MONITOR UNIT CALCULATIONS FOR ELECTRON BEAMS

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June 9, 1999

1.0 Introduction

Determination of monitor units requires a clear understanding of dose prescription and the factors affecting dose output. This process is unique for electron beams. The purpose of this paper is to identify these factors and to demonstrate how they affect the monitor unit calculations at The University of Texas M. D. Anderson Cancer Center Radiation Oncology Clinic.

Dose is normally prescribed to be the given dose or to some percentage, e.g. 90%, of it. Given dose is the maximum central-axis dose delivered in a water phantom at the same source-to-surface distance (SSD) and with the same field shape as that for the patient. If the treatment field is irregularly shaped, then a rectangular field representative of the irregular field is used. Dose output is the dose per monitor unit to the point at which the given dose is defined. The relationship between prescribed dose (D), the percent of given dose to which dose is prescribed ($\%D$), dose output (O), and the monitor units (MU) required to deliver this prescription is

$$MU = \frac{D}{\%D \cdot O} \ . \tag{1}$$

Dose output depends on the beam parameters, which include energy, applicator, field size, SSD, and skin collimation. Dose output does not depend on patient heterogeneity as the dose prescription point is relative to dose at a point in a water phantom. This is different from the case for photon beams. For electrons, the impact of patient heterogeneities on the dose distribution is to modify the coverage of the isodose surface defining the treatment volume and to create dose heterogeneity inside the treatment volume (Hogstrom, 1983). At M. D. Anderson the 90% isodose surface is usually selected to define the treatment volume.

This schema is consistent with the recommendations of AAPM Task Group 25 (Khan et al., 1991) and is directly applicable to single or abutted fixed fields. For some clinical applications determination of monitor units requires variations on this schema or a different schema. If the field shape is highly irregular, then other means are employed. Also, variations of the above schema are required for overlapping field techniques (e.g. electron arc and total skin electron irradiation). Extensions of this methodology are required to determine monitor units from dose distributions developed on a treatment planning system.

The material to follow will explain the methods used at M. D. Anderson to determine monitor units for fixed electron fields. This will include both calculations when dose is prescribed based on a water phantom and calculations when dose is based on output provided by a treatment planning system. Determination of monitor units for special treatment techniques and usage of other methods will be provided by others.

2.0 Dose Output Conventions at M. D. Anderson

This section specifies the conventions for dose output presently in use in the Radiation Oncology Clinic at M. D. Anderson. We will first discuss the field size dependence of output at the standard $SSD\,(SSD_o = 100\,cm)$. Then we will discuss its variation with SSD. Lastly the influence of bolus and skin collimation is discussed.

2.1 Fixed Applicators and Cones

Dose output (cGy per MU) is the maximum dose per monitor unit on the central axis of the field. It is a function of beam energy (E), applicator or cone (CS), insert field size (FS), and source-to-surface distance (SSD). At SSD_o, the output is related to the calibration output by

$$O(E,CS,FS) = OF(E,CS,FS) \cdot O_{cal}(E, CS = 10 \times 10, FS = 10 \times 10), \qquad (2)$$

where OF is the output factor, i.e. the output relative to the calibration output (O_{cal}). The calibration output is the output at SSD_o for the open 10×10 cm^2 applicator. Also, the depth of maximum dose (R_{100}) depends on energy and field size, i.e.

$$R_{100} = R_{100}(E,FS). \qquad (3)$$

2.1.1 Dependence on Field Size (SSD = 100 cm)

At the time of beam commissioning, the output factor is measured as a function of square field size for each energy and applicator combination at SSD_o, i.e. $OF = OF(E,CS,FS)$. If the applicator accepts inserts, then output is typically measured for square field sizes from 2×2 cm^2 up to the size of the open applicator. Table 1 illustrates this data for the 10×10 cm^2, 15×15 cm^2, 20×20 cm^2, and 25×25 cm^2 applicators at 9 MeV for a Varian Clinac 2100C.

In the case of circular cones on the Siemens ME electron intraoperative machine, collimating inserts are not an option. Table 2 illustrates the dose output at SSD_o for all energy-cone combinations for the straight cones on the Siemens ME electron intraoperative machine. For this machine, the reference field is the open 12-cm diameter cone.

Table 1. Output factor versus field size for various applicators for the 9-MeV beam on M. D. Anderson's Varian Clinac 2100C (SSD =100 cm).

Field Size (cm^2)	Applicator Size (cm^2)			
	10 × 10	**15 × 15**	**20 × 20**	**25 × 25**
2 × 2	0.843	0.846	0.824	0.812
3 × 3	0.908	0.908	0.891	0.871
4 × 4	0.963	0.954	0.936	0.911
5 × 5	0.991	0.982	0.962	0.936
6 × 6	1.003	0.997	0.976	0.951
7 × 7	1.005	1.004	0.983	0.960
8 × 8	1.003	1.006	0.986	0.964
10 × 10	1.000	1.003	0.985	0.965
12 × 12		0.997	0.980	0.962
15 × 15		0.992	0.976	0.955
20 × 20			0.981	0.950
25 × 25				0.955

Table 2. Output factor of straight cone versus cone diameter for all energies for the Siemens Mevatron ME electron intraoperative machine (SSD =100 cm), from Nyerick et al. (1991).

Cone Diameter (cm)	Energy (MeV)				
	6	**9**	**12**	**15**	**16**
5	0.900	0.960	0.998	1.010	1.030
6	0.951	0.990	1.021	1.022	1.023
7	0.980	1.001	1.027	1.020	1.021
8	1.005	1.011	1.035	1.029	1.026
9	1.000	1.000	1.021	1.010	1.011
10	0.997	0.995	1.007	1.002	1.000
11	0.992	0.995	1.002	0.995	1.003
12	1.000	1.000	1.000	1.000	1.000

2.1.2 Rectangular Field Inserts

For rectangular fields, the square root method of Mills et al. (1982) is utilized,

$$OF(E,CS,FS = L \times W) = \left[OF(E,CS,FS = L \times L) \cdot OF(E,CS,FS = W \times W)\right]^{1/2}. \quad (4)$$

Shiu et al. (1994) found this method to be more accurate than the equivalent square method $S_{eq} = 2LW/(L+W)$ reported by Meyer et al. (1984).

2.1.3 Irregular Field Inserts

The dose output for irregularly shaped fields can be determined in a number of ways. Many treatment centers measure the dose output for each irregularly shaped electron field, as their use of electron fields may be infrequent. If such a system is used, then the medical physicist usually files those results, slowly developing a library of different measurement conditions. These data can later be used for determining output of new, similarly-shaped treatment fields.

Another method is to calculate the dose output using an analytical algorithm that models electron scatter or using a Monte Carlo code. This will be the subject of a separate paper.

In our clinic, we typically approximate the irregular field by a rectangle and then utilize the dose output of the rectangle. It is important to follow some fundamental principles when determining an equivalent rectangle:

1. The equivalent rectangle for the dose output is determined for the field shape defined by the applicator insert, not by the skin collimator.

2. The maximum dose output usually occurs in the broadest region of the field, i.e. at the point surrounded by the greatest diameter circle that is enclosed by the field.

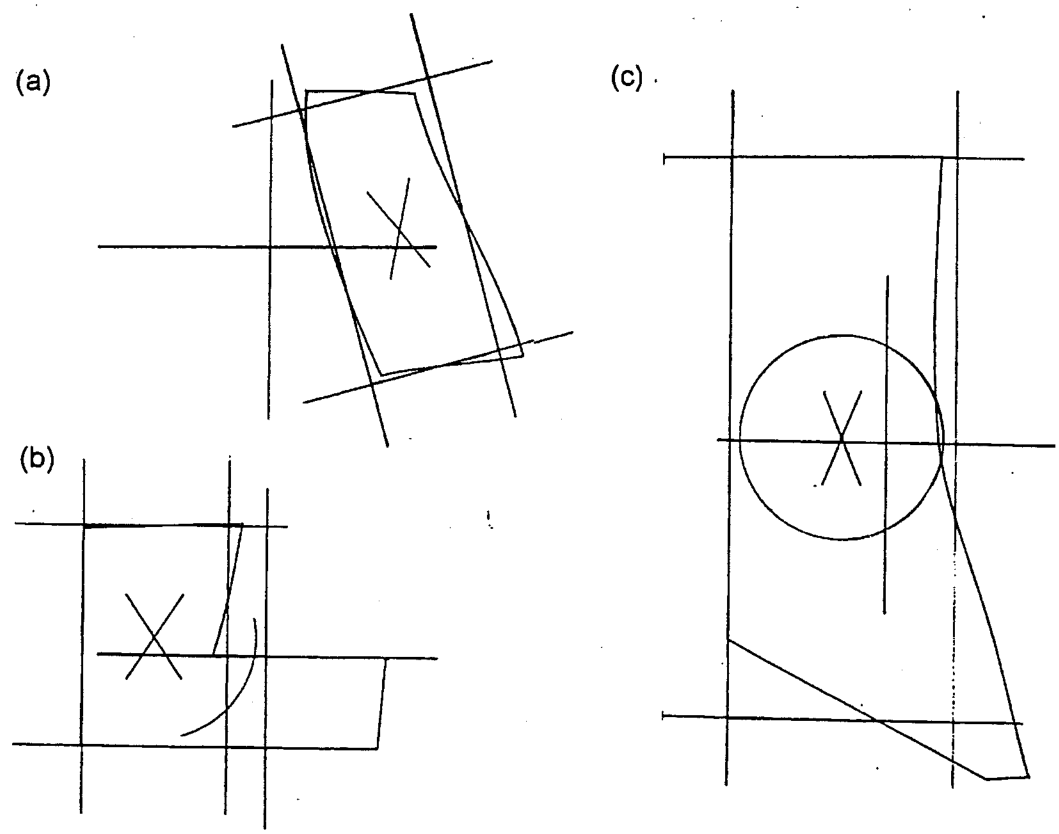
3. The dose output usually varies little beyond some minimum square field size. Hence, areas of the field located greater that one-half of that distance can be assumed to contribute insignificantly to the dose output. For example, the data in Table 1 shows little variation for field sizes greater than 6 x 6 cm^2. Hence, areas of the field greater than 3 cm from the estimated location for which dose output is being estimated are insignificant.

4. The rectangle should be constructed to minimize the difference in area between it and the irregular field less any regions ignored. Rotation about the beam central axis should be used.

5. The method may not be sufficiently accurate to be used for highly irregular fields. In such cases, measurement or other means of calculation are recommended.

Shown in Figure 1 are a few selected clinical examples that illustrate how the rectangular field is estimated.

Figure 1. Examples of constructing rectangular fields whose output approximates that of the irregular field for (a) a posterior cervical strip, (b) a posterior cervical strip plus submandibular nodes, and (c) an internal mammary chain. The irregular field is delineated by the irregularly shaped curve. The beam axes are delineated by orthogonal 10-cm line segments. The constructed rectangle is delineated by the four intersecting lines, the center delineated by "x." The arcs are a distance of 3 cm from the center of the rectangle.



2.1.4 <u>Off-Axis Fields</u>

Presently, we make no correction for rectangular fields that are centered at a point away from the central axis. Errors resulting from this omission are less than 3%, as our requirement for beam flatness, i.e. being within 3% of the central axis value, is met by all of our treatment accelerators. However, dose output could account for any beam non-uniformity by including a multiplicative factor, the off-axis ratio at position X,Y and depth R_{100} for the open applicator,

$$OAR(E,CS,SSD_o,X,Y) = OAR_{meas}\ (E,CS,SSD_o,X=\sqrt{X^2+Y^2},Y=0)\,. \qquad (5)$$

Note that a radial dependence of the OAR is recommended as the dual scattering foil system is radial symmetric. A major-axis scan (e.g. Y=0) measured at a depth near R_{100} can be used for OAR values for each energy-applicator combination.

2.2 Variable Collimators

Some treatment machines have variable collimators, and these collimators are either mechanically or electrically coupled to the x-ray jaws so that the two move synchronously. This movement causes significantly greater variation in dose output with field size than with fixed field applicators. Mills et al. (1985) showed that the square root method was insufficiently accurate as the x-ray jaws and electron trimmers in the X-Z plane were located at a distance from isocenter different from those in the Y-Z plane. This results in differences in electron scatter for equal field dimensions; hence, output for rectangular fields depends on the orientation of the collimator, e.g. $OF(10\times 4)\neq OF(4\times 10)$. To resolve this issue, a method referred to as the 1-D method, was implemented:

$$OF(E,X,Y)=\left[OF(E,X,10)\cdot OF(E,10,Y)\right]+CF(E,X,Y)\ , \tag{6}$$

where

$$CF(E,X,Y)=0 \qquad \Delta\leq 0 \tag{7a}$$

$$=C(E)\cdot\Delta \qquad \Delta>0, \tag{7b}$$

where

$$\Delta(X,Y)=\frac{(X-10)\cdot(Y-10)}{\left|(X-10)\cdot(Y-10)\right|^{1/2}}\ . \tag{7c}$$

It had been our experience on an AECL Therac 20 that CF equaled zero at energies of 17 and 20 MeV, but required energy corrections at 6, 9, and 13 MeV (Mills et al. 1985). Contrary to this, correction factors for the variable applicator on the Siemens Mevatron 6740D (Hogstrom et al., 1985), which has a maximum electron energy of 11 MeV, were not required. Hence, for this machine,

$$O(A\times B)=\left(0.960\ cGy/MU\right)\cdot OF(A\times 10)\cdot OF(10\times B), \tag{8}$$

where $OF(A\times 10)$ is the output factor for the $A\times 10\,\mathrm{cm}$ field, $OF(10\times B)$ is the output for the $10\,\mathrm{cm}\times B$ field, and 0.960 cGy/MU is the output for the reference field size (10×10 cm^2). Table 3 shows the output factors for the 8-MeV beam on the Mevatron 6740D.

2.3 Dependence on SSD

The output at extended SSD decreases due to both the inverse square effect and the loss of side scatter equilibrium (Hogstrom, 1991). Therefore, we use the convention originally described by Meyer et al. (1984) and found in AAPM Task Group 25 Report (Khan et al., 1991),

$$O(SSD) = O(SSD_o) \cdot \left[\frac{SSD_o + R_{100}(E,FS)}{SSD + R_{100}(E,FS)}\right]^2 \cdot f_{air}(E,FS,SSD). \qquad (9)$$

Table 3. 1-D output factors for the variable electron collimator versus field size at 8 MeV on the Siemens Mevatron 6740D (SSD = 100 cm).

Field Size A × 10.0 (cm²)		**Field Size 10.0 × B (cm²)**	
4.0 × 10.0	0.925	10.0 × 4.0	0.948
5.0 × 10.0	0.959	10.0 × 5.0	0.974
6.0 × 10.0	0.979	10.0 × 6.0	0.987
8.0 × 10.0	0.997	10.0 × 8.0	0.997
10.0 × 10.0	1.000	10.0 × 10.0	1.000
12.0 × 10.0	1.003	10.0 × 12.0	1.001
15.0 × 10.0	1.005	10.0 × 15.0	1.002
20.0 × 10.0	1.012	10.0 × 20.0	1.007
25.0 × 10.0	1.018	10.0 × 25.0	1.015
29.4 × 10.0	1.017		

The air gap factor (f_{air}) is determined by solving the above equation using square field output data measured at both extended SSD and standard SSD_o. We assume that air gap factors are applicator independent, and we measure them for the smallest applicator that is greater than the insert field size.

It should be noted that nominal, not measured virtual, SSD values are used. For example, on a Varian Clinac 2100C we measured the virtual source to be approximately 92 cm above isocenter. Nonetheless, for output determination we utilize the nominal source position of 100 cm above isocenter, as this is the value used by the radiation therapists. We could carry an offset between the actual and nominal source position, as all of our monitor unit calculations are done by computer. However, this is not necessary as small source position differences make little difference in the inverse square factor of equation (9), and what little difference exists is absorbed into the air gap factor. This guarantees accurate values of output at extended SSD. Table 4 lists air gap factors as a function of field size and SSD for the 9-MeV beam on the Varian Clinac 2100C. It should be clarified that the field size in the tables is the field size at the extended SSD demagnified to the standard SSD (SSD_o). The values in the table span the range from 100-cm to 120-cm SSD, and linear interpolation is utilized. For SSD less than 100-cm, f_{air} is assumed to equal 1.0; SSD greater than 120-cm SSD are highly unlikely, but if used, require additional measurements.

The air gap factor is less than unity, becoming increasingly smaller for smaller field size and larger SSD. Once f_{air} < 0.95, this indicates the beam is essentially all penumbra. In such cases our monitor unit program requests the user (medical dosimetrist, radiation therapist, or radiation oncology resident) to consult with the medical physicist who ensures this beam fulfills the intent of the treatment plan.

Table 4. Air gap factor versus field size (defined at SSD = 100 cm) and SSD for the 9-MeV beam on the Varian Clinac 2100C. The shaded area indicates values for f_{air} < 0.95.

Field Size (cm^2)	*SSD* (cm)				
	100	105	110	115	120
2×2	1.000	0.882	0.743	0.611	0.505
3×3	1.000	0.946	0.878	0.792	0.715
4×4	1.000	0.942	0.902	0.862	0.815
6×6	1.000	0.978	0.954	0.940	0.915
10×10	1.000	0.984	0.972	0.961	0.955
15×15	1.000	0.988	0.978	0.970	0.963
20×20	1.000	0.987	0.975	0.971	0.963
25×25	1.000	0.987	0.982	0.974	0.970

For rectangular fields the square root method of Hogstrom et al. (1981) also holds, so that

$$f_{air}(E, FS = L \times W, SSD) = \left[f_{air}(E, FS = L \times L, SSD) \cdot f_{air}(E, FS = W \times W, SSD)\right]^{1/2}. \quad (10)$$

The accuracy of this method has been demonstrated by Shiu et al., (1994) for the Varian Clinac 2100C.

2.4 Other Clinical Considerations Regarding SSD

In some cases the patient's surface is irregularly shaped, and the radiation oncologist will prescribe an off-axis SSD point. In such cases, we utilize the SSD value at the prescribed off-axis point in our calculation, not that of central axis. We do not presently make corrections for the OAR (equation 5). If desired, one could measure OAR at extended SSD or possibly project the off-axis point back to SSD_o and use equation 5.

Another clinical issue is the SSD in the presence of slab bolus (e.g. Superflab™). If bolus is on the skin surface, then the SSD to the bolus is used. For example, if 1-cm of

Superflab™ is placed on a chest wall located at 100-cm SSD, the SSD for output calculation will be 99 cm (100 cm – 1 cm).

2.5 Skin Collimator

Another consideration for electron therapy is how a skin collimator impacts the output factor. Skin collimators primarily impact the depth dose, as its field-size dependence is due to electron scatter in the patient or water phantom. On the other hand, the field-size dependence of dose output is primarily due to electron scatter in air and from collimator edges above the patient or water phantom. Therefore, to estimate dose output in the presence of a skin collimator, we use the following approximation:

$$O_{skin\_coll}(FS, FS_{skin\_coll}) = O(FS) \cdot \%DD(R_{100}(FS_{skin\_coll}), FS) \ , \qquad (11)$$

where $O(FS)$ is the dose output in the absence of the skin collimator and $\%DD(R_{100}(FS_{skin\_coll}), FS)$ is the percent dose for the field size in the absence of the skin collimator at the depth of R_{100} for the field size defined by the skin collimator. If there is no difference in R_{100} for the two field sizes, then the impact of a skin collimator on the dose output can be neglected. This is frequently the case.

2.6 Special Collimators

In some circumstances there are special collimators used in the treatment. For example, we utilize a custom set of brass collimators to boost the skin over larynx. Because these are such small field sizes, we have measured and recorded their output for the full range of clinical geometries (versus field size and SSD) for each beam energy.

3.0 Determination of Monitor Units from Treatment Planning Systems

We use a variety of treatment planning systems to assist in developing patient treatment plans. For these systems to be most useful, it is necessary to have a relationship between the dose distribution, dose prescription, and monitor units.

The bulk of our electron beam treatment planning is performed on a General Electric Target 2 treatment planning system. The system utilizes the Hogstrom pencil beam algorithm (Hogstrom et al., 1981 and Hogstrom et al., 1983) to calculate dose. The relationship between monitor units (MU) beam weight (W), and dose (D) prescribed to an isodose contour of value $\%D$ is given by (Hogstrom and Steadham 1996):

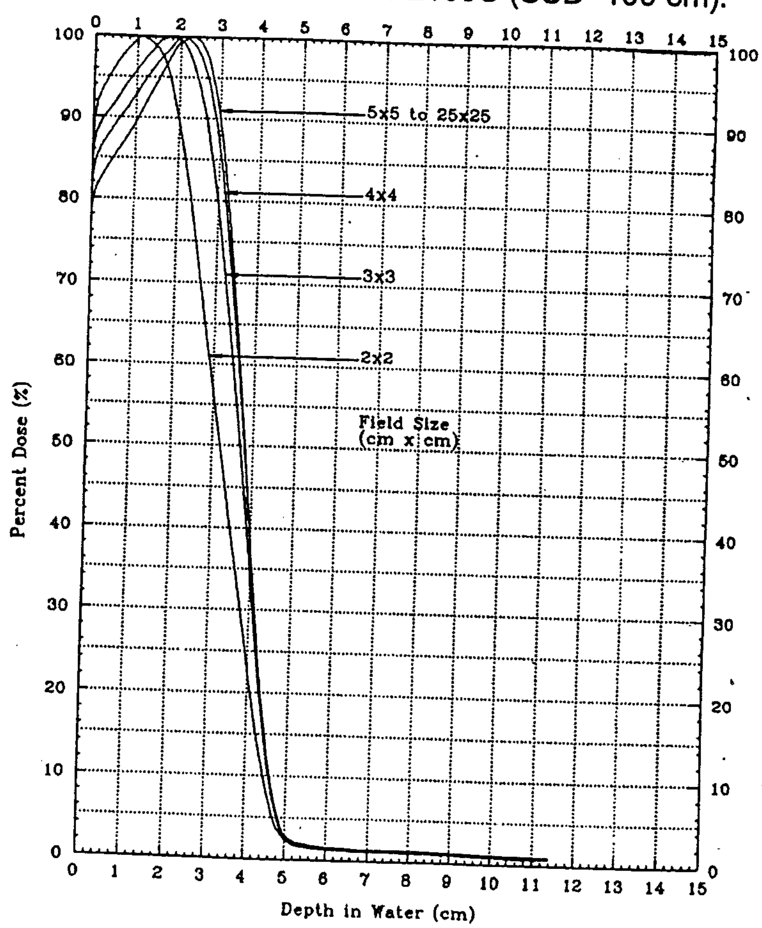
$$MU = \frac{W \cdot D}{\%D \cdot O(E, CS, FS) \cdot (N/100)} \ , \qquad (12)$$

where N is the value of relative dose to which 100 is normalized. The beam weight (W) is the percent dose relative to the central-axis dose maximum at the standard SSD (SSD_o equals 100 cm), and the output is at SSD_o. This methodology is usually sufficiently accurate, but errors greater than 2% are possible for small field sizes at extended SSDs, particularly for the low energy beams (Hogstrom et al. 1997). This problem would not occur if the beams were weighted at the SSD of treatment. Procedures to get around this limitation are beyond the scope of this document.

4.0 Example Calculations

In the radiation oncology clinic at M. D. Anderson, we utilize an in-house computer code to perform monitor unit calculations. The output lists the dose prescription beam parameters, individual factors used to determine output, output in cGy per monitor unit, and the monitor units. Given below are four problems that illustrate the procedures used in our clinic. The field sizes are specified on the patient's surface, i.e. at the treatment SSD. All problems utilize the 9-MeV beam on the Varian Clinac 2100C for which tables of required data, except for depth-dose data, have been presented earlier. 9-MeV depth-dose curves are provided in Figure 2.

Figure 2. Percent dose versus depth in water for various square field sizes of the 9-MeV beam on the Varian Clinac 2100C (SSD=100 cm).



Problem #1: Standard SSD, Rectangular Field
Dose Prescription is 200 cGy to the 90% dose

Energy	9 MeV
Applicator	20×20 cm^2
Field Size	6×16 cm^2
SSD	100 cm

Monitor Units = ___________

Problem #2: Extended SSD, Rectangular Field
Dose Prescription is 180 cGy to the 100% dose

Energy	9 MeV
Applicator	15×15 cm^2
Field Size	5×12 cm^2
SSD	110 cm

Monitor Units = ___________

Problem #3: Standard SSD, Rectangular Field, Surface Bolus
Dose Prescription is 180 cGy to the 90% dose

Energy	9 MeV	
Bolus Thickness	1 cm	
Applicator	20×20 cm^2	
Field Size	8×18 cm^2	(defined on bolus surface)
SSD	100 cm	(to patient's skin surface)

Monitor Units = ___________

Problem #4: Standard SSD, Rectangular Field, Skin Collimator
Dose Prescription is 400 cGy to the 90% dose

Energy	9 MeV
Applicator	10×10 cm^2
Field Size	6×6 cm^2
Skin Collimator	3×3 cm^2
SS	100 cm

Monitor Units = ___________

The computer output used to solve these problems is shown in Figure 3.

Figure 3. Computer output for solutions to four example problems provided above.

```
                         The University of Texas
                       M. D. Anderson Cancer Center
                      Division of Radiation Oncology
                Electron Beam Data Calculation - Version 2.1

                        Therapy Machine: 2100C #2
                           Patient Name: Test
                           Hospital No.: 000001
                              Physician: Radonc
```

Field Description	01 Standard SSD	02 Extended SSD	03 Bolus	04 Skin Coll.
ENERGY	9 MeV	9 MeV	9 MeV	9 MeV
SSD	100.0 cm	110.0 cm	99.0 cm	100.0 cm
Cone Desc	20 x 20 cone	15 x 15 cone	20 x 20 cone	10 x 10 cone
Field_size A	6.0 cm++	5.0 cm++	8.0 cm++	6.0 cm++
Field_size B	16.0 cm++	12.0 cm++	18.0 cm++	6.0 cm++
Skin Coll A	---	---	---	3.0 cm++
Skin Coll B	---	---	---	3.0 cm++
Rx Dose	200 cGy	180 cGy	180 cGy	400 cGy
Rx Isodose	90 %	100 %	90 %	90 %
Depth dmax	2.3 cm	2.3 cm	2.3 cm	1.8 cm
Depth Rx Iso	3.0 cm	2.3 cm	3.0 cm	2.7 cm
Depth 90%	3.0 cm	3.0 cm	3.0 cm	2.7 cm
Depth 10%	4.6 cm	4.6 cm	4.6 cm	4.6 cm
Surface Dose	172.2 cGy	139.5 cGy	155.0 cGy	369.3 cGy
Photon Dose	2.7 cGy	2.2 cGy	2.4 cGy	5.8 cGy
Given Dose	222.2 cGy	180.0 cGy	200.0 cGy	444.4 cGy
Calib D.R.	1.000 cGy/MU	1.000 cGy/MU	1.000 cGy/MU	1.000 cGy/MU
Output Factor	0.976	0.985	0.983	1.003
Inv Sqr Factor	1.000	0.830	1.020	1.000
Air Gap Factor	1.000	0.944 *	1.000	1.000
Skin Coll Fact	---	---	---	0.977
Given D.R.	0.976 cGy/MU	0.772 cGy/MU	1.002 cGy/MU	0.980 cGy/MU
Backup Times	250mu2/0.68min	257mu2/0.70min	220mu2/0.60min	479mu2/1.36min
Machine Set	228 MU	233 MU	200 MU	454 MU

```
*WARNING!  The air gap factor for this field is less than 0.95 .
           Make sure that physics has been consulted regarding this
           calculation.
                                   Physics check by ____________________
++ Field sizes are at surface of patient.
These calculations are NOT based on output from the treatment planning computer.

     Calculated by:      Dosimetrist                 Date:     10-MAR-99
     Technologist verification: ______________

        Checked by: ___________________________      Date: ____________
```

5.0 Future Technology

What future technology would be of clinical benefit? First, there is promise that Monte Carlo calculations will be capable of providing beam dosimetric data required for calculating dose plans and for determining monitor units. Preliminary work (Rogers et al., 1995, Ma et al., 1997) has demonstrated how the BEAM code can be of use. The attractions of the Monte Carlo calculations are (1) their ability to calculate a more complete set of dosimetric data than that which we normally measure, hence reducing inaccuracies due to interpolation, (2) their ability to calculate dose output for patient specific irregular fields (Kapur et al., 1998), and (3) their requiring little access to the treatment machine, only to perform measurements for validating Monte Carlo results.

Second, manufacturers need to integrate the process for determination of monitor units into their treatment planning computer. For electron beams it is recommended that the beam weight be specified relative to the given dose in water at the treatment SSD.

6.0 Summary

This paper has provided the methods currently used in the M. D. Anderson Radiation Oncology Clinic for determination of monitor units for electron beams. The system is consistent with AAPM Task Group 25 Report, with the exception that we use the square root method rather than the equivalent square method for calculation of output for rectangular inserts in fixed beam applicators. The methods presented explicitly show how we accommodate a skin collimator, bolus, and irregular field into our schema. Example problems have been provided to help the reader comprehend our schema. Finally, future technologies that we believe would be of considerable benefit to the determination of monitor units have been presented.

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Warning:
M. D. Anderson's radiotherapy accelerators have been configured to meet its custom specifications. The data contained in this report is presented for example only. Under no circumstances should this data be used for patient dose calculations.