Linear Accelerators in Radiotherapy: A Brief Introduction to Physical and Technical Design Features.

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ABSTRACT

The most widely used instrument in radiotherapy treatment is the linear accelerator. Since the first introduction into radiotherapy in 1952 a tremendous technical and physical development took place. It started with a stationary röntgen-like megavoltage equipment and is presently a web-like network including a computerized dynamically operating multi-modality linear accelerator. New components, such as asymmetrical jaws, multi-leaf collimator, virtual wedge, double-photon and multi-electron capability, high dose rates, automatic set-up and dynamical beam delivery, etc., improve machine handling and treatment of patients significantly. At present only three vendors offering medical linear accelerators with quite different technical solutions to produce useful electron and photon beams.

KEYWORDS:

radiotherapy, linear accelerator,

INTRODUCTION

Since the early 50th of the last century linear accelerators are in use in radiotherapy for treating cancer patients. The first examples of this type were quite cumbersome in use because of size, limited maneuverability and susceptibility to interference, especially the electronic components as well as vacuum pumps.

In this early times also Co-60-units, betatrons and Van-de-Graaf accelerators were in clinical operation. But technical and physical developments and clinical demands forced the broad introduction of microwave linear accelerators into radiotherapy practice (Tab. 1).

Item	Invention or initiation	Country	First installation
Microwave cavities	1934	U.S.	-
Klystron	1937	U.S.	-
Magnetron	1939	U.K.	-
Microwave linac	1945-47	U.K./U.S.	-
Multi-megawatt klystron	1947-52	U.S.	-
Radiotherapy linacs:			
Stationary X-ray	1948	U.K.	1952
Orientable X-ray	1949	U.K.	1953
Orientable X-ray	1950	U.S.	1955
Stationary electrons	1952	U.S.	1955
360° isocentric X-ray	1958	U.S.	1962
Dual photon energies plus electrons	-	U.K.	1965
Double achr. bent beam, dual X-ray energy plus electrons	-	U.S.	1979
Sputter ion pump (getter p.)	1956	U.S.	-
SW acc. guide	1958	U.S.	-
Double achrom. bend. magn.	1963	U.S.	-
In-line 6 MeV-X-ray	-	U.S.	1977
Multi-leaf-collimator	1975	J/S	1985

Tab. 1 *Brief chronologically technical developments.*

The main components of a medical linear accelerator are: accelerator guide, cathode, modulator, magnetron or klystron, vacuum and cooling system, bending magnet, flattening filter and scattering foils, beam shaping devices, monitor chamber, auxiliary electronic cabinet, power supplies, and beam steering system.

For easier steering, controlling and hand ling of the system "linear accelerator" and for recording and verifying patient treatment data, the whole system is computerized and part of a network containing all relevant radiotherapy subsystems. In total the medical linear accelerator is a highly sophisticated and in general a reliable tool to treat cancer patients in the low, medium and high photon and electron energy mode.

CLINICAL REQUIREMENTS.

The main aim of curative radiotherapy is to devitalize tumor cells without excessive damage of surrounding tissue. Taking into account human anatomy as well as radiobiological insights, these basic goals require specific high precision of linear accelerator performance, ease of precise patient set-up and thorough quality assurance procedures (Tab. 2).

Item	Major criteria
Precise delivered dose throughout target volume	Flatness and stability of field vs. gantry angle and field size Stability of penetrative quality
Precise dimensions of target volume	Spatial precision of machine and beam
Minimal dose to normal tissue	Slope of fall-off of electron depth dose Sharpness of dose profile shoulder Scatter from beam modifiers
Wide variety of radiation modality	Low and high X-ray energies Low and high electron energies Small to large field sizes
Reliability	Minimal unscheduled down time
Moderate time to irradiate	High dose rate
Patient safety	Radiation injury avoidance Mechanical injury avoidance

Tab. 2 *Clinical requirements and medical linear accelerator design features.*

The linear accelerator performance includes precise position, orientation and size of treatment fields, wide variety of radiation modalities, compactness of machine, high dose rate with large fields, dose precision, treatment beam stability, uniform X-ray treatment beams with minimal contamination, uniform electron treatment beams with minimal contamination, energy stability, stable parameters during the initial seconds of treatment, and during rotation of gantry. Beside this clinical requirements special attention should be drawn to internal machine security philosophy, service, maintenance and repairing costs and possibilities.

BRIEF DESCRIPTION OF BASIC COMPONENTS.

A radiotherapy linear accelerator consists of several thousand of single parts and components. But fortunately it is not necessary to understand how each part works in detail in order to have a basic understanding of how an accelerator works. The main components can be combined to 8 groups: accelerator guide including gun, vacuum system and cooling system; RF-system; bending magnet; treatment head with target, scattering foils, monitor chamber, and collimator system; power supplies and auxiliary electronics; patien support assembly; control consoles; control computer (Fig. 1).

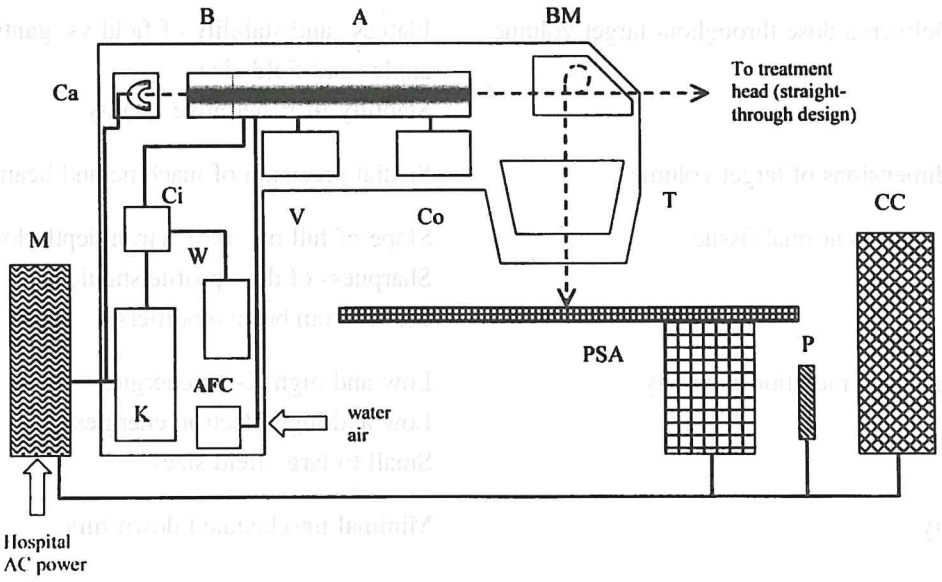


Fig. 1 *Main components of a linear electron accelerator*

M: Modulator; K: Klystron or magnetron; AFC: Automatic frequency control; Ci: Circulator; W: Water load; Ca: Cathode; B: Buncher section of acc. guide; A: Accelerator section of acc. guide; V: Vacuum system; Co: Cooling system; BM: Bending magnet; T: Treatment head; PSA: Patient support assembly (treatment couch); P: Hand pendant; CC: Control console.

The major component of a microwave linear accelerator is the so called waveguide. The waveguide is composed of a number of specially shaped copper microwave resonant cavities, that have been brazed together to form a single structure. The total length will vary from 30 cm to 250 cm, depending on the final electron energy and the type of structure utilized. Two different modes of accelerating the electrons are in use: traveling wave (TW) and standing wave (SW). The microwave for accelerating the electrons along the central axis of the waveguide structure is produced either by a magnetron or a klystron, which needs a separate oscillator. This RF-power sources operate at about 3 GHz, which is equivalent to a wavelength of 10 cm and is called S-band. This value characterizes also the length of the neighbouring cavities of the waveguide. These cavities have to couple and distribute microwave power between adjacent cavities, and provide an E-field with suitable axial distribution for accelerating electrons. Electrons injected along the axis of the structure would be accelerated by the moving E-field pattern. The phase velocity of wave exceeds that of light and is therefore unsuitable for accelerating charged particles. The wave is slowed down by inserting washerlike disks into the waveguide, so that the wave stays in step with the accelerated electrons. The shape of the resonant cavities, the fabrica tion technique, the method of injection of the electrons and the RF-power into the wave guide varies depending on manufacturer, but all structures are of two types: travelling wave or standing wave. Both show some common characteristics: The waveguide has two different regions, the buncher section and the accelerating sections. The continuously injected electrons are preaccelerated (≈18...50 keV) and grouped or bunched in the first section, and are accelerated up to the final energy in the second section. The energy E of the electrons emerging from an accelerator guide is given by the load line equation:

$$E = -A \cdot i + B \cdot (P \cdot L \cdot Z)^{1/2}$$

A, B	constants,
i	beam current,
P	RF power,
L	length of accelerating structure,
Z	impedance of structure.

Because of special internal structure features the Z-values of SW-type accelerators are about twice as high as in TW-structures: 100 ... 120 MΩ/m. In a TW-structure the RF-power is fed into the gun end of the accelerator guide and travels down to the end, where any excess power is dissipated in an absorbing load or recirculated for reuse. The electric field maxima are traveling down the structure, the electrons are riding on the crest of the wave and both are passing the same aperture holes that connect one resonant cavity to the next. This is a design conflict, because electrons and RF need different diameter for optimal transport.

In a SW-type accelerator the internal structure allows the RF to be reflected at both ends. This results in establishing nodal points and electric field maxima. Resonance cavities

containing the nodal points show a net electric field of zero, and do not participate in accelerat ing the electrons. Therefore these cavities can be moved off the centerline and thereby reducing the length of the structure. Off axis cavities and accelerating cavities can be optimized in shape for transport of RF power or accelerating electrons. In total a SW-guide is much more compact, shows a more efficient use of RF power, and in excess a 30 cm SW-guide allows electron energies of 6 MeV.

Both, the RF-power source and the gun or cathode are pulsed by the modulator: high voltage pulses at a rate of a few hundreds pulses per second synchronously power the cathode and the klystron or magnetron. These devices employ a number of RF cavities either in a circle (magnetron) or in a straight line (klystron) to produce pulsed high power RF, 2.5 ... 7.5 MW, which is fed via hollow metallic tube into the accelerator guide. Because this power can't be delivered continuously the RF-devices are pulsed: pulse width ca. 1 ... 5 μs; pulse interval 1 ... 5 ms.

The electron gun can be either a diode or a triode, in which the grid can be used to control the injected electron current, and hence, the dose rate. In commercially available linear accelerators the gun constructions are of different types: simple tungsten spirals, or bariumoxide coated concave surface emission type; directly or indirectly heated. The directly heated tungsten spiral tends to change mechanical form and emission characteristics during life time, which influences treatment beam parameter, such as symmetry or electron energy. As a consequence QA measurements are more time consuming than on machines equipped with other types of cathodes. In commercially available accelerators the gun and also the vacuum pumps are attached permanently to the accelerator guide, or are demountable for easy and cost-saving replacement.

In early linear accelerators the vacuum pumps have been of oil-diffusion type with the disadvantage of restrict rotation capability in clinical routine. Since the early 60th of the last century all linear accelerators are equipped with sputter ion pumps to maintain pressures lower than 10^{-7} hPa to prevent arcing within the RF source and the accelerating guide, and to increase cathode life time.

For a stable output of a linear accelerator in terms of energy, dose rate, symmetry and flatness two additional auxiliary systems are necessary: the automatic frequency control and the cooling system. The resonance frequency of approximately 3 GHz should coincide exactly with the dimensions of the resonant cavities, i.e. the temperature stability along the guide should be better than $\pm$ 0.5°C and the frequency excursions limited to $\pm$ 20 kHz.

To reduce mechanical and temperature stress to all parts of the accelerator it is recommended to keep the internal machine temperature constant, for instance at 40°C for day and night.

To vary the output energy of the electrons various techniques have been developed (Tab. 3). Especially in photon mode a high beam current is important to realize an accept

able clinical dose rate of at least 2 Gy/min, also when wedges are in use (Tab. 4). As a rule of thumb the following formula can be taken to estimate dose rate at isocenter distance for an unflattened field:

	Method	Structure	Advantage	Disadvantage
1.	Vary rf power and/or beam loading	single-section	simple	rapid spectrum deterioration
2.	Detune rf signal and/or accelerator cavity	single-section	simple	unstable and limited range
3.	Combination of two accelerators with phase shifter and attenuator	two-section	energy variable in wide range	complex
4.	Combination of single accelerator with 180^{o} bend magnet	single-section but double-path	high energy gain per given rf power	complex and limited range
5.	Energy switch	single-section	wide energy range without degradation of spectrum	add. component

Tab. 3 *Various techniques for varying the output electron energy.*

X - rays				Electrons			
E [MeV-X]	I_B (av) [μA]	T_F [%]	DR [cGy/min]	E [MeV]	I_B (av) [nA]	F_{Sc} [mil]	DR [cGy/min]
4	200	45	200				
6	100	35	400	6	100	3 Ta	500
10	70	30	500	9	97	8 Pb	500
15	50	25	500	12	67	+	500
18	30	18	500	16	42	7 Al	500
						button	
25	20	10	500	20	30		500

Tab. 4 *Linear accelerator operating parameters.*
E : Energy; I_B (av) : average beam current; T_F : Filter transmission; DR : Dose rate; F_{Sc} : Scatter foil.

$D_x = k \cdot i_{AV} \cdot V^n$ [cGy/min],

valid for 4 MeV-X ... 35 MeV-X : $k = 0.07$; $3 \geq n \geq 2.5$,

i_{AV} : average beam current [μA] ,

V : electron beam energy [MeV] .

The dose rate estimation in electron mode is done by the formula:

$D_e = 1.1 \cdot i_{AV} \cdot 10^7/A$ [cGy/min] ,

i_{AV} : average electron beam current [μA] ,

A : field area [cm^2] .

It is obvious, that the beam current has to be increased by 10^3 when switching from electron to photon mode. Only a narrow electron energy spectra guarantees a sufficient transmission rate through the bending system.

All techniques to vary electron energies show advantages, but also serious disadvantages, especially the degradation of output electron spectrum when switching to the low photon beam mode. The reason is the design of the buncher section optimized for only one specific electron energy (Fig. 2; Fig. 3) and the physics of producing bremsstrahlung:

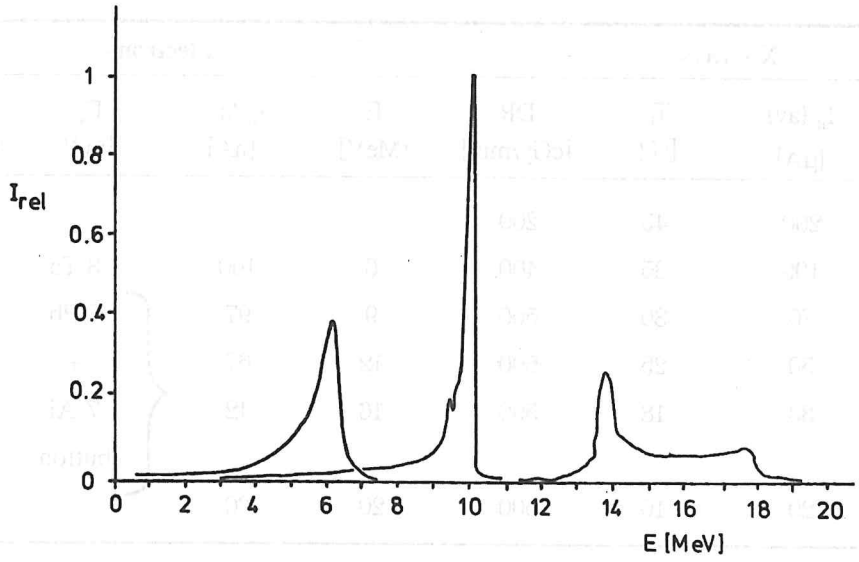


Fig. 2 Deterioration of energy spectrum by beam loading and/or power variation. Beam loading: (left) heavy; (centre) optimal; (right) light.

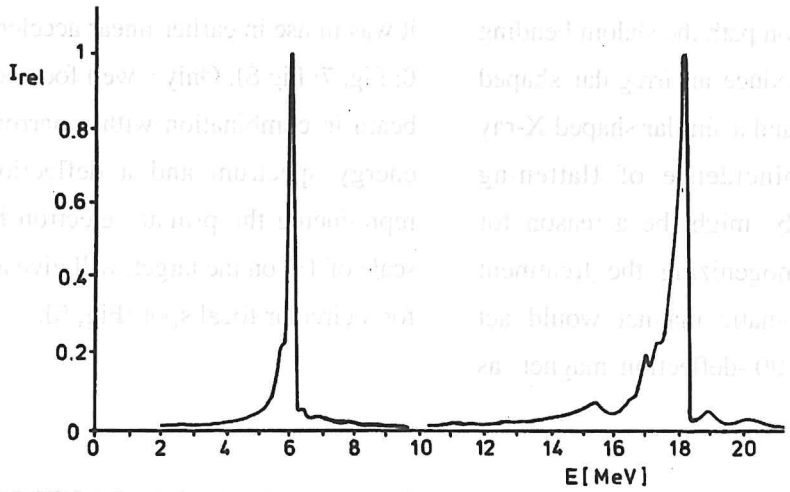


Fig. 3 *Energy spectra of electron beam in microwave switched 6 and 18 MeV-X-ray modes.*

$$I \sim C \cdot E^{2.7},$$

I relative intensity,

C constant,

E energy of electrons.

In electron mode this degradation of spectrum is no problem, because of the low beam current: mechanical slits within the bending system are normally set for a transmission of ±3% of the mean electron energy, and the metal used is tungsten with a melting point of about 3410°C. Additionally the slits as well as the target for X-ray production are part of the cooling system.

As all dual photon energy accelerators are optimized to deliver a high energy photon beam (10 ... 18 MeV-X), the main problem is the handling of the high beam current for the low energy photon beam (4 ... 6 MeV-X). The solution (5) in Tab. 3 ("energy switch") utilizes an air-pressure driven mechanical post to change coupling between adjacent accelerator cavities. The electric field in the buncher section is not affected and hence the narrow electron energy spectrum is capable for low and high energy X-ray beams, as well as a high dose rate for both energies.

For bending of the electron beam two different types of magnets are in use in commercially available linear accelerators: 270°-deflection magnet and slalom magnet. Both systems are achromatic. This means, that regardless of their energy over a certain range the distribution and orientation of rays in the electron beam bundle is the same after the bend as before. In combination with a not well designed focussing of the electron beam

along the acceleration path the slalom bending system tends to produce an irregular shaped focal spot (Fig. 4) and a similar shaped X-ray lobe. The non-coincidence of flattening filter and x-ray lobe might be a reason for difficulties in homogenizing the treatment field. A non-achromatic magnet would act like a prism, or a 90°-deflection magnet, as it was in use in earlier linear accelerators (Fig. 6; Fig. 7; Fig 8). Only a well focussed electron beam in combination with a narrow electron energy spectrum and a deflection magnet reproducing the primary electron beam on a scale of 1:1 on the target, will give a guarantee for a circular focal spot (Fig. 5).

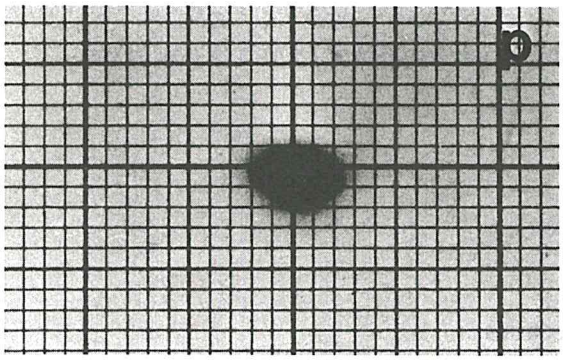


Fig. 4 *Focal spot : irregular shape.*

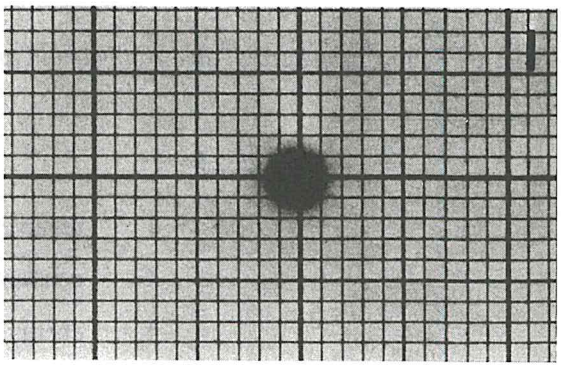


Fig. 5 *Focal spot : circular shape.*

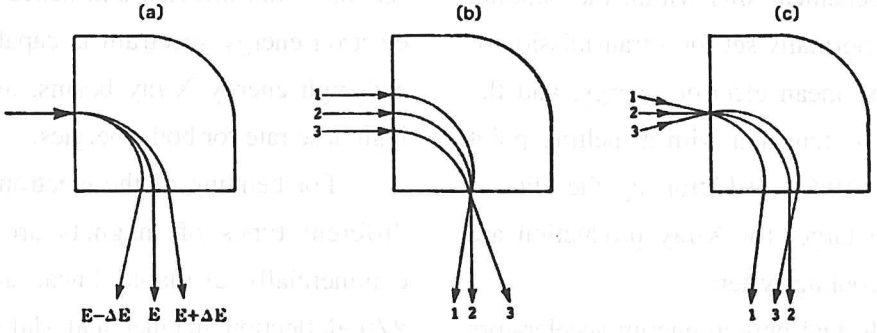


Fig. 6 *90° -bending magnet system.*

a. Energy spread; b. Radial displacement; c. Radial divergence.

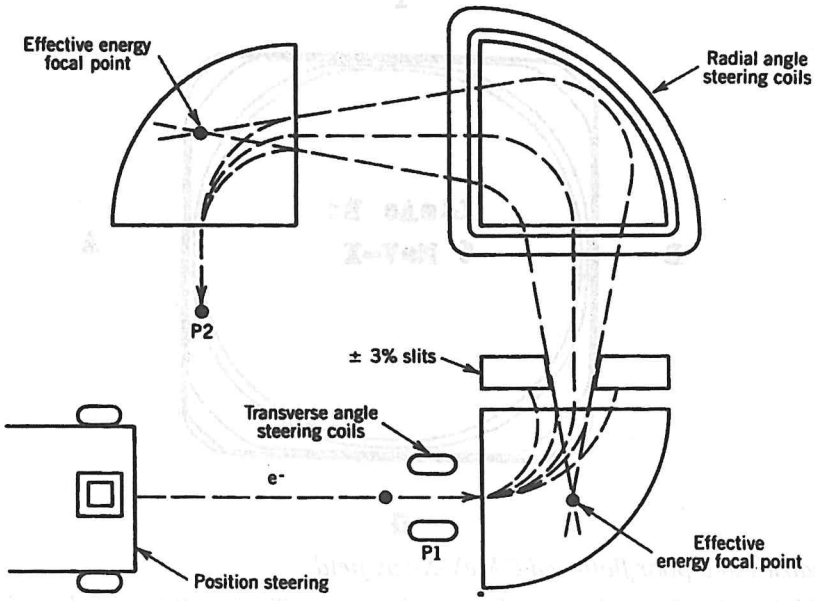


Fig. 7 *270° -bending magnet system.*

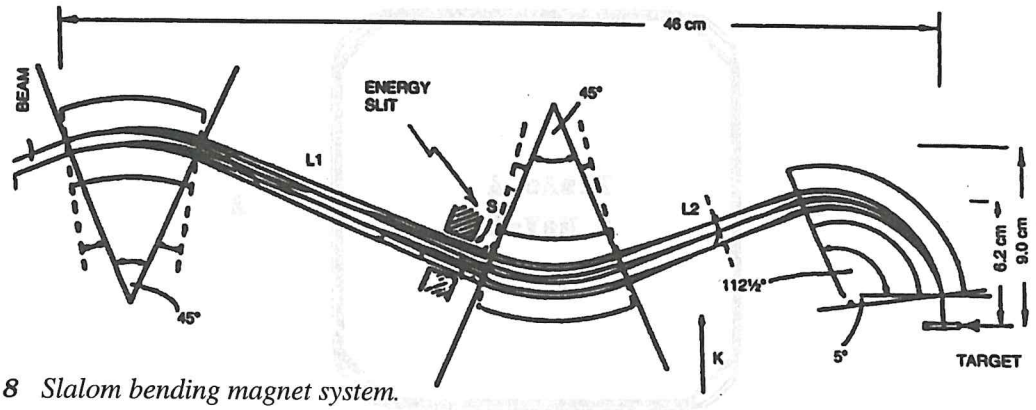


Fig. 8 *Slalom bending magnet system.*

When X-rays are produced in the target, they emerge in a forward directed lobe whose intensity distribution must be flattened for clinical use by a circularly symmetrical-shaped metal absorber, called flattening filter. Height and diameter of such a filter depend on material, beam energy and field area to be flattened. The absorbance varies from about 4 to 10 for X-ray energies of 6 to 18 MeV. About 80 % of the maximum field dimensions along the major axis and about 60 % along the diagonals have to be flattened and should fulfill the symmetry requirements stated in national and international recommendations. A closer look at maximum field size isodoses reveals large differences in flattening capabilities of different vendors of medical linear accelerators (Fig. 9; Fig. 10).

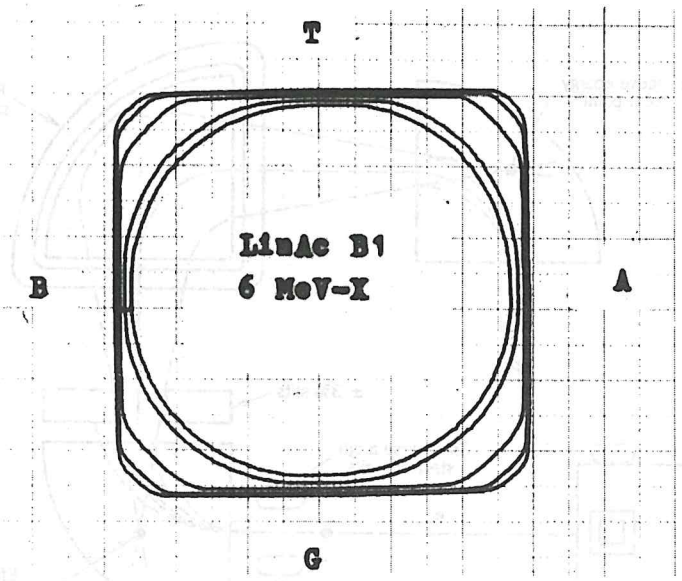


Fig. 9 *Isodoses in a poor flattened 6 MeV-X-ray field.*
Field size: 40 x 40 cm^2; Diameter of projected primary collimator: 49.5 cm; Isodoses: 90%; 80%; 50%; 20%; 10%; A, B, G, T: left-, right-, gun- and target direction.

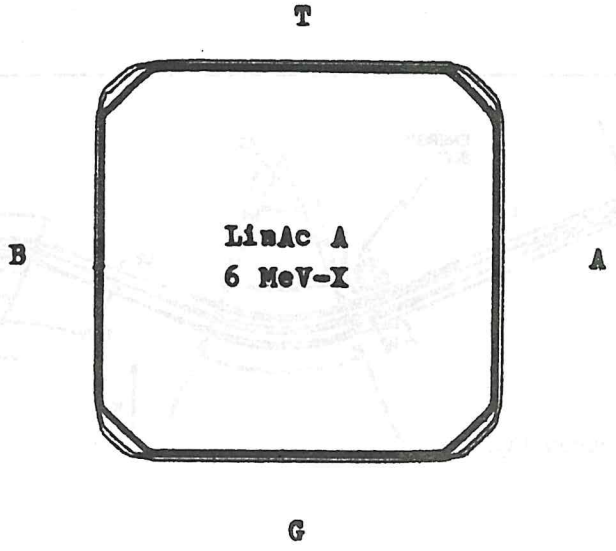


Fig. 10 *Isodoses in a well flattened 6 MeV-X-ray field*
Explanation: see Fig. 9

Electrons used directly for treatment are scattered by thin metallic foils and finally limited in spread by special applicators defining the clinical field size.

For field defining in photon mode the linear accelerator has two sets of independently adjustable jaws, that allow rectangular, square and asymmetrical field sizes with different overtravel across centerline for the X- and Y-collimator. This collimator should typically absorb more than 99.5% of the radiation to protect the patient from unwanted irradiation outside of the target volume.

To create an individually shaped treatment field linear accelerators are equipped with a multi-leaf collimator (MLC) consisting of 80 individually movable jaws with a projected width at isocenter of 1.0 cm. Special MLCs are available for small field treatments.

Three vendors offering three types of MLCs: replacement of the upper (1) or the lower jaws (2) or as an add-on-system (3). The practical use of MLCs and the introduction into daily routine is still under development: static use to replace blocks and apply sub-fields in IMRT-techniques; dynamical use in conformal treatment and dynamic IMRT.

One of the most important components of the system "linear accelerator" from the patient's security point of view is the monitor chamber as part of the beam steering system. There are three different types of beam steering and control philosophies implemented in commercially available medical linear accelerators: (1) stable subsystems; selected steps of beam energy and doserate; feed back control. (2) feed back control depending on computer based look-up tables. (3) well aligned equipment; oversized parameter (beam current; RF-power; energy spread; etc.); minimum feed-back control, because of low sensitivity for deviations in parameter set (Fig. 12). All three solutions have one component in common: the dose monitor chamber for measuring dose, dose rate, symmetry, flatness and in some cases the constancy of energy. For this purpose the flat ionization chamber is segmented to deliver individual signals for a number of treatment beam and field parameters. At least two independent means of terminating the radiation are provided, because the primary system could fail: This is called "dual dosimetry". When monitor chambers show an appropriate additional segmentation, the capability of detecting beam misalignment and energy variation is given (Fig. 11).

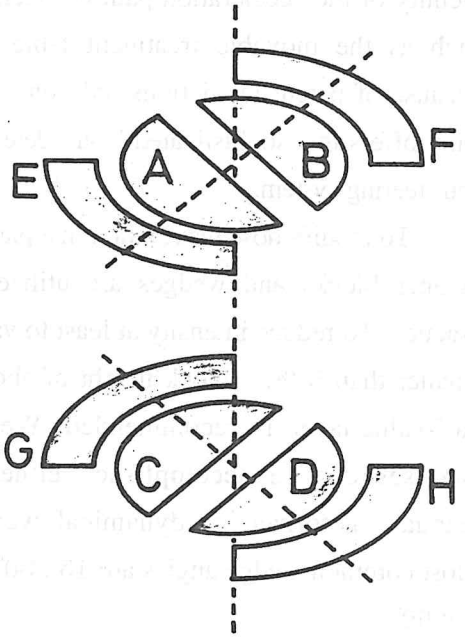


Fig. 11 Monitor chamber segmentation to detect shift-, angle- and energy error in photon lobe or scattered electron beam.

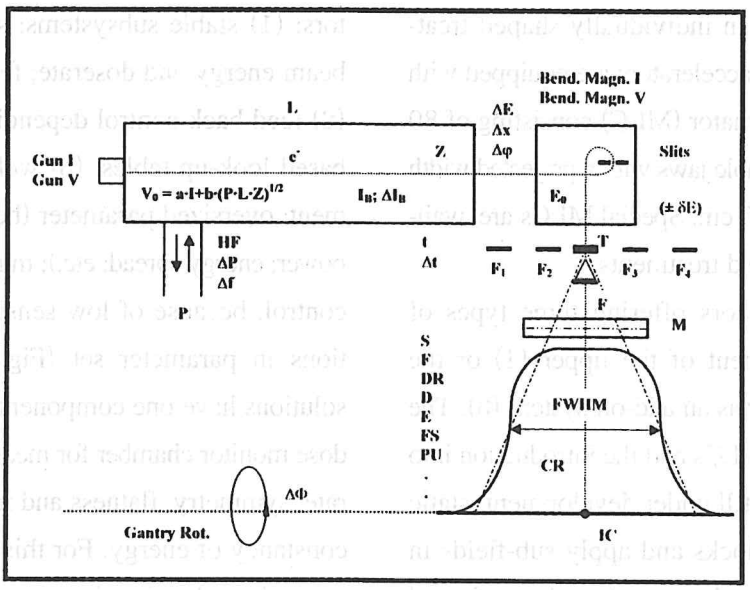


Fig. 12 *Basic acceleration parameter and treatment beam data.*
The abbreviations and explanations are self-explanatory.

Some dosimetry systems contain circuitries that feed back signals to steering coils to dynamically repositioning of the virgin electron beam to apply uniform treatment fields to the patient. For this purpose the system should be capable to detect selectively the three main errors of the primary electron beam hitting the target for X-ray production: (1) shift error; (2) angle error; (3) energy error. An optimal aligned electron beam results in a flat and symmetrical treatment field profile. The angle error influences the dose profile comparable to a wedge, the shift error causes a step in transversal distribution, additionally the whole profile is moved aside. Pairs of segments at the edges of the field and in the center are necessary for the selective identification of beam misalignments along the transversal and radial axis. Different combinations of center and peripheral segments are in some cases in addition used for energy control. To minimize field distortions such feed back systems are mandatory, also to compensate for the influence of ferromagnetic material in the vicinity of the acceleration path of electrons, such as the movable treatment table. But because of patent restrictions only one company offers this sophisticated beam detection and steering system.

To modify dose distribution in a planned manner blocks and wedges are utilized in practice. To reduce intensity at least to values smaller than 5 %, a block height of about 5 half-value-layer is recommended. Wedged isodoses can be accomplished either by manual, motorized or dynamical wedges. Most common wedge angles are 15°, 30°, 45° and 60°.

For convenience of the patient the treatment table (or patient support assembly PSA) should be capable of some minimum basic movements, to avoid restrictions in beam directions with regard to the patient: up and down (ca. 50 ... 170 cm above floor); right and left (± 25 cm); in and out (± 75 cm); rotation (± 90°); isocenter overtravel (>110 cm). Movable or C-shaped bars in the region of the isocenter guarantee an open dorsal view and an undisturbed treatment field, when table top is made of carbon fiber. Various accessories are available for all types of PSA for fixation of the patient.

To run such a high sophisticated complex unit, four minimum basic requirements are necessary: (1) stable and continuous public power supply; (2) air-conditioning to remove heat from treatment room; (3) cooling water system to stabilize internal temperature; (4) highly skilled and trained staff (physicists; technologists; dosimetrists; radiotherapists).

FUTURE DEVELOPMENTS AND TRENDS.

As mentioned above the medical linear accelerator is part of a complex radiotherapy network existing of a large number of components: (1) linear accelerator including accessories; (2) water phantom; (3) dosimetry system; (4) treatment planning system; (5) simulator; (6) computer tomopraph; (7) magnetic resonance tomograph; (8) virtual simulation unit; (9) auxiliary equipment for patient fixation; (10) hospital network; (11) hospital administration and billing department; (12) radiotherapy network.

The basic ideas of the linear accelerator have been developed some decades ago, but the computerization of some parts of the system "linear accelerator" has been added in last decade of the last century, because it made sense to have first a stable running and well designed linear accelerator and only just then add computer-steering and controlling. Some vendors had the idea to realize it the other way round. As a result only three companies survived in this special market.

Recent developments focus on automation and computer-controlling of beam delivery and dose application. In the same time the diagnostic possibilities to detect and to contour the tumour by visualization tools, such as CT, MRI, nuclear medicine, virtual simulation, reconstruction and contrast enhancement tools for CT, and ultra sound, grow tremendously. But nevertheless it is one of the biggest challenges for the future, to increase the biological and medical knowledge and information about tumour and healthy tissue for an adequate therapy for curing cancer patients. Individual information as well as topographical data of tumour, healthy tissue and organs at risk have to be investigated for the individual cancer patient. Otherwise 3-D calculation, biological dose planning and application of dose, high conformal irradiation techniques and intensity modulated radiotherapy have to be introduced and applied quite carefully in daily routine, because misapplication, over- or underdosage

with fatal consequences for the patient might be the result, since some basic biological related data are not known well in the moment. This was stated centuries ago by the French philosopher Voltaire defining his point of view about medical doctors: *A physician is a person, who pours drugs, of which he knows little, into a body, of which he knows less!*

LITERATURE

E. Dally: Information on Radiation Therapy: Comparison of Mevatron KD and Clinac 1800. Target-Siemens No. 4, 10/1985

E. Ginzton, C. S. Nunan: History of microwave electron linear accelerators for radiotherapy. Int. J. Rad. Onc. Biol. Phys. 11, 205-216 (1985)

D. Goer: Medical linear accelerators. In: Encyclopedia of Medical Devices and Instrumentation, J. G. Webster (Ed.), Vol. 3, 1772-1800 (1988)

D. Greene: Linear accelerators for Radiation Therapy. Medical Physics Handbook 17, Bristol 1986

HEW Publication (FDA) 76-8027: The Use of Electron Linear Accelerators in Medical Radiation Therapy: Physical Characteristics. San Diego, CA, USA 1980

J. M. Jensen: Anforderungen an Transmissions-Ionisationskammern in medizinischen Linearbeschleunigern. Medizintechnik 113, 171-79 (1993)

C. J. Karzmark: Advances in linear accelerator design for radiotherapy. Med. Phys. 11, 105-128 (1984)

C. J. Karzmark, R. J. Morton: A Primer on Theory and Operation of Linear Accelerators in Radiation Therapy. Medical Physics Publ. Corp. Madison, WI, 1989

C. J. Karzmark, C. S. Nunan, E. Tanabe: Medical Electron Accelerators. New York 1993

C. J. Karzmark, N. C. Pering: Electron linear accelerators for radiation therapy: History principles and contemporarly developments. Phys. Med. Biol. 18, 321-54 (1973)

C. S. Nunan: Design and performance criteria for medical electron accelerators. Nucl. Instr. Meth. Phys. Res. 10/11, 881-87 (1985)

C. Nunan: Criteria for optimal design of modern multi-mode medical linear accelerators. Proc. 12th Varian Users Meeting, Pebble Beach, Ca., 1988

J. A. Purdy, D. A. Goer: Dual energy X-ray beam accelerators in radiation therapy: An overview. Nucl. Instr. Meth. Phys. Res. B10, 1090-95 (1985)

E. Tanabe, R. W. Hamm: Compact multi-energy electron linear accelerators. Nucl. Instr. Meth. Phys. Res. B10/11, 871-76 (1985)