250/100The main advantages of MOSFET devices are:
1. The MOSFET dosimeter is direct reading with a very thin active area (less than 2 m).
2. The physical size of the MOSFET when packaged is less than 4 mm2.
3. The post radiation signal is permanently stored and is dose rate independent.2

The MOSFET Dosimeter
The basic MOSFET structure is depicted in Figure 1. The type shown is a P channel enhancement MOSFET which is built on a negatively doped (n-type) silicon. When a sufficiently large negative voltage is applied to the polysilicon gate a significant number of minority carriers (holes) will be attracted to the oxide/silicon surface from both the bulk silicon substrate and the source and drain regions.

Once a sufficient concentration of holes have accumulated there, a conduction channel is formed, allowing current to flow between the source and drain (I*ds*). The voltage necessary to initiate current flow is known as the device threshold voltage (V*TH*).
When a MOSFET device is irradiated, three things happen within the silicon dioxide layer (sensitive region):

- a build-up of trapped charge in the oxide;
- the increase in the number of interface traps;
- the increase in the number of bulk oxide traps.

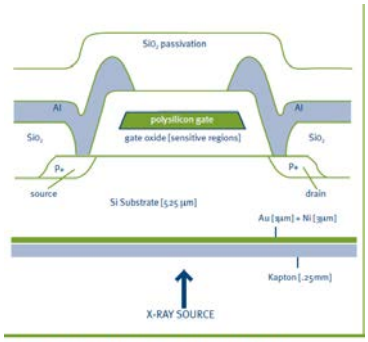


FIGURE 1. Schematic cross section of a P-channel MOSFET

Electron-hole pairs are generated within the silicon dioxide by the incident radiation. Electrons, whose mobility in SiO2 at room temperature is about 4 orders of magnitude greater than holes, quickly move out of the gate electrode while holes move in a stochastic fashion towards the Si/SiO2 interface where they become trapped in long term sites, causing a negative threshold voltage shift (V*TH*), which can persist for years.

The difference in voltage shift *before and after exposure* can be measured, and is proportional to dose.

At energies below 50 keV photoelectric absorption is the dominant process while between 60 keV and 90 keV both photoelectric and compton scattering become important. For both of these mechanisms the energy deposited to the oxide (sensitive region) of the MOSFET is delivered mostly by secondary electrons.

The photoelectric cross-section depends strongly on the target material, approximately Z4. The result of this is that interface dose enhancement occurs because the photoelectric cross section is different in the Si than in the SiO2. More electrons will be produced in the Si, the higher cross section material, so more will cross from Si into SiO2 than will cross the interface in the other direction. Dose enhancement and the increasing dominance of photoelectric absorption, as the incident energy decreases from 150 keV to 10 keV, results in an increase in device sensitivity. This energy dependence is typical of most dosimeters.

The range of photon energies used in fluoroscopy is narrow and the observed MOSFET energy dependence is 10%. This is small compared to other errors inherent in skin dose measurements, making it possible to use the dosimeter in the fluorographic energy range 40kVp to 140kVp.

Oncology procedures consist of irradiating an immobilized patient with a high energy photon (1 to 20 MeV)or electron beam(6 to 20 MeV). In these energy ranges the observed energy dependence does not exceed 5% and the MOSFETs respond identically to electrons and photons.3,4

Conclusions
The physics of MOSFET dosimeters is now well understood so that these devices can be characterized and used for many applications in radiation dosimetry. Small size and radio-transparency make the device an excellent choice for dosimetry in fluorography. Low energy dependence, high sensitivity and immediate read out make the MOSFET a good replacement for TLD in radiation therapy dosimetry.

Normal whole organ tissue tolerances (cGy)(adapted from Emami 1991 PMID 2032882)

Organ	TD 5/5			TD 50/5		
	Whole	2/3	1/3	**Whole**	2/3	1/3
Bladder	6500	8000	N/A	8000	8500	N/A
Brachial plexus	6000	6100	6200	7500	7600	7700
Brain	4500	5000	6000	6000	6500	7500
Brainstem	5000	5300	6000	6500	–	–
Cauda equina	6000	–	–	7500	–	–
Colon (obstruction,perforation,ulceration)	4500	–	5500	5500	–	6500
Ear (acute serous otitis)	3000	3000	3000	4000	4000	4000
Ear (chronic serous otitis)	5500	5500	5500	6500	6500	6500
Esophagus (stricture,perforation)	5500	5800	6000	6800	7000	7200
Femoral head	5200	–	–	6500	–	–
Heart (pericarditis)	4000	4500	6000	5000	5500	7000
Kidney	2300	3000	5000	2800	4000	–
Larynx (necrosis)	7000	7000	7900	8000	8000	9000
Larynx (edema)	4500	4500	–	8000	–	–
Lens	1000	–	–	1800	–	–
Liver	3000	3500	5000	4000	4500	5500
Lung	1750	3000	4500	2450	4000	6500
Optic chiasm	5000	–	–	6500	–	–
Optic nerve	5000	–	–	6500	–	–
Parotid gland	3200	3200	–	4600	4600	
Rectum (severe proctitis,necrosis,fistula,stenosis)	6000	–	–	8000	–	–
Retina	4500	–	–	6500	–	–
Rib cage	–	–	5000	–	–	6500
Skin	($100\ cm^2$) 5000	($30\ cm^2$) 6000	($10\ cm^2$) 7000	($100\ cm^2$) 6500	–	–
Spinal cord	(20 cm) 4700	(10 cm) 5000	(5 cm) 5000	–	(10 cm) 7000	(5 cm) 7000
Small intestine (obstruction,perforation)	4000	–	5000	5500	–	6000
Stomach (ulceration,perforation)	5000	5500	6000	6500	6700	7000
TMJ mandible	6000	6000	6500	7200	7200	7700

Thyroid	4500			8000		

The institution should compare the manufacturer's stated value with the institution's standard. If the two are within acceptable limits (Table I), either the manufacturer's or institution's value may be used.
We recommend that if the institution's verification of source strength disagrees with the manufacturer's data by more than 3%, the source of the disagreement should be investigated. We further recommend that an unresolved disparity exceeding 5% should be reported to the manufacturer

We support the earlier AAPM recommendations on source QA tests, their frequency, and tolerances as reproduced here in Table I. It should be noted that the recommended 3% tolerance between manufacturer and institution calibrations discussed above applies to the mean of a batch of sources. Since individual sources may differ from the mean by a greater amount, we recommend a maximum deviation from the mean of 5% for individual sources

Labeling of packages containing radioactive materials is required by the Department of Transportation (DOT) if the amount and type of radioactive material exceeds the limits for an excepted quantity or article as defined and limited by DOT regulations 49 CFR 173.403 (m) and (w) and 173.421-424.

TABLE I. Physical considerations involved in a TSET program,

1. Treatment field size.

   The field size of the composite electron beam at the patient treatment plane must be approximately 200 cm in height by 80 cm in width to encompass the largest patient. Within this rectangle, a vertical uniformity of ± 8% and a horizontal uniformity of ± 4% over the central 160 cm x 60 cm area of the treatment plane are achievable goals for most techniques.

2. Beam penetration depth.

   A penetration depth range from approximately 5 mm to 15 mm or more at the 50% isodose surface encompasses most lesions. It appears advantageous to provide more than one TSET beam energy to cover this range of depths. Since many electrons enter body surfaces obliquely, the energy required at the patient treatment plane for a specified average penetration depth is significantly greater than that obtained from invoking the simplistic energy loss approximation of 2 MeV/(g/cm2).

3. Electron energy at treatment plane.

   As the beam passes through the exit window and different materials between the exit window and the phantom surface, the energy will decrease and the energy spread will increase. The energy fluence distribution of such a beam arriving at the treatment plane (phantom surface) is characterized by its peak, or most probable energy Ep,o, and a lower mean energy Eo. The incident mean electron beam energy Eo at the patient treatment plane is usually in the range 3 to 7 MeV with accelerator energies, Ea, ranging from about 4 to 10 MeV.

4. Field flatness in treatment plane. (10% uniformity)

   Clear Lucite scatterer-energy degrader panel about 1 cm in thickness and 2 m x 1 m in cross section placed about 20 cm in front of the patient and contributes to large-angle scatter of the emergent electrons This improves dose uniformity, particularly on oblique body surfaces, but reduces penetration and the depth dose falls off at a shallower depth.

   Electrons reaching the patient from the scatterer placed near the accelerator exit window will have a significantly narrower angular spread than those from the scatterer placed at or near the patient surface. The wider angular spread of the latter distribution results in a higher surface dose and a shallower depth dose due to the decreased practical range because the mean angle of incidence is increased.

   Additional scatterer-degraders may be placed at the front surface of the treatment head. If improving uniformity is the criterion, a high-Z material is desirable since it maximizes scatter per unit of energy loss. Scattering and X-ray production exhibit a similar dependence on Z. Kumar, et al. find that for a 6 MeV accelerator electron beam, a relative X-ray intensity of > 15% results when a 9.6 mm Plexiglass Scatterer-

degrader is mounted on the collimator front surface When mounted as a large panel 15-29cm from the patient, the X-ray intensity is reduced to < 2%.

5. X-ray background.

The average X-ray dose can be reduced by angling the beam axes (around 10-20 degree) so that the peaks of the forward-directed X-rays lie outside the body. A desirable X-ray background level averaged over the body volume is 1% or less of the total mean electron dose at dose maximum.

6. Prescribed dose

It is recommended that TSET absorbed dose be evaluated at the calibration point located at (0,0,0) as shown in Fig. la. This dose is called the calibration point dose

The treatment skin dose is defined as the mean dose along a circle at or near the surface of a cylindrical polystyrene phantom 30 cm in diameter and 30 cm high which has been irradiated as a hypothetical patient with all six dual fields.

The cylindrical phantom, with appropriate dosimeters attached, usually film or TLD, is exposed with its proximal surface placed in the treatment plane, its cylindrical axis vertical and placed so that its front vertical surface midpoint coincides with the calibration point (x = 0, y = 0, and z = 0) as shown in Fig. la. Six dual-field exposures, each identical to the single dual-field calibration dose exposure described in Section 6.5, are given with the phantom progressively rotated 60° about its vertical axis between exposures. The dose at dmax below the phantom surface exhibits a periodicity and has a maximum value every 60° coinciding with the six angular phantom orientations intersecting the plane containing the two beam axes of each dual-field. Because of dose contributions to these maxima from the other five exposures, primarily the two dual fields ± 60° on either side, the six maxima occur at shallower depths than for a single dual-field exposure; possibly at the surface of the phantom (see Figs. 3b and 3c). The treatment skin dose has been defined as the mean dose in soft tissue evaluated along the circle passing through these six dose maxima. The dosimeter used for this averaging process is calibrated with a single dual field using the identical exposure at the calibration point.

This enables the calibration point dose to be related to the treatment skin dose by multiplication with a factor B. The electron monitor may be set so that the calibration-point dose is 1cGy/MU for the normal dual-field exposure. If D Gy is given for each of six dual fields (0.5 with the machine pointing up and 0.5 with the machine pointing down), the average skin dose given during a complete six-dual-field treatment cycle is B x D Gy. Typically, B is the range 2.5 to 3.1 for the example described but is difficult to determine with precision. It results from significant dose contributions from three dual fields and small contributions from others. The uncertainty in this factor, and thus in the mean skin dose determination, should be assessed and stressed to the responsible clinician.

7. Dose rate at treatment plane.

Average dose rates from 0.25Gy/min to several grays per minute at the depth of dose maximum are used. The six-dual-field TSET technique provides a dose rate of 0.25 Gy/min (0.60 Gy/min for the straight-ahead beam) at the treatment plane 3 m from the ion chamber-scatterer located on the front of the treatment head. The dose rate at the front of the treatment head, 50 cm from the electron source, is about 180 Gy/min for Stanford's particular ion chamber with a collection efficiency of approximately 80%. The ratio of these dose rates, 300, decreases with increasing energy.

8. Boost fields.

The areas generally boosted will be the soles of the feet, the perineal area, the dorsal surface of the penis, the skin in the peri-anal region, and the inframammary region in females with large breasts.

9. Patient positioning.

10. Special patient needs.

The lens of the eyes will generally be shielded. If the eyelids are to be treated, internal shields placed under the lids must be employed. Other body parts such as the finger nails and toe nails may be shielded by shaped sheet lead when these portions of the anatomy can be safely excluded from the treatment.

As treatment progresses, various skin reactions are to be expected. The loss of hair, finger nails and toe nails (if not shielded) is to be expected when the dose exceeds about 10 Gy. The skin may become erythematous, and there may be swelling of feet, ankles, and hands.

11. In-vivo dosimetry:

TLDs calibrated with ^{60}Co may differ by about 10% from TLDs calibrated with 4 MeV electrons in a polystyrene phantom. Hence, calibration of TLD with a parallel-plate ion chamber and electrons at TSET energy is recommended. The accuracy of the TLD dose data is adequate for patient dosimetry, since the day-to-day and patient-to-patient variation are much greater than the ± 5% accuracy that can be obtained with careful TLD dosimetry procedures

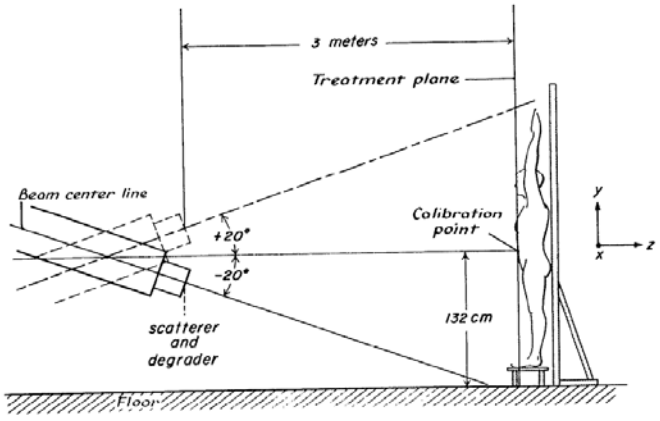


Fig. 1a. Geometrical arrangement of the symmetrical dual-field treatment technique. Equal exposures are given with each beam. The calibration point dose is at (x=0, y=0) in the treatment plane.

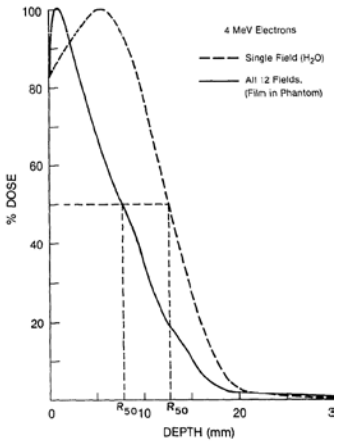


Fig. 3c. Depth doses in water for a single field with θ= 0 and for all 12 fields (six dual fields) using film in a humanoid phantom (Fraass[5]).

TABLE II. Mycosis fungoides (MF) TSET treatment prescription.
This example prescription is not to be construed as representative of all MF patient treatment prescriptions.
1. Dose - 36 Gy/9 weeks.
2. Fractionation - 4 Gy/week.
   - 4 days/week.
   - 3 dual fields/day.
3. Eyes shielded throughout.
4. Scalp shielded after 25 Gy if no involvement above neck.
5. Protect feet with 20 cm high Pb shield after first 10 Gy when sole boost starts. (Otherwise, 250 kV boost blisters tops of feet.)
6. Boost soles and perineum - Orthovoltage, 100 kV (0.5 Al HVL).
   - After first 10 Gy.
   - Rate, 1 Gy/day.

a) High dose total body irradiation. High dose total body irradiation (TBI) with megavoltage photon beams is frequently used to destroy the bone marrow and leukemic cells, to immunosuppress the patient prior to receiving a bone marrow transplant (BMT), or both. Aplastic anemia, and a number of leukemias and lymphomas, respond to this treatment. Ewing's sarcoma, advanced non-Hodgkin's lymphoma, oat cell carcinoma of the bronchus and lymphosarcoma have also been treated by TBI as an adjuvant. Usually this treatment regimen is combined with a comprehensive chemotherapy program prior to the TBI and bone marrow transplant. Total doses ranging from 300 to 1000 cGy have been given in a single fraction. In recent years, increased usage has been made of doses ranging from 1000 to 1400 cGy given in several fractions per day for several days.

b) Low dose total body irradiation. Low dose TBI with megavoltage photons giving about 10 to 15 cGy per day for 10 to 15 days is used for treatment of lymphocytic leukemias, blastoma.

c) Half body irradiation. High dose half body irradiation (HBI) for the palliation of widely disseminated metastatic disease as an adjuvant Ewing's sarcoma form of primary therapy for and bronchogenic carcinoma. Lower HBI has also been used for ovarian ablation in metastatic breast cancer. The aim of the half body technique is to give a sufficiently high dose to alleviate the effects of symptomatic disease while at the same time maintaining a sufficiently low dose to minimize complications.

d) *Total lymphoid irradiation.* Total lymphoid irradiation (TLI) has been shown to be a powerful immunosuppressive agent and, therefore, has been suggested as an adjunctive therapy for organ transplantation and a number of autoimmune diseases such as rheumatoid arthritis, aplastic anemia, multiple sclerosis and systemic lupus erythematosus. Total lymphoid irradiation using 3600 cGy in 16 fractions has produced favorable results for patients with rheumatoid arthritis.

There is no standard treatment technique for TBI, HBI or other large field treatments. As a result, significant differences in the dose distributions exist with different treatment methods. An approach uses a single point prescription, but also specifies limits for the highest and lowest dose acceptable for any point in the body. In addition, dose limits are set for certain specific tissues such as the lungs. Using this technique, a typical prescription might read:

> "The dose to the midpoint at the level of the umbilicus is 800 cGy. All points in the body must fall within the limits 840 cGy and 720 cGy (+5% and -10%). The dose to more than half the lung volume must not exceed 800 cGy. The dose rate at the prescribed point must not exceed 10 cGy/min."

Problems with detectors for *in vivo* dosimetry

	Cable	High voltage	Delay in results	Response dependent on			
				Energy	Dose rate	Dose	Temperature
Ionization chambers	XX	XXX	0	X	X	0	XX
Semi-conductor	X	0	0	XXX	0	XX	X
Thermo-luminescent dosimetry (TLD)	0	0	XX	XX	0	X	0

0 no concern
X mild concern
XX moderate concern
XXX serious concern

1. Choice of irradiation method

a) For the energy range between cobalt-60 and 25 MV x rays, the higher energies will give a more uniform dose distribution (not considering the build-up region).

b) AP/PA parallel opposed fields are preferred although under some conditions (e.g. pediatric cases, higher energies, a ±10% uniformity can be achieved with bilateral fields.

c) If the build-up region is of concern, beam spoilers or electron filters should be employed. For parallel opposed fields, the contribution of exit dose should not be neglected.

2. Basic phantom dosimetry

a) Phantoms

i) Water is the material of choice.

ii) Minimum phantom size should be 30 x 30 x 30 cm3. Larger phantoms are preferred and can be obtained by placing

water equivalent phantom materials about the minimum phantom size.

iii) Plastic phantoms will need corrections to convert to water as per TG21.

iv) Smaller phantoms will need corrections for the lack of full scatter. These corrections are dependent on phantom size, field size and energy.

b) Dosimeters

i) Dosimeter response should be energy independent.

ii) Stem and cable effects should be checked and must be minimal.

c) Dose calibration

i) In-air measurements should be avoided for the dose calibration.

ii) Use procedures as recommended by TG21 with the following changes :

A. Distance and field sizes should be representative of large field geometry.

B. Large field chamber factors should be applied.

C. A water or plastic phantom no smaller than 30 x 30 x 30 cm3 should be used. Larger phantoms are preferred. For plastic phantoms appropriate corrections must be made to determine the dose to water.

D. The measurements should be corrected to provide results representative of full scattering conditions.

E. If trays for shielding or compensators or if beam spoilers are used routinely, then these should be in position when the dose calibration is performed.

d) Central ray data

i) "In phantom" dose ratios, such as percent depth doses, tissue-phantom ratios or tissue-maximum ratios should be determined . If limited phantom sizes are used, corrections should be made for the lack of full scatter.

ii) Use cylindrical chamber at depth.

iii) Use parallel plate chamber in build-up region.

e) Inverse square test

i) Inverse square should be tested over the range of possible treatment distances that may be used for large field treatment.

ii) If deviations from the inverse square law are greater than 2% then dose calibrations should be performed at a number of distances.

3. Output factors

i) Output factors measured "in air" should be avoided unless additional scatter from the floor or walls can be accounted for.

ii) If a range of field sizes are to be used , then dose calibrations should be performed for a number of field sizes.

iii) If trays or filters are used routinely, then these should be in position for each field size.

4. Beam profiles

i) Dose ratios as measured on the central ray should also be measured along several lines which intercept the long and short field axes and are parallel to the central ray. This will provide dose profiles at any depth along the principal planes assuming all the data are normalized to a reading at some depth along the central ray.

ii) Flattening filters may have to be designed to provide a sufficient beam homogeneity.

5. Attenuation data

i) Attenuation coefficients should be measured for at least three field sizes under treatment conditions. A plot of broad beam attenuation coefficient versus side of square field will allow interpolation for any field size.

ii) The broad beam coefficient to be used for treatment calculations will depend on the size of the absorber or the field size if the absorber covers the entire beam.

6. Dose prescription

i) The dose should be prescribed to one point.

ii) Dose limits should be specified.

iii) The dose limit to critical organs may have to be specified separately.

iv) The therapy unit dose rate as well as the total duration of the treatment may need to be specified depending on the clinical requirements.

7. Corrections for patient size

i) Equivalent patient dimensions should be determined and central ray data for full scattering conditions should be corrected to account for the patient dimensions both in the lateral and depth directions.

ii) Contour corrections should be made using methods that are field size dependent i.e. ratio of tissue-air ratio method or ratio of tissue-phantom ratio method.

iii) For treatment time or monitor unit calculations either the field size or the patient size should be used depending on which one is smaller.

iv) The lack of backscatter can result in noticeable dose reduction especially for small patient thicknesses. The use of bolus or back scattering material will help to reduce this effect.

8. Compensators for missing tissue

i) The use of tissue equivalent bolus is the easiest if the loss of skin sparing is not a problem.

ii) If skin sparing is to be maintained compensators-should be built.

9. Inhomogeneities: Lung

i) When the dose to lung is critical, inhomogeneity corrections should be made.

a) For manual calculations, methods which are field size dependent should be used i.e. ratio of TAR's or ratio of TPR's (the power-law method should not be used with TAR's since it is in error for large fields) .

ii) Inhomogeneity corrections should be made using CT data to provide anatomical information.

a) The best is to perform pixel based calculations.

b) Alternately, an average density can be assumed.

c) Without CT, transmission measurements can provide equivalent thickness data.

d) Radiographs combined with an assumption about average density will provide an estimate of equivalent thickness.

e) The simple relation between dose correction factor and patient thickness can be used if no other information is available.

10. Inhomogeneities: Bone

i) The dose to bone can be calculated from equation (5) for lower energy photons in section 4.4.2 using the data of Table 7 or equation (6) for higher energy photons.

$$D_{bone} = D_{water} \left(\frac{\bar{\mu}_{en}}{\rho} \right)_{water}^{bone}$$

$$D_{bone} = D_{water} \left(\frac{\bar{\mu}_{en}}{\rho} \right)_{water}^{tissue} \quad \overline{\overline{S}}_{tissue}^{bone}$$

11. Methods of compensating for lung tissues

i) Produce three dimensional compensator from full dose distributions with inhomogeneity corrections.

ii) Use constant thickness lung attenuator.

iii) Use shields for part of the treatment time.

12. Anthropomorphic phantom measurements

i) TLD measurements should be performed in anthropomorphic phantoms using the treatment techniques and doses that are representative of patient treatments. The TLD response should be calibrated such that absolute doses can be determined.

13. *In vivo* measurements

i) The treatment technique should be verified by performing TLD measurements on several representative patients. The TLD response should be calibrated to determine absolute doses. To obtain an indication of the dose at dmax, the dosimeters should have sufficient tissue equivalent material such that full build-up is achieved. Exit dosimeter readings should be corrected for the lack of backscatter if they are to be used for determination of midplane doses.

Clinical photon beams emanating from a medical linac are produced in an X ray target and flattened with a flattening filter. At electron energies below 15 MeV (photon beam energies 15 MV) optimal targets have a high atomic number *Z*, while at electron energies above 15 MeV (photon beam energies above 15 MV) the optimal targets have a low atomic number *Z*. Optimal flattening filters have a low *Z* irrespective of beam energy. (filter usually made of lead, although tungsten, uranium, steel, Al or a combination has been used)

Hyperfractionated radiotherapy

From NCI: hyperfractionated radiation therapy
Radiation treatment in which the total dose of radiation is divided into small doses and treatments are given more than once a day. Also called hyperfractionation and superfractionated radiation therapy.

The delivery of radiation in small-dose fractions 2 to 3 times/day to minimize damage to normal tissue typical of smaller fractions. Hyperfractionation is generally expected to allow an escalation of total dose, thereby increasing tumour control rate, without increasing the risk of late complications.

From Khan' RTP book:

A hyperfracionated course of radiotherapy is one in which more than one fraction is delivered each day but the overall treatment time remains the same as for conventional fractionation. Typically this means 1.2 to 1.3Gy/fx, two fx per day, with an increase in total dose of order of 20-30% to account for repair at the lower dose per fraction. The major rationale is to take max advantage of the difference in repair capacity of late-reacting normal tissues compared with tumors.

Hypofractionated radiotherapy

From NCI:
Radiation treatment in which the total dose of radiation is divided into large doses and treatments are given less than once a day. Also called hypofractionated radiation therapy.

Pulsed radiation fields are encountered in the vicinity of electron Linacs (McCall and Ipe 1988). Most therapy Linacs are operated at repetition rates varying from 100 to 400 pulses per second with pulse widths of about 1 to 10 microseconds (AAPM 1986). The fraction of operating time (that is, pulse width x repetition rate) during which the beam is on is called the duty factor. For example, the duty factor for an electron Linac operating at 100 pulses per second with a pulse width of 10 μs is 0.001. These small duty factors impose severe limitations on the radiation detection instruments. The peak intensity is equal to the average intensity divided by the duty factor, which in this case will be 1,000 times higher than the average intensity. An instrument that normally responds well to the average dose rate spread out evenly in time will not be able to cope with such a high dose rate. The intense photon pulse usually overwhelms any active detector (that detects particle events electronically). Instruments which have long dead times, such as the GM and proportional counters, tend to become saturated in such fields and only count the repetition rate.
Thus GM counters should not be used to perform radiation surveys outside exterior walls of the Linac room.
Scintillation survey meters may become nonlinear at higher dose rates because photomultipliers cannot handle the high instantaneous currents. Ionization chambers are less influenced; however they must be operated with adequate voltage to overcome recombination losses.

Q: How will the guard electrode in an ionization chamber prevent the leakage current from the high-voltage collector (central electrode)?

A: You do not specify the type of ionization chamber you are concerned with. Among the most common types are the cylindrical chamber that contains a cylindrical central electrode along the longitudinal central axis of the chamber and the parallel plate chamber in which the two major electrodes are often disc-shaped and parallel to each other. The orientation and functioning of the guard rings (or guard electrodes) in these two designs are somewhat different. The guard rings play a major role in reducing the leakage of extraneous charge to the collecting electrode.

In the cylindrical design the central electrode, often the high-voltage electrode, is separated from the outer electrode, represented by the inner wall of the ionization chamber, by insulating material, often a plastic material. Because the ionization chamber is often used to measure quite small currents induced by the radiation field, small amounts of charge leakage through the insulator may contribute to the measured current and provide false readings. Even very good insulators may not have sufficient electrical resistance to reduce such leakage to a negligible level. To get around this problem such ionization chambers are often designed with a guard electrode, which is usually a cylindrical conductor that wraps around the insulator that is in contact with the inner central electrode. A second insulator wraps around the outside of the guard electrode and separates the guard from the outer electrode. **The guard electrode is maintained at the same potential as the central electrode.** With this arrangement the major voltage drop across the insulator is across the outer insulator between the outer electrode and the guard ring so that any charge leaking in does not get to the central electrode. Since the potentials of the guard electrode and the central electrode are the same, there is virtually no driving force to induce charge leakage across the inner insulator.

In the parallel plate chamber the charge-collecting electrode is surrounded by an annular ring. The annular ring represents the guard ring (or guard electrode) and is separated from the collecting electrode by a narrow insulating gap, and the applied voltage to the guard ring is the same as that to the collecting electrode. The other parallel plate

electrode has a diameter at least as great as the outer diameter of the guard electrode. With this geometry and voltage arrangement the guard electrode **serves two major purposes, although they are directly linked. First, it ensures that the electric field lines near the edge of the collecting electrode remain straight, perpendicular to both the collecting electrode and the second electrode so that the collecting volume is accurately defined by the area of the collecting electrode and the electrode separation. (If no guard electrode is used, the field lines near the edge of the collecting electrode will tend to bow out, resulting in a poorly defined charge collecting volume.) Secondly, maintaining the same voltage on the collecting electrode and the guard ring minimizes the extent of charge leakage from the volume outside of the collecting volume because there is no significant driving potential to promote this additional charge collection by the collecting electrode in preference to collection by the guard electrode.**

The principal sources of dose outside a treated volume are (1) photon leakage through the treatment head of the machine, (2) radiation scattered from the collimators and beam modifiers, and (3) radiation scattered within the patient from the treatment beams. For higher-energy (>10 MV) photon beams there is an additional contribution from neutrons emanating from the treatment head, neutrons produced from photoneutron interactions in the patient, and radioactive isotopes produced in photoneutron interactions.

The most important determinant of the peripheral dose is the distance from the radiation field edge, with the dose decreasing approximately exponentially with distance from the field edge.

In summary, the dose outside a beam is a function of the distance from the beam edge and the field size and depends on primary radiation energy and depth within the patient. Although variation exists among machines, the major components of the out-of-beam dose within 10 cm of the beam edge are typically scatter off the collimator and scatter from the useful beam within the patient. In the region 10 to 20 cm from the field edge, collimator scatter decreases so that the major component of the out-of-beam dose is scatter within the patient; however, collimator scatter and head leakage also contribute to the dose. At about 30 cm, scatter in the patient and head leakage are approximately equal, and beyond that point, head leakage predominates. Scatter from special blocking devices, such as wedge filters, increases the dose near the beam edge by a factor of 2 to 5. Measurements to separate the dose components in a formal way are not necessary, but the medical physicist should be aware that the out-of-beam dose includes radiation from several sources, some that can be reduced by shielding (head leakage and collimator scatter) and some that cannot (scatter within the patient).

Section 35.75, “Release of Individuals Containing Unsealed Byproduct Material or Implants Containing Byproduct Material,” of 10 CFR Part 35, “Medical Use of Byproduct Material,” permits a licensee to “authorize the release from its control any individual who has been administered unsealed byproduct material or implants containing byproduct material if the total effective dose equivalent to any other individual from exposure to the released individual is not likely to exceed 5 millisieverts (0.5 rem).”

Licensees may release patients to whom radionuclides have been administered in amounts greater than the activities listed in Column 1 of Table U.1, provided the measured dose rate at 1 meter (from the surface of the patient) is no greater than the value in Column 2 of Table U.1 for that radionuclide.

§ 35.75 Release of individuals containing unsealed byproduct material or implants containing byproduct material.

(a) A licensee may authorize the release from its control of any individual who has been administered unsealed byproduct material or implants containing byproduct material if the total effective dose equivalent to any other individual from exposure to the released individual is not likely to exceed 5 mSv (0.5 rem).

(b) A licensee shall provide the released individual, or the individual's parent or guardian, with instructions, including written instructions, on actions recommended to maintain doses to other individuals as low as is reasonably achievable if the total effective dose equivalent to any other individual is likely to exceed 1 mSv (0.1 rem). If the total effective dose equivalent to a nursing infant or child could exceed 1 mSv (0.1 rem) assuming there were no interruption of breast-feeding, the instructions must also include

Proportional counter

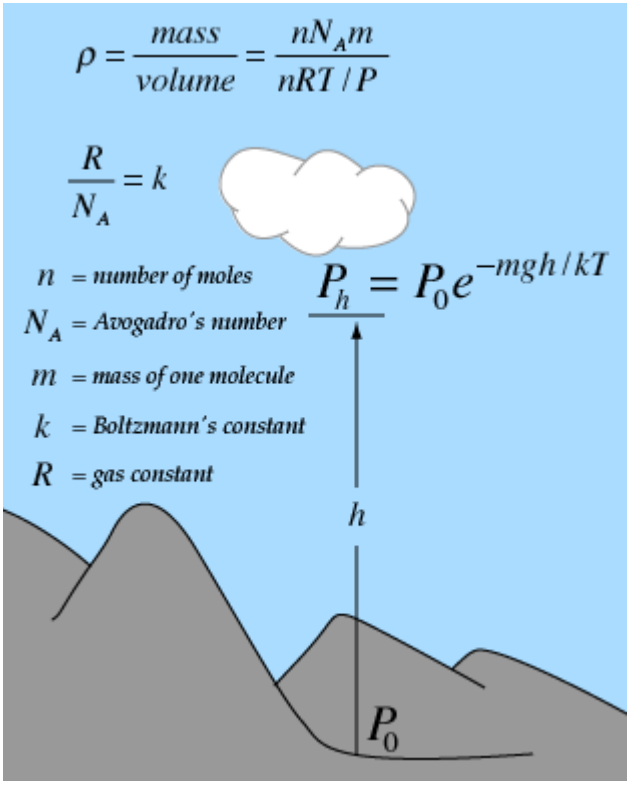
A **proportional counter** is a measurement device to count particles of ionizing radiation and measure their energy.

A proportional counter is a type of gaseous ionization detector - it works on the same principle as the Geiger-Müller counter, but uses a lower operating voltage. An inert gas is used to fill the tube, with a quench gas added as a stabilizer. A common proportional gas mixture is 90% argon, 10% methane, known as P-10. An incoming ionizing particle, if it has sufficient energy, liberates electrons from the atomic orbitals of the gas atoms (see ionization potential), leaving an electron and positively charged atom, commonly known as an ion pair. As the charged particle travels through the chamber it leaves a trail of ion pairs along its trajectory. The electrons created in this process drift toward a readout electrode, known as the anode, under the influence of an applied electric field. At the same time, the positive ions drift towards the cathode, at much lower speed; in practical devices, the drift times are measured in microseconds and milliseconds, respectively.

A proportional counter differs from an ionization chamber in that the operating voltage is sufficiently high that the drifting electrons gain enough energy over a mean free path to create further ion pairs when they collide with other neutral atoms of the gas. The electrons created in these new events also drift toward the readout electrode and can create further ion pairs themselves. In this manner, a cascade of ion pairs can be created, this is known as a Townsend avalanche. If the operating voltage is chosen carefully, each avalanche process occurs independently of other avalanches which derive from the same initial ionizing event. Therefore, even though the total number of electrons liberated can increase exponentially with distance, the total amount of charge created remains proportional to the amount of charge liberated in the original event.

By measuring the total charge (time integral of the electric current) between the electrodes, we can find out the particle's kinetic energy, because the number of ion pairs created by the incident ionizing charged particle is proportional to its energy.

The Barometric Formula



The equation for the variation of barometric pressure with height has the form

$$\frac{dP}{dh} = \frac{-mg}{kT} P$$

which has the formal solution

$$\frac{dP}{dh} = \frac{-mg}{kT} P$$

Substituting the solution gives

$$\left[bAe^{bh}\right] = -\frac{mg}{kT}\left[Ae^{bh} + C\right]$$

Since this equation must be valid for all values of h , forcing the solution to fit the physical boundary conditions yields:

$C = 0$ since the other terms vary with h

$b = -\frac{mg}{kT}$ by setting coefficients equal

$A = P_0$ since that must be the value for h=0.

introduced the concepts of $TD_{5/5}$ and $TD_{50/5}$ (the NTCP at 5% and 50%, respectively, within 5 years after radiotherapy) and published normal tissue tolerance data, in terms of $TD_{5/5}$ and $TD_{50/5}$, for a number of normal tissues and organs.

Stereotactic surgery or **stereotaxy** (not to be confused with the virtuality concept of *stereotaxy*) is a minimally-invasive form of surgical intervention which makes use of a three-dimensional coordinates system to locate small targets inside the body and to perform on them some action such as ablation (removal), biopsy, lesion, injection, stimulation, implantation, radiosurgery (SRS) etc. "Stereotactic" in Greek (another accepted spelling is "stereotaxic") means "solid ordering".
From M-W dictionary: involving, being, utilizing, or used in a surgical technique for precisely directing the tip of a delicate instrument (as a needle) or beam of radiation in three planes using coordinates provided by m
edical imaging in order to reach a specific locus in the body

The penetrating ability of beta particles in air as a function of energy is shown in Fig. 2.12

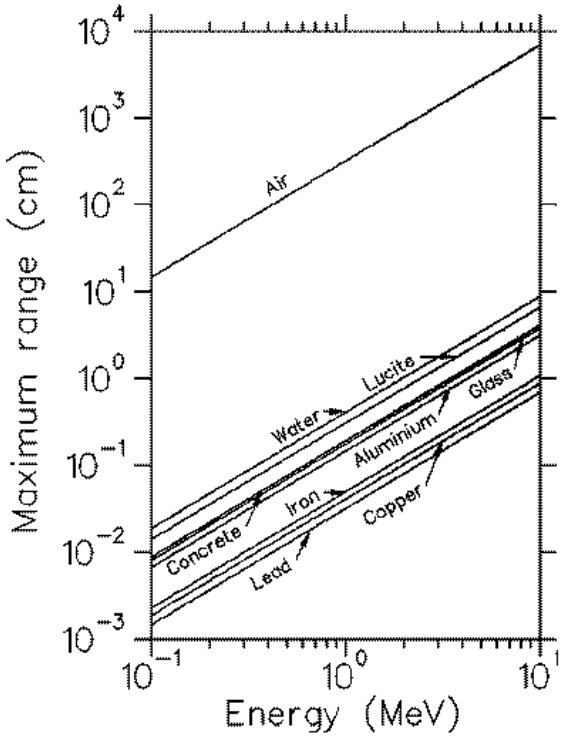


Figure 2.12: Range of beta particles in air

A rule of thumb, from the 1970 edition of the "Rad Health Handbook," is that the range of a beta particle, in g/cm2, is approximately equal to half the maximum energy, in Mev, i.e. R ~ E/2.

For more precise determinations, I recommend consulting "Introduction to Radiological Physics and Radiation Dosimetry," by Frank H. Attix (1986, John Wiley & Sons).

It requires a beta particle of at least 70kev to penetrate the protective layer of skin (0.07mm)
It requires a alpha particle of at least 7.5MeV to penetrate the protective layer of skin

Average E: $E(\beta^-) = 1/3$ Emax
$E(\beta^+) = 0.44$ Emax

Range in Air: Range-air = 12ft per MeV =30.48cm*12ft per Mev = 3.66m per MeV
Example: P^{32} Emax = 1.71MeV , max range in air = 20ft

Range in matter: $R(g/cm^2) = Emax / 2$ where R = range in g/cm2 (range in cm times the density g/cm^3)
E = maximu energy in MeV (1-4MeV energy range)

For example, Emax =1MeV, in water, R = $0.5g/cm^2$ = 0.5cm * ($1g/cm^3$ for water) = 0.5cm in water, match the plot above

PDD and Skin dose with wedges:
Surface dose of EDW appear to be slightly higher (up to 3%) than open field
By comparison, surface doses for physical wedge were reduced 7-12% compared to open field, indicating the filtration effects of physical wedge beam modifier.
PDD for EDW are similar to the depth dose for open field. Agreement to within 2% for depths from dmax to 30cm for nearly every field.
Depth doses using physical wedge filter differ because the introduction of the wedge filter into the beam may change the spectrum. For 6MV, 15x15 field, the wedge dose versus open field dose ration shoed a progressive increase to more than 10% at a depth of 30cm, compared to 2% effect in EDW.

NRC CFR-title10- part71:
TI definition:
Transport index (TI) means the dimensionless number (rounded up to the next tenth) placed on the label of a package, to designate the degree of control to be exercised by the carrier during transportation. The transport index is the number determined by multiplying the maximum radiation level in millisievert (mSv) per hour at 1 meter (3.3 ft) from the external surface of the package by 100 (equivalent to the maximum radiation level in millirem per hour at 1 meter (3.3 ft)).

Labeled with a Radioactive White I, Yellow II, or Yellow III label as specified in U.S. Department of Transportation regulations, 49 CFR 172.403 and 172.436-440.

The proper label to affix to a package of Class 7 (radioactive) material is based on the radiation level at the surface of the package and the transport index. The proper category of label must be determined in accordance with paragraph (c) of this section. The label to be applied must be the highest category required for any of the two determining conditions for the package.
RADIOACTIVE WHITE-I is the lowest category and RADIOACTIVE YELLOW-III is the highest.
For example, a package with a transport index of 0.8 and a maximum surface radiation level of 0.6 millisievert (60 millirems) per hour must bear a RADIOACTIVE YELLOW-III label.

(c) Category of label to be applied to Class 7 (radioactive) materials packages:

Transport index	Maximum radiation level at any point on the external surface	Label category [1]
0[2]	Less than or equal to 0.005 mSv/h (0.5 mrem/h).	WHITE-I.
More than 0 but not more than 1	Greater than 0.005 mSv/h (0.5 mrem/h) but less than or equal to 0.5 mSv/h (50 mrem/h).	YELLOW-II.
More than 1 but not more than 10	Greater than 0.5 mSv/h (50 mrem/h) but less than or equal to 2 mSv/h (200 mrem/h).	YELLOW-III.
More than 10	Greater than 2 mSv/h (200 mrem/h) but less than or equal to 10 mSv/h (1,000 mrem/h).	YELLOW-III (Must be shipped under exclusive use provisions; see 173.441(b) of this subchapter).

[1] Any package containing a "highway route controlled quantity" (§ 173.403 of this subchapter) must be labelled as RADIOACTIVE YELLOW-III.
[2] If the measured TI is not greater than 0.05, the value may be considered to be zero.

IMRT optimization:

Steve Webb also introduced the optimization method of simulated annealing, which had success in image processing before, into the field of IMRT (Webb 1989). Simulated annealing simulates the technical process of annealing a metal. With a carefully designed cooling scheme, the properties of the metal can be optimized, and so can the properties of a treatment plan if simulated annealing is used in IMRT. Themain advantage of simulated annealing is that it can escape local minima while searching for a global optimum. However, it quickly became clear that the problem of optimizing intensity maps using a typical (for example quadratic) objective function does not have local minima, so much faster downhill methods can be used. Nevertheless, simulated annealing has been implemented in the commercial Peacock planning system (Carol 1995) and was used for the planning of the first patient treated with IMRT in 1994, see also figure.

I realized that this problem with a quadratic objective function does not have local minima, so fast gradient descent methods can be used to find its solution. I developed the algorithm, borrowing heavily from image reconstruction methods, and implemented the code, both in 2D and later in 3D. Even on the hardware of that time the algorithm achieved respectable execution times (several minutes for a not too large case (Bortfeld *et al* 1990)). Various forms of gradient descent algorithms have since become the standard of inverse planning algorithms in commercial treatment planning systems. Using the initial algorithm, I analysed the effect of the number of beams on the resulting dose distribution and found that less than 10 intensity-modulated beams are often enough to get to a point of diminishing returns (Bortfeld *et al* 1990). This finding was relevant because most or all investigators at that time assumed the use of many more than 10 beams to simulate a rotational treatment. The reduction of the number of beams paved the way for a practical implementation of IMRT on treatment machines equipped with a multileaf collimator (MLC), which just began to appear on the horizon around 1991.