CHAPTER 10.

ACCEPTANCE TESTS AND COMMISSIONING MEASUREMENTS

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10.1. INTRODUCTION

Following the installation of a therapy machine, be it an orthovoltage x-ray unit, cobalt unit, linear accelerator or a brachytherapy machine, in a radiation therapy clinic, the medical physicist must perform a series of measurements and tasks prior to placing the unit into clinical operation. These duties include *acceptance testing* and *commissioning*. Although calibration of the treatment beams are a part of the acceptance tests and commissioning, calibration will not be discussed in this chapter as it is fully covered in Chapter 9.

10.2. MEASUREMENT EQUIPMENT

10.2.1. Radiation survey equipment

A Geiger counter and a large volume ionisation chamber survey meter are required for radiation survey for all treatment rooms. For facilities with a treatment unit operated above 10 MeV, neutron survey equipment such as Bonner spheres, long counters and BF_3 counters are necessary. However, it may be appropriate to contract neutron measurements to a medical physics consulting service. This may be a less expensive option than developing the specialized skills and knowledge required for most neutron measurements and acquiring the expensive neutron detection equipment that is typically required only during the acceptance tests.

10.2.2. Ionometric dosimetry equipment

A variety of ionisation chambers are required to compile the radiation beam properties measured during acceptance testing and commissioning of a radiation treatment unit. Thimble ionisation chambers with volumes on the order of 0.1 - 0.2 cm^3 are used to measure a number of relative quantities and factors. These relative factors, including central axis percentage depth doses, output factors, and penumbra, may exhibit a rapidly changing dose gradient. In this situation small volume ionisation chambers are preferred to reduce the uncertainty in the effective point of measurement. For measurements in the build up region that exhibits the greatest change in dose gradient, a parallel plate or extrapolation chamber is required. Calibration measurements are typically performed with a thimble ionisation chamber with a volume on the order of 0.5 cm^3 to increase the signal-to-noise ratio. A single electrometer that can be used with all these ionisation chambers is a wise choice.

10.2.3. Film

Radiographic film has a long history of use in radiation therapy physics measurements. It has been used most successfully for quality control and electron beam measurements. However, the composition of radiographic film is very different from that of tissue, which makes it difficult for use in photon beam dosimetry.

In the past decade radiochromic film has been introduced into radiotherapy physics practice. This film is more tissue equivalent than radiographic film and is becoming more widely used for photon beam dosimetry.

Film dosimetry also requires a densitometer to evaluate the darkening of the film and to relate the darkening to the radiation received. It should be noted that different densitometers are suggested for radiochromic film than for conventional radiographic film as the absorption peaks occur at different wavelengths for these different films.

10.2.4. Diodes

Because of their small size silicon diodes are convenient for measurements in small photon radiation fields. Diodes are also used for electron beam measurements because the stopping power ratio of silicon to water is almost constant over the energies measured in radiation therapy. The response of diodes should be checked against ionometric measurements before routine use.

10.2.5. Phantoms

Radiation field analyzer and water phantom

A water phantom that scans ionisation chambers or diodes in the radiation field is required for acceptance testing and commissioning. This type of water phantom is frequently referred to as a radiation field analyzer (RFA) or an isodose plotter. Although a two dimensional RFA is adequate, a three dimensional RFA is preferable, as it allows the scanning of the radiation field in orthogonal directions without changing the phantom setup.

The traversing mechanism for the ionisation chambers or diodes may also be used to move the film densitometer. The traversing mechanism should have an accuracy of movement of 1 mm and a precision of 0.5 mm. A three dimensional scanner of the RFA should be able to scan 50 cm in both horizontal dimensions and 40 cm in the vertical dimension. The water tank should be at least 10 cm larger than the scan in each dimension.

The RFA should be filled with water and then positioned with the radiation detector centered on the central axis of the radiation beam. The traversing mechanism should move the radiation detector along the principal axes of the radiation beam. After the gantry has been leveled with the beam directed vertically downward, leveling of the traversing mechanism can be accomplished by scanning the radiation detector along the central axis of the radiation beam indicated by the image of the cross-hair. Any deviation of the radiation detector from the central axis, as the detector is moved away from the water surface, indicates that the traversing mechanism is not leveled.

Plastic phantoms

For ionometric measurements in the build-up region a polystyrene or water equivalent plastic phantom is convenient. A useful configuration for this phantom consists of 10 blocks 25 cm × 25 cm × 5 cm. One block should be drilled to accommodate a Farmer-type ionisation chamber with the center of the hole, 1 cm from one surface. A second block should be machined to place the entrance window of a parallel plate chamber at the level of one surface of the block. This arrangement allows measurements with the parallel plate chamber with no material between the window and the radiation beam. An additional seven blocks of the same material as the rest of the phantom should be 25 cm × 25 cm. These blocks should be 0.5, 1, 2, 4, 8, 16 and 32 mm thick. These seven blocks combined with the 5 cm thick blocks allow measurement of depth ionisation curves in 0.5 mm increments to any depth from the surface to 40 cm with the parallel plate chamber and from 1 cm to 40 cm with the Farmer chamber. The depth of 40 cm is the limit because 10 cm of backscatter should be maintained downstream from the measurement point.

A plastic phantom for film dosimetry is also required. It is convenient to design one section of the phantom to serve as a film cassette. Other phantom sections can be placed adjacent to the cassette holder to provide full scattering conditions.

Use of ready pack film irradiated parallel to the central axis of the beam requires that the edge of the film be placed at the surface of the phantom and that the excess paper be folded down and secured to the entrance surface of the phantom. Pinholes should be placed in a corner of the downstream edge of the paper package so that air can be squeezed out before placing the ready pack in the phantom. Otherwise air bubbles will be trapped between the film and the paper. Radiation will be transmitted un-attenuated through these air bubbles producing incorrect data.

Plastic phantoms are also commonly used for routine quality control measurements. The design of these phantoms will depend on the requirements of the quality control program.

10.3. ACCEPTANCE TESTS

Acceptance tests assure that the specifications contained in the purchase order are fulfilled and that the environment is free of radiation and electrical hazards to staff and patients. The tests are performed in the presence of a manufacturer's representative. Upon satisfactory completion of the acceptance tests, the physicist signs a document certifying these conditions are met. When the physicist accepts the unit, the final payment is made for the unit, owner-ship of the unit is transferred to the institution, and the warranty period begins. These conditions place a heavy responsibility on the physicist in correct performance of these tests.

Acceptance tests may be divided into three groups:

(1) *safety checks*,
(2) *mechanical checks*, and
(3) *dosimetry measurements*.

A number of national and international protocols exist to guide the physicist in the performance of these tests.

10.3.1. Safety checks

Acceptance tests begin with safety checks to assure a safe environment for staff and public.

Interlocks, warning lights, patient monitoring equipment

The initial safety checks should verify that all interlocks are functioning properly. These interlock checks should include the door interlock, all radiation beam-off interlocks, all motion-disable interlocks, and all emergency-off interlocks.

The door interlock prevents irradiation from occurring when the door to the treatment room is open. The radiation beam-off interlocks halt irradiation but they do not halt the motion of the treatment unit or patient treatment couch. The motion-disable interlocks halt motion of the treatment unit and patient treatment couch but they do not stop machine irradiation. Emergency-off interlocks typically disable power to the motors that drive treatment unit and treatment couch motions and power to some of the radiation producing elements of the treatment unit. The idea is to prevent both collisions between the treatment unit and personnel, patients or other equipment and to halt undesirable irradiation.

The medical physicist must verify the proper function of all these interlocks and assure that all personnel operating the equipment have a clear understanding of each. After verifying that all interlocks and emergency off switches are operational, all warning lights should be checked. Next the proper functioning of the patient monitoring audio-video equipment can be verified. The audio-video equipment is often useful for monitoring equipment or gauges during the acceptance testing and commissioning involving radiation measurements.

Radiation survey

After completion of the interlock checks, the medical physicist should perform a radiation survey in all areas outside the treatment room. For cobalt units and linear accelerators operated below 10 MeV a photon survey is required, for linear accelerators operated above 10 MeV the physicist must survey for neutrons in addition to photons. The survey should be conducted using the highest energy photon beam. To assure meaningful results the physicist should perform a preliminary calibration of the highest energy photon beam before conducting the radiation survey. Photon measurements will require both a Geiger counter and an ionisation chamber survey meter. Neutron measurements will require a neutron survey meter. Several types are available including Bonner spheres, long counters and BF_3 counters.

The fast response of the Geiger counter is advantageous in performing a quick initial survey to locate areas of highest radiation leakage through the walls. After location of these "hot-spots" the ionisation chamber-type survey meter may be used to quantify the leakage values.

- All primary barriers should be surveyed with the largest field size, with the collimator rotated to 45º, and with no phantom in the beam.

- All secondary barriers should be surveyed with the largest field size with a phantom in the beam.

- The first area surveyed should be the control console area where an operator will be located to operate the unit for all subsequent measurements.

Collimator and head leakage

Shielding surrounds the target on a linear accelerator or the source on a cobalt-60 unit. Most regulations require this shielding to limit the leakage radiation to a 0.1% of the useful beam at one meter from the source. The adequacy of this shielding must be verified during acceptance testing.

This verification may be accomplished by closing the collimator jaws and covering the head of the treatment unit with film. The films should be marked to permit the determination of their position on the machine after they are exposed and processed. The exposure should be long enough to yield an optical density of one on the films.

For example, assume an exposure of 10 cGy yields an optical density of one on the film and the films are secured to the head of the treatment unit at a distance of 25 cm from the source. Then the expected radiation level at the position of the films is 1.6% of the useful beam (0.1% of the useful beam at one meter inverse-squared to 25 cm). An exposure of 625 cGy at isocenter (10 cGy divided by 1.6%) should yield an optical density of one on the film.

Any hot spots revealed by the film can be quantified by using an ionisation chamber-style survey meter. The survey meter can be positioned a meter from the hot spot with a ring stand and clamps. The reading may be viewed remotely with the closed circuit television camera to be used for patient monitoring.

10.3.2. Mechanical checks

The mechanical checks establish the precision and accuracy of the mechanical motions of the treatment unit and patient treatment couch.

Collimator axis of rotation

The photon collimator jaws rotate on a circular bearing attached to the gantry. The central axis of the photon, electron, and light fields should be aligned with the axis of rotation of this bearing and the photon collimator jaws should open symmetrically about this axis. This axis is an important aspect of any treatment unit and must be carefully determined.

The collimator rotation axis can be found with a rigid rod attached to the collimator housing. This rod should terminate in a sharp point and be long enough to reach from where it will be attached to the collimator housing to the approximate position of isocenter.

The gantry should be positioned to point the collimator axis vertically downward and then the rod is attached to the collimator housing. Millimeter graph paper is attached to the patient treatment couch and the treatment couch is raised to contact the point of the rod. With the rod rigidly mounted, the collimator is rotated through its range of motion. The point of the rod will trace out an arc as the collimator is rotated. The point of the rod is adjusted to be near the center of this arc. This point should be the collimator axis of rotation. This process is continued until the minimum radius of the arc is obtained. This minimum radius is the precision of the collimator axis of rotation. In most cases this arc will reduce to a point but should not exceed 1 mm in radius in any event.

Photon collimator jaw motion

The photon collimator jaws should open symmetrically about the collimator axis of rotation. A machinist dial indicator can be used to verify this. The indicator is attached to a point on the collimator housing that remains stationary during rotation of the collimator jaws. The feeler of the indicator is brought into contact with one set of jaws and the reading is recorded. The collimator is then rotated through 180º and again the indicator is brought into contact with the jaws and the reading is recorded. The collimator jaw symmetry about the rotation axis is one half of the difference in the two readings. This value projected to the isocenter should be less than 1 mm. This procedure is repeated for the other set of collimator jaws.

The two sets of collimator jaws should be perpendicular to each other. To check this, the gantry is rotated to orient the collimator axis of rotation horizontally. Then the collimator is rotated to place one set of jaws horizontally. A spirit level is placed on the jaws to verify they are horizontal. Then the spirit level is used to verify that the vertically positioned jaws are vertical.

With the jaws in this position the collimator angle indicators are verified. These indicators should be reading a cardinal angle at this point, either 0, 90, 180, or 270º depending on the collimator position. This test is repeated with the spirit level at all cardinal angles by rotating the collimator to verify the collimator angle indicators.

Congruence of light and radiation field

At this point the alignment of the light field is to be verified. With millimeter graph paper attached to the patient treatment couch, the couch is raised to nominal isocenter distance. The gantry is oriented to point the collimator axis of rotation vertically downward. The position of the collimator axis of rotation is indicated on this graph paper. The projected image of the cross-hair should be coincident with the collimator axis of rotation and should not deviate more than 1 mm from this point as the collimator is rotated through its full range of motion. The projected images of the jaws should open and close symmetrically about this point. The symmetry of the collimator jaw images about this point should be better than 1 mm at all cardinal angles of the collimator.

The congruence of the light and radiation field can now be verified. A ready-pack of radiographic film is placed perpendicularly to the collimator axis of rotation. The edges of the light field are marked with radio-opaque objects or by pricking holes with a pin through the ready pack film in the corners of the light field. The film is positioned near z_{max} by placing plastic on top of it and is irradiated to yield an optical density between 1 and 2. The light field edge should correspond to the radiation field edge within 2 mm. This test should be repeated over the range of field sizes and at two different distances to verify the virtual light and photon sources are the same distance from isocenter.

At this point the light field has been aligned to the collimator axis of rotation. Any misalignment between the light and radiation field may indicate that the central axis of the radiation field is not aligned to the collimator axis of rotation. The alignment of the photon field is a complex procedure that should only be performed by factory personnel. Any misalignment must be evaluated for its magnitude, effect on treatment and whether or not factory personnel should be called in to verify and correct the problem.

Gantry axis of rotation

The gantry axis of rotation can be found with any rigid rod that has a telescoping capability. Many treatment machines are supplied with a mechanical front pointer that may be used for this purpose. The axis of the front pointer should be aligned along the collimator axis of rotation and its tip should be at nominal isocenter distance. The gantry is positioned to point the central axis of the beam vertically downward. Then affix a second rigid rod that terminates in a small diameter tip off the end of the patient treatment couch and adjust the treatment couch to bring the rod affixed to the treatment couch to contact the point of the rod fixed to the gantry. Then the treatment table is shifted along its longitudinal axis to move the point of the rod out of contact with the rod affixed to the gantry. Care should be taken not to change the vertical or lateral positions of the rod.

The gantry is rotated 180º and the treatment couch is moved back to a position where the two rods contact. If the front pointer correctly indicates the isocenter distance, the points on the two rods should contact in the same relative position at both angles. If not, the treatment couch height and length of the front pointer are adjusted until this condition is achieved as closely as possible. Because of flexing of the gantry, it may not be possible to achieve the same position at both gantry angles. If so, the treatment couch height is positioned to minimize the overlap at both gantry angles. This overlap is a "zone of uncertainty" of the gantry axis of rotation. This zone of uncertainty should not be more than 1 mm in radius. The tip of the rod affixed to the treatment table now indicates the height of the gantry axis of rotation. This procedure is repeated with the gantry at parallel-opposed horizontal angles to establish the right/left position of the gantry axis of rotation.

The collimator axis of rotation indicated by the cross-hair image, aligned before, should pass through this point. Patient positioning lasers are then aligned to pass through this point.

Patient treatment couch axis of rotation

The patient treatment couch axis of rotation can be found by positioning the gantry with the collimator axis of rotation pointed vertically downward. Millimeter graph paper is attached to the treatment couch and the image of the cross-hair marked on this graph paper. As the treatment couch is rotated, the movement of the cross-hair image on the graph paper is noted. The cross-hair image should trace an arc with a radius of less than 1mm.

The collimator axis of rotation, the gantry axis of rotation, and the treatment couch axis of rotation should all intersect in a sphere. The radius of this sphere determines the isocenter uncertainty. This radius should be no greater than 1 mm and for machines used in radiosurgery should not exceed 0.5 mm.

Radiation isocenter

Radiation isocenter should be determined for all photon energies. To locate radiation isocenter a ready-pack film is taped to one of the plastic blocks that comprise a plastic phantom. The plane of the film should be placed in the plane traced out by the central axis of the x-ray beam as the gantry is rotated. The film should be perpendicular to and approximately centered on the gantry axis of rotation. A pin prick is made in the film to indicate the gantry axis of rotation. Then a second block is placed against the film sand-wiching it between the two blocks and the collimator jaws are closed to approximately 1 mm × 1 mm.

The film is then exposed to produce an optical density of 0.3 to 0.5. Without touching the film, the film is exposed again at a number of different gantry angles in all four quadrants. Gantry angles that are not 180º apart should be chosen to avoid having the entrance of one beam overlap the exit of another. The processed film should show a multi-armed cross, referred to as a "star shot." The point where all central axes intersect is the radiation isocenter. Because of gantry flex, it may be a few millimeters wide but should not exceed 4 mm. This point should be within 1 mm to 2 mm of the mechanical isocenter indicated by the pin-prick on the film.

Verification of the radiation isocenter can be accomplished by centering an ionisation chamber with an appropriate build-up cap on this point. The ionisation collected for a fixed number of monitor units on a linear accelerator or time on a cobalt-60 unit should be independent of gantry angle.

Optical distance indicator

A convenient method to verify the accuracy of the optical distance indicator (ODI) over the range of its read-out is with the plastic phantom discussed in section 10.2.4. With the gantry positioned with the collimator axis of rotation pointing vertically downward five of the 5 cm thick blocks are placed on the treatment couch with the top of the top block at isocenter. The ODI should read isocenter distance. By adding and removing 5 cm blocks the ODI can be easily verified at other distances in 5 cm increments.

Gantry angle indicators

The accuracy of the gantry angle indicators can be determined by placing a spirit level across the rails that hold the blocking tray. At each of the cardinal angles the level should indicate level. Some spirit levels also have an indicator for 45º angles that can be used to check angles of 45°, 135°, 225°, and 315º. The gantry angle indicators should be accurate to within 0.5º..

Collimator field size indicators

The collimator field size indicators can be checked by comparing the indicated field sizes to values measured on a piece of graph paper fixed to the treatment couch with the top of the couch raised to isocenter height. The range of field size should be checked for both symmetric and asymmetric field settings.

Patient treatment couch motions

The patient treatment couch should move in vertical and horizontal planes. The vertical motion can be checked by attaching a piece of millimeter graph paper to the treatment couch and with the gantry positioned with the collimator axis of rotation pointing vertically downward, mark the position of the image of the cross-hair on the paper. As the treatment couch is moved through its vertical range, the cross-hair image should not deviate from this mark. The horizontal motion can be checked in a similar fashion with the gantry positioned with the collimator axis in a horizontal plane. A piece of graph paper is affixed to the treatment couch, the position of the cross-hair is marked and the treatment couch is moved through its range of lateral motion. By rotating the treatment couch 90 degrees from its "neutral" position, the longitudinal motion can be verified with the collimator axis oriented in a horizontal plane.

10.3.3. Dosimetry measurements

Dosimetry measurements establish that the central axis percentage depth doses and off axis characteristics of clinical beams meet the specifications. The characteristics of the monitor ionisation chamber of a linear accelerator or a timer of a cobalt-60 unit are also determined.

Photon energy

The "energy" specification of an x-ray beam is usually stated in terms of the central axis percentage depth dose. Typical specifications are in terms of the value of the 100 cm *SSD* central axis percentage depth dose for a 10×10 cm^2 field at a depth of 10 cm in a water phantom. This value is compared to values given in the British Journal of Radiology, Supplement 25 to determine a nominal energy for the photon beam. During acceptance testing this value will be determined with a small volume ionisation chamber in a water phantom according to the acceptance test protocol.

Photon beam uniformity

The uniformity of a photon beam is typically specified either in terms of transverse beam profiles or the uniformity index. For the case in which transverse beam profiles are used, the flatness and symmetry of the beam are specified over the central 80% of the beam profile at a depth of 10 cm in a water phantom. The beam uniformity should also be specified at z_{max} in a water phantom. Specification at $z_{\max}$ prevents the off axis peaking of the beam profile becoming too large at this depth. The off axis peaking is a product of the design of the flattening filter to produce a flat profile at a depth of 10 cm. The flattening filter also produces a differential hardening across the transverse direction of the beam that results in the off axis peaking at depth shallower than 10 cm. Beam profiles are measured along the principal planes as well as along a diagonal of the beam.

The uniformity index is a measure of the beam uniformity over the entire area of the beam, not just the principal planes. The uniformity index is measured in a plane perpendicular to the central axis. It is defined to be the area enclosed by the 90% isodose curve divided by the area enclosed by the 50% isodose curve.

The International Electrotechnical Commission (IEC) definition of the flattened area of the beam depends on field size. According to the IEC, the flattened area is defined by straight lines joining points on the major axes and diagonals of square fields given in Table 10.I.

TABLE 10.I. IEC'S DEFINITION OF THE FLATTENED AREA OF THE BEAM.

Side of square field a (cm)	d_m	d_d
$5 \le a \le 10$	1 cm	2 cm
$10 < a \le 30$	0.1 a	0.2 a
$30 < a$	3 cm	6 cm

where d_m is the distance from contour of the 50% of the absorbed dose on the beam central axis to the flattened area of the beam. It is on a major axis of the beam. Similarly d_d is defined on a beam diagonal.

Photon penumbra

The photon penumbra is typically defined as the distance between the 80% and 20% dose points on a transverse beam profile measured 10 cm deep in a water phantom. However, there are also other definitions of the penumbra, such as the distance between the 90% and 10% dose points on the beam profile at a given depth in phantom. Whenever penumbra values are quoted the depth of profile and the spread in percentage dose should be stated.

Electron energy

The electron energy is typically determined from measurements of the practical range in a water phantom. The most probable electron energy at the phantom surface $E_{p,0}$ can be determined from the practical range with the following equation:

$$E_{p,o} = 0.0025\, R_p^2 + 1.98\, R_p + 0.22 \quad , \tag{10.1}$$

where R_p is the practical range.

Another energy of interest for calibration purposes is the average energy on the phantom surface. Further discussion of the determination of the average energy is found in Chapters 8 and 9. Although the manufacturer states nominal electron energies, the central axis percentage depth dose characteristics of electron beams are really the values of clinical interest.

Electron beam bremsstrahlung contamination

The radiation measured beyond the practical range of the electrons in the phantom material is the bremsstrahlung contamination of the electron beam. All electron beams have bremsstrahlung contamination that results from interactions between the electrons and materials in the scattering foils, collimators, air, and patients. The bremsstrahlung contamination increases with electron energy, as discussed in Section 1.3.2.

Electron beam uniformity

The beam uniformity of an electron beam is typically specified either in terms of transverse beam profiles or the uniformity index. Beam profiles are measured along the principal planes and along a diagonal of the beam. For the case in which beam profiles are used, the flatness and symmetry of the beam are typically specified over the central 80% of the beam profile at a stated depth in a water phantom. The depth of measurement will depend on machine specifications. If the vendor-supplied specifications are inadequate, the physicist should propose an alternate set. The IEC definition of electron field uniformity includes measuring beam profiles at depths of 1 mm, the depth of the 90% dose, and one half the depth of the 80% dose.

Electron penumbra

The electron penumbra is usually defined as the distance between the 80% and 20% dose points along a major axis at a given depth. The IEC defines this depth as one half of the depth of the 80% dose on the central axis. Machine vendors specify other depths such as depth of z_{max}, depth of the 90%, etc. for definition of electron penumbra.

Monitor characteristics

The amount of radiation delivered by a treatment unit is determined by setting of monitor unit (MU) device on the treatment unit. A timer serves this purpose on a cobalt unit and an ionisation chamber that intercepts the entire treatment beam is used on a linear accelerator. This monitor unit device should be calibrated according to an appropriate national or international protocol for all energies, dose rates and modalities that will be used clinically.

The *linearity* of the monitor unit device should be verified by placing an ionisation chamber at a fixed depth in a phantom and recording the ionisation collected during irradiations with different time or monitor unit settings over the range of the monitor. The collected ionisation can be plotted on the y-axis and the monitor or time setting on the x-axis. These data should produce a straight line indicating a linear response of the monitor unit device or timer.

If these data produce a straight line that does not pass through the origin, then the monitor is linear but has an end effect. A negative x-intercept indicates that more radiation is delivered than indicated by the monitor unit setting. Similarly a positive x-intercept indicates less radiation is delivered than indicated by the monitor unit setting. The end effect should be determined for each energy and modality on the treatment unit. For teletherapy units and orthovoltage x-ray units this effect is referred to as the shutter error.

An alternate means of determining the end effect is the multiple start-stop method. With this technique, an ionisation chamber is placed in the beam and irradiated for a given time or number of monitor units. The irradiation is repeated for the same time or number of monitor units, but with the irradiation interrupted a fixed number of times. If there is no end effect, the collected ionisation should be the same for both irradiations. If the collected ionisation is less for the irradiation that was interrupted, less radiation is delivered than indicated by the monitor setting. The end effect can be calculated from the following relationship:

$$\alpha = \left(\frac{I_n - I_1}{nI_1 - I_n} \right) T, \tag{10.2}$$

where

α is the end effect,
I_n is the ionisation collected after (n-1) interruptions,
I_1 is the ionisation collected after no interruptions, and
T is the total monitor units or timer setting.

Note that a negative end effect determined with the multiple start-stop method corresponds to a positive x-intercept determined from plotting the data for different monitor settings. In both instances less radiation is delivered than indicated by the monitor setting.

Most linear accelerator manufacturers design the monitor chamber to be either sealed so that the monitor chamber calibration is independent of *temperature-pressure fluctuations* or the monitor chamber has a temperature-pressure compensation circuit. The effectiveness of either method should be evaluated by determining the long-term stability of the monitor chamber calibration. This evaluation can be performed during commissioning by measuring the output each morning in a plastic phantom in a set up designed to reduce set up variations and increase precision of the measurement.

Linear accelerators usually provide the capability of irradiating at several different *dose rates*. Different dose rates may change the collection efficiency of the monitor ionisation chamber, which would change the calibration (cGy/MU) of the monitor ionisation chamber. The calibration of the monitor ionisation chamber should be determined at all available dose rates of the treatment unit. The constancy of output with gantry angle should also be verified.

Arc therapy

The verification of the arc or rotational therapy specification is accomplished by setting a number of monitor units on a linear accelerator or time on a cobalt-60 unit and a number of degrees for the desired arc. Then radiation is initiated. Termination of radiation and treatment unit motion should agree with the specification. Typical values are within 1 monitor unit and 3 degrees of the set values. This test should be performed for all energies and modalities of treatment and over the range of arc therapy geometry for which arc therapy will be used.

10.4. COMMISSIONING

Commissioning of an external beam therapy or brachytherapy device includes a series of tasks that generally should consist of the following:

(1) acquiring all radiation beam data required for treatment;
(2) organizing this data into a dosimetry databook;
(3) entering this data into a computerized treatment planning system;
(4) developing all dosimetry, treatment planning, and treatment procedures;
(5) verifying the accuracy of these procedures;
(6) establishing quality control tests and procedures; and
(7) training all personnel.

An abbreviated commissioning will be required following any major repair of the unit.

10.4.1. Photon beam measurements

Central axis percentage depth doses

Typically the first commissioning measurements are the central axis percentage depth doses. To measure these, the surface of the water phantom is placed at the nominal *SSD* or at the isocenter. The vertical depth of the ionisation chamber in the water phantom is determined by measuring from the bottom of the meniscus of the water to the center of the chamber.

Central axis percentage depth dose values should be measured over the range of field sizes from 4×4 cm^2 to 40×40 cm^2. Increments between field sizes should be no greater than 5 cm but are typically 2 cm. Measurements should be made to a depth of 35 cm or 40 cm. Field sizes smaller than 4×4 cm^2 require special attention. Although 0.1 cm^3 chambers typically have diameters of 3-4 mm, the length is on the order of 1.5 cm. Because of lack of lateral electronic equilibrium and penumbral effects for ionisation field sizes smaller than 4×4 cm^2, the dose varies significantly across the length of the chamber. Detectors of small dimensions are required for these measurements and several solutions are possible. A 0.1 cm^3 chamber oriented with its central electrode parallel to the central axis of the beam or a diode may be used in a water phantom.

Alternatively, it may be possible to use a polystyrene phantom with a parallel plate chamber that has a small collecting electrode. These techniques should be validated by first measuring a central axis percentage depth dose of a 10×10 cm^2 field and then comparing these results to the results determined with conventional measurement techniques. By comparing the 10× 10 cm^2 depth dose curves obtained with the two methods one can ascertain the validity of the method and the effective point of measurement of the diode or of the ionisation chamber.

Many photon central axis percentage depth doses reveal a shift in the depth of maximum dose (z_{max}) toward the surface as the field size increases. This shift results from an increasing number of secondary electrons in the beam generated from the increasing surface area of the collimators as well as flattening filter viewed by the detector.

Output factors

The radiation output at $z_{max,}$ in cGy/MU for a linear accelerator and cGy/min for a cobalt unit, increases with an increase in collimator opening or field size. This increase in output can be measured at z_{max} of each field size. Alternatively, the increase in output can be measured at a fixed depth for each field size and the output at z_{max} determined by using the appropriate central axis percentage depth dose values.

Regardless of which measurement technique is used, the increasing output is normalized to the radiation output of the calibration field size, typically a 10×10 cm^2 field. The resulting ratios are referred to as output factors (or relative dose factors or total scatter factors).

Output factors are usually presented graphically as a function of equivalent square fields. This approach assumes the output for rectangular fields is equal to the output of its equivalent square field. This assumption must be verified by measuring the output for a number of rectangular fields at their z_{max}. Outputs for rectangular fields with high and low aspect ratios should be measured. If the outputs of rectangular fields vary from the output of their equivalent square field by more than 2%, it may be necessary to have a table or graph of output factors for each rectangular field.

This matter can be further complicated as linear accelerators may exhibit a dependence on jaw orientation. For example, the output of a rectangular field may depend on whether the upper or lower jaw forms the long side of the field. This effect is sometimes referred to as the collimator exchange effect and should be investigated as part of the commissioning process.

Most modern linear accelerators have collimators that open asymmetrically about the central axis of the x-ray beam. Treatment with asymmetric fields requires knowledge of the output factors of these fields, if this effect is not accounted for in the treatment planning system. The output factors for asymmetric fields can usually be approximated by:

$$\left[OF\left(r\right)\right]_{a,y} = \left[OF\left(r\right)\right]_{s} \; OAR\left(z_{max}, y\right) \quad , \tag{10.3}$$

where $[OF(r)]_{a,y}$ is the output factor for an $r \times r$ field formed with an asymmetric collimator opening. The central ray of this field is y centimeters from the central axis of the symmetric field. $[OF(r)]_s$ is the output factor for an $r \times r$ field formed with symmetric collimator opening and $OAR(z_{max},y)$ is the off axis ratio measured at z_{max} and y centimeters from the central axis of the symmetric field.

The collimator scatter factor is measured “in air” with a build-up cap large enough to provide electronic equilibrium. Typically, these values are normalized to a 10×10 cm^2 field. A problem arises for small high-energy photon field sizes as the size of the build-up cap approaches or exceeds the size of the field. A significant portion of the measured signal represents scatter occurring in build-up cap. This scatter has been estimated to be in the range of 1% to 10% of x-ray energies between 2 and 30 MV.

Using a build-up cap constructed of higher density material such as aluminum or copper may solve the problem. This stratagem reduces the size of the cap permitting measurement of fields down to 4×4 cm^2.

Alternatively the collimator scatter correction factor may be determined by placing the ionisation chamber at an extended *SSD* but with the field defined at the nominal *SSD*. With the chamber at 200 cm the collimator scatter correction factor can be measured for fields with dimensions down to 4×4 cm^2 at 100 cm. These relative measurements should all be performed under the same conditions. In other words, if one chooses to measure with a high-density build-up cap, measurements for all field sizes should be performed with same build-up cap.

As the output factor is the product of the collimator scatter correction factor and the phantom scatter correction factor, the phantom scatter correction factor may be found by dividing the output factor by the collimator scatter correction factor.

Blocking tray factors

Most treatment units have collimators that form rectangular fields. Because treatment volumes are rarely rectangular, high-density shielding blocks are used to protect normal critical structures within the irradiated area. The blocks are either individually designed blocks fabricated of a low melting point alloy, such as Lipowitz's alloy, or standard "library" blocks that may be purchased from the vendor of the treatment machine. In either case these blocks are supported on a plastic tray to correctly position them within the radiation field. This tray attenuates the radiation beam. The amount of beam attenuation provided by the tray must be known to calculate the dose received by the patient. The attenuation for solid trays is easily measured by placing an ionisation chamber on the central axis of the beam at 5 cm depth in phantom in a 10×10 cm^2 field. The ratio of the ionisation chamber signal with the tray in the beam to the signal without the tray is the blocking tray transmission factor.

Although the tray transmission factor should be measured for several depths and field sizes this factor usually has only a weak dependence on these variables and typically one may use one value for all depths and field sizes.

Multileaf collimators (MLCs)

On most current treatment machines multileaf collimators (MLC) are finding widespread application for conventional field shaping as a replacement for shielding blocks. The advantages of an MLC include a reduction in the amount of storage space in the treatment room, elimination of the need for the treatment technologists to lift heavy blocks, and the ability to treat multiple fields without re-entering the treatment room. Disadvantages include the discrete step size of the leaves and additional quality assurance requirements. Additional data are also required to characterize the output factors, central axis percentage depth doses, and penumbra of the MLC fields and the leakage through and between the leaves.

Typically the central axis percentage depth doses of MLC-defined fields are not significantly different from fields defined with the collimator jaws. The penumbra of MLC-defined fields should be measured for both the leaf ends and edges. The penumbra will depend on the leaf design and whether the leaves are singly or doubly focused. Generally, the MLC penumbra is within 2 mm of the penumbra of fields defined with the collimator jaws, with the greatest difference being for singly focused MLC fields not centered on the collimator axis of rotation.

The output factor for fields shaped by MLC systems added downstream from the conventional four jaw collimator system are closely approximated by the product of the collimator scatter factor for the collimator setting and the phantom scatter factor for the irradiated area. This relation is the same as for fields formed with conventional blocks. Some MLC systems replace at least one set of conventional jaws. For these MLC systems the product of the collimator scatter factor and phantom scatter factor for the irradiated area approximates the output factor. Of course, the physicist should verify these relationships for central axis percentage depth doses, penumbra and output factors on each machine.

Leakage through the MLC consists of transmission through the leaves and leakage between the leaves. Leakage between the leaves is easily demonstrated by exposing a film placed perpendicularly to the collimator axis of rotation with the leaves fully closed. Leakage through the leaves can be determined by comparing the umbra region of transverse beam profiles for fields defined by the MLC to fields defined by the collimator jaws. Typical values of MLC leakage through the leaves are in the range of 3% to 5% of the isocenter dose.

Central axis wedge transmission factors

Wedges are used to shape the dose distribution of radiation treatment fields. The central axis wedge transmission factor is the ratio of the dose at a specified depth on the central axis of a specified field size with the wedge in the beam to the dose for the same conditions without the wedge in the beam. Central axis wedge transmission factors determined for one field size at one depth are frequently used to calculate beam-on time or monitor unit settings for all wedged fields and depths. However, the central axis wedge transmission factors may be a function of both depth and field size.

The field size variation may depend not only on the width of the field along the gradient of the wedge but also on the length of the field. In other words, the central axis wedge transmission factor for a given wedge for a 10×10 cm^2 field may differ from the central axis wedge transmission factor for a 10×20 cm^2 field even when the 10 cm is along the wedge gradient in both cases. These dependencies require measuring central axis percentage depth doses with the wedge in the beam for the range of field sizes. The dose with the wedge in the beam can then be related to the calibrated dose rate by measuring the central axis wedge transmission factor at one depth for each field size.

To measure the central axis wedge transmission factor for a given field size at one depth the ionisation chamber should be placed on the central axis of the beam with its axis aligned parallel to a constant thickness of the wedge. Measurements should be performed with the wedge in its original position and with the wedge rotated through 180°. This set of measurements verifies that the wedge and ionisation chamber are correctly positioned. The wedge position may be rotated through 180 degrees by rotating the collimator or by rotating the wedge itself.

Rotation of the wedge itself reveals whether or not the side rails are symmetrically positioned about the collimator axis of rotation. Rotation of the collimator verifies that the ionisation chamber is positioned on the collimator axis of rotation. The measured values should be the same for the two wedge orientations. If the values differ by more than 5% for a 60° wedge or 2% for a 30º wedge, then the wedge or ionisation chamber is not positioned correctly and the situation should be corrected. Otherwise it is usually adequate to take the average value of the two wedge orientations as the correct value.

Dynamic wedge

Linear accelerators are available with an option allowing independent movement of the collimator jaws. This option may be used to create wedged shaped dose distributions by moving one of the independent collimator jaws while the opposite jaw remains stationary during irradiation. This technique is referred to as a dynamic wedge. Clinical implementation of dynamic wedges requires measurement of central axis percentage depth doses, central axis wedge transmission factors, and transverse beam profiles of the dynamic wedges. These measurements are complicated by the modulation of the photon fluence during the delivery of the radiation field.

The central axis percentage depth dose may be measured by integrating the dose at each point during the entire irradiation of the dynamic wedge field. The central axis wedge transmission factors are determined by taking the ratio of the collected ionisation at a specified depth for the dynamic wedge field to the collected ionisation at the same specified depth for the open field with the same collimator and monitor unit settings.

It is important to note that the central axis wedge transmission factors for dynamic wedges may have much larger field size dependence than physical wedges and the field size dependence for dynamic wedges may not be asymptotic. Some manufacturer's implementations of the dynamic wedge technique show a significant change in the trend of the central axis wedge transmission factor as the field width changes between 9.5 cm and 10 cm. This change in the central axis wedge transmission ratio has been demonstrated to approach 20%. This characteristic should be carefully investigated on each machine. Dynamic wedge transverse beam profiles can be measured with a detector array or an integrating dosimeter such as radiochromic film. When a detector array is used, the sensitivity of each detector must be determined.

Transverse beam profiles/off-axis energy changes

The distribution of dose at any point within a radiation beam is required for treatment planning. Transverse beam profiles are measured to characterize the dose at points off the central axis. Frequently off-axis data are normalized to the dose on the central axis at the same depth. These data are referred to as *off-axis ratios* (*OAR*). *OARs* are combined with central axis data to generate isodose curves.

The number of profiles and the depths at which these profiles are measured will depend on the requirements of the treatment planning system. Some systems require these profiles at a few equally spaced depths, others require several profiles at specified depths, and some require only one off-axis profile for the largest field size measured "in-air" with a build-up cap. Transverse beam profiles should be measured in addition to those on which the beam model was determined to verify the accuracy of the treatment planning system algorithms.

Of course, these profiles should be measured for both open and wedged fields. The profiles of the wedged field can then be combined with the central axis percentage depth dose values for wedged fields to generate wedged isodose curves. Any change in wedge factor with depth is then included in the isodose curves.

Entrance dose and interface dosimetry

Knowledge of interface dosimetry, such as the entrance dose between the patient surface and z_{max} is important in a variety of clinical situations. Other areas of interface dosimetry that may be important are interfaces at small air cavities, such as nasopharynx, at the exit surface of the patient, at bone-tissue interfaces, and between a metallic prosthesis and tissue.

These measurements are usually time-consuming because they are not easily automated with a water phantom and scanner. The rapidly changing dose gradient demands measurement with a thin window parallel plate chamber. The requirement for a thin window makes water phantom measurements difficult because of the need to waterproof the chamber and to avoid deformation of the window by hydrostatic pressure.

Measurements are typically carried out in a polystyrene phantom in a constant *SSD* geometry. They begin with the block containing the chamber upstream, backed by two 5 cm blocks of backscattering material with all the buildup sheets placed downstream from these blocks. The first measurement is made with no buildup material upstream from the chamber. The next depth is measured by moving the appropriate sheet of buildup material from the bottom to the top of the phantom. This scheme maintains a constant *SSD* as build-up material is added.

Interface dosimetry measurements should always be performed with both polarities on the entrance window of the ionisation chamber. Large differences in the signal at the interface will be observed when the polarity is reversed. Measurements farther from the interface exhibit smaller differences than measurements nearer the interface. For depths beyond the transition zone readings with either polarity should be the same. The true value Q_T of the measured ionisation at each depth is:

$$Q_T = (Q_+ - Q_-)/2 \quad . \tag{10.4}$$

The positive or negative signs refer to polarity of the signal and the sign of the signal is maintained in this operation. The value computed with Eq. (10.5) is the same as the average of the absolute magnitudes of Q_+ and Q_- unless they have the same sign. This will occur in low signal-to-noise situations where the cable or stem contribute a significant spurious current that does not change sign with a change in polarity while the true signal from the sensitive volume of the chamber does change sign with change in polarity.

Virtual source position

Knowledge of the virtual source position is required for treatment at extended *SSD*. A common technique to determine the virtual source position is to make “in-air” ionisation measurements at several distances from the nominal source position to the chamber. The data are plotted with the distance to the nominal source position on the x-axis and the reciprocal of the square root of the ionisation on the y-axis.

This data should follow a straight line, if not the radiation output does not follow inverse square. If the straight line passes through the origin the virtual and nominal source positions are the same. If the straight line has a positive x-intercept, the virtual source position is downstream from the nominal source position while a negative x-intercept indicates an upstream virtual source position.

For example, consider a machine with a nominal *SAD* of 100 cm. Assume for this machine these measurements demonstrated that the reciprocal of the square root of the measured ionisation followed a straight line but the x intercept was +1 cm. This situation reveals that the inverse square law applies but the virtual source is 99 cm from isocenter. In this case the inverse square calculation should be from 99 cm rather than 100 cm.

This analysis should be performed for the range of field sizes as collimator scatter may change the virtual source position. Of course, if the data do not follow a straight-line, the inverse square law is not applicable and special calibrations will be required at each distance.

Measurement of the beam divergence at various distances from the source is a less commonly used technique to determine virtual source position. This measurement is performed by exposing films oriented perpendicularly to the central axis of the beam at a depth of z_{max}. The full width at half maximum (FWHM) of the beam is determined at each distance. The FWHM at each distance can be plotted and will form a straight line if the inverse square law is valid. The x-intercept indicates the virtual source position. This measurement should be performed for a range of field sizes. One problem is that the range of field sizes and distances where this technique may be used is limited by the size of the film.

10.4.2. Electron beam measurements

Central axis percentage depth dose

Electron central axis percentage depth dose values have been measured with cylindrical and parallel plate ionisation chambers, diodes, and film, however, the ionometric method remains the “gold standard”.

- The effective point of measurement for parallel plate chambers is the inside surface of the entrance window. The effective point of measurement with cylindrical chambers, on the other hand, is shifted from the center of chamber and the shift is one half the inside radius of the cavity toward the source.

- Cylindrical chambers also perturb the electron fluence more than parallel plate chambers. This perturbation is corrected with the replacement correction. This factor is less than one for cylindrical chambers and the value of the factor decreases (further from unity) as the energy of the electron beam decreases and the depth in phantom increases.

- Most thin window parallel plate chambers with a plate separation of 2 mm or less have a replacement correction of unity. However, some of these chambers have a replacement correction different from unity. The replacement factor is dependent on guard ring design, as well as on plate separation distance. Chambers with narrow guard rings tend to have replacement factors further removed from unity than those with wider guard rings.

- Parallel plate chambers can also be difficult to waterproof, if used in a water phantom. Hydrostatic pressure in a water phantom can also deform a thin entrance window of a parallel plate chamber, if a thin waterproof sheath is used. This deformation changes the chamber's sensitivity.

- Use of a parallel plate chamber in phantom can lead to a dosimetric mismatch, if the phantom material differs from the material of the chamber. This mismatch can result in a change in the number of backscattered electrons with the chamber in place from what would occur in a homogeneous phantom. Depending on the materials involved this change may be either an increase or decrease.

- Medical physics societies recommend calibration of low energy electrons with specially designed parallel plate chambers, because the replacement correction factor is much more significant for cylindrical chambers for electrons less than 10 MeV. Higher energy electrons can be measured with cylindrical chambers.

- Water is the phantom material generally recommended for high-energy electrons, because it is nearly tissue-equivalent and it has uniform composition regardless of its origin.

- Plastic phantoms are recommended for low energy electron measurements with a thin window parallel plate chamber that cannot be readily waterproofed to prevent hydrostatic deformation of the window. Plastic phantoms are also useful for film dosimetry measurements.

- Several plastic materials are acceptable for phantoms. However, these plastics are not exactly water-equivalent, i.e., they do not necessarily have the same linear collision and radiative stopping powers and the same linear angular scattering power as water. This lack of exact water-equivalence requires that depths of measurements made in plastic phantoms be corrected to water equivalent depths by scaling. The AAPM TG-25 dosimetry protocol recommends a scaling factor based on the ratios of the depth of the 50% ionisation measured in the two materials, i.e.,

$$z_{\text{water}} = z_{\text{med}} \left(R_{50}^{\text{water}} / R_{50}^{\text{med}} \right) , \tag{10.5}$$

where

z_{water} is the depth in water that is equivalent to the measurement depth z_{med},
R_{50}^{water} is the depth of the 50% ionisation in water, and
R_{50}^{med} is the depth of the 50% ionisation in phantom medium.

- In reference to plastic phantoms, it must be noted that polystyrene is an ambiguous term. Some medical physicists refer to clear polystyrene as polystyrene and to white polystyrene with a 3% loading of $Ti0_2$ as "high impact" polystyrene. Other physicists refer to the white polystyrene as polystyrene and the clear version as "clear" polystyrene. When using tables for depth-scaling factors one should ascertain which polystyrene is listed. Also the density of any plastic should be verified, as it can vary between production batches.

- Additionally, unlike photons, electron percentage depth ionisation curves are not equivalent to percentage depth dose curves. Electron ionisation measurements must be multiplied by the replacement factor and the restricted mass stopping power ratio to determine dose. These factors are energy dependent, and thus depth dependent because the electron beam loses energy as it penetrates the phantom.

- The therapeutic dose is frequently chosen to be the 90% dose at a depth beyond the depth of dose maximum z_{max}. For fields with dimensions similar to or smaller than the range of the electrons, loss of side scatter equilibrium will result in the depth of z_{max} shifting toward the surface and a decrease in the depth of the 90% dose. The range will remain approximately the same as for larger fields. For field sizes larger than the range, the depth of the therapeutic dose remains constant.

- Electron percentage depth dose should be measured in field size increments small enough to permit accurate interpolation to intermediate field sizes. Although skin sparing is much less than for photon beams, skin dose remains an important consideration in many electron treatments. Surface dose is best measured with a thin-window parallel-plate ion chamber. Central axis percentage depth dose should be measured to depths large enough to determine the bremsstrahlung contamination in the beam.

Output factors

The radiation output in cGy per MU is a function of field size and is determined at z_{max} at the standard *SSD*. The output is measured with a small volume ionisation chamber at z_{max} on the central axis of the field. The output depends on the method used to define the field. Three types of collimation are used to define an electron field. These are: (i) secondary collimators (cones) in combination with the x-ray jaws, (ii) irregularly shaped lead or low melting point alloy metal cutouts placed in the secondary collimators, and (iii) skin collimation.

Secondary collimators

Cones, or electron collimators, are available in a limited number of square fields typically 5×5 cm^2 to 25×25 cm^2 in 5 cm increments. Circular and rectangular cones are available but they are not as common as the square cones. The purpose of the cone depends on the manufacturer. Some use cones only to reduce the penumbra, others use the cone to scatter electrons off the side of the cone to improve field flatness. The output for each cone must be determined for all electron energies. These values are frequently referred to as cone ratios rather than output factors.

Metal cutouts

Irregularly shaped electron fields are formed by placing metal cutouts of lead or low melting point alloy in the end of the cone nearest the patient. The penumbra produced by these cutouts is essentially the same as the penumbra produced by the cones themselves. A thickness of 12 mm of a low melting point alloy, such as Lipowitz's metal, is adequate for electrons up to 20 MeV. The output factors for fields defined with these cutouts depend on the electron energy, the cone and the area of the cutout. The dependence of output should be determined for square field inserts down to 4×4 cm^2 for all energies and cones.

As with photons, fields smaller than 4×4 cm^2 require special precautions because the size of the ionisation chamber may approach the size of the field, and smaller detectors are required. A parallel plate chamber with a small collecting electrode may be used in a polystyrene phantom or a diode in a water phantom. In either case the same set up should be used to measure both the small field and the 10×10 cm^2 field.

Since z_{max} shifts toward the surface for electron fields with dimensions smaller than the range of the electrons, it must be determined for each small field size when measuring output factors. The output factor is the ratio of dose at z_{max} for the small field to dose at z_{max} for the 10×10 cm^2 field. For ionometric data this requires converting the ionisation to dose at each z_{max} before determining the output factor, rather than simply taking the ratio of the ionisations. If central axis percentage depth dose data measured with diodes agrees with the central axis percentage depth dose data determined from ionometric data, the diode data can be used directly to determine depth of z_{max}.

Film is an alternate solution. It can be exposed in a polystyrene or water equivalent plastic phantom in a parallel orientation to the central axis of the beam. One film should be exposed to a 10×10 cm^2 field; the other film to the smaller field. The films should be scanned to find the central axis z_{max} for each field. The ratio of dose at z_{max} of the small field to dose at z_{max} of the large field is the output factor. This requires that the dose has been determined from the net optical density with a characteristic curve and that good agreement has been demonstrated between percentage depth dose measured with film to that determined from ionisation chamber data for a 15×15 cm^2 field.

The output factor (or cone ratio) is a function of energy, cone size, and insert size. Typically all values are normalized to an open 10×10 cm^2 cone. For rectangular fields formed by placing inserts in cones the equivalent square can be approximated with a square root method. The validity of this method should be checked on each machine for which the approximation is used.

$$OF\left(E,x,y,f\right) = \left[OF\left(E,x,x,f\right)\times OF\left(E,y,y,f\right)\right]^{1/2} , \tag{10.6}$$

where

f	stands for the *SSD*,
OF(E,x,y,f)	is the output factor for a x-cm × y-cm rectangular field of energy E,
OF(E,x,x,h)	is the output factor for a x-cm × x-cm rectangular field of energy E,
OF(E,y,y,h)	is the output factor for a y-cm × y-cm rectangular field of energy E.

Skin collimation

Skin collimation is used to minimize penumbra for very small electron fields, to protect critical structures near the treatment area, and to restore the penumbra when treatment at extended distance is required. When designing skin collimation the cone insert chosen should be larger than the area to be treated. The skin collimation then collimates this larger field to the treatment area. The skin collimation should also extend a distance beyond the area collimated by the cone insert to protect the patient from scattered electrons.

The thickness required for any electron shielding can be estimated by:

$$\begin{aligned} t_{\mathrm{Pb}}\left(\mathrm{mm}\right) &= 0.5\, E_{\mathrm{p,o}}\left(\mathrm{MeV}\right) + 1 \qquad \text{for lead and} \\ t_{\mathrm{LM}}\left(\mathrm{mm}\right) &= 1.2\, t_{\mathrm{Pb}}\left(\mathrm{mm}\right) \qquad \text{for Lipowitz's metal} \end{aligned} \quad , \tag{10.7}$$

where

$E_{\mathrm{p,o}}$ *(MeV)* is the most probable electron energy at the surface of the patient,
t_{Pb}*(mm)* is thickness of lead,
t_{LM}*(mm)* is thickness of Lipowitz's metal.

Some clinical situations may require minimizing the weight of the skin collimation on the patient resulting in somewhat thinner masks. In these situations it is recommended that the degree of shielding be assessed. This assessment can be performed with a thin-window parallel-plate chamber in a polystyrene phantom at a depth of 1 mm.

As for any small field, skin collimation may affect the percent depth dose as well as the penumbra, if the dimensions of the treatment field are smaller than the electron range. The field size dependence of the percent depth dose is principally a result of scattering in the patient.

The percentage depth dose for a field defined by skin collimation can be approximated as the percentage depth dose of the field determined by secondary collimation, such as a cone. The field size dependence of the output results from electron scattering from the x-ray jaws and in air. In most cases for cones that are 5 cm or more from the skin, the output for a field defined with skin collimation is the same as the output defined by the secondary collimator for that treatment. However, if the skin collimation defines a field so small that the percent depth dose changes then the output may be affected and a measurement may be required.

Transverse beam profiles

As for photon beams, transverse electron beam profiles are measured to determine the off axis dose distribution of electron beams. This information is combined with the central axis percentage depth dose to yield the isodose distribution. The number of transverse profiles and depths at which they must be measured depend on the requirements set by the treatment planning computer.

These profiles are measured in a water phantom with a small volume ionisation chamber. The surface of the phantom is placed at 100 cm or the nominal SSD and the ionisation chamber is scanned perpendicularly to the central axis.

An alternate film dosimetry technique is to measure isodose curves rather than beam profiles. The film is exposed parallel to the central axis of the beam. Optical isodensity is converted to isodose. However, the percent depth dose determined with film is typically 1 mm shallower than ionometric determination for depths greater than 10 mm, and for depths shallower than 10 mm the differences may be as great as 5 mm. Isodose curves may also be measured with small volume ionisation chambers or diodes.

Virtual source position

Frequently electron fields must be treated at extended distances because the surface of the patient prevents positioning the electron applicator at the normal treatment distance. A common example of this occurs during the treatment of posterior neck fields for head and neck carcinoma. The shoulder typically interferes with positioning the electron applicator at the normal treatment distance to the neck. Additional scattering in the extended air path increases the penumbral width and decreases the output (cGy/MU). Knowledge of the virtual electron source is required to predict these changes. The virtual source position is the point from which the electrons appear to emanate.

Determination of the virtual source position is similar to the verification of inverse square law for photons. Treatment planning computers use the virtual source position to calculate the divergence of the electron beams at extended *SSD*s. Correction of the output at an extended *SSD* requires an air gap correction factor in addition to the inverse square factor. The air gap factor corrects the deviations from inverse square resulting from the collimator and air scatter of the electrons. The air gap correction factor may be either greater or less than unity as the output may either increase or decrease at extended distances depending on collimator design, electron energy, field size and air gap. However, for *SSD*s up to 110 cm and energies up to 25 MeV this correction is typically less than 2%.

There can be significant changes in the percent depth dose at extended *SSD* if the electron cone scatters electrons to improve the field flatness. For these machines it may be necessary to measure isodose curves over a range of *SSD*s.

Treatment at an extended *SSD* will also increase the penumbra width. At lower energies the width of the penumbra (80%-20%) increases approximately proportionally with air gap. As electron energy increases the increase in the penumbra width is less dramatic at depth than for lower energies but at the surface the increase in penumbra remains approximately proportional to the air gap. Because a large number of clinical situations demand treatment at an extended *SSD* the reader is advised to measure a sample of isodose curves at extended *SSD*s to evaluate the algorithms in the treatment planning system in use. The penumbra can be restored when treating at extended distances by use of skin collimation as discussed previously.

10.5. TIME REQUIRED FOR COMMISSIONING

Acceptance testing and commissioning of megavoltage treatment units has been discussed in this chapter. The completion of all the tasks associated with placing a treatment unit into clinical service can be estimated to require from 1.5 weeks to 3 weeks per energy following completion of the acceptance tests. The time will depend on machine reliability, amount of data measurement, sophistication of treatments planned and experience of the physicist. Highly specialized techniques, such as, stereotactic radiosurgery, intraoperative treatment, intensity modulated radiotherapy, total skin electron treatment, etc. have not been discussed and are not included in these time estimates.

Commonly used methods to estimate data that has not been measured from measured data have been discussed. The accuracy of all these techniques must be verified on each machine as variations exist between machines.

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