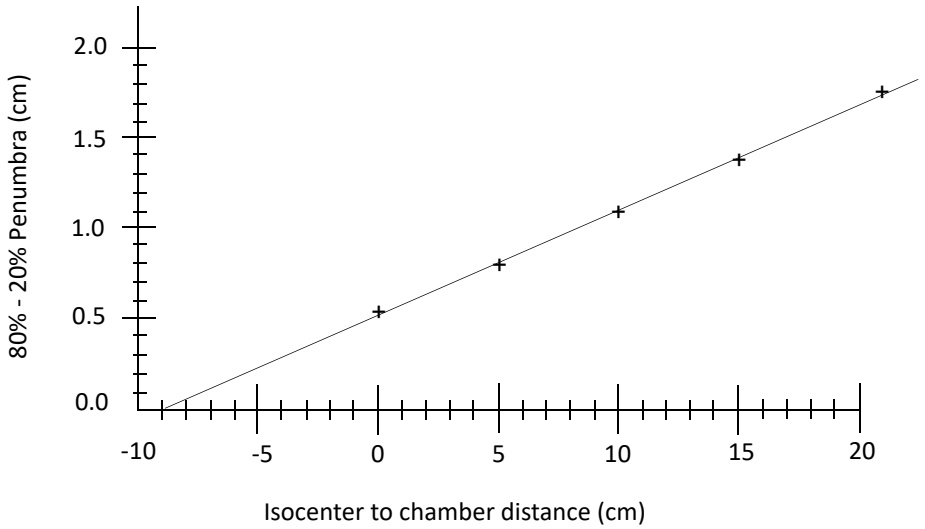
To determine the position of the virtual SSD, plot the beam width at the 50% dose positions (FWHM) versus the source to chamber distance for the *in-air* cross-axis profiles. Project the best fit line to the width of zero (FWHM = 0) and record the source to chamber distance. Then subtract the source to chamber distance from the calibration setup SSD to obtain the virtual SSD. For example,

for the plot shown below, the virtual SSD would be approximately 90.1 cm. You must calculate the virtual SSD for each electron energy that you commission.



Determining Sigma-Theta-X

This parameter characterizes the angular scattering of electrons in air. Sigma-theta-x can be calculated from the 80% - 20% penumbra width as described by Hogstrom (1981, 1987). Plot the 80% - 20% penumbra width in air versus the isocenter to chamber distance (ICD) for a large field.



Determine the slope of this plot and use it in the following equation to calculate sigma-theta-x.

$$\sigma_{\theta x} = 0.595 \bullet slope_{80\% - 20\%}$$

For example, for the plot above the slope is equal to

$$\frac{1.75 - 0.5}{20 - 0} = 0.0625$$

so

$$\sigma_{\theta x} = 0.0372$$

0.0372 is the product of 0.595 and 0.0625.

Dose measurements in water

For each energy and field size, a number of cross-axis dose profiles must also be measured in water. These scans are used to determine the photon contamination dose and for tuning the off-axis ratios and the water scatter correction factor. Select a detector that adequately resolves the edge of the electron dose distribution in order to accurately model the rapid fall-off in this region.

Required cross-axis dose profiles

For the determination of the photon contamination dose of the electron beam, you must measure the X cross-beam profiles in water at a depth of R_p + 2 cm for each energy and field size. We recommend that you use field sizes of 2 cm, 3 cm, 4 cm, 6 cm, 8 cm, 10 cm, 15 cm, 20 cm, and 25 cm and your maximum field size.

These profiles must be measured in order to model the electron beam.

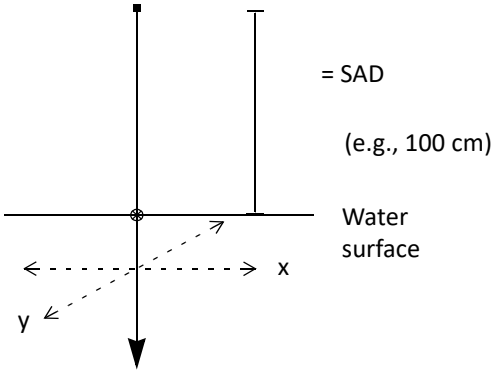
To adjust the off-axis ratios, you must obtain an X and Y cross-beam profile at (1/2) R_{90} (one-half the depth of the second occurrence of the 90% dose along the central-axis depth dose) for each energy and field size. To tune the water scatter correction factor (FMCS), you must measure cross-beam profiles at depths of R_{90}, R_{70}, and R_{50} for each energy and cone size.

For quality assurance, you can gather the data described above at extended SSDs (i.e., 105, 110, 115, and 120) and check the fit in the penumbra for these profiles. Use this data to compare plan profiles. This data cannot be used for modeling. Only 100 cm SSD data is used for modeling.

Required Data	Description
Measurement Type	Open-field orthogonal (horizontal and vertical) dose profiles in water through the central axis.
Field/Cone Sizes	2 cm, 3 cm, 4 cm, 6 cm, 8 cm, 10 cm, 15 cm, 20 cm, and 25 cm and your maximum field size.
Depths	$R_{90/2}$, R_{90}, R_{70}, R_{50}, and R_p + 2 cm

Required Data	Description
Measurement Limits	Measurements should extend as far outside the field boundary as possible.
Setup	Water surface at SSD = SAD

NOTE
You do not need to measure these field sizes in each cone. You only need to measure them in a clinically relevant cone, for example 2 cm, 3 cm, 4 cm, and 6 cm in a 6 cm x 6 cm cone, 8 cm and 10 cm in a 10 cm x 10 cm cone, and then the open 15 cm, 20 cm, and 25 cm cones.



Cutout material transmission factor measurement

For each energy, you will need to determine the photon dose that is transmitted through the cutout material so that the dose contribution from contamination photons can be accounted for in the dose computations. The dose at R_p + 2 cm is considered to be due only to photon contamination. Therefore, measurements for photon transmission through the cutout material should be made at this depth.

Measure the dose at a depth of R_p + 2 cm for a 10 cm x 10 cm open field and, at the same depth, measure the dose for a completely blocked 10 cm x 10 cm field. The ratio of these two measurements is the cutout material photon transmission factor.

$$\text{Cutout material transmission factor} = \frac{Dose_{R_p + 2cm}\text{blocked field}}{Dose_{R_p + 2cm}\text{open field}}$$

Output factor measurements

We recommend that you measure output factors at d_{max} for your clinical range of SSDs starting with the first SSD and in 5 cm increments for each cone size at exposed field sizes of 2 cm, 3 cm, 4 cm, 6 cm, 8 cm, 10 cm, and your maximum field size with a maximum field size for each cone being the cone size. Each field size must be made as a cutout.

If your clinical range of SSDs is 100-120, you would measure output factors at 100, 105, 110, 115, and 120 SSD. And, for a cone size of 6 cm, you would measure the output factor at all the SSDs for field sizes of 2 cm, 3 cm, 4 cm, and 6 cm. So, for the 2 cm and 3 cm cone sizes you would obtain ten output factor measurements.

Cone Size	Field Size	SSD
6 cm	2 cm	100 cm
		105 cm
		110 cm
		115 cm
		120 cm
	3 cm	100 cm
		105 cm
		110 cm
		115 cm
		120 cm

All output factors are relative to the open 10 cm x 10 cm field at 100 SSD and are measured at d_{max}. Monitor units will not be available outside of the range of measurements.

> **NOTE**
> While we recommend that you measure output factors at d_{max}, we do not require it. If you measure output factors at something other than d_{max}, ensure that the measurement geometry that you enter is correct for all relative output factors.

See the *Electron Physics & Physics Utilities* chapter of the *Pinnacle³ Physics Instructions for Use* for more information.

Electron beam data modeling

The *Electron Physics & Physics Utilities* chapter in the *Pinnacle³ Physics Instructions for Use* describes the Electron Physics tool and the electron dose algorithm that it uses. This section provides information about electron beam data modeling.

Adjusting computed profiles

When modeling an electron energy, you enter parameter information and compute profiles as described in the *Electron Physics & Physics Utilities* chapter in the *Pinnacle³ Physics Instructions for Use*. After you have computed profiles for all available measurement geometries, use the following steps to assess and adjust the profiles accordingly in the **Machine Electron Model** window.

1 Compare the measured and computed depth dose curves. If the curves do not match well, re-examine the plot from which you determined the practical range of electrons (R_p). Errors in the computed percent depth dose curves indicate that incident energy and the photon

contamination depth may be incorrect. Make any necessary corrections and recompute all profiles.

2 Compare the X and Y profiles measured and computed at a depth of $R_{90}/2$ to determine whether the off-axis ratios need to be adjusted. To see the profiles in more detail, click the **Detail** button in the **Machine Electron Model** window. If the profiles do not match well, you must adjust the off-axis ratios for the cone size.

> **NOTE**
> Do not use profiles at depths other than $R_{90}/2$ to tune the off-axis ratios. Doing so may cause erroneous results.

3 Using the worksheet entitled *Electron beam measured data worksheet 5* in the *Physics Data Worksheets* chapter of the *Pinnacle³ Physics Instructions for Use*, record the off-axis ratios for points in the shoulders of the profile where the measured and computed profiles do not match well. Record all off-axis ratios for all cone sizes.

4 Enter the off-axis ratios in the Cone X Ratio and Y Ratio tables for all square field sizes for which you have measured data. You must enter the points in the same order in both tables.

5 Compute the profiles again to verify that the changes to the off-axis ratios improved the fit between the measured and computed profiles.

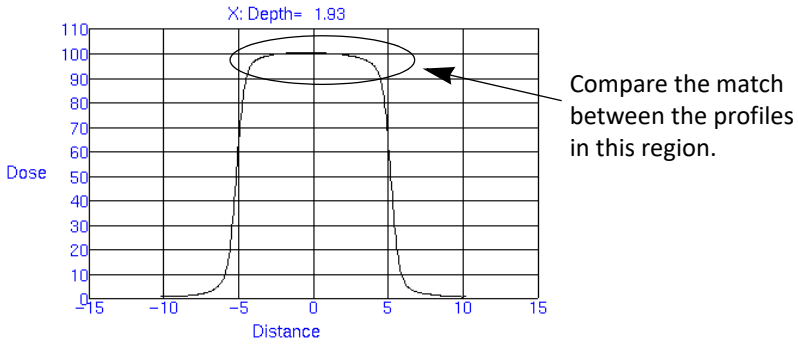
If the profiles still do not match well, repeat step 2. For each point that needs to be changed, multiply the new off-axis ratio by the previous one and enter the value in the off-axis ratio table. Continue to make adjustments until you obtain a good fit between the measured and computed profiles.

6 After adjusting all the other parameters, you can tune the FMCS parameter. Watch the penumbra region of cross-beam profiles at depths greater than R_{90} for the goodness of fit between the measured and computed profiles. Typically the FMCS ranges from 1.2 to 1.4. If you need to exceed 1.4 to obtain a good fit, something else is wrong with the model.

7 When you finish, return to the **Electron Physics Tool** window and click the **Save All Machines** button to save the computed dose lookup tables.

Guidelines for tuning the off-axis ratios

Use the off-axis ratios to shape the “shoulders” of the computed profiles to account for beam asymmetries.

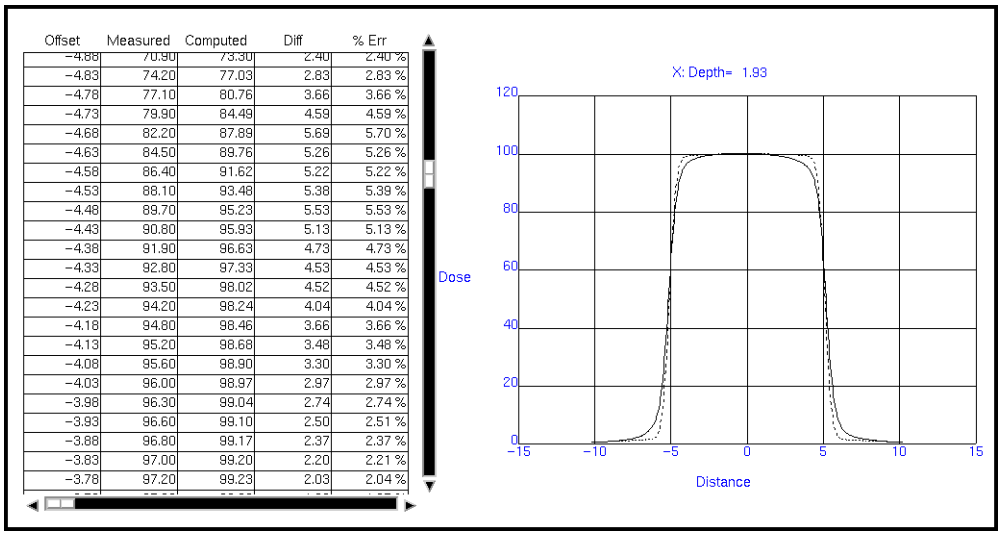


CAUTION
When tuning the off-axis ratios, do not extend your corrections into the high dose gradient region. Doing so within this region could cause dose artifacts for the interpolated field sizes.

To determine how to adjust the off-axis ratios (OAR), first compute the profile at $R_{90}/2$ with off-axis ratios of 1.0 at x = -30, 0, and 30. X defines the off-axis distance. Then, in regions of the profile where the measured and computed profiles do not match well, calculate the off-axis ratios for a number of points using the following equation.

$$OAR_{(x)} = \frac{Measured_{(x)}}{Computed_{(x)}}$$

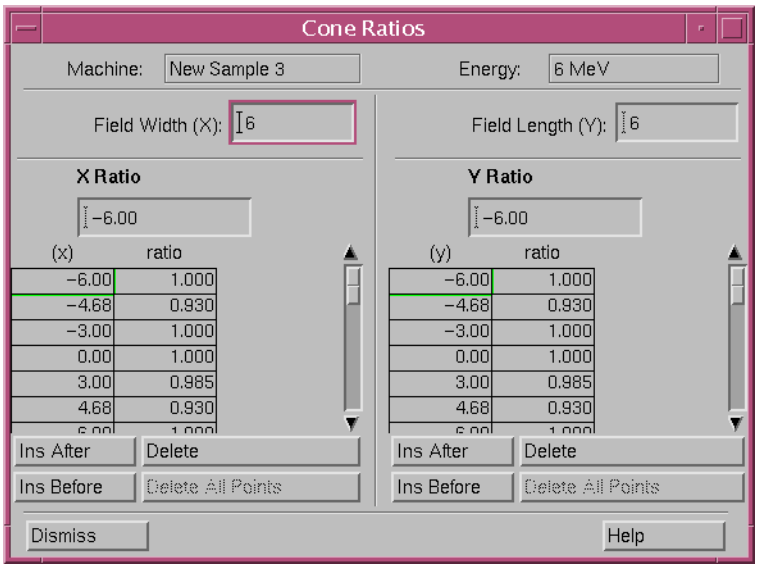
The following example shows the measured and computed profile for a 10 cm cone.



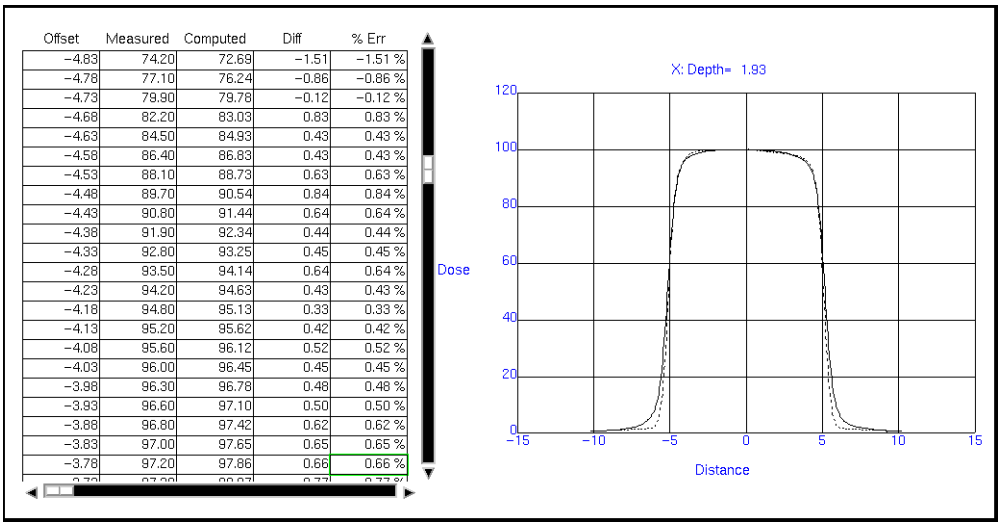
Offset	Measured	Computed	Diff	% Err
−4.88	70.90	73.30	2.40	2.40 %
−4.83	74.20	77.03	2.83	2.83 %
−4.78	77.10	80.76	3.66	3.66 %
−4.73	79.90	84.49	4.59	4.59 %
−4.68	82.20	87.89	5.69	5.70 %
−4.63	84.50	89.76	5.26	5.26 %
−4.58	86.40	91.62	5.22	5.22 %
−4.53	88.10	93.48	5.38	5.39 %
−4.48	89.70	95.23	5.53	5.53 %
−4.43	90.80	95.93	5.13	5.13 %
−4.38	91.90	96.63	4.73	4.73 %
−4.33	92.80	97.33	4.53	4.53 %
−4.28	93.50	98.02	4.52	4.52 %
−4.23	94.20	98.24	4.04	4.04 %
−4.18	94.80	98.46	3.66	3.66 %
−4.13	95.20	98.68	3.48	3.48 %
−4.08	95.60	98.90	3.30	3.30 %
−4.03	96.00	98.97	2.97	2.97 %
−3.98	96.30	99.04	2.74	2.74 %
−3.93	96.60	99.10	2.50	2.51 %
−3.88	96.80	99.17	2.37	2.37 %
−3.83	97.00	99.20	2.20	2.21 %
−3.78	97.20	99.23	2.03	2.04 %

In this profile, the regions from -5.0 to -3.0 and from 3.0 to 5.0 have the most error between the measured and computed profiles. By determining the ratios of the measured values to the computed values at a number of points in these regions and entering the ratios in the Off-Axis Ratio table, you can reshape the computed profile to correct the errors. Enter points at the edges of these regions in the table with Off-Axis Ratios of 1.0.

To correct the profile in the previous example, the following Off-Axis Ratio was entered.



The resulting computed profile is shown below.



You can use *Electron beam measured data worksheet 5* in the *Physics Data Worksheets* chapter of the *Pinnacle³ Physics Instructions for Use* to record the X and Y off-axis ratios that should be used for each cone size.

Electron beam output correction factor

The *Electron Physics & Physics Utilities* chapter in the *Pinnacle³ Physics Instructions for Use* describes how to enter output factors for electron beams. This section provides details about the output correction factor for electron beams.

Determination of scaling factor OF_c

Some effects are not accounted for in the pencil beam algorithm, including the difference in the jaw settings between cone sizes as well as the effect of electrons scattered off the cutout defining the field shape. In the method described below, these effects are included in the correction factor (OF_c). The OF_c depends on the cone selection, the field shape, and the SSD used in the dose computation. The OF_c for a specific beam is multiplied by the raw pencil beam computation to convert the relative dose distribution to dose per monitor unit.

First determine the calibration constant (A_{cal}) to obtain the dose per monitor unit delivered for the reference beam.

$$OF_{cal} = A_{cal} D_{pb}(10 \times 10, 10 \times 10, 100, d_{max})$$

$$A_{cal} = \frac{OF_{cal}}{D_{pb}(10 \times 10, 10 \times 10, 100, d_{max})}$$

where

OF_{cal} = the measured output factor for the calibration condition, and

D_{pb}*(cone size, cutout size, SSD, depth)* = the value computed by the pencil beam algorithm in a flat water phantom along the central axis of the beam.

Next tabulate the correction factors for a range of square field sizes and SSD. These correction factors (OF_c) relate the pencil beam dose computed for square fields back to the calibration condition:

$$OF_p(Cone, W, SSD, d_{max}) = A_{cal} D_{pb}(Cone, W, SSD, d_{max})$$

$$OF_c(Cone, W, SSD) = \frac{OF_{rel}(Cone, W, SSD, d_{max})}{OF_p(Cone, W, SSD, d_{max})}$$

where

OF_{rel} = the measured output factor relative to the calibration condition for the particular square field of interest (W).

Then for any square field (W), cone, and SSD, by applying the correction factors you can compute the actual dose per monitor unit within the dose volume obtained by the pencil beam algorithm.

$$\frac{D}{MU(r)} = OF_c A_{cal} D_{pb}(r)$$

If the pencil beam computation were perfect, the OF_c value would be constant over all SSD and field sizes. We know that the accelerator output changes with jaw settings, and each electron cone uses

different jaw settings; therefore, we can reasonably expect the OF_c to be dependent on the electron cone selection.

In the planning tool, the pencil beam algorithm is scaled by the OF_c to obtain the dose per monitor unit in the beam's dose grid.

Scaling final dose by OF_c for arbitrary field

The dose per monitor unit for an arbitrary field is determined using the following equation.

$$\frac{D}{MU(r)} = OF_c A_{cal} D_{pb}(r)$$

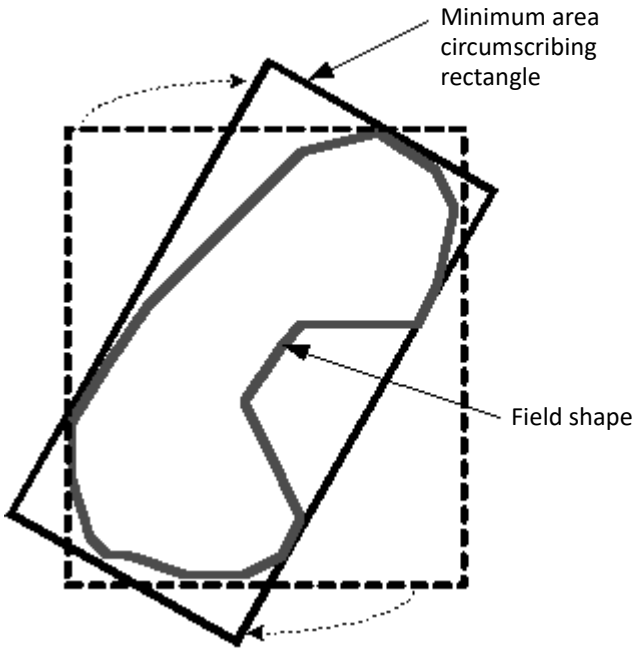
The OF_c is tabulated for each cone at discrete SSDs, and linearly interpolated over SSD to determine the appropriate value for any intermediate SSD. Linear interpolation is used because the inverse square effects are incorporated into the pencil beam algorithm and are accounted for in the OF_p value.

The OF_c is also tabulated for a discrete set of square field sizes. The OF_c value for rectangular fields is found by combining the OF_c values for the square fields matching the size of the rectangular field edges through the square root relationship described in Khan and TG25 (6,7).

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

For irregular field shapes, the minimum area circumscribing rectangle is calculated by rotating a rectangle around the field shape and finding the rotation that provides the minimum area. The OF_c is determined the same way as above, but using the circumscribing rectangular size. For most clinical shapes this is sufficient, however, complicated shapes (severe or multiple protrusions)

present some difficulty due to the relative amount of unexposed area within the minimum circumscribing rectangle.



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5 Stereotactic Radiosurgery Physics & Data Requirements

This section describes the stereotactic radiosurgery dose calculation algorithm and covers the measured data required for stereotactic radiosurgery.

WARNING
The accuracy of the treatment planning dose calculation depends on the quality of the data entered in the physics tool. If the quality of this data is poor, or if the computed dose is not verified with measured data, the dose calculation may be inaccurate.

Stereotactic radiosurgery dose calculation

The Pinnacle[3] stereotactic radiosurgery dose algorithm computes the dose to the target based on interpolation of absorbed dose as measured in water (referred to as “pencil beams”) for all stereotactic collimator sizes. The algorithm is very efficient in computing dose from a large number of pencil beams because interpolating dose values from a table is significantly faster than computing each pencil beam from first principles.

The stereotactic dose calculation uses lookup tables based on measurements in water and assumes that the patient's head is convex and homogenous. As with other dose algorithms, Pinnacle[3] determines the surface of the patient based on the patient-air threshold. If the patient CT image contains a head mask or other positioning aids, this threshold may be crossed by material outside of the patient, potentially impacting the calculation of the SSD and dose. To ensure that the patient surface is detected correctly, you can adjust the patient-air threshold to exclude the positioning aids or override the density in these regions using ROIs. For more information, see *Stereotactic radiosurgery lookup table calculation* in the *Physics Instructions for Use.*

The pencil beam dose algorithm

The fundamental stereotactic dose computation involves determining the dose per monitor unit *D/MU (SSD, d, r, W_c)* at a point from one of the stereotactic pencil beams that make up an arc:

$$\frac{D}{MU}(SSD, d, r, W_c) = \left[\frac{D_{penc}(SSD, d, r, W_c)}{D_{penc}(SSD, d_{iso}, (r = 0), W_c)}\right] \cdot TAR(d_{iso}, W_c) \cdot d_I(W_c)$$

where:

$D_{penc}(SSD, d, r, W_c)$ is the stored pencil beam dose distribution in water at *SSD*, depth *d* and position *r* for a collimator of size W_c as measured at the isocenter,

D_{penc} *(SSD,* d_{iso}*, r = 0,* W_c*)* is the stored pencil beam dose distribution in water at *SSD*, at isocenter on the central axis for a collimator of size W_c as measured at the isocenter, where the isocenter is at depth d_{iso} from the patient surface,

TAR(d_{iso}*,* W_c*)* is the tissue-air-ratio at the isocenter,

d_l *(*W_c*)* is the dose per monitor unit to a small mass of tissue in air at the isocenter.

D_{penc}*(SSD, d, r,* W_c*)* is the product of the fractional depth dose *FDD(SSD, d,* W_c*)* and the off-axis factor *OAF(SSD, d, r,* W_c*)*.

The above equation can then be expressed in terms of more standard dosimetric quantities:

$$\frac{D}{MU}(SSD,d,r,W_c) = \left[\frac{FDD(SDD,d,W_c)\cdot OAF(SSD,d,r,W_c)}{FDD(SSD,d_{iso},W_c)}\right]\cdot TAR((d_{iso},W_c)\cdot d_l(W_c))$$

where:

FDD(SSD, d, W_c*)* is the *fractional depth dose* at *SSD* and depth *d* for a collimator of size W_c as measured at the isocenter:

$$FDD(SSD,d,W_c) = \frac{D(SSD,d,r=0,W_c)}{D(SSD,d_{max},r=0,W_c)}$$

OAF(SSD, d, r, W_c*)* is the *off-axis factor* and is the dose profile of a beam with a given SSD at a given depth normalized to the value at the central axis:

$$OAF(SSD,d,r,W_c) = \frac{D(SSD,d,r,W_c)}{D(SSD,d,r=0,W_c)}$$

TAR(d_{iso}*,* W_c*)* is the *tissue-air-ratio* at the isocenter:

$$TAR(d_{iso},W_c) = \frac{D(d_{iso},W_c)}{D_{air}(W_c)}$$

Small field tissue-air-ratios can be computed using measured fractional depth dose *FDD(SSD,* d_{iso}*,* W_c*)* and small-field back scatter factors *BSF(*W_c*)*:

$$TAR(d_{iso},W_c) = FDD(SSD,d_{iso},W_c)\cdot BSF(W_c)\cdot\left(\frac{SDD+d_{iso}}{SDD+d_{max}}\right)^2$$

where d_{max} is the depth of maximum dose along the central axis.

In this equation, the BSF is expressed as a function of W_c, the field size at the depth of the isocenter. This differs from other common BSF definitions such as the BSF with the field size at the surface of the phantom (Khan) or at the depth of maximum dose (Johns).

$d_l(W_c)$ is the *dose per monitor unit to a small mass of tissue in air at the isocenter* and is determined by:

$$d_l(W_c) = \left(OF(W_c) \cdot \frac{BSF(10 \times 10)}{BSF(W_c)} \cdot \frac{D_{air}}{MU}\right)(10 \times 10)$$

where the *relative output factor* $OF(W_c)$ is given by:

$$OF(W_c) = \frac{D(d_{max}, W_c)}{D(d_{max}, 10 \times 10)}$$

Measured data requirements for stereotactic radiosurgery beams

The stereotactic radiosurgery dose model is based on interpolating dose from stored tabular circular field dose distributions measured in water. The measurements described in the following sections are required for each machine, energy and circular collimator size.

WARNING
In order to correctly generate the tabular dose data, the depth dose and cross-axis measured data profiles must contain one and only one dose value for each position coordinate. In addition, all depth/distance coordinates in the profiles must increase or decrease monotonically.

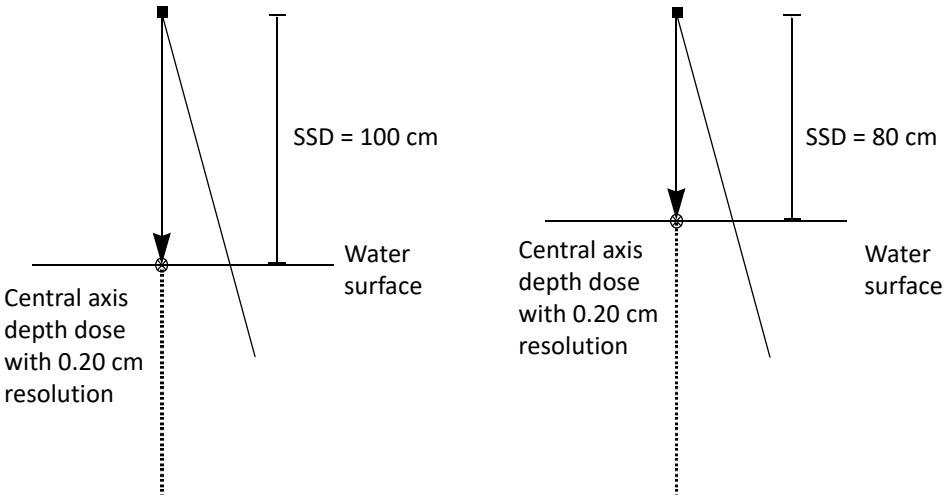
Depth dose measurements

For every collimator used with a given beam energy, you must measure two depth dose curves: one with an SSD of 80 cm and one with an SSD of 100 cm. These depth dose curves will be used to generate the stereotactic radiosurgery dose lookup tables. Additional measurement geometries may be measured, imported into the Stereotactic Radiosurgery Physics tool and compared to the computed profiles for quality assurance purposes. However, only profiles measured at 80 cm SSD and 100 cm SSD will be used to generate the dose lookup tables.

WARNING
The depth dose data is assumed to have a depth value that is a positive measure of distance from the surface of the water. For example, a measurement taken at one centimeter below the surface of the water is assumed to have a depth of 1.0 and not -1.0. Do not enter depth dose data with negative depth coordinates or you will get erroneous results.

Required Measured Data for a Given Energy and Collimator Size	
Measurement Type	Central axis depth dose

Required Measured Data for a Given Energy and Collimator Size	
Measurement Limits	0 to 30 cm
Resolution	0.20 cm increments (minimum)
Setup	SSD = 80 cm and SSD = 100 cm
Modifiers	None (open field)
Total Measurements	Two (2) depth dose curves



Cross-beam dose profiles

For each energy and collimator size, you must measure cross-beam profiles at 1 cm, 5 cm, 10 cm, 20 cm and 25 cm depths using both 80 cm and 100 cm SSDs. The resolution for the cross-beam profiles should be at least 0.20 cm. For smaller collimator sizes (less than 2 cm collimator diameter) a higher resolution should be used.

Required Measured Data for a Given Energy and Collimator Size	
Measurement Type	Cross-beam profiles
Depth	1, 5, 10, 20 and 25 cm
Resolution	0.20 cm or higher
Setup	SSD = 80 cm and SSD = 100 cm
Modifiers	None (open field)
Total Measurements	Ten (10) profiles (5 for each SSD)

Stereotactic radiosurgery output factors

A tissue-air-ratio type of formula is used in the stereotactic dose per monitor unit calculation. (TMR can also be used, as described in the following section.)

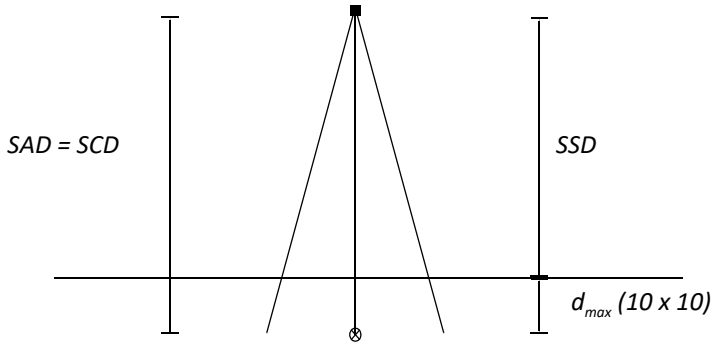
Measurement depth for all measurements

The reference field measurements should be taken isocentrically. So, the Source to Surface Distance (SSD) should be equal to the Source to Axis Distance (SAD) minus the depth of maximum dose for the 10 x 10 field.

SSD = SAD - d_{max}(10 x 10)

The Source to Calibration point Distance (SCD) should be equal to the Source to Axis Distance (SAD).

SCD = SAD



Reference field measurement

D_{air}/MU (10 x 10)—Dose per monitor unit to a small mass of tissue in air at isocenter for the 10 cm x 10 cm field.

BSF (10 x 10)—The back scatter factor (i.e., the tissue-air-ratio at d_{max}) for the 10 cm x 10 cm field.

For each energy and collimator size

ROF (W_c)—The relative output factor for collimator diameter W_c.

$$ROF = \frac{\text{Reading at } d_{max}(W_c)}{\text{Reading at } d_{max}(10 \times 10)}$$

BSF (W_c)—The back scatter factor which can either be measured or interpolated between the 0 cm x 0 cm and the 10 cm x 10 cm field size back scatter factors.

Using TMR instead of TAR

For stereotactic collimators, it is straightforward to use *TMR* instead of *TAR* when utilizing energies where air measurements are difficult.

We begin with the assumption that at high energies and with small field sizes, the backscatter factor (BSF) for the circular collimator will be very close to unity.

(1) $$BSF(W_c) = 1.0$$

The term backscatter factor is the tissue-air ratio at the depth of maximum dose on the central axis of the beam. It is defined as the ratio of the dose on the central axis at the depth of maximum dose to the dose at the same point in free space (Khan, 1994).

Because

(2) $$TMR(d, W_c) = \frac{TAR(d, W_c)}{BSF(W_c)}$$

It follows that for high energy, small fields:

(3) $$TMR(d, W_c) = TAR(d, W_c)$$

Thus, a collimator backscatter factor of 1.0 can be specified when the energy is high, and the *TAR* values are then equivalent to *TMR*.

Pinnacle[3] uses a quantity labelled d_l which is defined by the following equation:

(4) $$d_l(W_c) = \left(OF(W_c) \cdot \frac{D_{air}}{MU}\right)\left((10 \times 10) \cdot \frac{BSF(10 \times 10)}{BSF(W_c)}\right)$$

Because $BSF(W_c) = 1.0$, we have:

(5) $$d_l(W_c) = OF(W_c) \cdot \frac{D_{air}}{MU}(10 \times 10) \cdot BSF(10 \times 10)$$

The backscatter factor is the dose at d_{max} divided by the dose in free space at the same point. Thus, the second two terms of this equation can be multiplied, and the air measurement cancels, yielding the following equation:

(6) $$d_l(W_c) = OF(W_c) \cdot \frac{D}{MU}(d_{max}, 10 \times 10)$$

The Pinnacle[3] dose equation is given as follows:

$$\frac{D}{MU}(SSD,d,r,W_c) = \left[\frac{D_{penc}(SSD,d,r,W_c)}{D_{penc}(SSD,d_{iso},r=0,W_c)}\right] \cdot TAR(d_{iso},W_c)\cdot d_l(W_c) \quad (7)$$

where:

$D_{penc}(SSD, d, r, W_c)$ is the dose at depth d, and off axis distance r from the stored pencil beam dose distribution in water at SSD, for a collimator of size W_c,

$D_{penc}(SSD, d_{iso}, r = 0, W_c)$ is the dose at isocenter for a collimator of size W_c at SSD, where the isocenter is at depth d_{iso} from the patient surface,

$TAR(d_{iso}, W_c)$ is the tissue-air-ratio at the isocenter,

$d_l(W_c)$ is the dose per monitor unit to a small mass of tissue in air at the isocenter.

Replacing *TAR* with *TMR* and replacing d_l with the reduced d_l defined in equation 6 above yields the familiar *TMR* equation:

$$\frac{D}{MU}(SSD,d,r,W_c) = \left[\frac{D_{penc}(SSD,d,r,W_c)}{D_{penc}(SSD,d_{iso},r=0,W_c)}\right] \cdot TMR(d_{iso},W_c)\cdot d_l(W_c) \quad (8)$$

So, to avoid in-air measurements you can do the following:

- Enter a collimator backscatter factor $(BSF(W_c))$ of 1.0.
- Enter a reference field backscatter factor ($BSF(10 \times 10)$) of 1.0.
- For the $D_{air}/MU(10 \times 10)$ value, enter $D/MU(d_{max}, 10 \times 10)$.

In the planning software, the **Monitor Units** window reports d_l as defined by the equation:

$$d_l(W_c) = OF(W_c)\cdot \frac{D}{MU}(d_{max}, 10\times 10) \quad (9)$$

If the machine is calibrated such that $D/MU(d_{max}, 10 \times 10) = 1cGy/MU$, then

$$d_l(W_c) = OF(W_c) \quad (10)$$

6 Brachytherapy Physics

This chapter describes the Pinnacle³ brachytherapy dose algorithm and the required source characteristic data.

Brachytherapy dose calculation

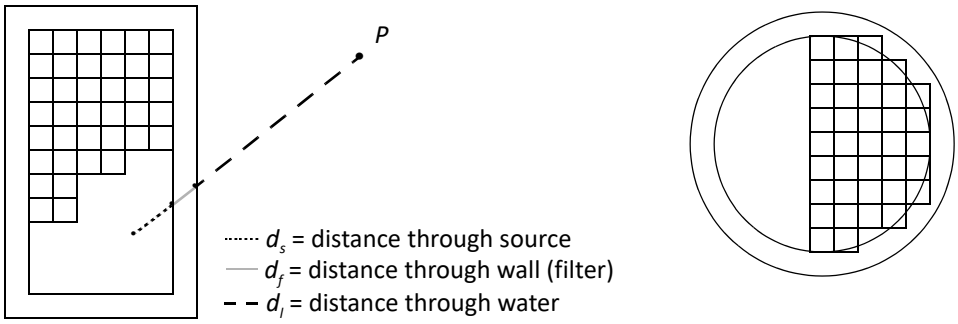
Pinnacle³ uses a two-dimensional, symmetric dose lookup table for determining dose from brachytherapy sources. The dose calculation is based on a homogeneous medium.

CAUTION
No heterogeneity corrections are applied, regardless of whether you plan from CT or film.

The Pinnacle³ brachytherapy physics tool provides two methods of calculating the dose lookup tables: Geometric and TG43. Geometric calculation is based on the Quantization method described by Cassell (1983). The TG43 calculation method uses the TG43 dose formalism (Nath, et al., 1995) when calculating the lookup table.

The geometric dose calculation method

In the Pinnacle³ software, the Cassell calculation method is called the Geometric method. When using the geometric calculation, the source is divided into equally sized elements, as shown in the diagram below.



Each element is treated as a point source when the dose rate for a line source is computed. The source is divided into a minimum of 20 voxels in each dimension. Sources that are larger than 0.5 cm use voxels of 0.025 cm in size.

The absorbed dose rate at a time *t* to point *p* from a single shielded line source, $D_p(t)$, can be determined from the following equation.

$$D_p(t) = f_{med} \cdot \Gamma \cdot A(t) \cdot \frac{1}{N} \sum_{i}^{N} e^{(-\mu_s d_s - \mu_f d_f)} \cdot ((T(d_l)) / d_i^{\,2})$$

where:

f_{med} is the exposure-to-dose conversion factor for soft tissue.

G is the specific exposure rate constant (R cm^2 mCi^{-1} h^{-1}) for the source.

A(t) is the activity, in mCi, of the source at the time of treatment. If activity is specified in units other than mCi, the system internally converts the activity to mCi for the dose calculation.

N is the number of point sources (elements) used to model the source.

μ_s is the linear absorption coefficient of the source material.

d_s is the path length of the radiation through the source.

μ_f is the linear absorption coefficient of the wall material.

d_f is the path length of the radiation through the wall.

d_l is the path length of the radiation through water.

d_i is the distance from the *i*th point source to the dose point *P*.

$d_i = d_s + d_f + d_l$

$T(d_l)$, the tissue correction, is the polynomial (i.e., Meisberger coefficients):

$$T(d_l) = \frac{\text{Exposure in water}}{\text{Exposure in air}} = A + Br + Cr^2 + Dr^3$$

A lookup table is computed which stores the “shape” of the dose. This table is calculated using the summation:

$$\frac{1}{N} \sum_{i}^{N} e^{(-\mu_s d_s - \mu_f d_f)} \cdot ((T(d_l)) / d_i^{\,2})$$

CAUTION
The dose algorithm used by the Pinnacle³ software requires the entry of a tissue correction factor, $T(d_l)$, represented by Meisberger polynomial coefficients. This factor is different from the radial dose function, g(r), described by the TG43 dose formalism (Nath, et al.).

CAUTION
Because labeling in the Pinnacle³ software once referred to Radial Dose Function data entry, it was possible to conclude that the radial dose function coefficients published for the TG43 dose

formalism could be used. The values for the radial dose function coefficients provided for the TG43 formalism should be used only with a TG43 dose algorithm. If they are used with any other dose formalism, such as the Quantization Method implemented in the Pinnacle³ software, significant dose calculation errors may result (Cassell, "A fundamental approach to the design of a dose-rate calculation program for use in brachytherapy," *British Journal of Radiology*, 1983).

If the source is a point source rather than a line source, the distance through the source (d_s) and the distance through the wall (d_f) are both 0, and $exp(0) = 1$. Additionally, since there is only one point source, activity summation is not necessary.

Therefore, the dose calculation equation

$$D_p(t) = f_{med} \cdot \Gamma \cdot A(t) \cdot \frac{1}{N} \sum_{i}^{N} e^{(-\mu_s d_s - \mu_f d_f)} \cdot \left((T(d_l)) / d_i^{\,2} \right)$$

becomes

$$D_p(t) = f_{med} \cdot \Gamma \cdot A(t) \cdot \left((T(d_l)) / d^2 \right)$$

Source strength decay calculation

In the Pinnacle³ software, the activity for a source is specified during treatment planning rather than in the physics tool. In addition, the calibration date and time and the treatment date and time are specified during treatment planning. However, the half-life for an isotope is specified in the physics tool.

The source activity at time *t* is calculated by the software using the following equation:

$$A(t) = A(t_{cal}) \cdot e^{-\beta(t - t_{cal})}$$

where

t_{cal} is the time of calibration

t is the time of treatment and

β = ln(2) / half life

In Pinnacle³'s source strength decay calculations, time is handled in seconds. The time conversions into seconds are as follows:

1 year = 365 • 24 • 60 • 60 seconds

1 month = (365/12) • 24 • 60 • 60 seconds

1 day = 24 • 60 • 60 seconds

1 hour = 60 • 60 seconds

Activity conversions

In all internal calculations, the system defines activity (*A (t)*) in mCi. When the activity is defined in units other than mCi, it is converted to mCi using the following methods.

Units	Conversion to mCi
mCi	1.0
Ci	1000.0 mCi/Ci = 1000.0
Bq	(1Bq / 10^6 MBq) * (1mCi /37.0 MBq) = 1 / (37 * 10^6)
MBq	(1.0 mCi / 37.0 MBq) = (1 / 37)
mgRaEq	8.25 R cm^2 mCi^{-1} hr^{-1}/ Γ
R*m^2/h	(100 cm / m)^2 / Γ
R*cm^2/h	1.0 / Γ
cGy*cm^2/h	1.0 / ((W/e) * Γ)
cGy*m^2/h	(100 cm / m)^2 / ((W/e) * Γ)
U	1.0 / ((W/e) * Γ)

where:

Γ is the exposure rate constant in units of R cm^2 mCi^{-1} hr^{-1}, and

W/e is the air exposure to dose constant expressed in cGy/R

TG43 dose computation method

The TG43 computation method uses the dose formalism proposed by the AAPM Task Group No. 43. This protocol allows for two-dimensional dose calculations around cylindrically symmetric sources. The dose formalism is described by Nath, et al., as follows.

General formalism for two-dimensional case

The TG43 formulas consider the source to be cylindrically symmetrical. For such sources, the dose distribution is two-dimensional and can be described in terms of a polar coordinate system with its

origin at the center, where *r* is the distance to the point of interest and θ is the angle with respect to the long axis of the source. The dose rate, $\dot{D}(r, \theta)$, at point (r, θ) can be written as

$$\dot{D}(r,\theta) = S_k \Lambda [G(r,\theta)/G(r_0,\theta_0)] g(r) F(r,\theta)$$

where:

S_k is the air kerma strength of the source (units of U)

Λ is the dose rate constant (units of cGy h^{-1} U^{-1})

$G(r,\theta)/G(r_0,\theta_0)$ is the geometry factor normalized to the reference point

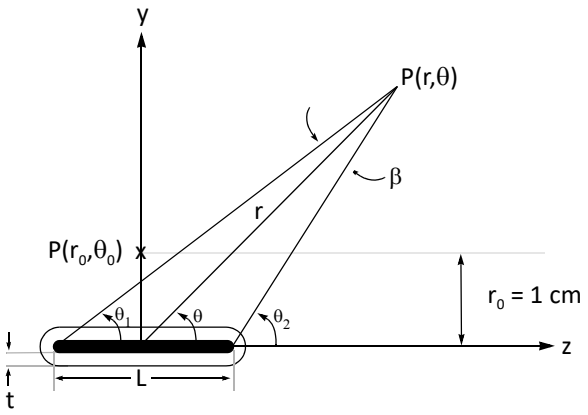
$g(r)$ is the radial dose function

$F(r, \theta)$ is the anisotropy function.

Each of these quantities or functions and the reference point (r_0, θ_0) are described in the following sections.

Reference point for dose calculations

The reference point (r_0, θ_0) lies on the transverse bisector of the source at a distance of 1 cm from its center (r_0 = 1 cm and $\theta_0 = \pi/2$).



Air kerma strength, S_k

Air kerma strength is a measure of brachytherapy source strength, which is defined as the product of air kerma rate at a calibration distance in free space, *K(d)*, measured along the transverse bisector of the source, and the square of the distance, *d*.

$$S_k = \dot{K}(d) d^2$$

The calibration distance must be large enough that the source may be treated as a mathematical point. In actual practice, air kerma rate standardization measurements are performed in air and corrections for air attenuation are applied if needed. While the measurements for source strength calibration may be performed at any large distance, *d*, it is customary to specify the air kerma

strength in terms of a reference calibration distance, d_0, which is usually 1 m. The physicist typically does not perform the in-air calibration; a standardization laboratory does. However, the physicist should verify the accuracy of source strength provided by the vendor. Typically, you should have a well-type ionization chamber with a calibration traceable to the national standards for each type of brachytherapy source.

If kerma, time, and distance are assigned units of μGy, h, and m, respectively, S_k will have units of μGy m^2 h^{-1}, as recommended by the TG43 report. This unit is denoted by the symbol U:

1 U = 1 unit of air kerma strength = 1 $\mu Gy\ m^2\ h^{-1}$ = 1 $cGy\ cm^2\ h^{-1}$

The geometric relationship between the point of output determination and an arbitrary filtered source have been described previously by Williamson and Nath.

Dose rate constant, Λ

The dose rate constant is the dose rate to water at 1 cm on the transverse axis of a unit air kerma strength source in a water phantom. The dose rate constant is an absolute quantity, unlike the normalized (relative) quantities that follow in this section. Liquid water is the reference medium used to specify the dose rate constant as well as relative dose distribution parameters. The 1 cm distance is specified along the transverse axis of the actual source (rather than an idealized point source) relative to its geometric center. Mathematically, the dose rate constant is

$$\Lambda = \dot{D}(r_0, \theta_0) / S_k$$

The constant includes the effects of source geometry, encapsulation, the spatial distribution of radioactivity within the source, self-filtration within the source, and scattering in water surrounding the source. This quantity also depends on the standardization measurements to which the air kerma strength calibration of the source is traceable. If the air kerma strength standard for a given source provided by NIST is changed in the future, the value of Λ also changes.

Geometry factor, G(r, θ)

The geometry factor accounts for the variation of relative dose due only to the spatial distribution of activity within the source, ignoring photon absorption and scattering in the source structure. It is defined as

$$G(r, \theta) = \frac{\int v[\rho(r')dV' / |r' - r|^2]}{\int v\rho(r')dV'}$$

where $\rho(r')$ represents the density of radioactivity at the point $\rho(r') = \rho(x', y', z')$ within the source and V denotes integration over the source core. dV' is a volume element located at r' in the source. Because the three-dimensional distribution of $\rho(r)$ is uncertain for many sources such as ^{125}I and because the choice of $G(r, \theta)$ influences only the accuracy of interpolation, the line source approximation to $G(r, \theta)$ is used.

When the distribution of radioactivity can be approximated by a point source or by a line source of length, *L*, the $G(r, \theta)$ reduces to r^{-2} for point source approximation, and to the following for line source approximation:

$$G(r, \theta) = \frac{\beta}{Lr \sin \theta}$$

where *L* is the active length of the source and β is the angle subtended by the active source with respect to the point (r, θ) $[\beta = \theta_2 - \theta_1]$.

Pinnacle[3] reports $G(r,\theta)/G(r_0,\theta_0)$ when it computes the Geometry function, which normalizes the displayed table to the reference point (r_0,θ_0).

CAUTION
If you manually enter TG43 geometry factors, Pinnacle[3] will automatically normalize them to be 1 at a distance of 1 cm from the center of the source. This function is indicated by the following equation, which appears when you manually enter TG43 geometry factors: G(r, theta)/G(1 cm, 90 deg).

Radial dose function, g(r)

The radial dose function, *g(r)*, accounts for the fall-off of dose rate along the transverse axis due to absorption and scattering in the medium. It is defined as

$$g(r) = \dot{D}(r,\theta_0)G(r_0,\theta_0) / \dot{D}(r_0,\theta_0)G(r,\theta_0)$$

The radial dose function applies only to transverse axis (only for points with an angle of θ_0, which is equal to $\pi/2$). It can be influenced by filtration of photons by the encapsulation and source materials.

The function *g(r)* is similar to a normalized transverse-axis tissue-attenuation factor or an absorbed dose to kerma in free space ratio. The above definition of radial dose function is different from the older (Dale) definition.

Anisotropy function, F(r,θ)

This function accounts for the anisotropy of dose distribution around the source, including the effects of absorption and scatter in the medium. It is defined as

$$F(r,\theta) = \dot{D}(r,\theta)G(r,\theta_0) / \dot{D}(r,\theta_0)G(r,\theta)$$

This two-dimensional function gives the angular variation of dose rate about the source at each distance due to self-filtration, oblique filtration of primary photons through the encapsulating material, and scattering of photons in the medium. The role of the geometry factor is to suppress the influence of inverse square law on the dose distribution around the source.

Due to the large dose rate gradients encountered near interstitial sources, it is difficult to measure dose rates accurately at distances less than 5 mm from the source. In addition, the large dose rate variation arising from inverse square law makes it difficult to accurately interpolate intermediate

dose rate values without an excessively large table of measured data. By suppressing inverse square law effects, you can more accurately extrapolate to small distances from dose rate profiles measured at distances of 5 and 10 mm, as well as interpolate between sparsely distributed measured values.

Point isotropic source approximation

Some clinical treatment planning systems for interstitial brachytherapy use the one-dimensional isotropic point source model to compute interstitial source dose distributions. In this approximation, dose depends only on the radial distance from the source. If a large number of seeds are randomly oriented, or the degree of dose anisotropy around single sources is limited, the dose rate contribution to tissue from each seed can be well approximated by the average radial dose rate, as estimated by integrating the single anisotropic seed source with respect to solid angle:

$$\dot{D}(r) = \frac{1}{4\pi}\int_0^{4\pi}\dot{D}(r,\theta)d\Omega$$

where $d\Omega = 2\pi \sin\theta\, d\theta$ for a cylindrically symmetric dose distribution.

Anisotropy factor, $\phi_{an}(r)$

If we substitute the general TG43 equation below into the one above:

$$\dot{D}(r,\theta) = S_k\Lambda[G(r,\theta)/G(r_0,\theta_0)]g(r)F(r,\theta)$$

and then rearrange, we can devise the following equation:

$$\dot{D}(r) = S_k\Lambda\frac{G(r,\theta_0)}{G(r_0,\theta_0)}g(r)\phi_{an}(r)$$

where $\phi_{an}(r)$ is the anisotropy factor

$$\phi_{an}(r) = \frac{\int_0^{\pi}\dot{D}(r,0)\sin\theta d\theta}{2D(r,\theta_0)}$$

The factor $\phi_{an}(r)$ is the ratio of the dose rate at distance r (averaged with respect to solid angle) to dose rate on the transverse axis at the same distance. For the sources, $\phi(r)$ is less than 1, having values ranging from 0.91 to 0.97 depending upon the source. For distances greater than the source active length, the equation for dose around a source using point-source approximation simplifies to:

$$\dot{D}(r) = \langle S_k\Lambda r_0^2/r^2\rangle g(r)\phi_{an}(r)$$

Anisotropy constant, ϕ_{an}

The anisotropy factor $\phi(r)$ may be approximated by a distance-independent constant, ϕ_{an}, which we call the anisotropy constant. It is usually less than 1.00. Thus, the anisotropy factor, $\phi_{an}(r)$, in the above equation can be replaced by a constant ϕ_{an} without a significant loss in accuracy. Point source approximation, as shown in the two previous equations, gives a dose rate at the reference point in the medium on the transverse bisector at 1 cm from the source, equal to $\Lambda\phi_{an}(r)$ for a unit air kerma strength source. Thus, dose rate on the transverse axis in the medium is somewhat lower using the point-source approximation than the actual dose rate by 3% to 9% for the sources.

References and recommended reading

Cassell, K.J. 1983. A fundamental approach to the design of a dose-rate calculation program for use in brachytherapy. *The British Journal of Radiology* 56:113-119.

Dale, R.G. 1983. Some theoretical derivations relating to the tissue dosimetry of brachytherapy nuclides, with particular reference to iodine-125. *Medical Physics* 10:176-183.

Meisberger, L.L., R.J. Keller, and R.J. Shalek, 1968. The effective attenuation in water of the gamma rays of gold 198, iridium 192, cesium 137, radium 226, and cobalt 60. *Radiology* 90:953-957.

Nath, R., L.L Anderson, G. Luxton, K.A. Weaver, J.F. Williamson, and A.S. Meigooni. 1995. Dosimetry of interstitial brachytherapy sources: Recommendations of the AAPM Radiation Therapy Committee Task Group No. 43. *Medical Physics* 22(2):209-234.

Williamson et al, 1999. Guidance to users of Nycomed Amersham and North American Scientific, Inc., I-125 Interstitial Sources: Dosimetry and calibration changes: Recommendations of the American Association of Physicists in Medicine Radiation Therapy Committee Ad Hoc Subcommittee on Low-Energy Seed Dosimetry. *Medical Physics* 26(4): 570-573.

Williamson, et al, 2000. Recommendations of the American Association of Physicists in Medicine on ^{103}Pd interstitial source calibration and dosimetry: Implications for dose specification and prescription. *Medical Physics* 27(4): 634-642.

7 Additional Reference Information

This chapter explains the following aspects of file handling with Pinnacle³:

- The first section lists manufacturers whose water phantom data is compatible with Pinnacle³.
- The next section explains how to import measured data files into Pinnacle³ using the full or simple ASCII format for dose profiles.

Compatible water phantom formats

The manufacturers in the following list make water phantoms whose software and data are compatible with Pinnacle³:

- Scanditronix Wellhöfer—When importing this data into Pinnacle³, use the legacy Wellhöfer 600 or 700 series ASCII format, or the full ASCII format. Do not use Binary format. For more information, contact Customer Support and ask for Application Note 2000-09.
- MultiData Full ASCII—Before importing this data into Pinnacle³, save the files in Comma Separated Value (CSV) format. Depending on how your profiles were generated, the scaling of the profile data will not reflect the changing scale factor with depth.For more information, contact Customer Support and ask for Application Note 2000-08.
- RFA Scanditronix—Their software outputs data in Pinnacle³ full ASCII format, which is described in detail later in this appendix. Contact your Scanditronix Wellhöfer vendor to obtain the software.
- CMS Dynascan—Their software outputs data in Pinnacle³ full ASCII format, which is described in detail later in this appendix. Contact your CMS Dynascan vendor to obtain the software.
- PTW scanning system—Their software outputs data in Pinnacle³ full ASCII format, which is described in detail later in this appendix. Contact your PTW vendor to obtain the software.
- XL-Plan—For details on data compatibility, contact your XL-Plan vendor.

Use the tabular format if you enter values by hand.

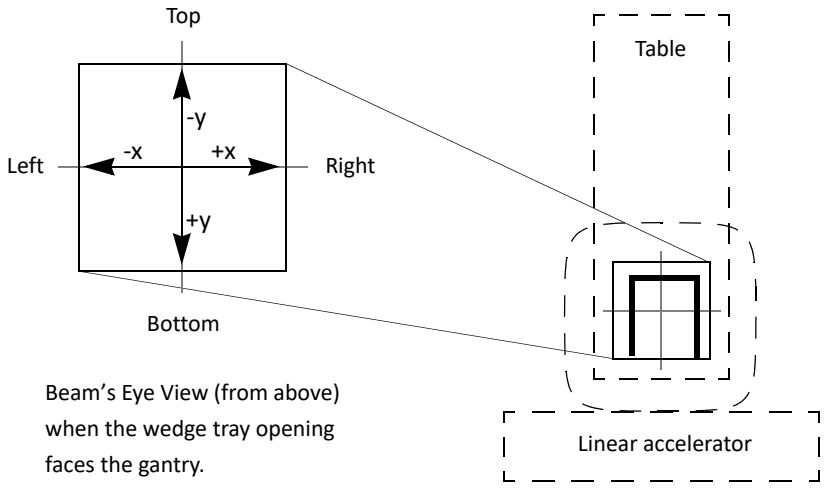
> **NOTE**
> If your phantom manufacturer is not listed, check with them. It is possible that their format is compatible with Pinnacle³.

Measured data file formats

Pinnacle[3] can import measured data using the following ASCII file formats:

- The full ASCII file format includes the energy, SSD and field size in the file. When using the full ASCII format, multiple profiles can be stored in a single file. This format also supports wedge name and circulator collimator size (optional).
- The simple ASCII format can only be used for individual profiles and does not include the energy or setup geometry in the file.

Offsets for X and Y profiles are measured out from the central axis as shown below.



Full ASCII file format for dose profiles

The full ASCII file format used by Pinnacle[3] allows you to import measured data files containing either a single profile or multiple profiles. The full ASCII file format includes information on the energy and geometry used for the measurement in the file. When using the full ASCII file format, all profiles in a given file must have the same energy, SSD and field size.

The full ASCII format requires the data to be stored as numeric values or unquoted character strings separated by spaces using the following format. Variables in bold typeface are described on the following page. Wedges and circular collimators are optional, but when they are used, the wedge name and circular collimator size must be prefaced with the text strings shown below in italics, and the wedge and collimators must be defined for the machine prior to importing the files. If the name or diameter values contain spaces, the name or diameter value must be surrounded by double quotation marks. For example, for a wedge called *45 degree*, the format for the wedge name line in the file would be:

WedgeName “45 degree”

The Energy, Number of Profiles and Number of Points must be integers. The rest of the values may be integer or floating point.

FileType

Energy SSD

Left Jaw Right Jaw Top Jaw Bottom Jaw

WedgeName Name

CircularCollimatorSize Diameter

Number of Profiles

Profile Type Measurement Parameter A Measurement Parameter B

Number of Points

Depth or **Offset** Measured Dose

..

..

..

Depth or **Offset** Measured Dose

Profile Type Measurement Parameter A Measurement Parameter B

Number of Points

Depth or **Offset** Measured Dose

..

..

..

Depth or **Offset** Measured Dose

The **FileType** must be *PinnDoseProfile*.

Left Jaw, **Right Jaw**, **Top Jaw**, and **Bottom Jaw** must be expressed in centimeters.

Profile Type must be *DepthDose*, *XProfile* or *YProfile*.

Measurement Parameter A and **Measurement Parameter B** vary depending on the Profile Type, as follows:

	Parameter A	**Parameter B**
For DepthDose profiles:	X Offset	Y Offset
For XProfile profiles:	Profile Depth	Y Offset
For YProfile profiles:	Profile Depth	X Offset

Depth and **Offset** values must be expressed in centimeters.

NOTE
It is not necessary to normalize the dose values because all dose measurements are normalized within Pinnacle[3].

The following example shows a portion of a full ASCII format file containing a depth dose curve and an X profile curve:

`PinnDoseProfile`	(indicates file type)
`18 100`	(18 MV energy, 100 SSD)
`5 5 5 5`	(10 x10 cm field size)
`2`	(two profiles)
`DepthDose 0.010 0.00`	(depth dose data, X offset = 0.01 cm, Y offset = 0 cm)
`141`	(141 points in the depth dose)
`0.000 21.900`	(first depth = 0 cm, first dose = 21.900)
`0.250 27.100`	(second depth = 0.25 cm, second dose = 27.100)
.	
.	
.	
`35.000 27.799`	(last depth = 35 cm, last dose = 27.799)
`XProfile 10.000 0.000`	(X profile data, Profile depth = 10 cm, Y offset = 0 cm)
`121`	(121 points in the second profile)
`0.250 81.700`	(first offset = 0.25 cm, first dose = 81.7)
`0.500 82.300`	(second offset = 0.5 cm, second dose = 82.3)
.	
.	
.	
`30.000 0.130`	(last offset = 30 cm, last dose = 0.13)

Simple ASCII file format for dose profiles

Individual measured data profiles can be written out in a simple ASCII file format and imported into the physics tools. The simple ASCII file format requires the total number of measured points in the file to be listed on the first line and the points and dose values to be listed one per line and separated by a space, as follows:

Total number of points

Depth or offset (in cm) Measured dose

It is not necessary to normalize the dose values, as all dose measurements are normalized within Pinnacle^3.

The following example shows a portion of an ASCII format file for a depth dose curve containing 141 measurements (from 0 to 35 cm depth using 0.25 cm spacing).

```
141
0.000000 21.900000
0.250000 27.100000
0.500000 44.299999
0.750000 57.000000
1.000000 69.300003
1.250000 76.900002
1.500000 83.900002
1.750000 88.500000
2.000000 92.000000
2.250000 94.699997
.
.
.
34.000000 28.900000
34.250000 28.700001
34.500000 28.500000
34.750000 28.299999
35.000000 27.799999
```

www.Philips.com/Pinnacle

The online repository of is www.philips.com/ifu.

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