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College on Medical Physics: Radiation Protection and Imaging Techniques

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Principles of Equipment in Radiotherapy

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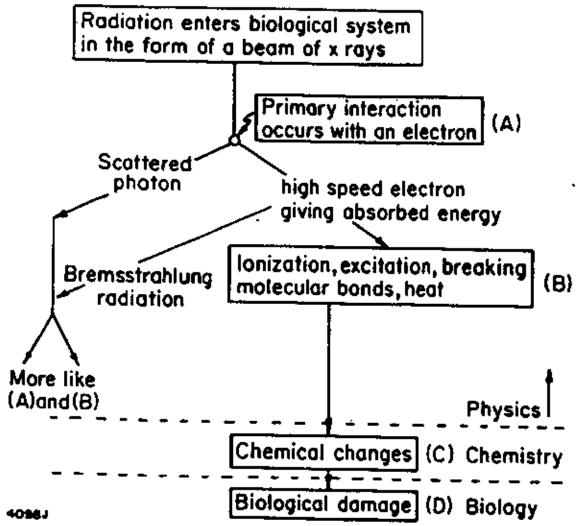
DOSIMETRY

When a x-ray or γ-ray beam passes through a medium, interaction between photons and matter can take place with the result that ***energy transfer to the medium***.

If the absorbing medium consists of body tissues, sufficient energy may be deposited within its cells, destroying their reproductive capacity (***biological damage***). However most of the absorbed energy is converted into heat, producing no biological effect.



In dosimetry two distinctly point of view are taken in consideration to characterize the interaction of a radiation beam with a medium:

- ***description of the radiation beam,***

- ***description of the amount of the energy deposited in some medium.***

QUANTITIES AND UNITS USED IN DOSIMETRY

FLUENCE or PHOTON FLUENCE: $$\Phi = \frac{dN}{da} \quad \frac{\text{number of photon}}{\text{area}}$$

dN is the number of photons that would cross an area, da, at right angles to the beam

FLUENCE or ENERGY FLUENCE: $$\Psi = \frac{dN \cdot h\nu}{da} \quad \frac{\text{energy}}{\text{area}}$$

$dN \cdot h\nu$ is the amount of energy crossing an area, da, at right angles to the beam

QUANTITIES AND UNITS USED IN DOSIMETRY

EXPOSURE: $$X = \frac{dQ}{dm} \quad \frac{\text{charge}}{\text{mass}}$$

dQ is the absolute value of the total charge of the ions of one sign produced in air when al the electrons liberated by photons in a volume of air having mass dm are completel stopped in air.

The SI (Systems Internationale d'Unites) unit for exposure is Coulomb per kilograr (C/kg), redefined as Roengten (R):

$$1\ R = 2.58 \times 10^{-4}\ C/kg_{\ air}$$

ABSORBED DOSE: $$D = \frac{dE_{ab}}{dm} \quad \frac{\text{energy}}{\text{mass}}$$

dE is the mean energy imparted by the ionizing radiation to a mass, dm, of matter.

The SI (Systems Internationale d'Unites) unit for absorbed dose is called **Gray** a is defined as:

$$1\ Gy = 1\ J/kg$$

RELATION BETWEEN THE EXPOSURE AND ABSORBED DOSE IN AIR

Under equilibrium condition, since W_{air}/e is the average energy absorbed per unit charge of ionization produced in air (its value is equal to 33.85 J/C), the absorbed dose in air may be expressed as:

$$D_{air} = X\ W_{air} / e$$

$$D_{air}\ (\text{rad}) = 0.873\ \ (\text{rad/R})\ X(\text{R})$$

The roengten-to-rad conversion factor for air is 0.873

RELATION BETWEEN THE EXPOSURE AND ABSORBED DOSE IN ANY MEDIUM

Under conditions of electronic equilibrium, the dose in air is related to the dose in the medium by the following relationship:

$$\frac{D_{med}}{D_{air}} = \frac{(\mu_{en}/\rho)_{med}\ A}{(\mu_{en}/\rho)_{air}}$$

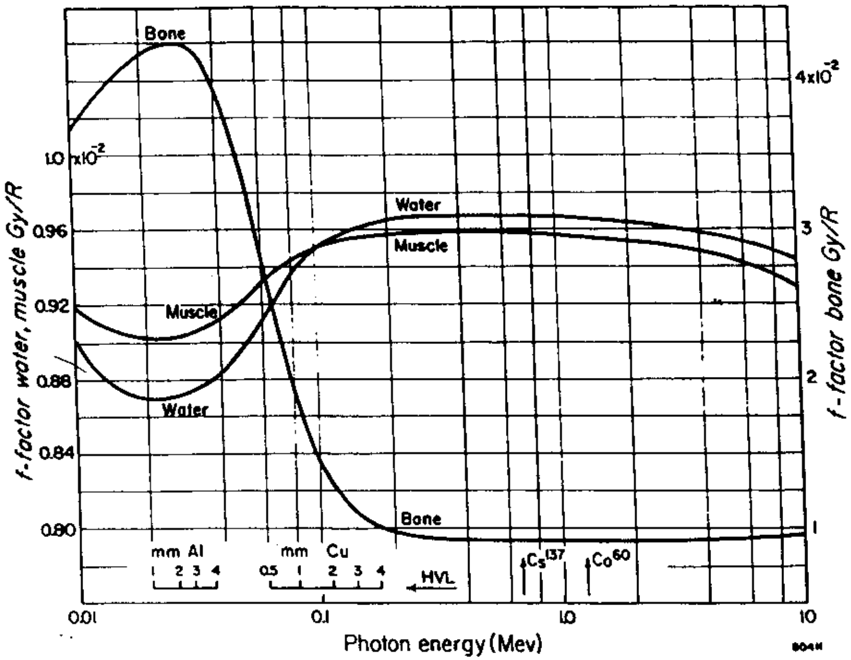
where A is a trasmission factor which equals to the ratio of the respectively energy fluences at the point of interest.

From the relationship between the exposure and the absorbed dose in air, the absorbed dose in a medium is related to exposure in the following way:

$$D_{med} = 0.873\ \frac{(\mu_{en}/\rho)_{med}}{(\mu_{en}/\rho)_{air}}\ X\ A$$

The f_{med} depends on the mass energy absorption coefficient of the medium relative to the air. Thus the f_{med} factor is a function of the medium composition as well as the photon energy.

Diagram of *f* factor for water, bone and muscle as a function of photon energy



ENERGY TRANSFER: A TWO STAGE PROCESS

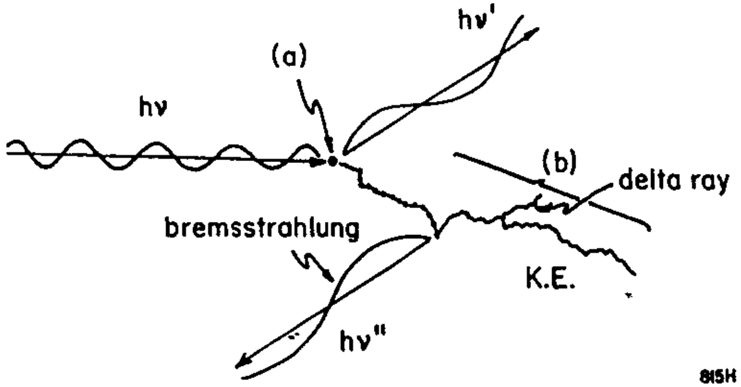
The transfer of energy from a photon beam to the medium takes place in two stages.

a) *Interaction of the photon with an atom by three important mechanism known as photoelectric process, Compton incoherent scattering and Pair production,* ***causing an electron or electrons to be set in motion,***

b) *Transfer of the energy* ***from high energy electron to the medium through excitation and ionization.***

The transfer of energy along the stage a) is called *Kerma*

and along the stage b) is called *Absorbed Dose*



hν' is the photon scattered from event a.

hν" is a Bremsstrahlung photon produced with the collision between the electron and a nucleus.

The delta ray is another electron track resulting from a relatively violent electron-electron collision.

KERMA

Kerma stands for Kinetic Energy Released in the Medium

$$K = \frac{dE_{tr}}{dm} \quad (\text{MeV/kg})$$

*where the dE_{tr} is the sum of the **initial kinetic energies** of all the charged ionizing particles (electrons) liberated by uncharged ionizing particles (photons) in a material of mass dm.*

If we have a beam of photons with energy hν and photon fluence Φ, the kerma is given by:

$$K = \Phi\,(\mu / \rho)\,E_{tr}$$

where (μ / ρ) is the mass attenuation coefficient for the medium and E_{tr} is the average amount of energy transferred to electrons

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ABSORBED DOSE

$$D = \frac{dE_{ab}}{dm}$$

dE_{ab} is the mean energy imparted by the ionizing radiation to a mass, dm, of matter.

KERMA AND ABSORBED DOSE

Energy is transferred to an electron in the first interaction but not all of it is retained in the medium: some is radiated away as bremsstrahlung. The energy actually retained in the medium is brought about by the ionization and excitations that take place along the electron track.

Because of the length of the electron tracks may be appreciable, kerma and absorbed dose don't take place at the same location:

> ***The absorbed dose is equal to the Kerma less the energy carried away by Bremsstrahlung.***

Types of photon interactions in water, as a function of energy

$h\nu$	% Interactions by Each Process				% Energy Transferred			% Energy Lost to Bremsstrahlung
	a	b	c	d	e	f	g	h
(keV)	Coh	Compton	Photo	Pair	Compton	Photo	Pair	
10.0	4.5	3.1	92.4	0.0	0.1	99.9	0.0	0.0
15.0	8.5	10.8	80.7	0.0	0.4	99.6	0.0	0.0
20.0	11.6	23.3	65.1	0.0	1.3	98.7	0.0	0.0
30.0	13.0	50.7	36.3	0.0	6.8	93.2	0.0	0.0
40.0	11.0	69.6	19.4	0.0	19.3	80.7	0.0	0.0
50.0	8.6	80.4	11.0	0.0	37.2	62.8	0.0	0.0
60.0	6.8	86.6	6.6	0.0	55.0	45.0	0.0	0.0
80.0	4.5	92.6	2.9	0.0	78.8	21.2	0.0	0.0
100.0	3.1	95.3	1.5	0.0	89.6	10.4	0.0	0.0
150.0	1.6	97.9	0.5	0.0	97.4	2.6	0.0	0.0
200.0	1.0	98.8	0.2	0.0	99.0	1.0	0.0	0.0
300.0	0.5	99.4	0.1	0.0	99.7	0.3	0.0	0.1
400.0	0.4	99.6	0.0	0.0	99.9	0.1	0.0	0.1
500.0	0.3	99.7	0.0	0.0	99.9	0.1	0.0	0.1
600.0	0.2	99.8	0.0	0.0	100.0	0.0	0.0	0.1
800.0	0.1	99.9	0.0	0.0	100.0	0.0	0.0	0.2
(MeV)								
1.0	0.1	99.9	0.0	0.0	100.0	0.0	0.0	0.2
1.5	0.0	99.8	0.0	0.2	99.9	0.0	0.1	0.4
2.0	0.0	99.2	0.0	0.8	99.3	0.0	0.7	0.5
3.0	0.0	97.1	0.0	2.9	96.7	0.0	3.3	0.8
4.0	0.0	94.5	0.0	5.5	93.3	0.0	6.7	1.1
5.0	0.0	91.6	0.0	8.4	89.6	0.0	10.4	1.4
6.0	0.0	88.9	0.0	11.1	86.2	0.0	13.8	1.6
8.0	0.0	83.1	0.0	16.9	79.0	0.0	21.0	2.3
10.0	0.0	77.0	0.0	23.0	71.9	0.0	28.1	2.9
15.0	0.0	65.6	0.0	34.4	59.3	0.0	40.7	4.6
20.0	0.0	56.0	0.0	44.0	49.3	0.0	50.7	6.5
30.0	0.0	43.2	0.0	56.8	37.1	0.0	62.9	10.0
40.0	0.0	35.1	0.0	64.9	29.7	0.0	70.3	13.6
50.0	0.0	29.3	0.0	70.7	24.6	0.0	75.4	16.8
60.0	0.0	25.3	0.0	74.7	21.1	0.0	78.9	19.8
80.0	0.0	19.7	0.0	80.3	16.4	0.0	83.6	25.3
100.0	0.0	16.0	0.0	84.0	13.3	0.0	86.7	30.1

a. $100\ \sigma_{coh}/\mu$
b. $100\ \sigma_{inc}/\mu$
c. $100\ \tau/\mu$
d. $100\ \pi/\mu$
e. $100\ \sigma_{tr}/\mu_{tr}$
f. $100\ \tau_{tr}/\mu_{tr}$
g. $100\ \pi_{tr}/\mu_{tr}$
h. $100\ (1 - \bar{E}_{ab}/\bar{E}_{tr})$

BEHAVOUR OF KERMA AND ABSORBED DOSE IN A MEDIUM

(PATIENT OR PHANTOM)

As the high energy photon beam enters the medium, high speed electrons are ejcted from the surface and the subsequent layers.

These electrons deposit their energy a significant distance away from their site of origin.

The electron fluence and hence the absorbed dose increases with depth until it reachs a maximum.

However the photon energy fluence continuously decreases with depth and as a result the production of secondary electrons also decreases with depth.

NET EFFECT

1) the kerma is maximum at the surface and decreases with depth (due to the decrease of photon energy fluence)

2) the absorbed dose reachs the maximum at a depth approximately equal to the range of electrons in the medium, and then decreases with depth.

DOSE DISTRIBUTION

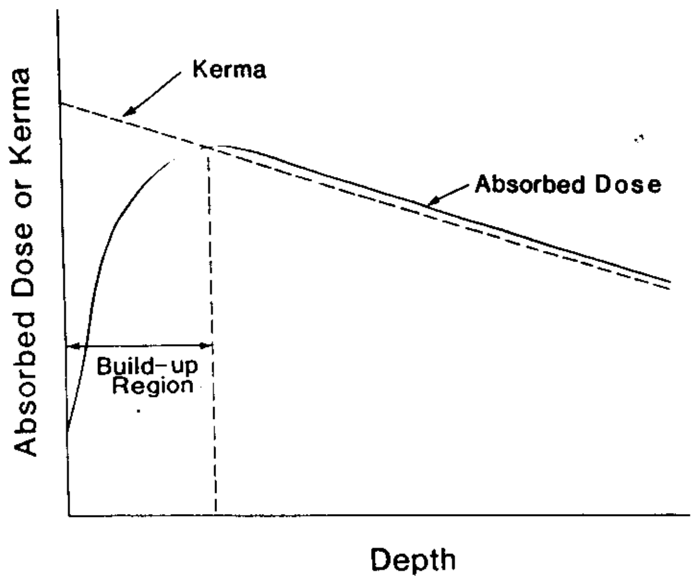
Basic dose distribution data are usually measured in a water phantom, whic closely approximates the radiation absorption and scattering properties of musc and other soft tissues.

Dose distribution data can be derived from measurements in **solid di phantoms**: ***tissue or water equivalent materials***, large enough in volume to provic full scatter conditions and the same electron density as that of water.

The Compton effect is the most predominant mode o interaction for megavoltage photon beams in the clinical range.

The region between the surface and the maximum is called:

Dose Build-Up Region



Anthropomorphic Phantoms

In addition to the homogeneous phantoms, *anthropomorphic phantoms* are frequently used for clinical dosimetry.

A commercially available system, called **Alderson Rando Phantom**, incorporates materials to simulate various body tissue: muscle, bone, lung and air cavities, and it's shaped into a human torso and it is sectioned into transversal slices.

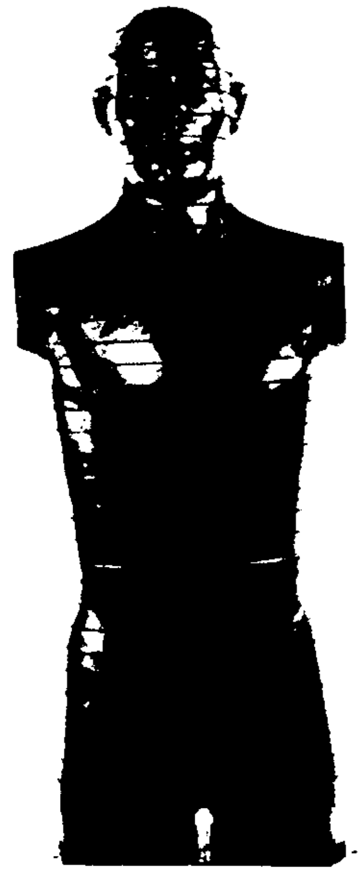


Table of some physical properties of various phantom material.

Material	Chemical Composition	Mass Density (g/cm^3)	Number of Electrons/g ($\times 10^{23}$)	Z_{eff}* (photoelectric)
Water	H_2O	1	3.34	7.42
Polystyrene	$(C_8H_8)_n$	1.03–1.05	3.24	5.69
Plexiglas (Perspex, Lucite)	$(C_5O_2H_8)_n$	1.16–1.20	3.24	6.48
Polyethylene	$(CH_2)_n$	0.92	3.44	6.16
Paraffin	C_nH_{2n+2}	0.87–0.91	3.44	5.42
Mix D	Paraffin: 60.8 Polyethylene: 30.4 MgO: 6.4 TiO_2: 2.4	0.99	3.41	7.05
M 3	Paraffin: 100 MgO: 29.06 $CaCO_3$: 0.94	1.06	3.34	7.35

Data are from Tubiana *et al.* (2) and Schulz and Nath (3).
* Z_{eff} for photoelectric effect is given by Equation 6.4.

DEPTH DOSE DISTRIBUTION

As the beam is incident on a patient or a phantom, the absorbed dose in the patient varies with depth.
The variation depends on many conditions:

- ***beam energy***
- ***depth***
- ***field size***
- ***distance from source***
- ***beam collimation system***

An essential step is to establish depth dose variation along the central axis of the beam.
For this purpose a number of quantities have been defined:

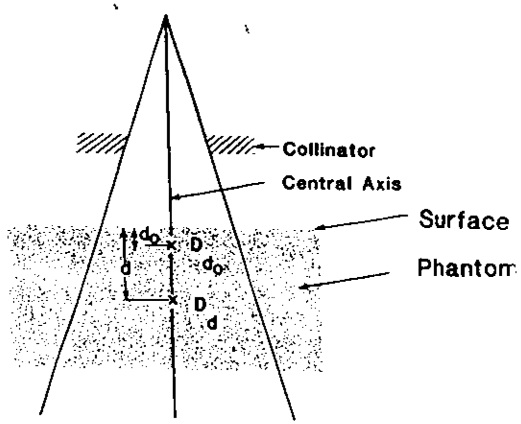
- ***percentage depth dose***
- ***tissue air ratio***
- ***tissue-phantom ratio***
- ***tissue-maximum ratio***

These quantities are usually derived from measurements made in water phantom using small ionization chamber or diodes

PERCENTAGE DEPTH DOSE

One way of characterizing the central axis dose distribution is to normalize dose at depth with respect to dose at the reference depth.

$$P = D_d / D_{d_o} \times 100$$

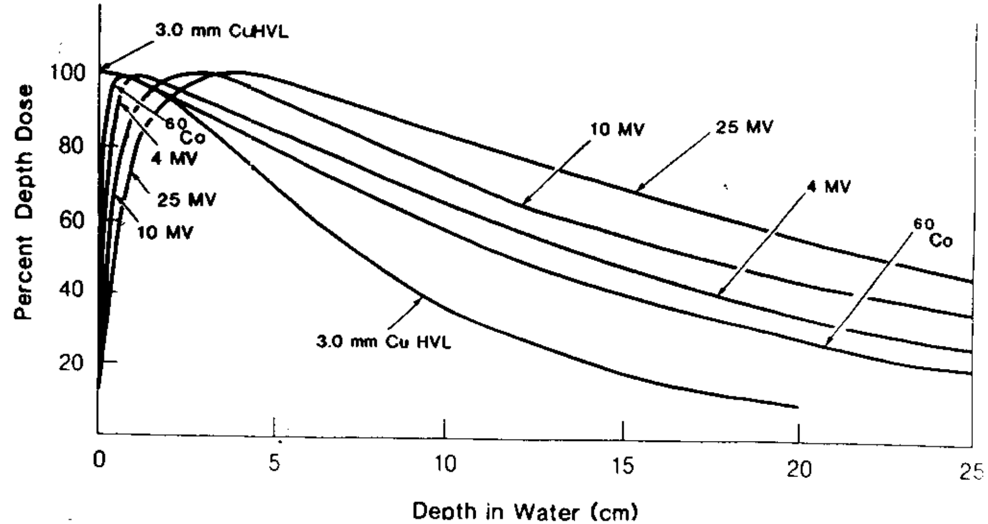


$E < 400\ kV_p$ $\quad d_o = 0$

E (high energy x-rays) $\quad d_o = d_m \quad$ (d_m is the position of the *peak absorbed dose*

PERCENTAGE DEPTH DOSE

DEPENDENCE ON BEAM QUALITY



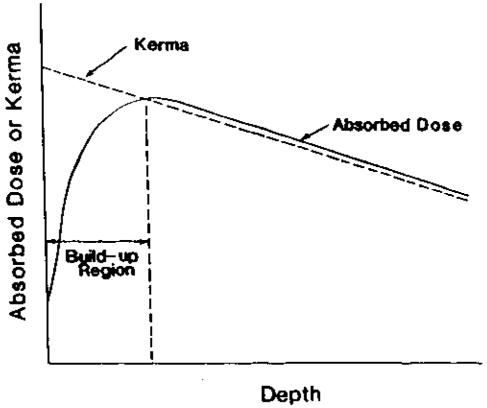
Beyond the depth of maximum dose

P increases with beam energy

Higher energy beams $\Rightarrow$ *Higher penetrating power*

PERCENTAGE DEPTH DOSE

DEPENDENCE ON DEPTH



Dose build-up region

↘

Exponential attenuation

SKIN-SPARING EFFECT

The dose build-up effect of the higher energy x-ray beams gives rise to the skin-sparing effect: higher doses can be delivered to the deep-seated tumors without exceeding the tolerance of the skin.

PERCENTAGE DEPTH DOSE

DEPENDENCE ON FIELD SIZE

*The **field size** increases*

↘

***Scattered-radiation contribution** increases*

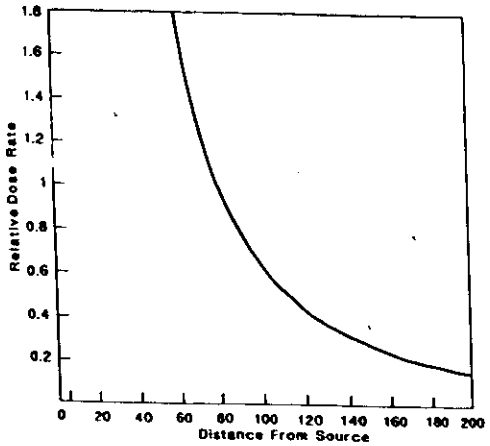
↘

***Percentage depth dose** increases*

The field size dependence of P is less pronounced for the higher energy beams than for the lower energy beams

PERCENTAGE DEPTH DOSE

DEPENDENCE ON SOURCE-SURFACE DISTANCE (SSD)



Plot of **relative dose rate** as <u>inverse-square-law function of distance</u> form a point source:

- *the **relative dose rate** at a point decreases with the SSD increase*
- *the **percent depth dose** increases with the SSD*

In clinical radiotherapy, the SSD is set at a distance which provides a compromise between the dose-rate and percent depth dose

ISODOSE CHART

In order to represent volumetric or planar variation in absorbed dose, distributions are depicted by means of ***isodose curves*** which are <u>lines passing through points of</u> <u>***equal dose***</u>.

***An isodose chart** for a given beam consists of **a family of isodose curves**, representing the variation in dose as a function of depth and transverse distance from the central axis.*

*NB: **the depth dose values of the curves are normalized either at the point of the maximum dose on the central axis or at a fixed distance along the central axis in the irradiated medium.***

ISODOSE CHART

SOME GENERAL PROPERTIES

1) The dose at any depth is greatest on the central axis and gradually decreases toward the edges of the beam

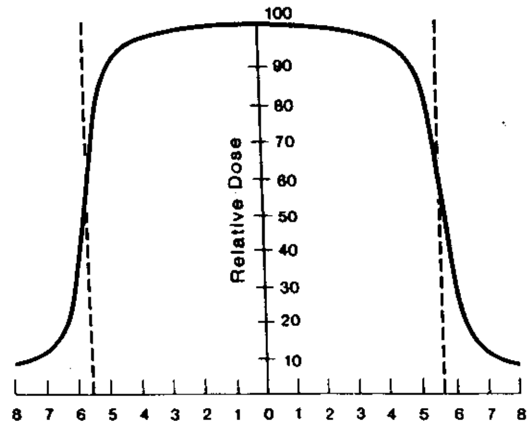
2) Near the edges of the beam (the penumbra region), the dose rate decreases rapidly as a function of lateral distance from the beam axis

3) The falloff of the beam is due not only to geometrical penumbra but also to reduced side scatter: the physical penumbra

4) Outside the geometric limits of the beam and penumbra, the dose variation is due to side scatter from the field and both the leakage and scatter from the collimator system

Depth Dose Profile

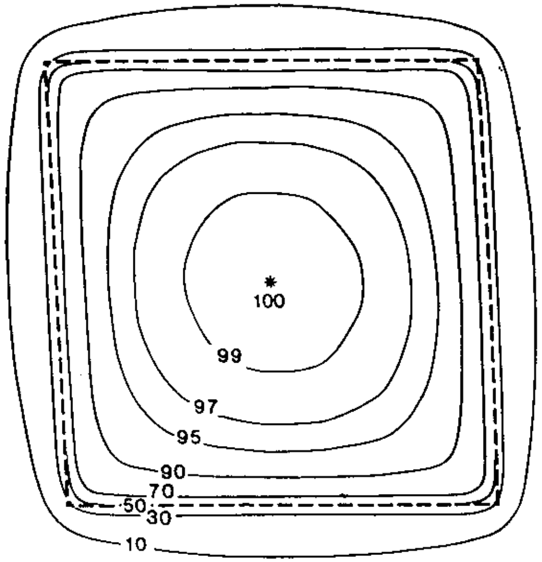
showing variation of dose across the field



*The **field size** is defined as the lateral distance between the 50% isodose lines at a reference depth*

Cross Sectional Isodose Distribution

showing variation of dose in a plane perpendicular to the central axis



- *Isodose values are <u>normalized to 100%</u> at the center of the field* •
- *The **physical penumbra** is defined as the lateral distance between 90% and 20% isodose lines at the depth of D_{max}* •

PARAMETERS OF ISODOSE CURVE

The parameters that affect the single beam isodose distribution are:

- ***beam quality***
- ***source size***
- ***beam collimation***
- ***field size***
- ***source surface distance***

BEAM QUALITY

The central axis depth dose distribution depends on the beam energy

⇓

The depth of a given isodose curve increases with beam quality

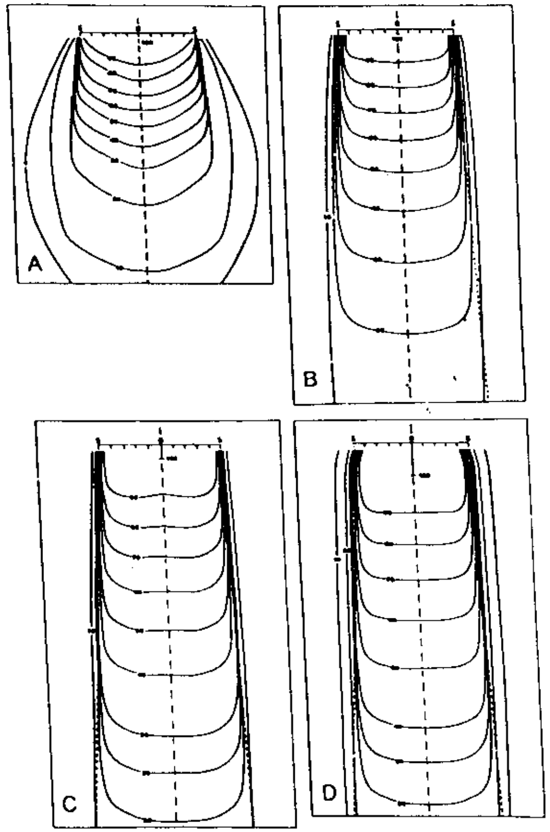
The absorbed dose in the medium outside the primary beam is greater for low energy beams than for those of higher energy (due to greater lateral scatter)

⇓

The isodose curves outside the field bulge out

BEAM QUALITY

Isodose distributions for different quality radiations



A: 200 kV_p, SSD=50 cm, field size=10x10 cm^2

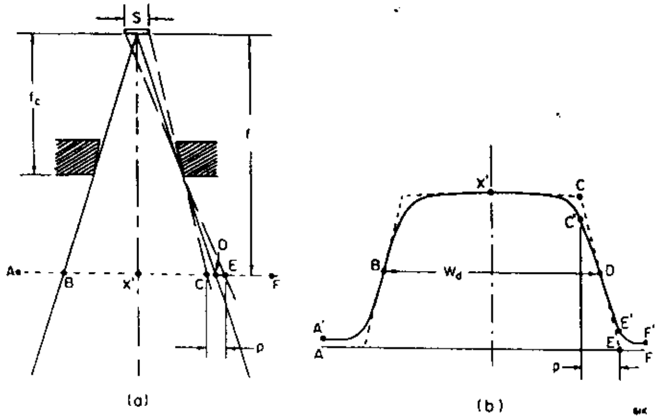
B: ^{60}Co, SSD=80 cm, field size=10x10 cm^2

C: 4 MeV x-rays, SSD=100 cm, field size=10x10 cm^2

D: 10 MeV x-rays, SSD=100 cm, field size=10x10 cm^2

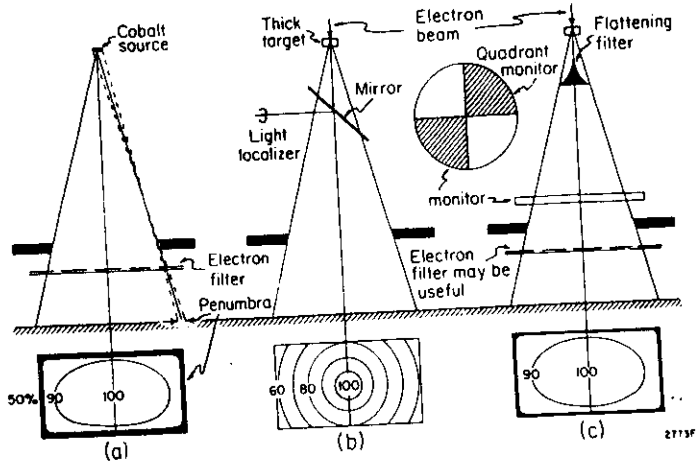
SOURCE SIZE, SSD, SDD: PENUMBRA EFFECT

*These three parameters affect the shape of isodose curves by virtue of the **geometric penumbra**.*



The SSD effects the percent depth dose and therefore the depth of the isodose curves

COLLIMATION AND FLATTENING FILTER



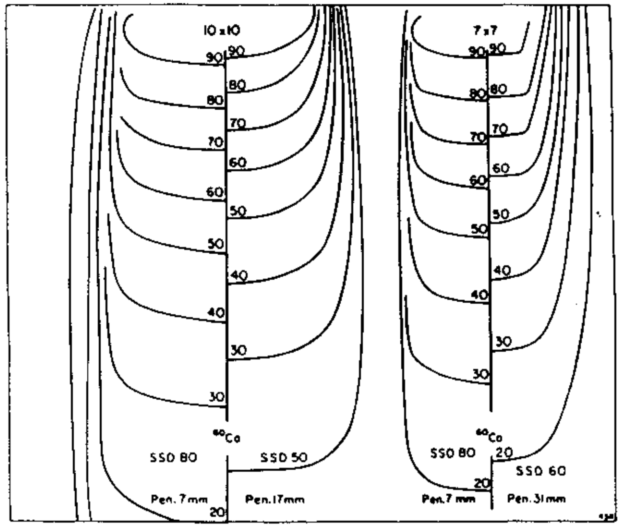
Collimator blocks: they give shape and size to the beam

Flattening Filter: it is used for megavoltage x-ray beams to make the beam intensity relatively uniform across the field, "flat"; the cross sectional variation of the filter thickness causes variation in beam quality across the field due to the selective hardening of the beam. **The change in quality causes the flatness change with depth.**

By careful design of the filter and accurate placement in the beam, it is possible to achieve flatness to within ± 3% of the central axis dose value at 10cm depth.

ISODOSE CURVES FOR ^{60}CO

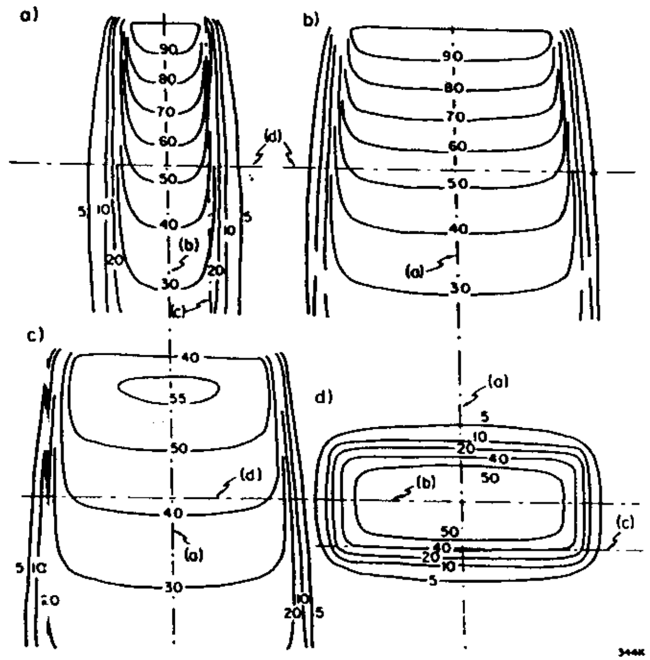
WITH DIFFERENT PENUMBRA



Left side: **10×10** cm field, **SSD=** 80 and 50 cm
Right side: 7×7 cm field, **SSD=** 80 and 60 cm

ISODOSE CURVES FOR ^{60}CO

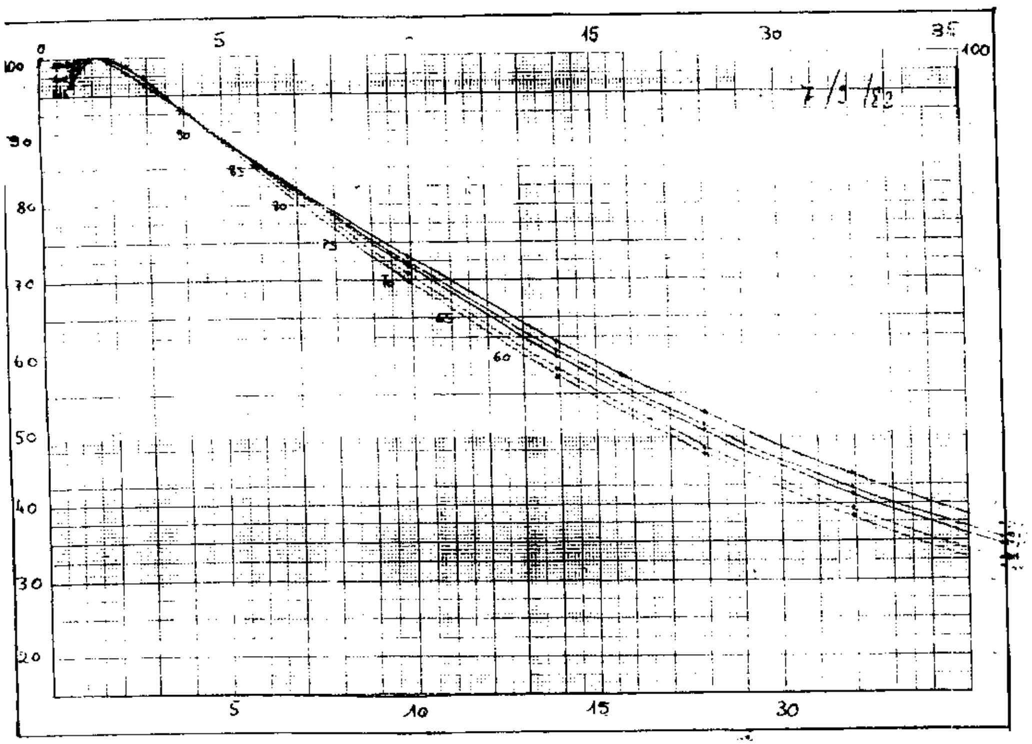
DISTRIBUTIONS IN 4 PLANES FOR A 6X15 BEAM



a) and **b)** isodose distribution in principal planes
c) isodose distribution in a plane parallel to b but 3 cm from it
d) isodose distribution in a cross section at 10-cm depth

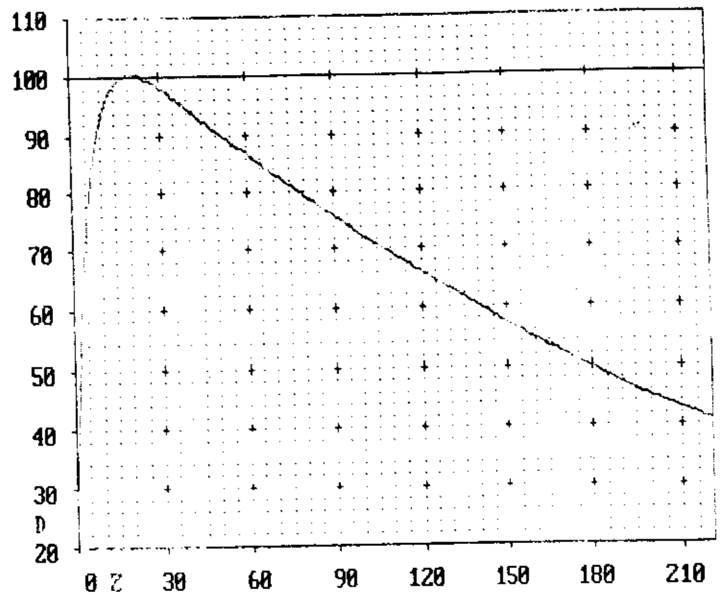
ISODOSE CURVES FOR
10 MeV x - rays

PERCENT DEPTH DOSE: DIFFERENT FIELD-SIZE BEAM



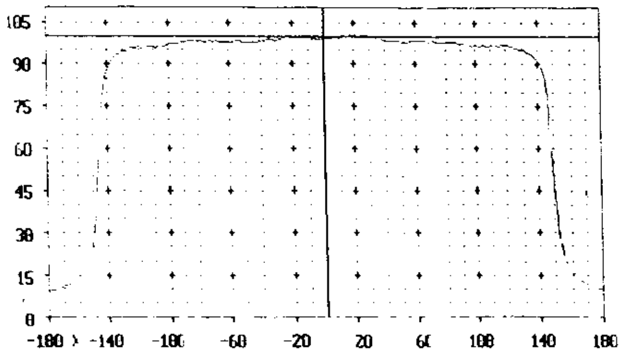
ISODOSE CURVES FOR
10 MeV x - rays

PERCENT DEPTH DOSE: 10 x 10 BEAM

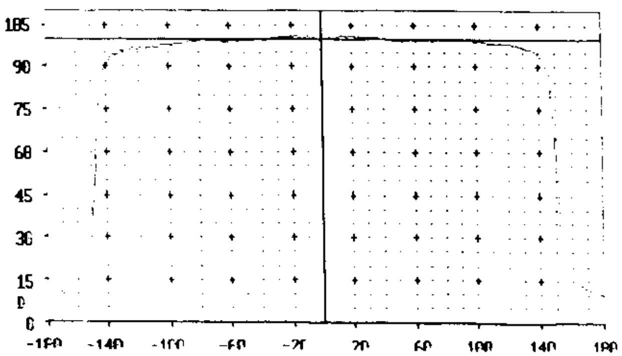


ISODOSE CURVES FOR
10 MeV x-rays

FIELD FLATNESS: 30 x 30 BEAM



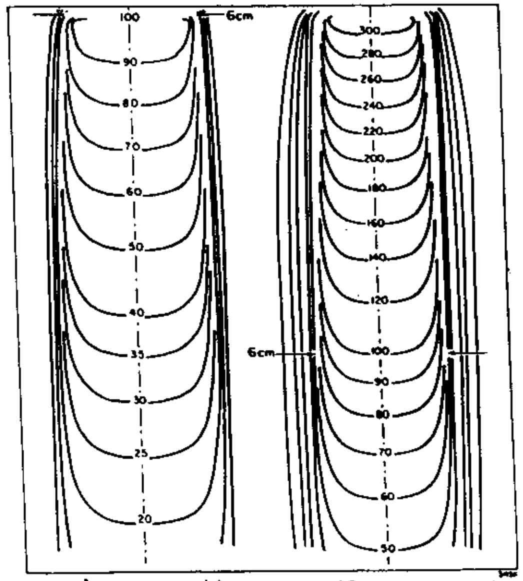
GT - AXIS



AB - AXIS

ISODOSE CURVES FOR ^{60}CO

FOR A 6X6 BEAM



LEFT SIDE: isodose pattern with percentage referred to B.U. (0.5 cm for ^{60}Co)

RIGHT SIDE: isodose pattern for ***isocentric use*** with percentage and field size referred to the axis of rotation

EQUIPMENT IN RADIOTHERAPY
HIGH ENERGY MACHINES

Since 1945, with the development of supervoltage machines and isotope teletherapy units, leading radiotherapists throughout the world demonstrated that many types of cancer can be better controlled using high energy radiations.

During these 50 years many types of machines have been developed or replaced by new and improved units. Further improvments in cure rate can be expected with refinements in present day machines and with the greater precision in their use.

Among the most in use high energy machines we will discuss in detail the ***linear accelerator*** and ***cobalt units*** and we compare some technical and physical properties.

Considerations in the design of high energy beams

*Firstly we remember some of the problems that must be solved in order that **high energy x-ray and electron beams may be used correctly in the treatment of a patient**.*

Depth Dose Curve

Electron contamination

Field flatness curve

Penumbra

Light localizer

Electron filter

Field flatness curve

It shows the behavour of the dose inside the patient perpendicularly to the beam axis.

The curve produced by a **cobalt source** is uniform from the center of the field to the edge of the beam: such a source emits uniformly photons in each directions about the axis.
The distribution of photons, produced by a **linac or a linear accelerator**, is circular and so quite useless for radiotherapy. This problem can be overcome by placing a *flattening filter* in the beam, will yield a dose distribution essentially constant.

Penumbra

It erises from the finite sizes of the beam source.

This problem is more relavant in cobalt units where the sources are quite large (about 3 cm in diameter).

Light localizer

It is required to show the technician the area of the field that will be treated.

The optical device is usually accomplished using a light and mirror combination: naturally the light field and the radiation field should correspond and periodically should be checked.

Electron contamination

It produces the initial rising part of the depth dose curve, called the build up region.

<u>The B.U. region is very sensitive to the design of the beam-defining collimating system: the electron contamination can be partially overcome with a filter.</u>

Electron filter

It is required to remove low energy electron and photon contamination, in particular for large fields.

The incorporation of all these features at the same time in a machine may create engineering problems, involving the sliding of various components in and out of the beam for the set-up and treatment sequence.

THE LINEAR ACCELERATOR - LINAC

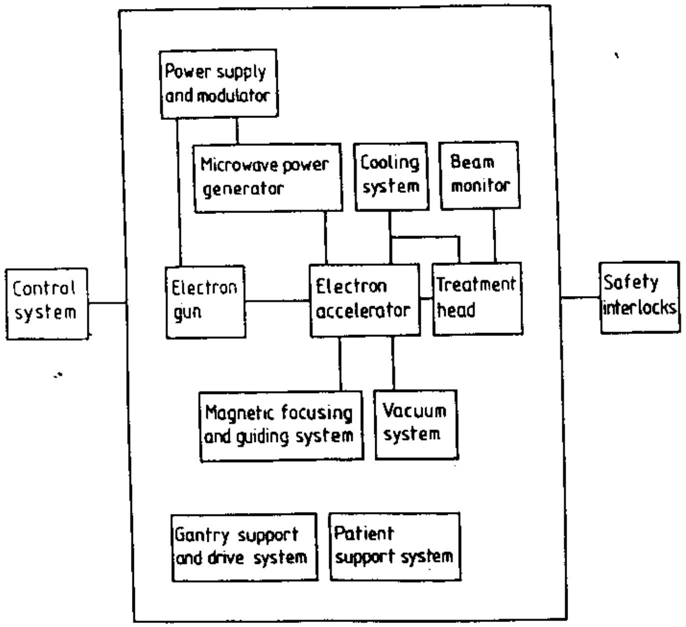
In strict terminology the term "linear accelerator" only applies to that part of the system where electrons are accelerated up to the required energy.

*In generai usage, this term is used as a description of the **whole system used for radioterapy treatments**.*

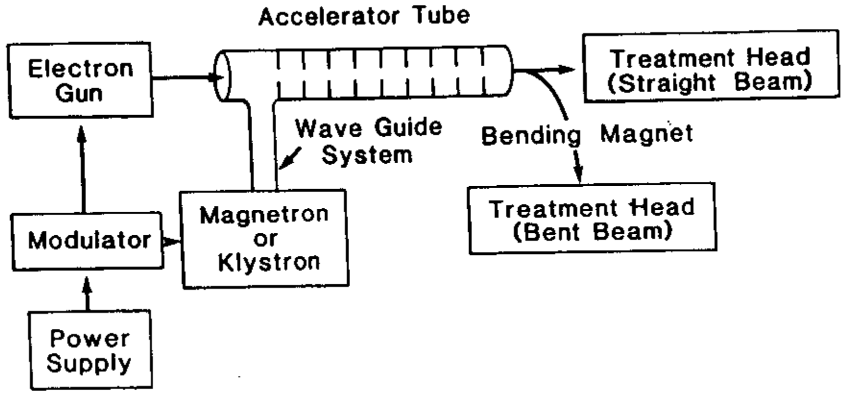
It is a device which uses high frequency electromagnetic waves to accelerate charged particles such as electrons to high energies through a linear tube.

The high energy electron beam itself can be used for treating superficial tumors or it can be made to strike a target to produce x-rays for treating deep seated tumors.

BLOCK DIAGRAM SHOWING THE COMPONENT SYSTEM OF A LINEAR ACCELERATOR



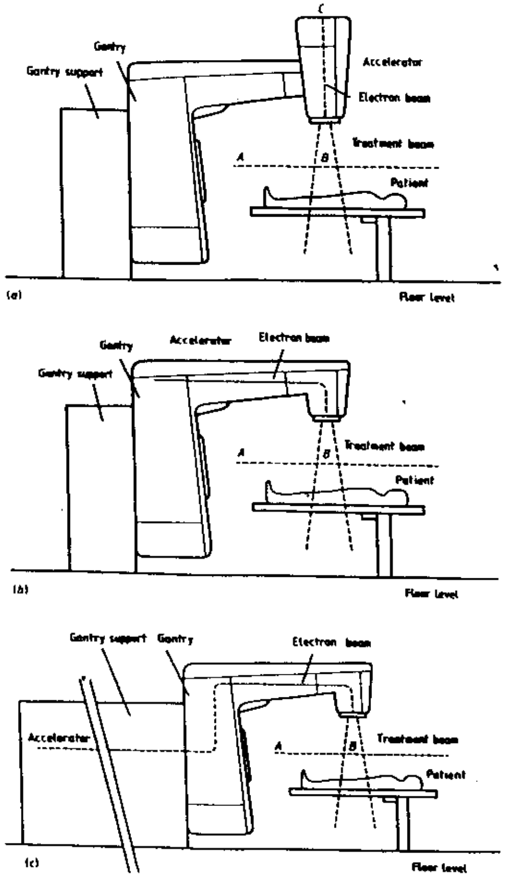
ELECTRON ACCELERATOR



*The **electron accelerator** is a waveguide structure which is energised at microwave frequency, most commonly at 3000 MHz . The **microwave radiation** is supplied in short pulses, a few µsecs long, and this is generated by supplying high voltage pulses of about 50 kV from the modulator to the microwave generator (**magnetron valve**). The **electron gun** is also pulsed so that high velocity electrons are injected into the accelerating waveguide at the same time as it is energised.*

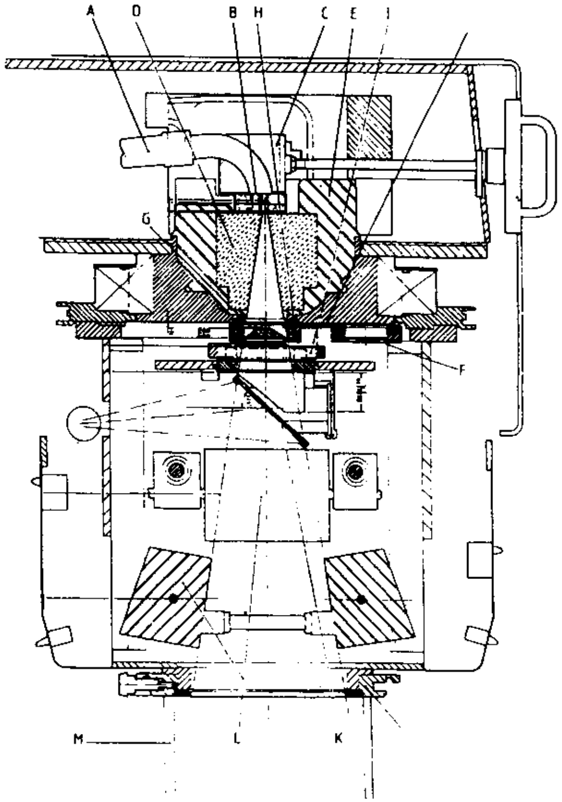
The electron gun and accelerating waveguide system have to be evacuated to a pressure such that the mean freee path of electrons between atomic collisions is long compared with the electron path through the system.

DIAGRAM SHOWING THREE METHODS OF MOUNTING THE ELECTRON ACCELERATOR



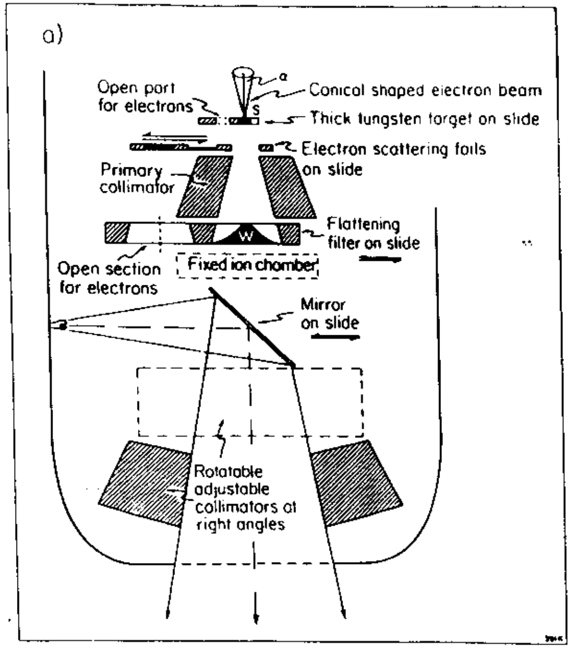
A) electron accelerator in line with treatment beam
B) electron accelerator in parallel to axis of rotation of gantry
C) electron acceleratoı in line with axis of rotation of the gantry

TREATMENT HEAD



*The accelerating electrons can be focused back on their straight path by the use of a coaxial magnetic field (**beam bending magnet C**).*
*The accelerated electrons enter in the "**treatment head**", through a window (B) and will generate either an x-ray or an electron beam.*
The treatment head consists of a thick shell of high density shielding material (such as lead, tungsten or lead-tungsten alloy). Lead blocks (D e E) reduce the radiation leakage.

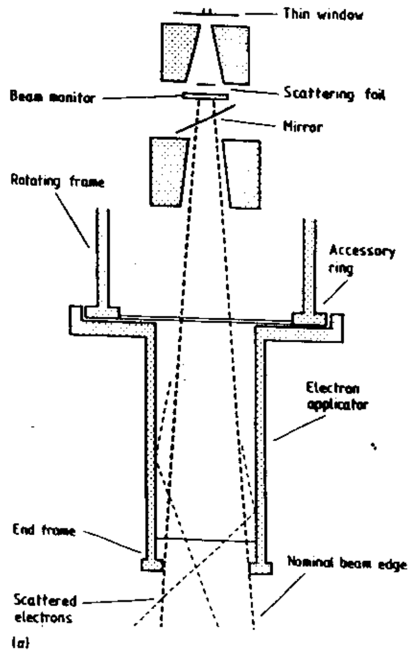
X - RAY BEAM THERAPY



In a x-ray generator, the treatment head will contain:

***x-ray target**: thick enough to stop the electrons and made of tungsten for the best compromise between good x-ray output and beam penetration*
***filtering system:** generally made of tungsten*
***beam monitor detectors:** a complex set of ionisation chambers which monitor dose rate, dose and dose distribution in the field*
***beam defining system:** movable beam defining collimators, removable wedge filters*

ELECTRON BEAM THERAPY

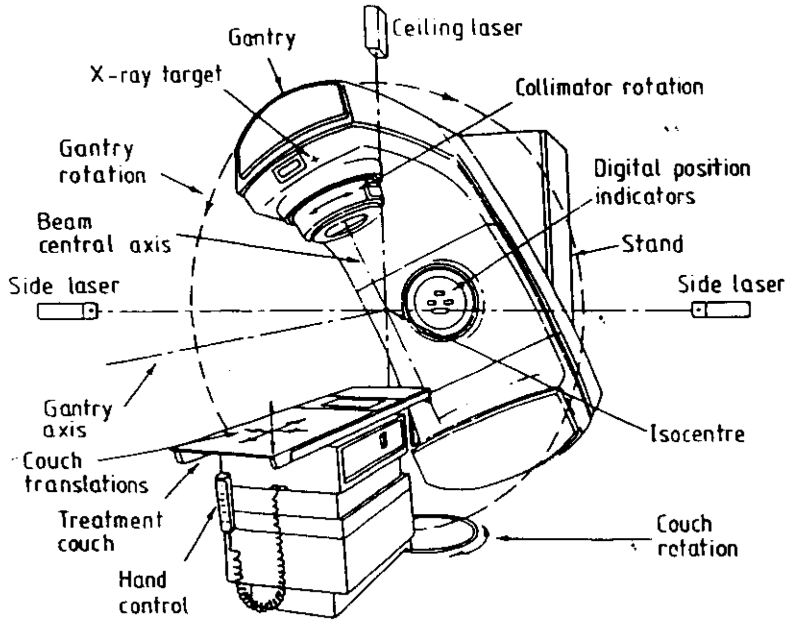


Electron beam applicator mounted on accessory ring

*If the **electron beam** is to used for treatment, it will emerge, as a narrow pencil of about 3 mm diameter, from the vacuum system through a thin window into the treatment head.*
Then it is again monitored and made to strike an electron scattering foil in order to spread the beam as well as get a uniform electron fluence across the treatment field.
*The **scattering coil** consists of a thin metallic foil, usually of lead, such that most of the electrons are scattered instead of suffering bremsstrahlung.*
*The electron beam is defined using **aluminium applicators**, which thickness is about one third of the electron range.*

*Some machines have the **dual purpose treatment heads for x-ray and electrons**: elaborate mechanical and electrical arrangements are required to allow the changeover from one mode to the other.*

***Major mechanical elements**: gantry support and drive systems, which serve to position the radiation source with respect to the patient, and the patient support system which will place the patient in the desired position.*



Schematic view of the treatment unit emphasing the geometric relationship of the linac and treatment couch motions

COBALT 60 UNITS

*Of all the radionuclides used as sources of γ rays for teletherapy, such as Radium-226, Cesium-137 and Cobalt-60, ^{60}Co has proved to be the most suitable for external beam radiotherapy, since higher possible **specific activity** (Curie per gram), greater **radiation output** and higher **photon energy**.*

Teletherapy source characteristics

Radionuclide	Half-life (years)	γ Ray energy (MeV)	Γ Value* $\left(\frac{Rm^2}{Ci-h}\right)$	Specific activity achieved in practice (Ci/g)
Radium-226 (filtered by 0.5 mm Pt)	1622	0.83 (avg.)	0.825	~0.98
Cesium-137	30.0	0.66	0.326	~50
Cobalt-60	5.26	1.17, 1.33	1.30	~200

* Exposure rate constant (Γ) is discussed in Section 8.5. The higher the Γ value, the greater will be the exposure rate or output per curie of the teletherapy source.

The ^{60}Co source is produced by irradiating ordinary stable ^{59}Co with neutrons in a reactor:

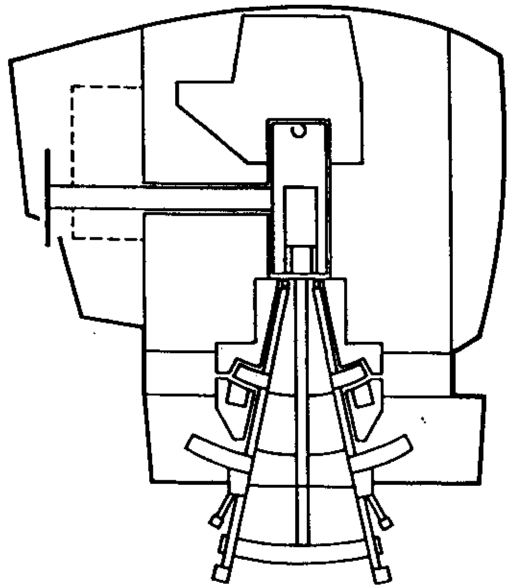
$$^{59}Co\ (n,\gamma)\ ^{60}Co$$

The ^{60}Co source, in the form of a solid cylinder, discs, is contained inside a stainless steel capsule and sealed by welding.
The ^{60}Co source decays to ^{60}Ni with the emission of β particles (E_{max}=0.32 MeV) and two photons per disintagration of energies 1.17 and 1.33 MeV.

The radiation spectrum is characterized also by lower γ ray produced by the interaction of primary γ radiation with the source itself, the surrouding capsule, the source housing and the collimator system.

TREATMENT HEAD

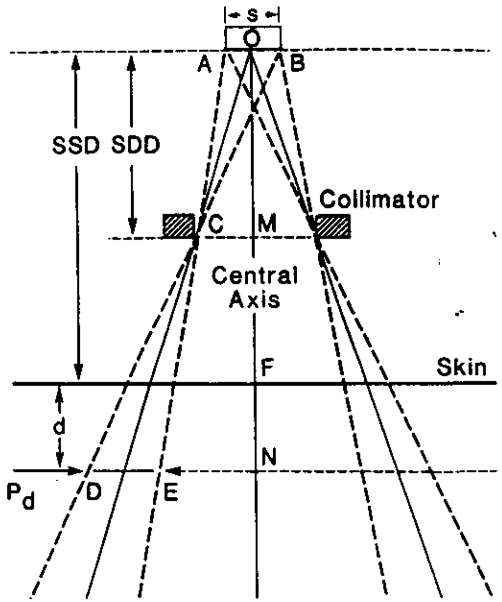
Siemens teletherapy machines
Machine types:
Gammatron R, 1, 2 and 3 – (Square draw)
Gammatron S – (Round draw)



The housing for the source is called the "sourcehead".

The source head consists of a steel shell filled with lead for shielding purposes and a device for bringing the source in front of an opening in the head from which the useful beam emerges.

COLLIMATOR SYSTEM: PENUMBRA



A collimator system is designed to vary the size and the shape of the beam to meet the individual requirements.

The presence of the collimating blocks produces the ***"trasmission penumbra"****.*
The source diameter, instead, produce the ***" geometric penumbra"****: it is dependent of SSD and depth.*