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UMI®
A Comparison Study of Multileaf and Micro-Multileaf Collimators

by

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January 2001

A Thesis submitted to the Faculty of Graduate Studies and Research in partial fulfillment of the requirements for the degree of Master of Science in Medical Physics

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Abstract

The dosimetric characteristics of a standard Varian 52-leaf multileaf collimator (MLC) and BrainLAB m₃ micro-multileaf collimator (micro-MLC) have been investigated for square, rectangular, and irregular fields for 6 MV and 18 MV photon beams provided by a Varian Clinac 2300 CD linear accelerator (linac). The percentage depth dose data and the conventional collimator factor are unaffected by the addition of MLC or micro-MLC shaped field unless, in the latter case, the tertiary field is much less than the jaw setting. However, relative dose factors for a given MLC or micro-MLC field size depend on the jaw setting. The penumbra is generally sharpest for fields defined by the micro-MLC and the least sharp for fields defined by the MLC. Average transmission values were found to be between 1.5% and 2.5%. Comparison and evaluation of two treatments, one delivered using the MLC and the other using the micro-MLC, for the same radiosurgical target volume are described.
Les caractéristiques dosimétriques d'un collimateur multilames (CML) standard à 52- lames de Varian et du collimateur micro-multilames (micro-CML) m3 de BrainLAB ont été étudiées pour des champs carrés, rectangulaires, et irréguliers dans le cas de faisceaux de rayons-X 6 MV et 18 MV produits par un accélérateur linéaire Varian Clinac 2300 C/D. Le pourcentage de dose en profondeur et le facteur conventionnel de collimation demeurent inchangés par l'ajout des champs formés par le CML ou le micro-CML à moins que, dans le dernier cas, le champ tertiaire soit beaucoup plus petit que la configuration de mâchoire du collimateur secondaire. Les facteurs relatifs de dose pour un champ donné par CML ou micro-CML sont dépendants de la configuration de mâchoire. La pénombre est généralement la plus aiguë pour des champs définis par le micro-CML et la moins aiguë pour des champs définis par le CML. Les valeurs moyennes de transmission se sont avérées être entre 1.5% et 2.5%. La comparaison et l'évaluation de deux traitements, le premier livré en utilisant le CML et l'autre livré en utilisant le micro-CML, sont décrits pour un même volume cible radiosurgical.
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Fundamentals of Medical Physics

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1.1 Introduction

Clinical medical physics is concerned with the application and measurement of ionizing radiation in the diagnosis and treatment of disease, typically malignant tumors. In the treatment of these tumors with radiation therapy, the aim is to deliver a high, yet uniform, radiation dose to the tumor volume while minimizing the amount of radiation received by the neighboring healthy tissue. However, the risk of any undesirable complications due to the inherent irradiation of the surrounding normal tissue is a limiting factor in the effectiveness of radiotherapy treatments. Additionally, certain organs, such as the eyes, spinal cord, lungs, and salivary glands to name a few, are relatively sensitive to radiation damage. These critical structures must be given special consideration during the treatment planning process.

Field shaping of the radiation beam is used to minimize the radiation received by the healthy tissues surrounding the irregularly-shaped target volumes. In addition,
multiple radiation fields are used to lower the dose to the tissue outside the target volume, while maximizing the dose to the target. Three dimensional conformal radiation therapy has been suggested to provide better local tumor control through allowing a larger target dose without an increase of dose to healthy tissues surrounding the target.\textsuperscript{1,2} Obviously, the dose calculation algorithms of the 3D treatment planning software must be capable of modeling the dose distribution accurately.

Evidently, it is crucial to know the radiation dose received at any point in the patient. The delivered dose is determined from the amount of energy absorbed by the medium; however, in order to determine this energy, the underlying physics on how ionizing radiation interacts in a medium must be understood. This chapter will therefore concentrate briefly on how ionizing radiation is produced, how it subsequently interacts with a medium such as tissue, how the resulting charged particles deposit their energy in the medium, and how the radiation dose is determined from this energy.

1.2 Production of Ionizing Radiation

Ionizing radiation is any radiation beam of sufficient energy to ionize an atom through the removal of one of its outermost electrons by Coulomb interactions. The ionization potential of all atoms is on the order of about a few electron-volts (eV), ranging from roughly 5 eV for alkali atoms to nearly 25 eV for helium. Evidently, the removal of a more tightly bound inner shell electron will require a larger energy. This energy increases as the atomic number of the material increases, amounting to about 150 keV for a K-shell electron in dubnium, which has an atomic number of 105.

Ionizing radiation is observed in one of two possible forms: either directly ionizing or indirectly ionizing. Directly ionizing radiation is characterized by charged particles, such as electrons, protons, and alpha particles, which ionize atoms through direct Coulomb interactions with the orbital electrons of the atoms. Indirectly ionizing radiation, on the other hand, is characterized by neutral particles, such as photons and neutrons, which must first undergo non-Coulomb interactions where charged particles are released. For instance, photon interactions with the medium, such as the photoelectric effect, Compton scattering, and pair production, release high energy electrons and positrons while neutron collisions with the nuclei of atoms release protons or heavier
charged particles. These charged particles may then proceed to ionize the medium directly provided that they are released with sufficient kinetic energy to be able to produce ionization events.

Indirectly ionizing photons originate primarily in one of two forms: either as characteristic (fluorescence) x-rays or bremsstrahlung (continuous) radiation.

A. Characteristic (or fluorescence) radiation

In 1913, Bohr proposed that the electrons of an atom orbit the nucleus in discrete allowed orbits or shells. The angular momentum of orbital electrons in each respective orbit or shell is quantized, meaning that each shell possesses a discrete electron binding energy. Bohr furthermore postulated that the electrons do not lose any energy while traveling in their orbits. If, by some means, the orbital electron is supplied enough energy to overcome its binding energy, it will be ejected from the atom with a kinetic energy equal to the difference between the energy supplied and its shell binding energy. The atom thus loses an orbital electron and is then said to be ionized. On the other hand, if the energy supplied to the orbital electron is not large enough to overcome its binding energy, yet it is sufficient to cause the electron to jump to a higher vacant yet bound energy level, the atom is said to be in an excited state. The atom will eventually return to its ground (or lowest energy) state through electronic transitions from higher to lower shells. As the electron falls from a higher to a lower shell, its loss of energy is accounted for by the emission of a photon of energy equal to the difference in binding energies between the two shell levels. The resulting photon is called a characteristic (or fluorescence) x-ray.

B. Bremsstrahlung radiation

Bremsstrahlung radiation occurs when an incident charged particle of a certain initial kinetic energy, most commonly an electron, passes near the nucleus of an atom. The charged particle will feel either an attractive or repulsive Coulomb force from the positive nucleus. Assuming the charged particle is an electron, it will partially orbit the nucleus and lose part or all of its kinetic energy through the Coulomb interaction with the nucleus. The loss of the electron’s kinetic energy will appear as a photon of energy equal
to the difference between the electron's initial and final kinetic energy. The resulting photon is referred to as bremsstrahlung radiation. The angular distribution of photons emitted through bremsstrahlung is proportional to the energy (or velocity) of the incident electron in the following manner:

$$\text{Angular distribution of photons} \propto \frac{\sin^2 \theta}{\left(1 - \frac{v}{c} \cos \theta\right)^5},$$

where $\theta$ is the angle between the point of observation and the acceleration of the incident charged particle, $v$ is the velocity of the incident particle, and $c$ is the speed of light in vacuum.

Low energy electron beams incident on a target will result in x-rays predominantly radiated at right angles to the direction of the electron beam, whereas the higher the energy of the electron beam, the more forward peaked is the production of x-rays. Since the velocity of the charged particle can never reach the speed of light, photons from a single event will never be emitted at an angle $\theta$ of 0° (i.e., in the direction of the incident electron beam). However, the envelope due to a large number of electrons interacting in a typical x-ray target can be considered to be forward peaked and centered about $\theta = 0°$.

The probability for the production of bremsstrahlung radiation is proportional to $Z^2$, where $Z$ is the target atomic number. Therefore, bremsstrahlung production for energetic electrons in high $Z$ materials is important; however, it is relatively insignificant in low $Z$ materials, such as tissue, for electron energies below 10 MeV.

1.3 Interaction of Ionizing Radiation in a Medium

Now that a beam of photons has been produced, the means by which the photon beam interacts with matter is of interest. As a monoenergetic photon beam passes through a medium, a certain fraction of photons will interact per unit distance into the medium. This fraction per unit thickness is called the linear attenuation coefficient $\mu$ and is given in units of cm$^{-1}$. For a photon beam initially containing $N_0$ particles, $N$ particles will emerge after passing through an absorber thickness $x$. The law of exponential attenuation
Chaplin states that

\[ N(x) = N_0 e^{-\mu x}. \tag{1-1} \]

There is no limit to the distance a photon may travel through a medium before undergoing an interaction. When an interaction does occur, it predominantly takes place in one of three ways. They are the photoelectric effect, Compton scattering, and pair production. In each occurrence, the photon transfers some or nearly all of its energy to an electron in the medium. An electron is ejected from the atom at a high speed and the incident photon may be either scattered or it may disappear completely. Rayleigh scattering is also a possibility although in this case the incident photon is scattered at a small angle with no energy loss and no production of energetic electrons. This effect is therefore of little importance in radiation dosimetry, since no energy is transferred from the photon to the medium.

A. **Photoelectric effect (photoeffect)**

The photoeffect occurs when an incident photon interacts with a tightly bound orbital electron of the absorbing material. An orbital electron is considered tightly bound when its binding energy is smaller yet comparable to the photon energy. The photon completely disappears and the orbital electron is ejected from the atom with a kinetic energy equal to the energy of the photon minus the electron's shell binding energy. The photoeffect is most likely to occur when the incident photon energy is slightly greater than the binding energy of the electron in a shell. The ejected electron, referred to as the photoelectron, creates a vacancy in the shell and, as discussed in Section 1.2.A, the atom may return to ground state concurrently with an emission of characteristic x-rays. Alternatively, the energy loss as the atom returns to ground state may not be accounted for entirely by characteristic x-rays but either by Auger electrons as well, or exclusively by Auger electrons. Therefore, the energy transferred to the medium \( E_p \) due to the photoelectric effect is the sum of the kinetic energies of the photoelectron plus the Auger electrons.
B. Compton effect

The Compton effect occurs when an incident photon interacts with a free and stationary orbital electron of the absorbing material. An orbital electron is considered essentially free if its binding energy in a shell represents a small fraction of the incident photon energy. The incident photon is scattered and the free electron recoils away with a large kinetic energy given by the difference between the energies of the incident and scattered photon. A maximum amount of energy is transferred to the recoil (or Compton) electron when the photon makes a direct hit with the free electron. In this case, the electron will travel straight forward while the photon will be backscattered with a minimum amount of energy at a scattering angle \(\theta\) of 180°. Conversely, no energy will be transferred to the recoil electron when the photon just grazes the free electron. The electron will recoil at 90° while the photon will be forward scattered at 0° with no loss of energy. The mean fraction of incident photon energy given to the recoil electron is used to determine the amount of energy transferred to the medium through the Compton effect.

C. Pair production

Pair production may occur when an incident photon encounters the strong field of a nucleus or, less frequently, the electric field of an orbital electron. If the photon has an energy of at least \(2m_e c^2\), it may disappear and conservation of energy into mass generates an electron-positron pair. Conservation of momentum is maintained by the recoil of the nucleus and the energy transferred to the medium is equal to the kinetic energy of the electron-positron pair.

The electron and positron deposit their energy in the absorbing medium through ionization and excitation of the atoms of the absorbing medium. Once the positron has dispensed all of its kinetic energy, it annihilates with a nearby free electron and produces two photons. The two photons, each of energy \(m_e c^2\), travel in opposite directions in order to conserve energy and momentum.

1.4 Predominance of Individual Interactions

Each of these four possible interactions has its own probability to occur at a
given photon energy and a given absorber atomic number. The linear attenuation coefficient $\mu$ is the sum of the probabilities for each interaction to occur for a given photon energy and absorber material. It is thus expressed as

$$\mu = \tau + \sigma_R + \sigma_C + \kappa,$$

(1-2)

where $\tau$, $\sigma_R$, $\sigma_C$, and $\kappa$ are the linear attenuation coefficients for the photoelectric effect, Rayleigh scattering, Compton scattering, and pair production, respectively.

Fig. 1.1 displays the regions for which a particular interaction is likely to prevail over a range of photon energies $\hbar \nu$ and absorber atomic numbers $Z$. The curves represent energies and atomic numbers for which two types of interactions are equally probable: the left hand curve signifies an equal probability for the photoelectric effect and the Compton effect to occur, while the right hand curve is for equal probability of Compton effect and pair production. It is evident that the photoelectric effect dominates at low photon energies, the Compton effect predominates in the mid range, while pair production takes over at high photon energies. Tissue is mainly made up of water which has an effective atomic number of 7.51. The Compton effect therefore dominates over a large range of energies, from about 20 keV to 30 MeV for low Z materials. Since clinical treatments are usually carried out with a photon energy anywhere between 1 MeV and 20 MeV, the Compton effect is thus the most important photon interaction in radiation therapy.

Fig. 1.1: Regions of dominant interactions for photon energies $\hbar \nu$ in the range from $10^2$ MeV to $10^3$ MeV and the atomic number $Z$ of absorbing materials.
Fig. 1.2 displays the total mass attenuation coefficient $\mu/\rho$ for water and lead for photon energies ranging from 10 keV to 100 MeV. As previously stated, at low energies the photoelectric effect predominates. Since the mass attenuation coefficient for the photoelectric effect is proportional to $(Z/h\nu)^3$, it has a high value for large $Z$ materials and a steep drop off with increasing energy. The K-edge of lead at 88 keV is evident from the spike in the curve. Below this energy the K-shell electrons do not contribute to the photoelectric effect because their binding energy is too large compared to the incident photon energy. The K-edge of water is not depicted on the graph since it occurs at an energy lower than the minimum energy on the scale. The mass attenuation coefficient due to Compton scattering is independent of $Z$. This independence is only approximate because, except for hydrogen, $Z/A$ varies from 0.5 down to 0.4 with increasing $Z$ whereas for hydrogen $Z/A$ is equal to 1. Hence, as expected, the mass attenuation coefficients for water and lead are roughly equal in the Compton region. Finally, the mass attenuation coefficient for pair production is proportional to $Z$. It is evident that, at high energies above 10 MeV where this interaction predominates, the mass attenuation coefficient increases with energy.

Fig. 1.2: Total mass attenuation coefficient $\mu/\rho$ as a function of photon energy for water and lead.
1.5 Energy Transfer and Energy Absorption Coefficients

When a photon interacts with a medium, part of its energy generally goes into photons that are scattered away while the remainder is transferred to electrons and positrons which are set in motion. These charged particles travel through the medium and deposit a portion of their kinetic energy in the medium while the remainder is radiated away as bremsstrahlung. It is impossible to predict the amount of energy that will be transferred from the incident photon to the high speed electron or positron in one particular photon interaction, but on average, the amount of energy transferred due to a large number of interactions is known. Analogous to the linear attenuation coefficient, an energy transfer coefficient $\mu_t$ and energy absorption coefficient $\mu_{ab}$ have been defined. These coefficients are very important in radiation dosimetry and they state, on average, the fraction of energy that will be transferred to electrons in the medium or absorbed by the medium at a particular photon energy and are given by

$$\mu_t = \frac{|E_{\nu}|}{h\nu} \quad \text{and} \quad \mu_{ab} = \frac{|E_{ab}|}{h\nu}. \quad (1-3)$$

As previously stated in Section 1.3, no energetic electrons are produced through Rayleigh scattering, and as a result, there is no contribution to the linear energy transfer coefficient from Rayleigh scattering. However, Section 1.3.A stated that the energy transferred to the medium through the photoelectric effect $|E_{\nu}|_{pe}$ is the sum of the kinetic energies of the photoelectrons plus Auger electrons. Therefore, if the loss of energy as the atom returns to ground state is accounted for solely by the emission of characteristic radiation, the energy transferred to the medium is a minimum and simply that of the photoelectrons. This is generally representative of higher atomic number materials such as lead. On the other hand, if the loss of energy as the atom returns to ground state is accounted for completely by the emission of Auger electrons, the energy transferred to the medium is a maximum and equal to the energy of the incident photon. This is generally representative of lower atomic number materials such as tissue. Thus,

$$|E_{\nu}|_{pe} = h\nu - \delta\mathcal{E}_{g}, \quad (1-4)$$
where \( \delta \) is a parameter between 0 and 1, accounting for the fluorescent yield and the probability for the photoelectric effect to occur in a given shell and \( E_B \) is the binding energy of the orbital electron.

The maximum and mean fraction of incident photon energy transferred to Compton recoil electrons are depicted in Fig. 1.3. The mean fraction of incident photon energy given to the recoil electron is used in the determination the amount of energy \( |E_{\nu}|_c \) transferred to the medium due to the Compton effect. Finally, the energy transferred to charged particles in the medium through pair production is equal to the kinetic energy of the electron-positron pair, thus

\[
|E_{\nu}|_c = h\nu - 2m_e c^2. \tag{1-5}
\]

Therefore, the linear energy transfer coefficient is given by the sum of the three effects such that

\[
\mu_{\nu} = \tau \left( 1 - \frac{\delta E_B}{h\nu} \right) + \sigma_c \frac{|E_{\nu}|_c}{h\nu} + \kappa \left( 1 - \frac{2m_e c^2}{h\nu} \right). \tag{1-6}
\]

Similarly, the average energy absorbed by the medium is obtained by the sum of the energies absorbed due to the three contributions. However, not all of the energy transferred to electrons and positrons in the absorbing material is absorbed by the medium. A fraction of the energy of the secondary charged particles goes into the production of bremsstrahlung photons that carry their energy out of the local volume. As a result, the energy absorption coefficient is defined as

\[
\mu_{ab} = \mu_{\nu} (1 - B), \tag{1-7}
\]

where \( B \) is the bremsstrahlung fraction. In general, for low Z materials at photon energies below about 1 MeV bremsstrahlung production is negligible so that the amount of energy transferred is roughly equal to the amount of energy absorbed. Furthermore, at very low photon energies, little radiation is scattered so that the energy transferred and absorbed is roughly equal to the incident photon energy.

For example, a 10 MeV photon in lead will, on average, transfer 8.45 MeV to the medium while 6.42 MeV will be absorbed by the medium. Thus, 10 MeV \(-\) 8.45 MeV
= 1.55 MeV will go into scattered radiation while \(8.45 \text{ MeV} - 6.42 \text{ MeV} = 2.03 \text{ MeV}\) is radiated away as bremsstrahlung. However, a 10 MeV photon in water will, on average, transfer 7.33 MeV to the medium while 7.07 MeV will be absorbed by the medium. Thus, 2.67 MeV will go into scattered radiation while 0.26 MeV is radiated away as bremsstrahlung.

![Graph](image)

**Fig. 1.3**: Maximum and average energy transferred to recoil electrons due to Compton scattering.

### 1.6 Interaction of Charged Particles in a Medium

Now that a high speed charged particle, usually an electron, has been set in motion, the manner in which this electron interacts with matter is of interest. Unlike a photon, which will typically lose its energy in a relatively small number of large energy transfer interactions, an electron set in motion will typically lose its energy in a large number of small energy transfer interactions. The high speed electron travels through the medium, losing energy along its track due to Coulomb collisional and radiative interactions with orbital electrons and nuclei of atoms, respectively, in the medium. Collisional interactions result in a transfer of energy from the high speed electron to atoms of the medium, creating excitation and ionization of the atoms. This energy will contribute to the radiation dose. On the other hand, radiative interactions between the high speed electron and nuclei of the absorber atoms result in the production of
bremsstrahlung photons that carry energy away from the electron's track. Thus, energy is transferred from the high speed electron either to atoms of the medium through collisional losses or into bremsstrahlung photons through radiative losses until the electron eventually comes to rest. Consequently, a charged particle has a specific range through which it is predicted to travel in a particular medium.

The linear stopping power $S$ of a charged particle describes the expected rate of energy loss of the particle per unit distance it travels. The stopping power depends on both the type of particle and its kinetic energy as well as on the medium through which it is traveling, and consists of two components: the collisional and radiative losses.

Fig. 1.4 illustrates the total mass stopping power of an electron in water which is a combination of the collisional and radiative mass stopping power contributions. As previously stated, bremsstrahlung interactions are more probable in high $Z$ materials and at high energies. Thus, radiative interactions are relatively insignificant in low $Z$ materials, such as water, for electrons below 10 MeV.

![Fig. 1.4: Collisional, radiative, and total mass stopping power as a function of electron kinetic energy for water.](image)

### 1.7 Determination of the Absorbed Dose

The absorbed dose $D$ is defined as the amount of energy from ionizing radiation that is absorbed or retained by a mass of medium and is given in units of J/kg or Gray.
(Gy). The mass of the medium is assumed to be sufficiently small so that the absorbed dose may be defined at a point.

For an external photon radiation beam striking a patient or phantom, the absorbed dose within the patient or phantom builds up to a maximum at a certain depth, beyond which the amount of energy absorbed declines. This is illustrated in Fig. 1.5 and can be explained as follows. As a high energy photon beam enters a patient or phantom, it interacts with the atoms and electrons of the medium. At the point of interaction, a certain fraction of the photon’s energy is transferred to electrons within the medium and these electrons are ejected from their atoms. These high speed electrons travel through the medium and deposit their energy, producing a track of either ionized or excited atoms. It is therefore evident that the electrons may deposit their energy a significant distance away from the initial point where energy transfer occurs. If the phantom is considered to be made up of a succession of layers, a large amount of energy will be transferred to electrons in the first layer; however, very little of the electrons’ energy will be absorbed in this layer. A larger amount of energy will be absorbed in the second layer due to the electrons that were ejected in the first layer. Meanwhile, additional photon interactions in which electrons are ejected will occur in the second layer. As a result, an even greater amount of energy will be absorbed in the third layer due to electrons that were ejected from both the first and second layers, and so forth. However, since the photon beam is attenuated exponentially, fewer interactions occur in each subsequent layer indicating that the overall amount of energy transferred to the medium also decreases with each subsequent layer. Therefore, the absorbed dose will initially increase, reach a maximum value at the depth of dose maximum $d_{\text{max}}$, and then decrease due to the attenuation of the photon beam at larger depths. The region between the surface and $d_{\text{max}}$ is referred to as the dose buildup region.
Fundamentals of Medical Physics

Chapter 1

X rays are emitted from a source primarily in one of two ways, either when an electron falls to a lower atomic energy level (characteristic radiation) or when an electron decelerates in a nuclear Coulomb field (bremsstrahlung).

The incident photon interacts with an atom or orbital electron of the medium by either the photoelectric effect, Compton scattering, pair production, or sometimes Rayleigh scattering. A summary of the characteristics of these four types of photon interactions is provided in Table 1.1. Except for Rayleigh scattering, which is coherent and pair production which occurs in the field of a nucleus, energy is transferred from the incident photon to the orbital electron so that the electron is ejected from the atom with a high kinetic energy $E_{kr}$. In the Compton effect the incident photon is scattered at a certain angle with an energy equal to $h\nu - E_{kr}$. The high energy electron travels through the medium, losing energy through collisional and radiative Coulomb interactions with other atoms or nuclei of the medium. The radiative interactions result in the emission of bremsstrahlung photons of energy $h\nu''$. These photons carry away some of the electron's energy so that this energy is no longer available for deposition in the medium. Therefore, the electron deposits an energy $E_{ab} = E_{kr} - h\nu''$ in the medium. This absorbed energy at a certain point in the medium determines the radiation dose at that point.

Charged particles traveling through the body tissues produce tracks of either
ionized or excited atoms. Once the atoms of the molecular cells within the body tissue are ionized or excited, chemical and biological changes result which modify their atomic structure. The cells may be damaged depending on the extent of the cell’s alteration, either killing the cell or ceasing its ability to reproduce.

Table 1.1: Summary of the characteristics of the four types of photon interactions.

<table>
<thead>
<tr>
<th>Photon interaction</th>
<th>Photoeffect</th>
<th>Rayleigh scattering</th>
<th>Compton effect</th>
<th>Pair production</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mode of interaction</td>
<td>photon disappearance</td>
<td>photon scattering</td>
<td>photon scattering</td>
<td>photon disappearance</td>
</tr>
<tr>
<td>Threshold</td>
<td>no</td>
<td>no</td>
<td>no</td>
<td>$2m_ec^2$</td>
</tr>
<tr>
<td>Attenuation coefficient</td>
<td>$\tau$</td>
<td>$\sigma_R$</td>
<td>$\alpha_c$</td>
<td>$\kappa$</td>
</tr>
<tr>
<td>Particles released</td>
<td>photoelectron</td>
<td>none</td>
<td>Compton (recoil) electron</td>
<td>electron/positron pair</td>
</tr>
<tr>
<td>Energy dependence</td>
<td>$(h\nu)^3$</td>
<td>$(h\nu)^2$</td>
<td>decreases with energy</td>
<td>increases with energy</td>
</tr>
<tr>
<td>Mass coefficient dependence on Z</td>
<td>$Z^3$</td>
<td>$Z$</td>
<td>independent</td>
<td>$Z$</td>
</tr>
<tr>
<td>Average energy transferred</td>
<td>$h\nu-\Delta E_{\beta}$</td>
<td>0</td>
<td>$</td>
<td>E_p</td>
</tr>
<tr>
<td>Subsequent effect</td>
<td>characteristic x-ray, Auger effect</td>
<td>none</td>
<td>characteristic x-ray, Auger effect</td>
<td>annihilation radiation</td>
</tr>
<tr>
<td>Significant energy region</td>
<td>0 to 0.5 MeV</td>
<td>&lt; 1 MeV</td>
<td>around 1 MeV</td>
<td>&gt; 1 MeV</td>
</tr>
</tbody>
</table>

1.9 References


Introduction

In radiotherapy it is crucial to know the radiation dose at any point in a patient. The absorbed dose distribution is determined from the energy absorbed per mass of medium at various points in the patient. Since it is impossible to measure the dose distribution of the radiation beam in a patient as various beam parameters are changed, phantoms are used to perform this study. In photon beam radiotherapy, a phantom must therefore be made of a material that can simulate tissue by absorbing and scattering photons in the same manner as tissue. As discussed in Section 1.4, the predominant type
of photon interaction with tissue in the energy range used in radiotherapy is Compton scattering. Therefore, the phantom must have the same electron density as tissue, and since tissue is mainly made up of water, water is obviously a suitable phantom material. However, problems with water phantoms may arise because ionization chambers, which are used to measure the radiation dose, exhibit leakage problems when immersed in water. Therefore, solid materials, such as polystyrene, Solid Water, and Lucite, were developed to overcome this problem.

### 2.2 Quality of an X-ray Beam

When an object is placed into the path of a photon beam, a change in the quality and quantity of the radiation beam will result because of absorption and scattering of the beam. Consequently, the dose received at a fixed point in the patient or phantom will be affected. The quality of an x-ray beam refers to a measure of its penetrating power and may be specified in terms of its half value layer in a particular material for low energy beams and by its nominal accelerating potential or percent depth dose at a depth of 10 cm for high energy beams in the megavoltage range.

The half value layer (HVL) is defined as the thickness of a given material that will attenuate the beam to half of its original intensity, and is given by

\[ HVL = \frac{\ln 2}{\mu}, \]  

(2-1)

where \( \mu \) is the linear attenuation coefficient.

Section 1.3 demonstrated that the attenuation of a monoenergetic photon beam is exponential with absorber thickness. Consequently, the intensity will be represented by a straight line on a semilogarithmic plot of \( \mu \) versus absorber thickness. Subsequent HVLs will therefore be the same as the first HVL. For example, cobalt-60, a common radioisotope source used in radiotherapy, emits monoenergetic gamma rays of average energy equal to 1.25 MeV. The first HVL, second HVL, and so on, are a constant value of 1.1 cm in lead. On the other hand, a heterogeneous x-ray beam is made up of a spectrum of energies ranging from zero to a nominal value and the semilogarithmic plot of its transmission versus absorber thickness is no longer a straight line.
Since interactions of low energy beams with tissues are dominated by the photoelectric effect in which the mass attenuation coefficient of the beam is roughly proportional to \((Z/h\nu)^3\), a high atomic number \(Z\) material placed into the path of the beam will preferentially attenuate the low energy portion of a heterogeneous photon beam energy spectrum. Beam hardening will result in a shift of the mean energy of the photon beam to a higher energy, therefore increasing the penetrating power of the beam. The greater the atomic number of the material, the greater the attenuation of the low energy portion of the beam and the greater the beam hardening. The same is true for absorber thickness. The greater the thickness, the greater is the beam hardening effect on the beam spectrum.

On the other hand, a heterogeneous photon beam in the mid energy range from a few 100 keV to a few MeV will be dominated by the Compton effect. In this region the mass attenuation coefficient is roughly independent of the atomic number of the material, and therefore, any material placed in the path of the beam will simply lower the overall energy spectrum of the photon beam.

Very high energy photon beams (a few MeV and above) rely on pair production. Since the mass attenuation coefficient for pair production varies approximately as \(Z\cdot\ln(h\nu)\), the higher energy portion of the photon beam's energy spectrum will be preferentially attenuated, and beam softening will result manifesting itself in a decrease of the average energy of the beam. As a result, an undesirable reduction in the penetrating power of the beam will occur.

While for low energy photon beams in the 100 kV range the HVL is a good descriptor of beam quality, the quality of high energy photon beams in the megavoltage range is generally specified by referring to the maximum energy of the photon beam spectrum in MV. This specification is used rather than HVL as a result of all energies being attenuated roughly the same in the Compton region. Therefore, little variation in the half value layers will be observed even though there may be a large difference in the beam energy.

However, the spectra of x-rays produced by treatment machines operating at the same nominal energies may be significantly different from one another due to degradation.
of the beam as it passes through the flattening filter and other beam collimating devices. Consequently, determination of the absorbed dose, which is a function of the stopping power ratio, which in turn is a function of the spectrum of electrons at the point of measurement, may result in dosimetric errors. Therefore, specification of clinical photon beams by the maximum energy in their spectrum is not adequate for reference beam dosimetry. The TG-21 protocol\textsuperscript{1} recommends that the quality of photon beams be described by their nominal accelerating potential (NAP) in MV. This is determined from the ratio of ionizations measured on the beam axis in a water phantom at depths of 20 cm and 10 cm, both at the nominal SAD of the treatment machine, for a field size of 10\times10 cm\textsuperscript{2}. TG-21 provides a conversion of the ionization ratio to the nominal accelerating potential. However, there may be a variation of up to 1.3\% in the stopping power ratio for a given nominal accelerating potential.\textsuperscript{2} 

More recently, the TG-51 protocol\textsuperscript{3} recommends the use of the percent depth dose, \(\%dd(10)\), due to photons only for a 10\times10 cm\textsuperscript{2} beam at an SSD of 100 cm measured at a depth of 10 cm in a water phantom as a descriptor of the quality of the photon beam. For high photon energies of 10 MV and above, electron contamination from the accelerator head may significantly affect the dose at depth of dose maximum and consequently lower the value of the percent depth dose, \(\%dd(10)\), when electron contamination is included. Measurements must therefore be taken with a 1 mm thick lead foil positioned in the beam to reduce the electron contamination from the accelerator head to a negligible level. The lead foil itself subsequently produces a known amount of electron contamination that can later be taken into account. The lead foil should be placed at 50 cm from the phantom surface to exclude electrons scattered from the flattening filter and collimators, or at 30 cm from the phantom surface if the accelerator does not permit the placement of the foil at 50 cm, as is the case for accelerators equipped with tertiary multileaf collimators. The following relations are used to determine the percent depth dose excluding electron contamination:\textsuperscript{4} 

\(i\) For beams with energies below 10 MV an assumption is made that the electron contamination of the beam is negligible, i.e., 

\[
\%dd(10)_x = \%dd(10). 
\] (2-2)
(ii) For beams with energies of 10 MV and above the electron contamination is accounted for as follows:

\[
\%dd(10)_x = [0.8905 + 0.00150\%dd(10)_{PS}]\%dd(10)_{PS}, \quad (2-3)
\]

(foil at 50 cm, \(\%dd(10)_{PS} \geq 73\%\))

\[
\%dd(10)_x = [0.8116 + 0.00264\%dd(10)_{PS}]\%dd(10)_{PS}, \quad (2-4)
\]

(foil at 30 cm, \(\%dd(10)_{PS} \geq 71\%\))

TG-51 provides a table in which the quality conversion factor, which converts the calibration factor for a Co-60 beam to that of a different quality beam, can be obtained from the value of \(\%dd(10)_x\). Ultimately, the value of \(\%dd(10)_x\) maintains better sensitivity to beam quality changes and provides a more precise specification for high-energy beams than does the nominal accelerating potential.

2.3 Dose Delivered by an X-ray Beam

The dose delivered by an x-ray beam depends on numerous factors, including the material in which the dose is being delivered, the depth of the point of interest in the phantom, the field size and energy of the beam, the distance from the source to the surface of the phantom, as well as the type of beam modifier placed in the path of the beam. However, if the dose is known at a reference point in the patient, the dose at any other point in the patient can be calculated using various functions, some of which will be discussed in this chapter.

Fig. 2.1(a) displays schematically a photon beam in air, while Fig. 2.1(b) shows a photon beam of the same energy incident perpendicularly on a phantom. The photons are assumed to be emitted from a point source S after which the beam is collimated to define the desired field size in both the x and y directions. In the diagram, a square field of size \(A\) (cm\(^2\)) is defined at the surface of the phantom. The source to surface distance (SSD, or sometimes referred to as \(f\)), the depth of maximum dose \(d_{max}\), and the depth \(d\) in the phantom are also illustrated in Fig. 2.1. Point P is defined at depth \(d_{max}\) while point Q
is an arbitrary point on the beam central axis at depth $d$. The primed notation implies conditions in air while the unprimed notation implies conditions in phantom.

![Diagram](image)

**Fig. 2.1**: Notation used in defining the dosimetric parameters, where part (a) represents a photon beam in air while part (b) represents a photon beam incident on a phantom surface.

### 2.4 Depth of Dose Maximum Dependence on Field Size

As stated in Section 1.7, the dose in phantom builds up to a maximum at a depth $d_{\text{max}}$ after which it falls off roughly in an exponential fashion. The magnitude of $d_{\text{max}}$ is a function of both the beam energy and the field size of the beam; however, in practice, this depth of dose maximum for a particular beam energy is taken to be a constant and essentially independent of field size. For example, the depth of dose maximum for an 18 MV photon beam ranges from roughly 2.5 cm for a $1\times1$ cm$^2$ field up to 3.5 cm for a $5\times5$ cm$^2$ field and back down to 2.2 cm for a $30\times30$ cm$^2$. For extremely small field sizes, in the limit of $0\times0$ cm$^2$, the dose to the medium is a result of only primary radiation. Since there is no scatter, the depth of dose maximum is at the point of electronic equilibrium. As the field size is increased the depth of dose maximum also increases as a result of scatter from within the phantom. However, at a field size of approximately $5\times5$ cm$^2$, the
value of $d_{\text{max}}$ reaches a peak and gradually declines as a result of increased scatter from within the collimator and flattening filter which brings $d_{\text{max}}$ closer to the surface of the phantom. The value of $d_{\text{max}}$ for large fields sizes approaches the value obtained when only primary radiation is considered.

### 2.5 Inverse Square Law

For a constant collimator setting $A_{\text{col}}$, doses at any two points in air are related by the inverse square of their distances from the source assuming that no other material is placed in the path of the photon beam. The relationship (see Fig. 2.1) is given by:

$$\frac{D'_{Q}(d, A_{\text{col}}, f, h\nu)}{D'_{P}(d_{\text{max}}, A_{\text{col}}, f, h\nu)} = \left(\frac{f + d_{\text{max}}}{f + d}\right)^2.$$

### 2.6 Peak Scatter Factor

For a given field size $A$ and beam energy $h\nu$, the peak scatter factor (PSF), sometimes referred to as the backscatter factor, is the ratio of the dose at depth dose maximum in phantom to the dose to a small mass of tissue at the same point in air, both defined along the beam central axis:

$$\text{PSF}(A, h\nu) = \frac{D_{P}(A, h\nu)}{D'_{P}(A, h\nu)}.$$

The peak scatter factor determines the amount by which the radiation dose will be increased due to radiation scattered back to point P in the phantom. For very low energy beams the absorbed dose in tissue can be increased by as much as 50% due to backscatter. However, Compton scattering theory demonstrates that, as the beam energy is increased, photons are increasingly scattered in the forward direction. As a result, the backscatter decreases for higher energy beams. For example, the peak scatter factor for $^{60}$Co for a 10×10 cm$^2$ field is only 1.054$^7$ indicating that the dose at point P will be only 5.4% higher than the dose to a small mass of tissue at the same point in air. By an energy of about 8 MV, the peak scatter factor reaches a minimum of 1.00 demonstrating that the amount of scatter at $d_{\text{max}}$ is essentially negligible. Furthermore, as the field size is increased for a given beam energy, more scatter is permitted to reach point P on the
central axis and therefore the peak scatter factor increases with field size. The dependence of the PSF on both beam energy and field size is evident in the plot of PSF versus field size for various beam energies given in Fig. 2.2.

![Graph showing PSF versus side of square (cm) for different beam energies.]

**Fig. 2.2**: The peak scatter factor for various beam energies. [Data from the British Journal of Radiology Supplement 25, “Central Axis Depth Dose Data for Use in Radiotherapy.” (London: British Institute of Radiology), 1996.]

For a given field size, on the other hand, PSF first increases with beam energy, reaches a maximum between 65 and 75 keV, depending on the field size, and then rapidly decreases. Although the greatest amount of backscatter occurs at very low photon energies, the decline in the peak scatter factor for decreasing energies is a result of the increasing value of \( \mu \). The backscattered photons are rapidly attenuated and few reach the point of interest \( P \) resulting in a decreasing peak scatter factor with decreasing energy. At relatively high photon energies, as stated earlier, the scatter becomes more forward peaked and beyond a certain energy the peak scatter factor begins to decrease with increasing energy.

### 2.7 Collimator Factor

The collimator scatter is attributed to photons scattered by all components of the machine head which are in the path of the beam, such as the flattening filter, the monitor chamber, and the beam modifiers. The amount of scatter from within the collimator head
is associated with the collimator factor \(CF(A)\), or sometimes designated as \(S_c(A)\), which is defined as the fraction of the output in-air for a given field size \(A\) (cm\(^2\)) to that of a reference field, usually 10x10 cm\(^2\), both defined at the isocenter of the machine. More explicitly,

\[ S_c(A) = CF(A) = \frac{D'(A)}{D'(10)}, \]  

(2-7)

where \(D'(A)\) and \(D'(10)\) are the measured outputs in air at nominal SAD for a field size \(A\) (cm\(^2\)) and 10x10 cm\(^2\), respectively. The collimator factor depends on the configuration of the treatment machine and may differ from one treatment unit to the next.

### 2.8 Scatter Factor

Phantom scatter originates from photons scattered within the phantom. The amount of scatter from within the phantom is associated with the phantom scatter factor \(SF(A)\), or sometimes referred to as \(S_p(A)\), which is a measure of the change in the amount of scattered radiation at the reference depth \(d_{\text{max}}\) that originates in the phantom as the field size is changed. It is given by the fraction of the in-phantom output at \(d_{\text{max}}\) for a given field size \(A\) (cm\(^2\)) to that of a reference field, again usually 10x10 cm\(^2\) but with the same collimator setting as the previous field. This implies that the reference field must be blocked in order to achieve a smaller field size incident on the phantom but with the same collimator setting used for the larger \(A\) (cm\(^2\)) field. This is illustrated in Fig. 2.3. The fixed collimator setting for both fields ensures that the measured variations in scatter are a result only of variations of scatter from within the phantom and not from variations of scatter originating in the collimator. The scatter factor is therefore independent of the treatment unit involved but depends on the beam energy.

The scatter factor will thus be defined by

\[ SF(A) = \frac{D_p(A)}{D_{p,\text{block}}(10)}, \]
where $D_p(A)$ is determined from the arrangement in Fig. 2.3(a), while $D_p^{\text{block}}(10)$ is determined from the arrangement in Fig. 2.3(b). The scatter factor can therefore be written as

$$SF(A) = \frac{D_p'(A) \cdot \text{PSF}(A)}{D_p^{\text{block}}(10) \cdot \text{PSF}(10)}$$

$$= \frac{D_p'(10) \cdot \text{CF}(A) \cdot \text{PSF}(A)}{D_p'(10) \cdot \text{CF}(A) \cdot \text{PSF}(10)}$$

$$= \frac{\text{PSF}(A)}{\text{PSF}(10)},$$

which suggests that $SF(A)$ is simply a normalized peak scatter factor with a reference field of $10\times10$ cm$^2$.

\[\text{Fig. 2.3 : Set-up required for the determination of the scatter factor where part (a) represents the arrangement for the measurement of the in-phantom output for a field size } A \text{ (cm}$^2$) and part (b) represents the arrangement for the measurement of the in-phantom output for a blocked field of size $10\times10$ cm$^2$ but with a collimator setting } A \text{ (cm}$^2$).\]

The larger the field size, the larger the volume of the irradiated material from which photons can be scattered toward the reference point. In other words, the scatter
factor is related to changes in the irradiated volume of the phantom while the collimator setting remains fixed.

Alternately, the scatter factor may also be defined as

$$SF(A) = \frac{D_p(A)}{D_p(10) \cdot CF(A) \cdot PSF(10)}$$

$$= \frac{D_p(A)}{D_p(10) \cdot CF(A)}$$

$$= \frac{RDF(A)}{CF(A)}$$

where $RDF(A)$ is the relative dose factor or output factor for a field size $A$ (cm$^2$). The relative dose factor, as shown below, is therefore a combination of the contributions of scatter from within the collimator and scatter from within the phantom, making it an easier quantity to measure than the scatter factor.

2.9 Relative Dose Factor or Output Factor

The relative dose factor ($RDF(A)$, or sometimes referred to as $S_{C.P}(A)$) is the ratio of the output at a reference depth in phantom, usually the depth of dose maximum $d_{max}$, for a given field size $A$ (cm$^2$) compared to the output at the same depth for a reference field size, again usually 10x10 cm$^2$. More explicitly,

$$RDF(A) = \frac{D_p(A)}{D_p(10)} = CF(A) \cdot SF(A),$$

where $D_p(A)$ and $D_p(10)$ are the measured outputs in phantom at the reference depth for a field size $A$ (cm$^2$) and 10x10 cm$^2$, respectively.

2.10 Percent Depth Dose

For a photon beam irradiating a patient or phantom, the absorbed dose in the patient or phantom will vary with depth in the medium. The central axis percent depth dose (PDD) is used to characterize this variation along the beam's central axis and is designated by the dose at a given depth $d$ in the medium normalized to the dose at depth
The percent depth dose is dependent upon the depth \( d \) of measurement in the phantom, the radiation field size \( A \) defined on the phantom surface by the collimator, the source-surface distance \( f \), and the energy \( h \nu \) of the beam, and is given by:

\[
PDD(d, A, f, h \nu) = \frac{D_p(d, A, f, h \nu)}{D_p(d_{max}, A, f, h \nu)} \times 100.
\] (2-11)

At points in the phantom starting at the surface and going down to the depth \( d_{max} \), the dose builds up until it reaches transient equilibrium at \( d_{max} \). Beyond \( d_{max} \) the dose continuously falls off due to photon attenuation of the beam. A few typical percentage depth doses for 10x10 cm\(^2\) fields of various photon beam energies are shown in Fig. 2.4. As the field size \( A \) of the beam increases, but \( d, f \), and \( h \nu \) are kept constant, more scattered radiation reaches the central axis and the percent depth dose increases accordingly. This effect, however, is less visible in higher energy beams since photons are scattered preferentially in the forward direction at higher energies. For constant \( d, A, \) and \( h \nu \), the percent depth dose dependence on the source-surface distance \( f \) is related to the inverse square law. Since the relative dose follows the inverse square law, the relative decrease in dose at two points close to the source will be greater than the relative decrease in dose at two points further from the source. Therefore, with constant depth \( d \), field size \( A \), and energy \( h \nu \), the percent depth dose will be greater for larger SSDs. Finally, for constant \( d, A, \) and \( f \), the higher the energy of the beam, the more penetrating the beam. Thus, as the energy of the beam increases, the dose at a fixed depth in the phantom will increase also increasing the value of the percent depth dose at that depth.

Higher energy beams not only have a deeper depth of maximum dose, they also have a lower surface dose. Ideally, the surface dose of megavoltage beams should be zero, however, electron contamination of the incident beam from photon interactions in air, collimator, and other objects in the path of the beam as well as backscattered radiation from within the medium contribute to the absorbed dose at the surface. Nevertheless, it is evident that the deeper the tumor, the higher is the photon beam energy required in order to deliver the highest relative dose at the depth of the tumor.
Fig. 2.4: Central axis percent depth dose curves in water for 10\times10 \text{ cm}^2 fields at an SSD of 100 cm for various energy beams. [Data from the British Journal of Radiology Supplement 25, "Central Axis Depth Dose Data for Use in Radiotherapy." (London: British Institute of Radiology), 1996.]

2.11 Beam Profiles

A radiation beam deposits its dose in a three dimensional volume. Determination of the percent depth dose only gives information about the beam's behavior along the central beam axis but gives no information for points off the central axis. However, the variation in dose across the field at a certain depth is measured by scanning across the radiation field at a fixed depth and the resulting dose distribution is referred to as a beam profile. For radioisotope beams, such as cobalt-60, the beam profiles generally exhibit a maximum on the central axis of the beam, a gradual decrease towards the edges of the beam, followed by a rapid decrease in the penumbra region at the outer edges of the beam. For linac x-ray beams, on the other hand, the profiles, as discussed below, are more complicated. Typical beam profiles for a 6 MV photon beam at various depths for a field size of 10\times10 \text{ cm}^2 are shown in Fig. 2.5.

From a beam profile, several beam parameters can be defined including the flatness and symmetry of the beam, the off-axis ratio, and the beam penumbra.
Fig. 2.5: Beam profiles for a 6 MV photon beam of field size 10×10 cm² at various depths in water.

A. Flatness and symmetry

Without a flattening filter in place on a linac, the isodose curves would be conical in shape due to the highly forward-peaked production of bremsstrahlung photons at high electron energies. However, a uniform dose across the entire field at a given depth in phantom is desired, and therefore, the conical shape of the isodoses is highly unwanted. The need for a uniform dose across the entire field infers that the x-ray beam produced in a linac target must be flattened in order to obtain a uniform x-ray intensity across the field at various depths in the phantom. Therefore the physical design of the flattening filter is thicker in the center and tapers off toward the edges of the field resulting in a conical shape. It should be noted that the radiation field size is usually defined as the lateral distance between the 50% isodose lines at a reference depth and can be obtained from a beam profile as depicted in Fig. 2.6. The field flatness of a photon beam is determined from the variation in dose values observed on the dose profile over the central 80% of the field size relative to the central axis, and is usually specified at a depth of 10 cm in water with a tolerance level of ±3% from the central axis value. However, to obtain the field flatness specifications at a depth of 10 cm, over-flattening of the beam at shallower depths and under-flattening of the beam at greater depths will be required. The over-
flattening of the beam, especially at $d_{\text{max}}$, will be observed as “horns” in the dose profile and special attention should be paid to the flatness of the profile measured at this depth to ensure that the “horns” do not exceed the central axis value by more than 5%.  

The symmetry of the beam may be considered by folding the beam profile about the central axis and comparing points from the two half-profiles. This is usually determined at $d_{\text{max}}$ and the values for any pair of off-axis points should not vary by more than 2%.

![Graph](image.png)

*Fig. 2.6: Measurement of the field flatness of a photon beam from a beam profile.*

**B. Off-axis ratio**

The off-axis ratio (OAR) is the ratio of the dose at a particular point of interest off the central axis to the dose on the beam's central axis at the same depth. Beam profiles are usually measured at a number of depths as required by the treatment planning system in order to obtain the appropriate off-axis ratios, and the data is stored in the treatment planning computer in the form of a matrix. This information along with the percent depth dose data is subsequently used to generate the isodose distributions of treatment plans.

**C. Penumbra**

At the edge of the radiation field the dose would ideally instantaneously drop to zero and the neighbouring healthy tissue would receive no radiation. However, in practice this is not the case, as depicted in Fig. 2.7. The penumbra is the term used to define the
region at the edge of a radiation beam where the dose changes rapidly with increasing
distance from the beam’s central axis. It consists of two components; the transmission
penumbra and the geometric penumbra.

The transmission penumbra is attributed to photons that have traversed part of
the thickness of the collimator. It varies as a function of the field size, the collimator and
phantom distance from the x-ray source, as well as the collimator edge shape. Evidently,
for larger field sizes, the angle between the ray extending from the source to the
collimator edge and the central axis increases, implying that the width of the transmission
penumbra will increase. At greater distances from the source the transmission penumbra
will also increase; however, in the case where the collimator edge shape is designed to
follow the divergence of the beam, the transmission penumbra is independent of these
factors and is consequently minimized.

The geometric penumbra, on the other hand, is due to the finite source size and
depends on the source diameter, the SSD, and the depth in the phantom, yet it is
independent of field size. A dose profile measured in air at a given distance from the
source will demonstrate the combined effect of the transmission and geometric penumbra.

Scattered radiation from within the patient or phantom will also contribute to the
dose fall-off at the field edges. For this reason, the physical penumbra is used to define
the region of high dose variation at beam edges and includes contributions from the
transmission penumbra, geometric penumbra, and scatter from within the patient. Thus,
the physical penumbra has been defined as the lateral distance between two specified
isodose curves, often 80% and 20% of the central axis dose at a specified depth. This physical penumbra determined from a dose profile at a given depth is illustrated in Fig. 2.7.

2.12 Summary

When an object is placed into the path of a photon beam, absorption and scattering of the beam results in a change of the quality and dose delivered by the beam. The absolute dose to any point in a patient or phantom can be calculated through various beam parameters, if the dose at a reference point is known. The variation of these parameters with beam energy, field size, depth in the patient or phantom, and distance from the source to the surface of the patient were discussed in this chapter.

The dose distribution in the irradiated volume is of interest in the treatment planning process. The selection and arrangement of radiation beams is crucial to the high dose delivery at the tumor, while maintaining as low a dose as possible to the surrounding healthy tissue. Tables of beam parameters allow for the facilitation of the tumor dose calculation. Furthermore, input of percent depth dose data for a range of field sizes as well as the off-axis ratios for a range of depths and field sizes into the treatment planning system for all available beam energies permits the observation of the isodose distribution of treatment plans. These parameters were derived for greater ease and accuracy of the treatment planning process.

2.13 References


3

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3.1 Introduction

Linear accelerators (linacs) are presently the most commonly used source of ionizing radiation in radiotherapy for the treatment of a range of malignant tumors. Since their introduction clinically in the early 1950's, advances in technology as well as computer power have enabled current medical linacs to provide greatly improved and more efficient treatment capabilities. In general, a linac produces a radiation beam by accelerating electrons to large kinetic energies of between 4 and 25 MeV. These electrons are subsequently used either directly to treat superficial lesions or they are made to strike a thick target in order to produce x rays for the treatment of deep-seated tumors. The various dosimetric parameters used to characterize photon beams were discussed in Chapter 2. Clearly, easy and accurate measurement of the amount of ionizing radiation produced by the radiation beam is crucial for the determination of these parameters, and measuring devices, referred to as dosimeters, are used for this purpose.

Consequently, this chapter deals with the way in which a clinical radiation beam is produced by a medical linac and also provides an explanation on how the amount of ionizing radiation was measured by the dosimeters used in this project. The dosimeters discussed here include the ionization chamber, film, and diode.
In order to accelerate a particle in a linac, three conditions must be met. First, the particle must have charge; second, there must be an electric field in the direction of propagation of the charged particle; and third, the velocity of the particle must be similar to the phase velocity of the wave in the waveguide. The charged particles accelerated by medical linacs are electrons and the electric field used is a high frequency electromagnetic travelling or standing wave, having a radiofrequency in the microwave region, typically 2856 MHz. This frequency is referred to as the S band and has a wavelength of 10.5 cm. A linac is a cyclic accelerator, meaning that even though the electrons travel in a straight trajectory, they cycle through the same fixed potential difference numerous times, thereby gaining kinetic energy with each cycle. In this fashion, a relatively low potential difference in the linac can generate electrons of very high kinetic energy.

The radiofrequency propagates through a hollow, cylindrical structure called a waveguide. The waveguide is either evacuated or contains a dielectric material such as sulphur hexafluoride, and the outer wall of the waveguide is made of a conducting material such as copper. Maxwell’s equations and boundary conditions as applied to uniform waveguides form the basis for the understanding of the propagation of the microwaves through waveguides. However, a uniform waveguide cannot be used to accelerate electrons due to the previously mentioned third condition in which the velocity of the particles must be similar to the phase velocity of the microwaves. Since the phase velocity of the waves in a uniform waveguide exceed the speed of light, yet particles cannot reach velocities greater than the speed of light, this condition cannot be met in a uniform waveguide. Consequently, obstacles in the form of irises (discs) are added to uniform waveguides causing interference with the microwaves and slowing the phase velocity down to a speed below that of the speed of light. Therefore, in a disc-loaded waveguide, referred to as accelerating waveguide, the electrons are able to follow the radiofrequency and acceleration of the electrons is possible.

The components of a medical linac essential for production of ionizing radiation as well as the modifications of this beam in order to generate a suitable clinical radiation
beam are depicted in Fig. 3.1. A description and explanation of the functioning of each component is given in subsequent sections.

Fig. 3.1: Schematic diagram of the components of a medical linear accelerator (linac).

A. Electron injection system

The electron injection system, referred to as an electron gun, is the source of electrons in a linac. The electron gun acts as an electrostatic accelerator that thermionically emits electrons from a heated cathode. The emitted electrons are accelerated by a potential difference through a perforated anode and enter the accelerating waveguide. A typical potential difference of 25 kV means that electrons exiting the electron gun will have a velocity equal to approximately one quarter of the speed of light. Even though the phase velocity of the disc-loaded waveguide has been lowered somewhat below the speed of light, the condition that the particle velocity equal the phase velocity cannot be met at the electron gun side of the waveguide. However, if a large enough peak electric field is provided by the radiofrequency waves, the waves can still capture the electron for acceleration. The minimum value of this peak electric field is governed by the so-called capture condition. For instance, according to the capture condition, an electron gun voltage of 25 kV requires a peak electric field of at least $1.2 \times 10^5$ V/cm. As the electron gun voltage decreases, the minimum value of the peak electric field in order
to still capture the electron increases and 25 kV is the current practical limit for the lowest possible potential on electron guns.

B. RF system

The RF system consists of a radiofrequency generator and the accelerating waveguide. The radiofrequency fields required by the waveguide for electron acceleration are produced by the microwave rf power generator, either a magnetron or klystron. While the magnetron acts as both the source and amplifier of microwaves, the klystron amplifies the low power microwaves produced by an rf oscillator, referred to as the rf driver. In either case, pulsed microwaves of 2856 MHz are produced through the following process. The cathode of the rf generator receives high voltage, high current, but short duration (on the order of a few microseconds) DC pulses which are produced by a pulsed modulator. Electrons are thermionically emitted from the cathode of the rf generator and accelerated toward the anode by the pulsed electric field. Resonant cavities cause the electrons to be either accelerated or decelerated, thereby grouping the electrons into bunches. Due to the conservation of energy, the kinetic energy of these bunches of electrons is converted into high frequency electromagnetic waves.

The peak power of operation for magnetrons and klystrons is typically on the order of several megawatts. Although klystrons are capable of generating larger peak power microwaves than magnetrons and are therefore generally used in higher energy linacs, they are also bulkier than magnetrons. Consequently, klystrons cannot be mounted in the gantry of a linac, as is the case with magnetrons and this makes the use of klystrons in linacs more cumbersome and complicated.

The large peak power requirements for the radiofrequency field to be capable of accelerating electrons to several MeV explain the necessity that linacs be operated in a pulsed mode rather than continuously. If the duration of the pulse is small compared to the pulse period, then the average rf power is on the order of several kilowatts. The ratio of pulse width and pulse duration is referred to as the duty cycle of a linac.

The high power radiofrequency is transmitted from the rf generator to the accelerating waveguide through a uniform waveguide. The transmitting waveguide is
Equipment

typically filled with a dielectric gas such as sulphur hexafluoride (SF₆). However, the accelerating waveguide must be evacuated so that accelerated electrons do not experience collisions with molecules of the dielectric gas and therefore must be separated from the transmitting waveguide by a special ceramic waveguide window. Additionally, a circulator mounted between the two waveguides absorbs any rf power that is reflected back toward the rf generator, thereby protecting the rf generator.

The high voltage pulses that are delivered to the magnetron or klystron by the modulator are also simultaneously delivered to the electron gun. This allows for electrons to be pulsed into the accelerating waveguide from the electron gun at the same time that the microwaves are also pulsed into the accelerating waveguide from either the magnetron or the klystron. As the electrons are injected into the accelerating waveguide, they interact with the electromagnetic field of the microwaves and gain kinetic energy as they propagate through the accelerating waveguide.

C. Auxiliary system

The auxiliary system consists of a vacuum pumping system, a water-cooling system, an air pressure system, and a gas pressure system. Due to the large electromagnetic fields that are used, the accelerating waveguide must be maintained at high vacuum to prevent collisions between accelerated electrons and any residual gas in the waveguide. A pressure on the order of 10⁻⁶ torr is sustained in the accelerating waveguide by the vacuum pumping system. The water-cooling system maintains components, such as the accelerating waveguide, bending magnets, x-ray target, and microwave power source at a relatively constant temperature. The transmitting waveguide is pressurized at about 2 atm with the dielectric gas sulphur hexafluoride (SF₆). This gas is chosen because it breaks down at a peak power of roughly ten times that of air or nitrogen. Finally, the air pressure system is used for the air driven mechanisms of the pneumatic system, such as the x-ray target.

D. Beam transport system

The high speed electrons that emerge from the accelerating waveguide are in the form of a pencil electron beam. This electron beam can either be made to strike a
scattering foil to spread out the pencil electron beam and produce a relatively large electron field, or the pencil electron beam can be made to strike a thick target to generate an x-ray beam. The electron beam transport system therefore consists of steering and focussing coils along with bending magnets which are used to transport the electron pencil beam from the accelerating waveguide to either the x-ray target or to the scattering foil.

The length of the accelerating waveguide depends on the required final kinetic energy of the electron beam, and ranges from roughly 30 cm for electrons of final kinetic energies of 4 MeV to roughly 150 cm for electrons of final kinetic energies of 25 MeV. For low energy linear accelerators up to 6 MV, the accelerating structure is short enough so that it can be mounted vertically in the gantry head and the electron pencil beam proceeds in a straight path to strike the x-ray target. However, it is impractical to mount the accelerating waveguide of high energy beams above 10 MeV in the gantry of the linac due to space restrains. An alternative solution is to mount the accelerating waveguide horizontally in either the gantry or in the gantry stand. Consequently, the electron beam does not follow a straight path in going from the waveguide to the x-ray target or electron scattering foil and a transport system that bends the electron beam through either 90° or 270° is required.

E. X-ray target

As previously discussed in Section 1.2.B, bremsstrahlung photons are produced when an electron beam strikes a target. To ensure that all incident electrons are stopped by the target and do not contaminate the photon beam, the target must be of a thickness greater than the practical range of the electrons striking that target. A heterogeneous bremsstrahlung spectrum of photon energies ranging from the maximum electron kinetic energy of the incident electrons down to zero results. For example, a medical linac can produce electron beam energies from 4 to 25 MeV. Consequently, photon beam energies of 4 to 25 MV result; however, it should be noted that a photon beam spectrum of 25 MV implies that the maximum photon energy is 25 MeV although the average photon beam energy is approximated to about a third of this value.
In the megavoltage energy range obtained with linacs, the forward peaked production of bremsstrahlung photons implies that the resulting photon beam is in the same direction as the electron beam just before it impinges on the target. It has been found that the production of x-rays in the forward direction is slightly greater and more energetic in low atomic number targets. However, for practical targets which must be reasonably thin to conserve space in the linac head, a compromise must be made between the low atomic number and high density of the target for high energy linacs resulting in copper as a suitable compromise.

F. Beam collimation and monitoring system

The beam collimation and monitoring system is made up of the components in the treatment head of the linac that shape and monitor the resulting clinical treatment beam. For the purposes of this project, only photon beams will be considered and therefore further discussions will be limited to the production of clinical photon beams.

The photon beam emerges from the x-ray target in all directions and is collimated by a primary collimator which restricts the beam to a large circular field. As previously stated, the bremsstrahlung production of photons is forward-peaked implying that the intensity of radiation is maximum along the central axis and declines towards the edges of the field. As the energy of the charged particles striking the x-ray target is increased, the production of photons is increasingly forward-peaked. However, a uniform intensity across the radiation field is desired and a flattening filter is therefore inserted into the path of the beam to achieve the desired flatness at a specific depth in the patient. The filter is thickest at the central axis and tapers off toward the edges of the field so as to attenuate the beam more in the central region and generate a more uniform intensity across the radiation field.

Section 2.2 explained that a high Z material placed in the path of a high energy photon beam (a few MeV and above) will soften the beam due to the Z dependence of the attenuation coefficient in the pair production region. Alternatively, a low Z material will hold the attenuation coefficient nearly constant and therefore have little effect on the penetrating power of the photon beam. Consequently, a low atomic number material, such as aluminum, is a preferred flattening filter material for high energy photon beams.
However, low atomic number materials also generally have a low density meaning that the large thickness required of the filter in order to achieve the desired flatness would increase the size of the treatment head of the linac. Examination of the effects of various $Z$ materials on the penetrating power of megavoltage photon beams\(^3\) demonstrate that copper presents a suitable compromise between relatively low atomic number and relatively high density.

Since the output of the linac is known to fluctuate throughout the course of the treatment, the flattened beam proceeds through dual transmission ionization chambers which continuously monitor the output of the beam. The chambers must be sealed in order to eliminate the effects of fluctuations in temperature and pressure of the outside air, and they must also have a minimal effect on the radiation beam. Due to the variations of the linac output, monitor units (MU) rather than units of time are used to set the desired dose to be delivered. The electrometer of the primary chamber is calibrated to read 100 cGy at a depth of dose maximum in water for a 10×10 cm\(^2\) beam at a source-surface distance (SSD) of 100 cm.

The flattened x-ray beam is further collimated by the secondary collimators consisting of a set of upper (Y) jaws and lower (X) jaws. These jaws are made of a high atomic number, high density material and are used to shape square or rectangular fields up to a maximum size of 40×40 cm\(^2\). The x-ray beam may be further modified by other beam modifying devices, such as wedges and beam blocks or, as further discussed in the following chapter, by multileaf collimators (MLCs).

A light localizing system consisting of a light source and a combination of mirrors projects the light beam as would be seen from the source. The light field is therefore equivalent to the radiation field and can be used to delineate the field size obtained by the linac collimators and beam modifying devices.

The linac used in the work reported in this thesis was a Clinac 2300 C/D (Varian, Palo Alto, CA) which is able to produce photons of energy 6 and 18 MV as well as electrons of energy 6, 9, 12, 15, 18, and 22 MeV. The linac is isocentrically mounted and can produce field sizes ranging from 0.5×0.5 cm\(^2\) to 40×40 cm\(^2\) through the use of the secondary collimator jaws. Additionally, the collimator jaws can be made to produce
asymmetric fields. The linac can generate a dose rate given in monitor units (MU) per minute between 100 MU/minute to 600 MU/minute in 100 MU/minute intervals for photon beams.

3.3 Dosimeters

Dosimeters are devices used to measure the radiation dose to a particular point in a medium and are subdivided into two categories. In the first category are absolute dosimeters in which the dose is determined directly, whereas the second group consists of relative dosimeters which must first be calibrated in a known radiation field. Absolute dosimeters consist of calorimeters, Fricke dosimeters, and the standard free air ionization chamber. Relative dosimeters, on the other hand, consist of calibrated ionization chambers, film, diodes, thermoluminescent dosimeters, optically stimulated luminescent detectors, and radioelectrets. In our measurements, ionization chambers, film, and diodes were used, and therefore further discussion will be limited to these devices.

A. Ionization chamber

The ionization chamber is the most commonly used device for clinical dosimetry measurements due to their relative ease of use, long term stability, high precision, and direct readout. The method in which they operate is as follows.

As a charged particle moves through a medium, it leaves behind a track of ion pairs resulting from the ionization of molecules within the chamber sensitive medium. The collection of these ion pairs creates an electronic signal that, when produced in a gas, can be measured by an ionization chamber. From the collected charge, it is possible to determine the absorbed radiation dose at the chamber reference point. However, the usefulness of an ionization chamber is based on the collection of all charges created by ionization within the chamber sensitive gas by all the charged particle tracks.

When a charged particle passes through the chamber gas medium, usually air, it ionizes and excites the molecules along its path. Once a neutral gas molecule is ionized, an ion pair consisting of a positive ion and a free electron results. In some gases, such as oxygen, there is a tendency for the free electrons to attach themselves to neutral gas
molecules to form negative ions. In order to ionize a molecule, the charged particle must transfer an energy at least equal to the ionization potential of the molecule which was stated in Section 1.2 to be on the order of a few eV. Electrons other than the least bound valence electrons may also be removed; however, for this to occur, a greater amount of energy must be transferred from the charged particle to the gas molecule. Additionally, the charged particle may lose some of its energy by excitation of the molecules and this does not contribute to the creation of an ion pair. Therefore, the average energy $\overline{W}$ that must be transferred from a charged particle to a molecule in order to create an ion pair is always greater than the actual ionization potential. For dry air the current value for $\overline{W}$ is taken as 33.97 eV/ion pair. Thus, if a 1 MeV electron is completely stopped in air, it will create about 30,000 ion pairs. Since $\overline{W}_{\text{air}}$ is assumed to be constant and independent of energy, the energy deposited in the medium will be proportional to the number of ion pairs formed and the radiation dose can therefore be determined by measuring the charge collected due to the ion pairs.

If the mass of air $m_{\text{air}}$ in the ionization chamber is known, then the absorbed dose to air $D_{\text{air}}$ in the chamber by definition will be:

$$D_{\text{air}} = \frac{Q}{m_{\text{air}}} \overline{W}_{\text{air}},$$

where $Q$ is the charge collected in the volume of air from the ionization chamber. The application of Bragg-Gray cavity theory$^5$ permits the determination of the absorbed dose in the medium in which the ionization chamber has been placed. The absorbed dose in water $D_{\text{water}}$ is obtained by:

$$D_{\text{water}} = \frac{Q}{m_{\text{air}}} \overline{W}_{\text{air}} S_{\text{air}} \left( \frac{\mu_{\text{abs}}}{\rho} \right)_{\text{water}},$$

where $S_{\text{air}}$ is the mean stopping power ratio in going from the mass of air in the ionization chamber to the wall material of the ionization chamber and $\left( \frac{\mu_{\text{abs}}}{\rho} \right)_{\text{water}}$ is the
mean ratio of mass energy absorption coefficients in going from the wall material of the ionization chamber to the water medium.

Fig. 3.2 displays a schematic diagram of a typical parallel-plate ionization chamber with two main electrodes: the measuring (collecting) electrode and the polarizing (biasing) electrode. An external voltage is applied to the two electrodes, creating oppositely charged plates. As a photon interacts with either an air molecule or the upper plate, it creates a track of positive and negative ions, expending 33.97 eV with the creation of each ion pair. The positive and negative charges drift to the oppositely charged plates, generating an electric current. The guard electrodes not only define the sensitive volume of the ionization chamber due to the straight electric field lines in this region but they also prevent the measurement of any leakage current between the two electrodes. Assuming that as many electrons enter the sensitive volume as leave, so that electronic equilibrium is maintained, the negative ions within the sensitive volume will grab positive charges from the lower plate to become neutral air molecules again. As a result, the lower plate has an excess of negative charges which are removed and go to ground due to the power supply. If the polarity of the power supply is reversed, this current will flow in the opposite direction. Furthermore, some photons may land and interact on the measuring electrode, ejecting electrons from the plate. These electrons are replaced by the power supply and a current \( i_c \), independent of the power supply polarity, is set up. Therefore, the overall electronic current detected by a sensitive ammeter, called an electrometer, is a combination of ionization current \( i \) and \( i_c \) and will not be the same for the two polarities. The true measurement is thus taken as the average value between the positive and negative polarities. However, when a build-up cap or thickness of material equal to the depth of dose maximum \( (d_{\text{max}}) \) of the photon beam is placed above the upper electrode, electronic equilibrium is achieved so that the number of photons that interact on the lower electrode and eject an electron is compensated by the number of electrons that just land on the lower electrode before losing all their energy. Thus, beyond the depth of dose maximum, the overall current detected will be the same irrespective of the chamber polarity. This trend is illustrated in Fig. 3.3 which shows schematically the measured energy deposition in water from a photon beam.
Fig. 3.2: Schematic diagram of a typical parallel-plate ionization chamber.

Fig. 3.3: Observed percent depth dose for the positive and negative polarity where \( i_+ \) and \( i_- \) represent the positive and negative polarities, respectively.

i. **Recombination**

The absorbed dose deposited in the ionization chamber gas is a function of the charge produced in the gas. However, if not all of the charges are detected by the measuring electrode, the measured dose will be lower than the actual absorbed dose. This occurs either when a positive ion and a negative ion collide, the extra electron being transferred to the positive ion, or when a positive ion and a free electron recombine. In
both instances, the ions return to neutral air molecules and the charge represented by the original ion pair is lost so that it cannot contribute to the signal of the detector.

Generally, if no voltage is applied across the chamber electrodes, no net current will be measured as a result of the absence of an electric field. The ion pairs created will simply recombine and disappear. However, as the applied voltage increases, it generates a stronger electric field, and in turn, the charges are collected more rapidly by their respective electrodes. The opportunity for recombination is reduced and the charge collected is more representative of the actual charge generated in the sensitive volume. However, above a certain voltage, the ionization current can no longer increase once all the charges created are collected and no recombination occurs. Thus, the ionization current first increases, almost linearly, and then more slowly as it approaches saturation. On the other hand, if the voltage is increased too much beyond the saturation region, the ions can gain enough energy to ionize the gas molecules on their own, resulting in charge multiplication. This detector is then known as a proportional counter. At even higher voltages, the electron multiplication is even greater, and the number of electrons collected is independent of the initial ionization. This detector is called the Geiger-Müller counter, in which the large output pulse is the same for all photons. At still higher voltages, electrical breakdown of the capacitor will occur. Evidently, in radiation dosimetry it is imperative to operate the ionization chamber in the saturation region. A voltage of typically 300 V/mm of plate separation is high enough so that all the charges produced are collected but not sufficiently high that the accelerated electrons themselves create more ionizations. In photon beam calibrations with ionization chambers any minute charge losses to ionic recombination must be accounted for. A recombination correction factor $P_{ion}$ corrects the ionization chamber reading for the uncollected charges produced and for pulsed beams is given by:

$$P_{ion} = \frac{1 - \frac{V_H}{V_L}}{M_H/M_L - \frac{V_H}{V_L}}.$$  

(3-3)

$V_H$ is the chamber's usual operating voltage and $M_H$ is the corresponding chamber reading. $V_L$ represents a reduced bias voltage, which must be at least a factor of two lower than $V_H$, and $M_L$ is the corresponding chamber reading once it has reached equilibrium.
ii. **Thimble chamber**

A cross-section of a typical thimble chamber is illustrated in Fig. 3.4. The chamber consists of a spherical or cylindrical air cavity surrounded by a solid, air-equivalent wall. The cavity, generally with a sensitive air volume between 0.1 and 3 cm³, is assumed to be uniformly irradiated by a photon beam. In addition, the wall thickness is assumed to be equal to the maximum range of electrons generated in air so that all electrons that produce ionizations in the air cavity originate in the wall. In order for electronic equilibrium to exist, the wall thickness must be such that the number of electrons entering the cavity is equal to the number of electrons leaving the cavity. In practice, thimble chambers are constructed with wall thicknesses of roughly 1 mm; however, buildup caps of the appropriate thickness are added to achieve the desired thickness depending on the energy of the photon beam. By measuring the ionization charge produced in the cavity by electrons which are liberated in the wall, the exposure at the center of the cavity can be determined, if the volume or mass of air inside the cavity is known.

![Fig. 3.4: Rough schematic diagram of a thimble ionization chamber.](image)

An insulating material must be used between the two electrodes as support. Since ionization currents are extremely small, on the order of $10^{-12}$ A, the leakage current through these insulators must be kept very small so as not to add to the measured ionization current and create erroneous readings. A guard ring is therefore used to reduce the effects of insulator leakage on measured currents or charge.
B. Film

Radiographic film consists of a thin transparent base of polyester or cellulose acetate coated with an emulsion made of very small silver bromide crystals. The crystals undergo a chemical change when the film is exposed to either visible, ultraviolet, or x-ray photons. This reduction of the crystals is called a latent image. When the exposed film with its latent image is processed, it goes through a developer which reduces the modified silver bromide crystals to small grains of metallic silver. The film then goes through a fixer which removes the unaffected silver bromide crystals leaving a clear film in their place. The metallic silver grains are unaffected by the fixing solution and cause a darkening of the film. The degree of darkening of the film is related to the concentration of silver granules deposited, and hence is related to the amount of radiation absorbed. The radiographic image is formed because of difference in attenuation between the different components of the object to be radiographed. For example, bone can be seen on radiographs because it attenuates or stops more x-rays per given thickness than the surrounding soft tissues, or the heart can be seen on thoracic radiographs because it attenuates more x-rays than the surrounding lungs.

The amount of darkening of the film can be measured with a densitometer which detects the optical density. The optical density \( OD \) is defined as:

\[
OD = \log \frac{I_0}{I},
\]

where \( I_0 \) is the amount of light incident on the film and \( I \) is the amount of light transmitted through the film. This implies, for example, that if the film transmits 1/10 of the incident light, it will have an optical density of 1.0. However, an unexposed developed film has a characteristic optical density referred to as the base+fog. Therefore, in radiation dosimetry, the net optical density which eliminates the base+fog is of interest. A plot of the net optical density versus known radiation doses characterizes the response of the film to radiation and is known as the sensitometric curve. As the exposure is increased, the optical density initially increases roughly linearly but eventually reaches saturation. Saturation occurs when the unexposed silver bromide crystals are used up. A fast film implies that it requires a lower exposure to achieve a certain optical density.
compared to a slow film. Faster film is generally used for diagnostic purposes while slower film is used for radiation therapy verification.

As stated in Section 1.3.A, the mass attenuation coefficient for the photoelectric effect is proportional to the cube of the atomic number. Therefore, for silver, which has an atomic number of 45, the film emulsion absorbs radiation very strongly by the photoelectric effect for photons of energy less than 150 keV, especially near the K-edge of silver at 25 keV.

Accurate representation of the dose in the penumbra region may be a problem because of the size of available ionization chambers which may be large relative to the penumbra width. Better spatial resolution can be obtained with film, although concern regarding the energy dependence of film due to the high atomic number components in film may create an over-response of film to low energy radiation. The over-response would result in systematic errors in the measured dose from scattered radiation in the tail of the penumbra and can amount to a few percent. Therefore, care should be taken when quoting relative doses lower than 10% of the central axis dose in the penumbra region.

In addition, changes in processing conditions can limit reproducibility of results and air pockets adjacent to the film emulsion may also cause errors. However, in the megavoltage range, film dosimetry has been found to agree with ionometric measurements to within 3%\(^7,8\), although variations of up to 5% may occur when film is developed at different times.\(^9\) The film must be tightly packed within a phantom and positioned parallel to the central axis of the beam. The edge of the film must be aligned with the surface of the phantom. The optical densities are correlated with the dose using a depth-dependent sensitometric curve. Depth-dependent curves are required because at large depths, the lower energy photons undergo greater attenuation and the photon energy spectrum shifts towards higher energies. Therefore, the higher sensitivity of the film to lower energy photons may create differences in the calibration curves at shallow and deeper depths. However, in megavoltage beams at shallow depths, that is, below 20 cm in phantom, no apparent sensitivity of the film to beam hardening is observed and the calibration curves are therefore independent of depth.\(^10\) Furthermore, investigations show
that for increasing beam energies, the depth dependence of the photon spectra decreasingly affects the film's response per unit dose with depth.\textsuperscript{11}

Film has the advantage of providing a convenient and rapid means for obtaining a complete set of isodose curves. It also is convenient for checking the size and shape of the radiation field, as well as for measuring the size of the penumbra around a field, the leakage radiation around collimators, the positioning of special radiation shields, and dose distributions in the buildup region of a beam.

C. Diode

The diode is a solid state detector which measures the amount of radiation by the ionization produced in a solid. The advantage of diodes is their small size and high sensitivity to radiation. Their instantaneous electronic response makes them suitable for scanning devices. Furthermore, their large signal makes the electrometer design much simpler than for ionization chambers. However, diodes exhibit an energy and temperature dependence and can be damaged by radiation, and therefore dose distributions with diodes should be verified against ionization chamber measurements.

The main component of a diode detector is a 1 mm thick slab of lightly doped silicon. The silicon, which has an atomic number of 14, causes the diode detector to have a much greater energy dependence than an air ionization chamber and is therefore not a practical device for measuring beams containing photons of many different energies. The energy dependence is a result of the decrease in the ratio of the energy absorption coefficient $\mu_{\text{abs}}$ of silicon compared to air by a factor of 4.6 for photon energies between 60 keV and 1.25 MeV of Co-60. However, the density of silicon is roughly 2000 times that of air and the average energy that must be transferred from a charged particle to silicon in order to create an ion pair is about 3.5 eV per ion pair, nearly a tenth of that for air. Therefore, a silicon diode detector is about 20,000 times more sensitive than an ionization chamber of the same volume.
3.4 Three Dimensional Water Scanner

A three dimensional water scanner provides a fast and convenient means for measuring the dose distribution in a water phantom. Beam profiles and percent depth dose curves can be plotted in real time through computer interface and stored as either curves or tables. The data can afterwards be exported to the treatment planning system or saved as an ASCII file.

The RFA-300 Radiation Field Analyzer (Scanditronix, Upsalla, Sweden) used in this work consists of a 50×50×50 cm³ water-filled plastic tank in which an arm to hold the radiation detector can move in the x, y, and z planes. It is a computer-based system running with Windows 3.1. Once the origin is set at the isocenter of the treatment machine, the detector can be moved to any desired coordinate within the water tank. However, attention must be paid in order to ensure that the specified range of the scan does not exceed the boundaries of the tank, otherwise the chamber will collide with the plastic wall.

The radiation detectors used in our experiments, both sealed and waterproof, were RK ion chambers as well as RFA diode detectors. For both types of detectors, the field chamber was mounted on the automatic arm while a reference chamber was placed at the edge of the radiation beam near the head of the linac. The ratio of the signals from the field and reference detector was used as the chamber reading to exclude effects of the variation of machine output with time.

3.5 Summary

Linear accelerators (linacs) are the most commonly used source of ionizing radiation in modern radiotherapy. In clinical linacs, high frequency electromagnetic travelling or standing waves accelerate electrons through a disc-loaded waveguide to energies between 4 and 25 MeV. The high speed electrons, after passing through various beam modifying devices, can be used to treat superficial tumors or the electron beam can be made to strike a thick target in order to generate a megavoltage photon beam for the treatment of deeper-seated tumors.
Measurement of the amount of ionizing radiation produced at a point in a water phantom is obtained by various devices referred to as dosimeters, each having their own advantages and disadvantages. Precision and accuracy of the measurement of beam parameters are essential for clinical applications.

3.6 References


4

Physical Characteristics of an MLC and micro-MLC

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4.1 Introduction

The goal of radiotherapy is to provide a high dose to the tumour while maintaining as low a dose as possible to tissues surrounding the tumour. Hence, in order to minimize the amount of radiation received by the neighboring healthy tissue, irregular field-shaping of the beam is usually required. Conventionally this has been achieved with the collimator jaws accompanying custom blocks that are placed by hand below the jaws. These blocks may be either a set of lead blocks that range in shape and size or custom-made cerrobend blocks that are each fabricated specifically for one given irregular field. Since three dimensional conformal radiation treatments are complex and frequently involve a large number of fields, a particular patient’s treatment may require the
moulding of multiple blocks thus making the block fabrication process time consuming and labour intensive. Following the fabrication, each block is mounted onto a tray. The tray fits into a slot on the head of the linac and enables the blocks to be mounted in a fixed position for each treatment field. However, between treatment fields the therapist must remove the mounted block from its slot and replace it with the block for the subsequent field. Furthermore, the trays may be heavy and injury to the patient and/or therapist may result if a block is accidentally dropped or falls. Modification to a block once it is fabricated means that the block must be sent back to the mould room involving both additional treatment room time and mould room time. Another drawback of the use of blocks is that cerrobend and lead are potentially toxic materials and can be hazardous to the machinists moulding these blocks.

To overcome some of these drawbacks, multileaf collimators (MLCs) were introduced in the early 1990s to replace conventional blocks for irregular field shaping. The MLC, as depicted schematically in Fig. 4.1, consists of a large number of narrow, closely-abutting leaves that are each positioned independently through computer control to form an arbitrarily-shaped field. For each treatment field an MLC computer file is generated containing the positional settings of each MLC leaf. Once the initial set-up of the patient in the treatment room is complete, all fields can be treated sequentially without the therapist having to re-enter the treatment room. Minor modifications to the field shape can be made on the spot by simply modifying the leaf setting in the computer file. All this considered, treatment efficiency was greatly improved with the introduction of multileaf collimators.\textsuperscript{1,2} However, on the down side, the use of MLCs requires a sophisticated computer control system as well as expanded dosimetry calculations. The finite leaf width in addition to the leaf end and side geometry must be analyzed and its ramifications incorporated into the treatment planning computer software.

In addition to their use as simple replacements to beam blocks for static field shaping, MLCs may also be used for dynamic beam shaping as the linac gantry is rotated or to achieve intensity modulation of the beam.\textsuperscript{3}
4.2 MLC as a Tertiary Collimator

The leaves of an MLC are permanently inserted into the head of the linac and, depending on the manufacturer of the linac, the placement of the leaves varies. For example, Philips, Siemens, Scandatronix, and GE insert the MLC leaves as a replacement to either the upper or lower photon jaw. Varian, on the other hand, uses the configuration illustrated in Fig. 4.2 in which the MLC leaves are placed below both the upper and lower secondary collimator jaws. The MLC is thus referred to as a third level or tertiary collimator. This configuration allows the leaves to be retracted and treatments to continue with standard upper and lower jaw collimation when there is a malfunction of the MLC system.

![Diagram of the MLC leaves as seen by the radiation source.](image)

*Fig. 4.1: Diagram of the MLC leaves as seen by the radiation source.*

The addition of the MLC below the standard jaws places the block tray at a distance of 34.6 cm from the isocenter while the distance from the radiation source to the bottom of the leaves is 53.4 cm. This leaves a clearance from the bottom of the leaves to isocenter of 46.6 cm. Placing the leaves of a tertiary MLC below the standard jaws further increases the distance from the leaves to the x-ray target compared to other configurations. As a result, a greater physical travel distance of the leaves is required in order for the leaves to move from one side of the field to the other. Consequently, the length of the leaves must be longer compared to other configurations. The end result is an increase in the size of the treatment head of the linac making some treatments no longer
possible. Thus, the major drawback of a tertiary collimator is the added bulk to the linac gantry and the reduced clearance to isocenter.

![Diagram of the Varian MLC as tertiary collimator](image)

_Fig. 4.2: Schematic of the Varian MLC as tertiary collimator._

### 4.3 Physical Properties of MLCs

The MLC leaves are made of a tungsten alloy. This material is used because it has a high density, it is hard, readily machinable, and fairly inexpensive. Tungsten alloy has the advantage of a low coefficient of expansion which allows it to be machined to exacting tolerances. This is an important factor with respect to the separation between adjacent leaves of the MLC since radiation leakage is highly undesirable yet a minimum spacing between the leaves is necessary in order to reduce friction as the leaves move adjacent to one another.

The width of each individual leaf determines how accurately a beam will conform to the continuous contour of the target shape. Evidently the narrower the leaf width, the better the conformity. On the other hand, the more narrow the leaf width, the greater the number of leaves required to achieve a certain field size. Since each leaf is
independently driven by a separate motor, the mechanical complexity of the MLC system will increase with the number of leaves. Consequently, a compromise must be made between the complexity of the mechanical device and the leaf width by setting a limit on the number of leaves of the MLC system.

The MLC leaf bank used in our project consists of 52 opposing tungsten alloy leaves. Each side of the leaf bank is denoted as either the A or B carriage and each leaf provides a 1.0 cm wide and 16.0 cm long projection at the isocenter. Each leaf can translate from 20 cm off the central axis to 16 cm across the central axis giving a maximum field size of 26×40 cm². The leaf span range is limited to 14.5 cm meaning that the distance between the most extended and most retracted leaves from either carriage cannot exceed this value. Thus, for example, no leaf from either carriage A or B can be set to a value greater than -16.0 cm (the minus sign denoting that the leaf is across the central axis). However, if one leaf or numerous leaves from carriage A are set to -16.0 cm, then the minimum setting of any other leaf from carriage A is -1.5 cm. A positive leaf setting implies that the leaf is retracted away from the central axis. Furthermore, if the X2 collimator jaw is not set to align with the central axis or extend over the central axis, a gap behind the carriage A leaves will appear that will allow radiation to be transmitted. In this case, an interlock will be tripped preventing the treatment from proceeding until the secondary collimator jaws appropriately shield the backends of the MLC leaves.

Since MLCs are used to replace custom made beam blocks, the combination of the transmission through the leaves and the leakage radiation between the leaves of the MLC system should not exceed transmission through a typical shielding block. However, the overall field size of a tertiary multileaf collimated field is defined by the secondary collimator jaws, just as is the case if a custom beam block is used instead. For this reason, the leaves of the MLC system are required to attenuate the primary beam to the same value as a custom beam block. Since custom blocks attenuate the beam to less than 5% of the primary beam, meaning between 4 and 5 half value layers of tungsten, the leaves of the MLC system must meet this criterion. However, since there is some leakage between adjacent leaves due to their finite spacing, the thickness of the leaves should compensate for this by having a transmission lower than 5% in order to ensure that the overall transmission meets this requirement. A leaf thickness of 5.40 cm of tungsten alloy has
been found suitable for this purpose. The transmission could be reduced by a further factor, however, the additional leaf thickness would add to the size of the gantry and further reduce the distance from the bottom of the leaves to the isocenter. Hence, a compromise between radiation transmission and the clearance to isocenter is made in modern linac design.

4.4 Leaf Side Design

Two requirements were addressed in the design of the cross-section of the leaves. First, the leaves must be focused in the plane orthogonal to their travel in order to follow the divergence of the beam. To meet this requirement, the leaves were designed with the truncated pie shape depicted in Fig. 4.3(a). This means that for a leaf thickness of 5.4 cm and a leaf width projection at isocenter of 1.0 cm, the physical width of the top and bottom of the leaf must be 0.48 cm and 0.53 cm, respectively. This implies that the width of the bottom of the leaf must be only 0.5 mm greater than the width of the top of the leaf in order to follow beam divergence. The second requirement addressed in the design of the leaf cross-section entails minimizing interleaf transmission by somehow overlapping adjacent leaves so as to prevent any ray from having a direct path between the leaves. This was met by the tongue and groove profile of the sides of the leaves as illustrated in Fig. 4.3(b).

4.5 Leaf End Design

The ends of the secondary collimator jaws follow beam divergence by moving along an arced path. This design, however, is difficult to implement when applied to the leaves of the MLC because each of the leaves must be individually controlled. Varian and Philips use the simple alternative of restricting the leaf movement to a plane perpendicular to the central axis of the beam. Since in this case the leaf ends do not follow beam divergence and only the sides of the leaves do, the system is single focused, generating two concerns. First, the leaf end penumbra will be larger than that of a divergent system, and second, the penumbra of the MLC system may increase as the leaves are moved away from the central axis. To counterbalance these two effects, the leaf ends are rounded in order to maintain the rays of the divergent beam tangent to the
leaf ends as they move in and out of the field. Actually, the design is not smoothly curved but has two 'flat' tangential portions as seen in Fig. 4.4.

Fig. 4.3: (a) End view of a Varian MLC leaf bank demonstrating the divergence of the leaves through a truncated pie shape; (b) Front view of the leaf demonstrating the tongue-and-groove design.

Fig. 4.4: Side view of a Varian MLC leaf demonstrating the curved end design.

4.6 Light Field vs. X-ray Field

Because of the rounded leaf ends, the physical positions of the tips of the leaf ends do not correspond to the projection of the light field edge or the radiation field edge at isocenter. The edge of the light field is determined by the ray tangent to the outermost
point of the leaf end as seen by the center of the source. As the leaf moves away from central axis, the outermost point of the leaf end seen by the source will move around the leaf end. On the other hand, the field size is defined by the position of the 50% isodose line from a dose profile. Thus, the radiation field edge is defined by the ray that passes through one half-value-layer thickness of the leaf material. Therefore a correction must be applied to the MLC control system when setting the field edge.

Fig. 4.5 demonstrates the difference between the projection at isocenter of the light field edge and the radiation field edge for three different leaf end positions. The lower case letters refer to positions along the plane through the true physical center of the leaves which also correspond to the position of the leaf tips, and the upper case letters refer to positions along the plane of projection of the leaves at isocenter. The distance from the x-ray source to the isocenter is given as SAD while the distance from the source to the center of the leaves is given as SCD. If the leaf is physically moved from position c to e, it travels a distance \( w_i \) along the plane at SCD while its projection travels a distance \( W_i \) along the plane at SAD. Considering the similar triangles, we get

\[
W_i = w_i \cdot \frac{SAD}{SCD}. \tag{4-1}
\]

However, from Fig. 4.5 it is evident that the edge of the light field projection will correspond to \( X_i \) not \( W_i \), and as the leaf tip is moved further off central axis the difference between \( X_i \) and \( W_i \) will increase. The light field edge projection has been found to be up to roughly 5 mm greater than the projection of the leaf tip at isocenter. To determine this non-linear difference the origin is set at the isocenter and the radius of curvature of the leaf ends to \( R \). If \( \theta' \) represents the angle Sdc then the angle between the line \( R \) and the plane at SCD is also \( \theta' \). Therefore, point e (where \( x_i' = w_i \)) is at \( z = SAD - SCD \) and point d (where \( x_i' = w_i + R - R\cos \theta' \)) is at \( z = SAD - SCD - R\sin \theta' \). If \( \theta \) represents angle Sce, then \( \theta \) can be assumed to be roughly equal to \( \theta' \). Consequently, if \( \tan \theta' = \tan \theta = \frac{W_i}{SAD} \), then the projection of the light field edge is given by:

\[
X_i = x_i' \cdot \frac{SAD}{SCD + R\sin \theta}. \tag{4-2}
\]

Substituting the expressions for \( x_i' \) and \( w_i \) into Eq. (4-2) we get
\[ X_i = \left( w_i + R - R \cos \theta \right) \cdot \frac{SAD}{SCD + R \sin \theta} \]

\[ = \left[ w_i \cdot \frac{SCD}{SAD} + R(1 - \cos \theta) \right] \cdot \frac{SAD}{SCD + R \sin \theta} \]

\[ = \frac{w_i \cdot SCD + R \cdot SAD(1 - \cos \theta)}{SCD + R \sin \theta}. \quad (4-3) \]

The position of the projection of the edge of the light field with respect to the position of the projection of the leaf tip, \( W_i \), will therefore be given by

\[ X_i = \frac{w_i \cdot SCD \pm R \cdot SAD \left(1 - \frac{SAD}{\sqrt{SAD^2 + W_i^2}}\right)}{SCD \pm \frac{R \cdot W_i}{\sqrt{SAD^2 + W_i^2}}}. \quad (4-4)\]

The positive sign is used when considering leaves from the A carriage (i.e., leaves moving toward the central axis from the positive x direction) while the negative sign is used when considering leaves from the B carriage (i.e., leaves moving toward central axis from the negative x direction). This expression decreases the difference between the light field edge projection and the projection of the tip of the leaf end to a maximum of around 1 mm. The Varian MLC system uses this relationship to make small position corrections that account for the non-linearity of the light field projection. However, as stated previously, the radiation field edge projection does not correspond to this relationship. It has been found that the difference between the radiation field edge and the light field edge is a small and nearly constant value of 1 mm and can therefore be considered clinically negligible.3

In the Varian MLC system, a look-up table that states the relative values for the radiation field edge and leaf travel is stored in the MLC controller. The amount of leaf travel required to move a leaf to the radiation field edge can be determined from the measured relationship.
4.7 Creation of MLC Files

The MLC files containing the leaf positions for each field are created either by hand using the Shaper software (Varian, Palo Alto, CA), through digitization of the target shape from a radiographic image (simulation film or digitally reconstructed radiograph), or directly through a 3D treatment planning system. The MLC leaf settings are stored in a computer file that can be transferred over the network system to the MLC controller and workstation in the treatment room prior to treatment. Although the MLC system is independent of the rest of the accelerator, the MLC system is aware of the jaw settings and collimator angle of the accelerator. For example, if there is a gap between the back of a leaf and the collimator jaws, an interlock will not allow the treatment to proceed. In addition, prior to treatment a verification that the light field projection of the MLC-shaped...
field corresponds within 2 mm of either the original shaped field drawn on the simulation film or the digitally reconstructed radiograph (DRR) placed at the appropriate SSD should be performed.

4.8 Leaf Positioning

Geometric methods within the software allow positioning of each leaf with respect to the continuous contour of the target volume. Three leaf coverage strategies may be employed: the ‘out-of-field’, the ‘in-field’, and the ‘cross-boundary’ strategy as shown schematically in Fig. 4.6. The ‘out-of-field’ technique avoids shielding any part of the target volume but some regions of healthy tissue near the boundary will not be shielded. The ‘in-field’ technique is conservative with respect to normal structures that abut the treatment volume but an underdosing of the target volume under the leaves will result. Finally, a compromise approach is the ‘cross-boundary’ technique. One condition for optimizing the leaf position has been the criterion that the ‘in-field’ area be equal to the ‘out-of-field’ area.

Fig. 4.6: The three classes of leaf placement strategies where (a) represents the out-of-field technique, (b) the cross-boundary technique, and (c) the in-field technique.

The physical linear position of each leaf is detected by position encoders within the MLC system. These position encoders are moved by the individual electric motors of each leaf and the leaves can be positioned to an accuracy of ±0.1 mm. An uncertainty of any leaf greater than this will trip the system interlock and treatment will be unable to proceed. An infrared beam within the MLC system allows the system to determine the position of each leaf and if a deviation of greater than 0.1 mm is detected between the measured and expected leaf positions, an error message will occur.
4.9 Optimization of MLC Field

The optimum closeness of fit of the MLC stair-step aperture shape compared to the continuous aperture shape of the target area is dependent on several factors, most important being the collimator angle. Optimization of the MLC field is achieved by placing the largest number of leaf ends tangent to the field edges without changing the target area. Fig. 4.7 demonstrates how this may be achieved by optimizing the collimator angle. This angle is obtained automatically through a computer algorithm that scans through all collimator angles and selects the best angle. Optimization may also involve varying additional treatment machine parameters such as couch and gantry angle.

Fig. 4.7: Diagram depicting the optimization of collimator angle where (a) demonstrates the optimized angle while (b) demonstrates the leaf placement prior to optimization of collimator angle.

4.10 Properties of the micro-MLC

It was previously stated that the width of the individual leaves determines how accurately a given beam shape can be resolved. Obviously, the narrower the leaf width the better the resolution yet a compromise between the mechanical complexity of the device and the number of leaves must be made. Consequently, a limit is set on the number of leaves that make up the multileaf collimating system. In order for the MLC to accommodate fairly large, irregular shaped tumors, a leaf width of 1.0 cm at isocenter is required to achieve an adequate overall field size. However, this large leaf width relative
to small targets such as in the brain or head and neck region make it difficult for the MLC to accurately conform to these targets.

Consequently, the need for a multileaf collimating system with a smaller leaf width to accommodate such fields is apparent. The recent advent of the micro-multileaf collimator (micro-MLC) is presently becoming an additional feature of modern megavoltage radiotherapy machines.

The BrainLAB micro-multileaf collimator designed for the Varian series linear accelerators was a joint development project between BrainLAB (Germany) and Varian (USA). Apart from the leaf width, the engineering design and operation characteristics of the micro-MLC are basically identical to those of the standard Varian MLC. The micro-MLC is detachable, weighing about 28 kg, and can be attached to the head of the linac either with or without the standard MLC already installed. The leaf width of the micro-MLC is smaller and varies with the intent of achieving superior shaping around smaller targets at the center of the field area. Like the standard MLC, each leaf is individually computer-controlled, however, the leaves of the micro-MLC move perpendicular to the conventional MLC. That is, the leaves of the standard MLC move parallel to the lower X jaws while the micro-MLC leaves move parallel to the upper Y jaws.

Like our standard Varian MLC, the BrainLAB micro-MLC has 26 pairs of tungsten leaves. The leaf widths are variable, with the central fourteen pairs providing a 3.0 mm projection at the isocenter, the middle six pairs giving 4.5 mm, and the outermost six pairs giving 5.5 mm.

Up to a certain limit, the smaller leaves can provide greater sparing to the normal tissue without compromising dose conformity to the target. The tungsten leaves can be precisely machined in widths of less than two millimeters and integrated into a micro-MLC allowing for minimal transmission through the leaves and minimal leakage between adjacent leaves.

Leaf ends and cross sectional shape of the micro-MLC are somewhat different compared to those of a standard MLC. A more sophisticated “tongue and groove” leaf cross sectional design is necessary to allow the drive shaft to be inserted into each leaf for
movement purposes (see Fig. 4.8). Adjacent leaves have this drive shaft inserted in vertical increments permitting optimum positioning of each 10 mm diameter driving motor.

**Fig. 4.8**: Front view diagram of the sophisticated tongue and groove design of the micro-MLC leaves.

The leaf end is milled to 3 angled straight edges each covering a third of the leaf edge length. These edges correspond to the fan line divergence of the beam when the leaf is fully extended (+5 cm), centered, and fully retracted (-5 cm). The central edge is parallel to the opposing leaf's central edge so that the leaves meet when they are closed. Each leaf is also constructed to diverge in cross section, the same as the divergence of the beam across the field area.

### 4.11 Summary

Multileaf collimators (MLCs) are becoming a standard feature for beam shaping of modern megavoltage radiotherapy machines. They consist of a large number of narrow, closely-abutting leaves, each independently controlled and positioned through the MLC controller computer. The physical characteristics of a tertiary multileaf collimator must ensure that its dosimetric characteristics resemble those of a standard beam block. Consequently, special consideration was given to the design of the leaf end and side shape. More recently, the advent of a micro-multileaf collimator (micro-MLC) with a smaller leaf width can better conform to the smaller targets of regions such as the brain and the head and neck.

### 4.12 References


5.1 Introduction

Implementation of new high technology equipment in a clinic requires that the device go through acceptance testing and commissioning before it is used clinically. Acceptance testing verifies that the new equipment, the MLC and micro-MLC in our case, performs to manufacturer's specifications. For example, the maximum travel of an MLC carriage across the central axis is stated to be 16 cm while the maximum leaf span range is stated to be 14.5 cm for a Varian MLC. Knowledge of these factors is important, since they determine the extent to which the MLC can define large, irregular fields, as well as the possible placement of abutting leaves off central axis. Leaf transmission is stated to be less than 4% and should be investigated for various leaf abutting positions. Some additional acceptance testing may include checks of light field and mechanical axes alignment, alignment of the MLC relative to the secondary collimators, leaf calibration,
leaf positional reproducibility, as well as a check of active interlocks for leaf and jaw positional tolerances. Rotational checks are necessary to ensure that the added weight to the head of the linear accelerator, due to the addition of the MLC, does not distort the isocenter position. Ideally, the isocenter of the machine would be an infinitesimally small point at which the gantry, collimator, and couch axes of rotation intersect for all gantry, collimator, and couch rotations. However, in practice these three axes do not intersect at a single point but instead define a sphere or ellipse whose radius should be within a specified tolerance. Additionally, any software used to create irregularly-shaped fields should be thoroughly tested before clinical implementation. Often many of the acceptance tests are the same as subsequent routine quality assurance tests. A typical routine quality assurance program for the MLC is summarized in Appendix A.

Next, commissioning of the new device must be performed. This involves obtaining a collection of beam data for all energies and various field size combinations. Beam profiles and depth dose data must be entered into the treatment planning system. Because of the physical characteristics of the MLC leaves, such as the finite leaf width and the rounded leaf ends, there are clearly inherent limitations of an MLC. Although numerous investigations have shown that these physical characteristics of the MLC do not seem to affect the basic properties of radiation beams such as flatness, symmetry, output, and percentage depth dose\textsuperscript{1,2,3}, an evaluation of these dosimetric parameters must still be performed for fields defined by the MLC. As a result, commissioning of MLCs involves verifying that the same dosimetry data used for conventional collimators applies to the MLC-shaped fields. In particular, attention should be paid to the penumbra regions. Evaluation of these factors can determine the benefits and potential drawbacks of the MLC system.

This chapter discusses the method and equipment used for determination of various dosimetric parameters investigated including the collimator factor, relative dose factor, percentage depth dose, penumbra, leakage and transmission, and finally, a comparison of two treatments; one involving the MLC for shaping of the x-ray beam and the other involving the micro-MLC, both providing a relatively small, irregular radiosurgical field. All measurements were performed for both 6 MV and 18 MV photon beams on a Clinac 2300 C/D linac (Varian, Palo Alto, CA).
5.2 Collimator Factor

Measurement of the collimator factor (CF), as stated in Section 2.7, is carried out at the source-axis distance (SAD) with field sizes defined at the nominal SAD of the treatment machine. A build-up cap of thickness equal to the depth of dose maximum for a particular photon beam is required to obtain measurements in the region of electronic equilibrium. Furthermore, for measurements to reflect accurately the relative photon fluences, the entire radiation field must cover the build-up cap. Thus, small field-size measurements may be acquired at greater distances from the source to ensure that even the smallest radiation field covers the entire build-up cap.

A. Open fields defined with linac jaws (no MLC or micro-MLC)

The collimator factor was initially measured for square, symmetric fields ranging from 4×4 cm² to 40×40 cm². The fields were centered on the central axis with the MLC completely retracted and in park mode, so that the fields were defined by the jaws alone. An ion chamber (Shonka Farmer chamber #781) was placed at the isocenter (SAD = 100 cm) and a build-up cap of 1.5 cm and 3 cm of tissue equivalent material for the 6 MV and 18 MV beams, respectively, was added to the chamber to ensure sufficient dose build-up. For the smaller field sizes, the chamber was placed at an extended SAD of 156 cm to ensure that the radiation field covered the entire build-up cap. The charge collected for a setting of 200 MU on the linac was read by the electrometer (Victoreen 530) and the collimator factor was obtained from the ratio of in-air outputs as stated in Eq. 2-8.

In order to minimize the size of the build-up cap for the 18 MV photon beam, an aluminum cap of 1.27 cm was used instead of 3 cm of acrylic. Due to the higher density of aluminum ($\rho_{al} = 2.699 \text{ g/cm}^3$ versus $\rho_{acrylic} = 1.180 \text{ g/cm}^3$), a smaller thickness of build-up material is required to achieve electronic equilibrium. This thickness $t$ was determined as follows:

$$\rho = \frac{\text{mass}}{\text{Volume}} = \frac{\text{mass}}{\text{area} \cdot \text{thickness}}$$

$$\therefore \rho \propto \frac{1}{t}$$
The collimator factor was measured for rectangular, symmetric fields defined by the linac upper (X) and lower (Y) jaws, of various elongation ratios but of the same equivalent square field. Due to the different distances of the X and Y jaws from the source, the collimator factor depends on which dimension is collimated by which pair of jaws for rectangular fields. This is a result mainly of the different extents to which the X and Y jaws prevent the scattered radiation originating in the primary collimator, flattening filter, and other parts of the machine head from reaching the point of measurement, and is referred to as the collimator exchange effect. The magnitude of this effect was investigated for our linac.

The equivalent square field was determined by Day's equivalent field method when dealing with rectangular fields\(^4,5\) and the Clarkson method when dealing with irregularly-shaped fields\(^6\). An equivalent field is a square or circular field that has the same dosimetric characteristics as a given irregular field.

**B. Irregular fields defined with MLC or micro-MLC**

The aim of this section is to show that the addition of an MLC as a tertiary collimator has no significant effect on the collimator factor, and therefore, the collimator factor can be determined by the rectangular field determined by the collimator jaws rather than the irregular field shaped by the MLC. Since scatter from the tertiary collimator is dependent on the irradiated area of the tertiary collimator, only when the size of the irregularly shaped field from the MLC is much smaller than the size of the rectangular field defined by the secondary collimators is it possible that the presence of the MLC will affect the collimator scatter.\(^7\) For the case of the Varian MLC installed on a Varian Clinac 2300 C/D, it has previously been calculated that the smallest field size that can be defined by the MLC without affecting the collimator factor is given by:

\[
\frac{\rho_{\text{acrylic}}}{\rho_{\text{Al}}} = \frac{t_{\text{Al}}}{t_{\text{acrylic}}}
\]

\[t_{\text{Al}} = 1.31 \text{ cm}\]

The equation above represents the relationship between the acrylic and aluminum density factors for the tertiary collimator. The collimator factor was measured for rectangular, symmetric fields defined by the linac upper (X) and lower (Y) jaws, of various elongation ratios but of the same equivalent square field. Due to the different distances of the X and Y jaws from the source, the collimator factor depends on which dimension is collimated by which pair of jaws for rectangular fields. This is a result mainly of the different extents to which the X and Y jaws prevent the scattered radiation originating in the primary collimator, flattening filter, and other parts of the machine head from reaching the point of measurement, and is referred to as the collimator exchange effect. The magnitude of this effect was investigated for our linac.

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\[
\frac{\rho_{\text{acrylic}}}{\rho_{\text{Al}}} = \frac{t_{\text{Al}}}{t_{\text{acrylic}}}
\]

\[t_{\text{Al}} = 1.31 \text{ cm}\]
x-direction: $X_{MLC} = 0.298X$ and y-direction: $Y_{MLC} = 0.164Y$, \hspace{1cm} (5-1)

where $X_{MLC} \times Y_{MLC}$ define the smallest permissible MLC field size for a collimator jaw setting of $X \times Y$.\(^8,9^\) This implies that for a collimator jaw setting of $25 \times 25 \text{ cm}^2$, the collimator factor should not be affected by the presence of the MLC unless the field size defined by the MLC is smaller than $X_{MLC} = 0.298 \times 25 \text{ cm} = 7.45 \text{ cm}$ and $Y_{MLC} = 0.164 \times 25 \text{ cm} = 4.1 \text{ cm}$. This was investigated with the secondary collimator jaws fixed at $25 \times 25 \text{ cm}^2$, while the charge collected for a range of MLC field sizes, both smaller and larger than the critical MLC field size of $7.45 \times 4.1 \text{ cm}^2$, was measured.

The extent to which tertiary collimators affect head scatter factors is dependent on their closeness to the focal spot of the x-ray source. Therefore, a larger critical field size should be observed when the micro-MLC is used to define the field compared to the MLC. However, the casing of the micro-MLC will create changes in the energy fluence of the beam because of both attenuation of the incident photons and scattered photons produced by the casing. The amount of scatter produced relies on the irradiated area of the casing that is visible from the point of measurement. The collimator factor was investigated for a range of micro-MLC field sizes with the secondary collimator jaws fixed at $9.8 \times 9.8 \text{ cm}^2$.

### 5.3 Relative Dose Factor

#### A. Open fields defined with linac jaws (no MLC or micro-MLC)

The RDF was initially measured for square fields ranging from $4 \times 4 \text{ cm}^2$ to $40 \times 40 \text{ cm}^2$. The fields were centered on the central axis with fields initially defined by the jaws alone. An ionization chamber (Shonka Farmer chamber #781) was placed at the depth of dose maximum $d_{max}$ in a water phantom at a source-surface distance (SSD) of 100 cm. The relative dose factor was determined from the ratio of in-phantom outputs as stated in Eq. 2-11.

The RDF was also measured for rectangular, symmetric fields defined by the jaws to investigate the magnitude of the collimator exchange effect in phantom.
B. Irregular fields defined with MLC or micro-MLC

In-phantom dosimetric parameters, such as the RDF, are determined by the irregular field shaped by the MLC or micro-MLC. Behavior of the RDF was therefore investigated for fixed square jaw settings of $25 \times 25 \text{ cm}^2$, $20 \times 20 \text{ cm}^2$, $16 \times 16 \text{ cm}^2$, $12 \times 12 \text{ cm}^2$, $8 \times 8 \text{ cm}^2$, and $4 \times 4 \text{ cm}^2$ with a series of diamond-shaped MLC fields and fixed jaw settings of $9.8 \times 9.8 \text{ cm}^2$, $6 \times 6 \text{ cm}^2$, and $3 \times 3 \text{ cm}^2$ with a series of circular micro-MLC fields. With the addition of MLC or micro-MLC shaped fields, the relative dose factor is determined by:

$$RDF(A_{\text{MLC}}) = \frac{D_p(A_{\text{MLC}})}{D_p(A_{\text{jaws}} = 10)}.$$  \hspace{1cm} (5-2)

In addition, the extent of the variation in RDF when switching between the in-field, cross-boundary, and out-of-field leaf placement strategies was investigated for the 18 MV x-ray beam.

5.4 Scatter Factor

With the addition of a tertiary MLC or micro-MLC, the collimator scatter (CF) is determined by the rectangular field shaped by the jaws ($A_{\text{jaws}}$) while the phantom scatter (SF) is determined by the irregular field shaped by the MLC or micro-MLC ($A_{\text{MLC}}$). Therefore the total scatter, or relative dose factor (RDF), is given by:

$$RDF(A_{\text{MLC}}) = CF(A_{\text{jaws}}) \times SF(A_{\text{MLC}}).$$  \hspace{1cm} (5-3)

From the results of the measurements described in Section 5.2 and Section 5.3, the scatter factor was calculated by means of Eq. 5-3 for the jaws alone, for the MLC, and for the micro-MLC.

5.5 Percentage Depth Dose

The aim here was to verify that the PDDs involving the linac jaws alone (open fields) can still be used even when the MLC or micro-MLC is used to define the field size. Clinically, the two PDDs measured for a given field size need to be within 2% to be
considered the same. However, because the MLC and micro-MLC are closer to the phantom surface than the secondary collimators, scatter from the MLC or micro-MLC and electron contamination may cause a small difference in the percent depth doses, particularly in the build-up region.

The central axis percentage depth doses were measured in a three-dimensional, computer-controlled water scanner. The water surface was positioned at the nominal source-surface distance (SSD) of 100 cm and the origin of the measuring device was positioned at the isocenter. The scan depth ranged from the water surface to a depth of 30 cm. Square and rectangular fields defined by the jaws alone as well as square, rectangular, and irregular fields defined by the MLC with the jaws constant at 25x25 cm² were measured using an RFA water scanner (Scanditronix, Uppsala, Sweden) and RK ion chambers (Scanditronix, Uppsala, Sweden).

Square, circular, and irregular fields defined by the micro-MLC with the jaws fixed at 9.8x9.8 cm² were measured using both a Wellhofer water scanner (Wellhofer Dosimetrie, Schwarzenbruck, Germany) and small graphite thimble ionization chamber (IC-04, Wellhofer Dosimetrie, Schwarzenbruck, Germany) of sensitive volume 0.028 cm³ as well as the RFA water scanner with a high spatial resolution p-type silicon diode detector (RFA-3, Scanditronix, Uppsala, Sweden) with active diameter of 2.5 mm. The energy response of the diode was linear over the photon energies of interest. In all measurements, the ratio of outputs from the field and reference chamber were normalized to the ionization measured at the depth of 1.5 cm and 3 cm for the 6 MV and 18 MV x-ray beams, respectively.

5.6 Penumbra

The physical penumbra of a field was defined in Section 2.11.C as the lateral distance between the 80% and 20% isodose curves at a specified depth. Measurement of the penumbral characteristics for single-focussed MLC systems is an important parameter because the rounded leaf ends create a concern that the penumbra width may change as the position of the leaf end changes with respect to the beam’s central axis. When a leaf is retracted away from central axis the field edge is defined by the lower part of the curved
leaf end, while when a leaf is extended beyond the central axis the field edge is defined by the upper part of the curved leaf end. Geometrically, the penumbra width is expected to be larger when the field edge is defined by the upper part of the curved leaf end. However, since previous investigations proved that a divergent collimator system always produces the smallest penumbra, the shape of the leaves was carefully chosen to approach that of a divergent collimator system. Furthermore, the leaf sides follow beam divergence, therefore, the penumbra width in the direction perpendicular to the motion of the leaves is expected to be less than the penumbra width in the direction of motion of the leaves. However, the tongue and groove profile, designed to decrease leakage between leaves, may degrade the profile for the leaf sides.

The physical penumbra was measured in an SSD set-up in phantom at a depth of dose maximum and at a depth of 10 cm. Square fields defined by the jaws alone as well as square, rectangular, and irregular fields defined by the MLC with the jaws maintained at 25×25 cm² were measured using the RFA water scanner and RFA-3 diode detector. Square, circular, and irregular fields defined by the micro-MLC with the jaws fixed at 9.8×9.8 cm² were measured using both the Wellhofer water scanner with IC-04 ionization chamber as well as the RFA water scanner with RFA-3 diode detector.

![Fig. 5.1: Positioning of the micro-MLC field for determination of the penumbra width as a function of leaf position.](image)

In addition, the penumbra width was evaluated as a function of field edge distance from the central axis by offsetting a 3×3 cm² field with the micro-MLC and a fixed jaw setting of 9.8×9.8 cm². The field was initially centered on the central axis with
additional scans taken when the field center was moved diagonally off axis as depicted in Fig. 5.1.

### 5.7 Transmission and Leakage

The rounded leaf ends give rise to radiation leakage even when two opposing leaves are set to zero. Furthermore, a minimum separation between opposing leaf pairs is required to prevent collision of the leaf pairs. This causes additional radiation leakage at the junction where opposing leaves abut. However, radiation leakage can be reduced significantly by setting the position where opposing leaves abut outside the treatment field. This will allow the junction to be shielded by the secondary collimator jaws. Another source of radiation leakage is the small gap between adjacent leaves whose function is to reduce friction as the leaves move. Although the tongue and groove design of the leaf sides will minimize this radiation transmission, interleaf transmission is not negligible and must still be considered.

Transmission through and leakage between the leaves was measured by closing the leaves completely with the jaws set to 10×10 cm² for the MLC and 9.8×9.8 cm² for the micro-MLC. All doses were measured relative to an open jaw setting of 10×10 cm² for the MLC and 9.8×9.8 cm² for the micro-MLC with the tertiary collimator leaves completely retracted. The transmission and leakage was measured at a depth of d_{max} and 10 cm in the RFA water tank at a source-surface distance of 100 cm. Scans along both the x and y axis were obtained, as indicated in Fig. 5.2, with a dose rate of 400 MU/min for the open beam scans and 600 MU/min for the closed field scans so as to magnify the intensity of the transmitted x-rays. These were then scaled accordingly.

Leakage and transmission were measured independently with therapy verification XV-2 film (Kodak Inc., Rochester, N.Y.) and compared with the water tank results. The film was irradiated in solid water and at a source-film distance of 100 cm at a depth of 1.5 cm and 3 cm for the 6 MV and 18 MV beams, respectively. A total of 1600 MU was delivered and a calibration curve was obtained by retracting the MLC or micro-MLC leaves and setting the jaws to 10×10 cm². The calibration curve was obtained by exposing the radiographic film to 0.10, 0.20, 0.30, 0.40, and 0.50 Gy. The films were
scanned using a scanning densitometer (Wellhofer Dosimetrie, Schwarzenbruck, Germany) and the optical density for each pixel was converted to absorbed dose.

![Image of film illustrating transmission and leakage through and between the MLC leaves where the leaves abut at (a) 0 cm, (b) 2.5 cm, and (c) 5 cm. Position of scans to measure the transmission through and leakage between leaves is indicated.]

**Fig. 5.2:** Image of film illustrating transmission and leakage through and between the MLC leaves where the leaves abut at (a) 0 cm, (b) 2.5 cm, and (c) 5 cm. Position of scans to measure the transmission through and leakage between leaves is indicated.

### 5.8 Comparison of Treatment Plans

Two single isocenter, 9-field conformal treatment plans on a Rando phantom head were compared; one where the conventional MLC was used to shape the irregular target volume and the other where the micro-MLC was used.

The treatment planning process was carried out by removing five to six plugs from two slices on the Rando phantom head to define the irregular target volume. A stereotactic localizing head frame was pinned onto the head below the slices containing the target volume. Virtual simulation of the phantom head was performed using the AcQsim CT simulator which takes standard computerized tomography (CT) scans and generates three-dimensional images on a sophisticated computer system. A total of 70 slices, each with a slice thickness of 2 mm, was obtained. The isocenter, or intersection of the lasers, was marked on the phantom head as a reference for future alignment of the treatment fields. The information taken from the AcQsim was used to generate digitally reconstructed radiographs (DRR) that could be viewed in the axial, sagittal, or coronal planes.

The MLC-defined treatment plan was done on the CadPlan 3D treatment planning system while the micro-MLC-defined plan was done using the BrainLAB
BrainSCAN treatment planning system. The reference point, where the lasers intersected as previously marked on the phantom head, was set as the (0,0,0) coordinate. The target volume of 19.6 cm$^3$ to be shaped by the MLC or micro-MLC was drawn in the 3D treatment planning system's beams-eye-view display. A 2 mm margin was added around the outlined target volume and the external area was defined. Nine static treatment fields, all of equal weighting, were used with both the MLC and micro-MLC as displayed in Table 5.1 to conform to the projected target volume shape as a function of couch rotation and gantry angle. The dose to isocenter was 2 Gy.

Table 5.1: Treatment parameters for the nine static fields used in the dose delivery for the two treatment plans.

<table>
<thead>
<tr>
<th>Beam number</th>
<th>Table angle</th>
<th>Gantry angle</th>
<th>Collimator angle</th>
<th>Monitor Units (MU)*</th>
<th>Monitor Units (MU) given**</th>
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<td>60°</td>
<td>90°</td>
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<td>5</td>
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<td>270°</td>
<td>90°</td>
<td>25</td>
<td>5</td>
</tr>
</tbody>
</table>

* prescribed for a target dose of 2 Gy  
** given in our experiment so as not to overdose the film

Following the creation of the two treatment plans, the head phantom was irradiated to verify the dose delivery according to the treatment plans. The head phantom was attached to a head ring and set up in the same way as a stereotactic radiosurgery treatment for a patient. The coordinates of the isocenter of the target volume relative to the laser reference marks, referred to as x, y, and z coordinates, were coincident with the isocenter of the linac. XV-2 therapy verification film was sandwiched between the two slices containing the target volume in the phantom head and used to record the dose delivery for each treatment. The photon jaws were set to 9.8×9.8 cm$^2$ and the head phantom was irradiated with a 6 MV photon beam for the 9 static fields listed in Table.
5.1. One fifth of the prescribed number of MU were delivered so as not to saturate the films.

Following the treatment delivery, each of the films was scanned with a Wellhofer scanning densitometer and an isodose distribution was obtained. The measurements were compared and used to verify that the 3D treatment planning system was capable of simulating the micro-MLC technique. The possible advantage of the micro-MLC technique was explored and compared to that of the MLC system.

5.9 Summary

The method for collection of beam data for two beam energies (6 MV and 18 MV) and various field size combinations was presented for the linac jaws alone (open fields), the conventional MLC, and the micro-MLC. Beam profiles and depth dose data was required for entry into the BrainLAB treatment planning system before treatment plans could be generated for the micro-MLC.

Due to the physical characteristics of the MLC leaves, such as the finite leaf width and the rounded leaf ends, an evaluation of these dosimetric parameters must be performed. Evaluation of these factors can determine the benefits and potential drawbacks of the MLC system. Results of our measurements are given in the next chapter.

5.10 References


6.1 Introduction

Results of measurements described in Chapter 5 for a 6 MV and 18 MV photon beam on a Clinac 2300 C/D linac (Varian, Palo Alto, CA) are presented here. Although both the 6 MV and 18 MV x-ray beams are discussed here, only the data for the 6 MV beam is displayed in this chapter. Since the 18 MV beam generally follows the same trends as the 6 MV beam, data for the 18 MV beam is presented in Appendix B.

6.2 Collimator Factor

A. Open fields defined with linac jaws (no MLC or micro-MLC)

The results of the collimator factor for square, symmetric fields, shaped by the
linac jaws and ranging from 4×4 cm² to 40×40 cm² for 6 MV and 18 MV photons are presented in Fig. 6.1. Each data point represents the average of three readings; however, the charge collected by the electrometer was very stable, and consequently, the error bars are too small to be visible on the scale displayed.

![Graph of Collimator Factor](image)

**Fig. 6.1:** Comparison of the collimator factor for 6 MV and 18 MV photon beams and various square, symmetric fields defined by the linac jaws.

A comparison of the collimator factor for rectangular, symmetric jaw settings of X×Y versus Y×X with elongation ratios ranging from to 1 to 7, each with an equivalent square field of 10×10 cm², is illustrated in Fig. 6.2 for the 6 MV photon beam. Evidently, the collimator factor for rectangular fields depends on which dimension is collimated by which pair of jaws because of the different distances of the X and Y jaws from the source. This variation of the collimator factor is referred to as the collimator exchange effect and is mainly caused by the different extents to which the X and Y jaws prevent scattered radiation originating in the primary collimator, flattening filter, and other parts of the machine head from reaching the point of measurement. For example, the difference in head-scatter photons is due to different projections of the upper and lower collimator jaws on the flattening filter when viewed from the point of measurement. The difference in monitor backscatter is a result of the different number of photons and electrons that are
backscattered from the upper and lower collimator jaws into the monitor chamber.

It was observed that when the Y jaw has the larger setting it produces a larger CF as a result of the surface of the Y jaw which is closer to the source, providing the limit on the amount of scattered radiation that contributes to the output. Therefore, as the elongation ratio of the Y jaw setting compared to the X jaw setting increases for a constant equivalent field, the CF also increases. On the other hand, when the elongation ratio of the X jaw setting compared to the Y jaw setting increases, the CF decreases but less rapidly compared to the increase in the CF with the Y jaw setting. For instance, the CF of the 6 MV photon beam with elongation ratio of 7 reveals a variation of +1.2% when the Y jaw has the larger setting yet only a -0.7% variation when the X jaw has the larger setting. However, the CF for the 18 MV photon beam, illustrated in Fig. B.1, with the same elongation ratio, exhibits a variation of 1.2% in both instances. Previous experiments have shown that the collimator exchange effect can result in a variation of up to 3% in the collimator factor.2

![Graph](image)

**Fig. 6.2**: CF measurements for a 6 MV photon beam defined by the linac jaws. The "collimator exchange effect" for rectangular, symmetric fields with a side of equivalent square equal to 10 cm and elongation ratios ranging from 2 to 7 is also presented.

Several approaches to this problem have been previously investigated, including
the derivation of analytic expressions to predict the CF for rectangular and irregular field combinations. For example, the x-ray source model models the output of the linac for photon beams as the sum of two components: primary radiation reaching the point of measurement directly from the target and the secondary radiation arising from photons scattered in the head of the linac.\(^1\)\(^2\) The first component is represented by a point source while the latter, referred to as extrafocal radiation, is represented by a broadly distributed source. However, only the part of the distributed source that can be seen through the collimators from the point of measurement contributes to the output.

Scatter from the flattening filter, collimators, and monitor chamber all contribute to the collimator scatter factor and each component has its own field size dependence characteristics. For instance, scatter due to the flattening filter is a function of the area of the flattening filter that is visible at the point of measurement. Monitor backscatter depends on the position of both the upper and lower collimator jaws with respect to the beam monitor chamber. These factors evidently depend on the configuration of the linac.

As previously stated, not all photons that are produced by a linear accelerator emanate from a point source. A fraction of these photons, called extrafocal radiation are produced outside the focal spot and are due to structures surrounding the target such as the monitor chamber, primary collimator, and flattening filter. As the secondary collimator jaws are opened, more extrafocal photons are permitted to reach the detector, therefore increasing the output. Measurements have shown that for a Clinac 2100C, extrafocal radiation accounts for approximately 8% of the output of the linac.\(^3\) Forward scattering in the flattening filter represents the major portion of scattered radiation and is limited by the primary collimator. Secondly, some photons are backscattered from the secondary collimators into the beam monitor chamber. As the secondary collimator jaws are opened, the area of the upper surface of the collimators that is exposed to the photon beam decreases. As a result, the number of photons that are backscattered into the beam monitor chamber decreases as the field size increases. A lower number of monitor units (MU) are therefore measured at the point of reference and the output increases. Measurements have shown that for a Varian C-series linac output can increase by more than 2% due to reduced backscatter as the field size varies from 5×5 cm\(^2\) to 40×40 cm\(^2\).\(^2\) Lastly, forward scattering from the secondary collimators affects the collimator factor.
when the area of the faces of the secondary collimator jaws that are exposed to the photon beam increases as the collimator jaws are opened. A greater number of photons are therefore scattered into the point of measurement. Previous measurements have shown that this scatter continues to increase as the field size varies from $1 \times 1 \text{ cm}^2$ to $40 \times 40 \text{ cm}^2$, although typically the increase is less than 1\%.

B. Irregular fields defined with MLC

The results of the collimator factor for the 6 MV photon beam for various MLC setting while the jaws were fixed at $25 \times 25 \text{ cm}^2$ are presented in Fig. 6.3 while the 18 MV photon beam results are displayed in Fig. B.2. The conventional collimator factor for 6 MV photons with a fixed jaw setting of $25 \times 25 \text{ cm}^2$ has a constant value of 1.0378. Evidently, the CF for MLC field sizes greater than the critical field of $7.45 \times 4.1 \text{ cm}^2$, when the jaws are maintained at $25 \times 25 \text{ cm}^2$, is not affected. However, MLC field sizes smaller than the critical field exhibit a large reduction in the CF. Any material placed between the target and the phantom produces scattered photons and secondary electrons that will affect the quality of the radiation beam. However, the amount of scattered radiation that reaches the phantom is dependent upon the proportion of linac head components that are visible from the point of measurement when looking towards the x-ray target. Therefore, only when the MLC field is much smaller than the jaw setting will the CF be affected.

The addition of an MLC as a tertiary collimator below the secondary collimator jaws does not have a significant affect on scatter from the flattening filter, collimators, and the monitor chamber. Firstly, the large distance between the source and the MLC restricts its ability to block out portions of the extrafocal photons. Secondly, the distance between the MLC and the beam monitor chamber is too large for the MLC to scatter a significant portion of the photons back into the beam monitor chamber. Finally, the solid angle presented to the source by the leaf ends or the leaf sides is too small to generate significant scatter onto the detector plane.

Previous experiments have illustrated that as the field is increasingly blocked, the relative in-air output starts to increase at first (up to 0.5\% for MLC blocked fields)
and then decreases as the field blocking becomes more extreme. This is consistent with the present findings and is attributed to increasing scatter from the tertiary collimator and decreasing head scatter as the field is progressively blocked. As the tertiary blocking becomes more extreme the screening of head scatter photon fluence by the tertiary collimator increases. Similar results were obtained for 18 MV photons also with a fixed jaw setting of 25×25 cm².

![Figure 6.3](image)

**Fig. 6.3**: Comparison of the conventional collimator factor for a 6 MV photon beam with jaws fixed at 25×25 cm² against the measured collimator factor for fields defined by the MLC.

**C. Irregular fields defined with micro-MLC**

The results of the CF for the 6 MV photon beam for various micro-MLC settings while the jaws were fixed at 9.8×9.8 cm² are displayed in Fig. 6.4. Below a square field size of 4.2 cm, the CF was observed to have a variation of greater than 2% compared to the conventional collimator factor defined by the jaws. Below this equivalent field size, the source of scattered x rays originating in the flattening filter is shielded by the micro-MLC. Since the photon jaws are located closer to the flattening filter than the micro-MLC, the analogous source shielding effect would occur at larger field sizes.

The relative output generated by a MLC shaped field is expected to be slightly
less than an identically shaped micro-MLC field because the leaves of the MLC are closer to the monitor chamber than the leaves of the micro-MLC. The MLC leaves will therefore produce more backscatter into the chamber. Furthermore, due to the greater distance of the micro-MLC from the focal spot, the magnitude of the collimator exchange effect for the micro-MLC will be greater than for the MLC.

![Graph showing collimator factor comparison]

**Fig. 6.4**: Comparison of the conventional collimator factor for the 6 MV photon beam with jaws fixed at 9.8×9.8 cm² with the measured collimator factor for fields defined by the micro-MLC.

### 6.3 Relative Dose Factor

**A. Open fields defined with linac jaws (no MLC or micro-MLC)**

The relative dose factor was initially measured for 6 MV and 18 MV photon beams for various field sizes defined by the secondary collimator jaws, as presented in Fig. 6.5. Again, each data point represents the average of three readings; however, the charge collected and read by the electrometer was very stable so that the error bars are too small to be visible on the scale displayed.
Fig. 6.5: Measurement of the relative dose factor for 6 MV and 18 MV photon beams for various square field sizes defined by the linac jaws.

A comparison of the RDF for rectangular, symmetric jaw settings of various equivalent square fields is illustrated in Fig. 6.6 for the 6 MV photon beam and Fig. B.3 for the 18 MV photon beam. The collimator exchange effect for the relative dose factor was observed to be less significant compared to that of the collimator factor.

Fig. 6.6: RDF measurements for a 6 MV photon beam defined by the linac jaws.
B. Irregular fields defined with MLC

The RDF for several fixed jaw settings for various cross-boundary, diamond-shaped MLC field sizes is presented in Fig. 6.7 and Fig. B.4 for 6 MV and 18 MV photons, respectively. It is evident, as in the case for fields shaped with custom blocks, that the relative dose factor for a given MLC field size depends on the jaw setting. When the jaws are decreased, a decrease in the overall output is observed largely due to changes in the collimator scatter. It should be noted that the relative dose factor by definition is normalized to 1.0 for a 10×10 cm² or 100 cm² field. However, the area implied in Fig. 6.7 is that of the field defined by the MLC while the charge collected for each of these MLC defined fields was normalized to 1.0 for a 10×10 cm² field defined by the jaws alone. This explains why the curves of Fig. 6.7 do not pass through 1.0 for a 100 cm² MLC field.

As observed, the RDF for a fixed jaw setting, while the MLC field is continuously decreased, follows the same shape as the RDF with no MLC field. However, as the photon jaw setting is decreased while the MLC field remains fixed, the RDF is observed to decrease at an increasing rate. For example, consider the 6 MV beam in Fig. 6.7. For an MLC field area of 50 cm², the RDF decreases from 1.0241 to 1.0164, or 0.75%, when going from a 25×25 cm² to 20×20 cm² jaw setting. When going from the 20×20 cm² to 16×16 cm² jaw setting, the RDF now decreases by 0.82%.

Fig. 6.7: Relative dose factor for 6 MV photons, 200 MU, MLC leaves in the “cross-field” position.
Boyer found that after blocking the corners of square fields set by the photon jaws with the MLC leaves to reduce the field area by 50%, agreement between measured and calculated dose was within 1.7% for 6 MV photons and up to 2.5% for 18 MV.

As expected, by using the various leaf placement strategies, the RDF changes for a constant target area as seen in Fig. 6.8 for 18 MV photons. When using the "in-field" technique, part of the leaves cover the target area, and therefore, the field defined by the MLC is smaller compared to the other two leaf placement strategies. Since the RDF decreases with smaller field sizes, the RDF due to the "in-field" leaf placement strategy will be smaller than the RDF for the other two leaf placement strategies. Therefore, it is important that the same leaf-placement strategy be used in designing the micro-MLC fields as the leaf-placement strategy used for the data input into the treatment planning system.

![Graph showing RDF for different leaf placements](image)

**Fig. 6.8**: Comparison of the "in-field", "cross-field", and "out-of-field" leaf placement for 18 MV photons with various MLC field sizes while the jaws are kept constant at 20×20 cm².

**C. Irregular fields defined with micro-MLC**

The RDF for several fixed jaw settings for various circular micro-MLC field sizes is presented in Fig. 6.9 for 6 MV photons and Fig. B.5 for 18 MV photons.
Fig. 6.9: Relative dose factor for 6 MV photons as a function of the field diameter defined by the micro-MLC for various jaw settings.

6.4 Scatter Factor

Fig. 6.10 displays the average scatter factor (SF) for the 6 MV and 18 MV x-ray beam for the linac jaws alone. Fig. 6.11 and Fig. 6.12 display the average scatter factor for the 6 MV x-ray beam for the MLC and the micro-MLC, respectively, while Fig. B.6 display the average SF for the 18 MV photon beam defined by the MLC. The error bars represent the standard deviation. The SF is independent of the jaw setting since the RDF is a function of the MLC setting and the CF is constant for various MLC settings. Therefore, every resulting SF will be the same.

The larger the field size, the larger the volume of irradiated material from which photons can be scattered toward the point of measurement, and in turn, the larger the SF. This increase in phantom scatter will increase the measured output and is independent of the treatment unit. 8

All fields for the 6 MV beam, even the smallest field of 8 cm² which has the largest standard deviation of 1.7%, was within 2%. Therefore, the conventional method of calculating the output factor can be used with the MLC. However, the smallest field for the 18 MV photon beam has a standard deviation of 2.3%, although the larger field sizes were all within 2%. Therefore, very small MLC fields inside relatively large jaw settings
must be used with caution.

**Fig. 6.10**: Scatter factor for 6 MV and 18 MV photons as a function of the field size defined by the jaws obtained using the results of Fig. 6.1 and Fig. 6.5.

**Fig. 6.11**: Scatter factor for 6 MV photons as a function of the field area defined by the MLC obtained using the results of Fig. 6.2 and Fig. 6.7. The error bars indicate the
maximum discrepancies between the scatter factor obtained with different square field settings.

Fig. 6.12: Scatter factor for 6 MV photons as a function of the field area defined by the micro-MLC obtained using the results of Fig. 6.4 and Fig. 6.9.

6.5 Percentage Depth Dose

The physical effects that contribute to the values of the percent depth dose include the inverse square of radiation intensity on distance from a point source, build-up between primary photons and locally-absorbed secondary electrons from the surface to an equilibrium depth, exponential attenuation of the primary photons with depth, and the scattered radiation from surrounding irradiated material.

Graphs for the comparison of the percentage depth dose curves for field sizes ranging from 4×4 cm² to 20×20 cm² defined by either the linac jaws alone or the MLC are plotted in Fig. 6.13 for a 6 MV photon beam. Small differences between the PDDs of identical field size incident on the phantom surface were observed due to the change in the amount of scattered radiation that results when the field size was defined by the jaws versus the MLC. Since the MLC is closer to the phantom surface than are the jaws, scatter from the MLC and electron contamination result in an increase in the percent depth dose curves, particularly in the build-up region. However, beyond the build-up region this variation was within, on average, 0.7% for the 10×10 cm² field and 0.9% for the 20×20
cm² field yet always within 2%. The larger field sizes always exhibited the greatest discrepancies, confirming that the main difference is due to an increase in scatter radiation.

Similarly, the PDDs for an 18 MV photon beam showed comparable results; however, the differences between the PDDs for the higher energy beam were less important. Graphs for the comparison of the percentage depth dose curves for field sizes ranging from 4×4 cm² to 20×20 cm² defined by either the jaws alone or the MLC are plotted in Fig. B.7 for the 18 MV photon beam.

These results agree with previously measured data where the differences observed between normalized PDD curves for 6 MV photons with respect to the collimator jaw fields and MLC fields were within 0.5% between depths of 1.0 cm and 32 cm.⁷ For the 18 MV MLC fields, the PDD curves in the build-up region and around the depth of maximum build-up showed an increase of roughly 1.5% over those produced by the jaws. Beyond a depth of 3 cm the differences were within 0.5%.

It was stated in Section 5.6 that two PDDs for a given field size must be within 2% to be considered clinically the same. Although slight differences were observed between the PDDs when the field size was determined by the linac jaws alone versus the MLC, in practice, the differences are insignificant. As a result, the PDD data for fields defined by the conventional jaws can be used in the treatment planning process even when the MLC is used to shape the field.
Fig. 6.13: Percentage depth dose curves for the 6 MV photon beam in water when the jaws and the MLC are used to define the various square field sizes: (a) 4x4 cm\(^2\), (b) 8x8 cm\(^2\), (c) 10x10 cm\(^2\), (d) 14x14 cm\(^2\), (e) 18x18 cm\(^2\), and (f) 20x20 cm\(^2\). In the case of the MLC, the jaws were fixed at 25x25 cm\(^2\).

Furthermore, no significant difference was observed in the percent depth dose distributions for various linac jaw settings when the MLC setting remained fixed, since the field size incident on the phantom surface remained the same. This was investigated for a fixed MLC setting of 10x10 cm\(^2\) while the jaws were increased from 10x10 cm\(^2\) to 25x25 cm\(^2\). An average difference of 0.2% was observed for the 6 MV photon beam. These results are plotted in Fig. 6.14 and Fig. B.8 for the 6 MV and 18 MV photon beams, respectively.
Results and Discussion

Jaw setting:

- 10x10 cm²
- 15x15 cm²
- 20x20 cm²
- 25x25 cm²

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Fig. 6.14: Comparison of the percentage depth dose for 6 MV photons with a constant MLC field size of 10x10 cm² and various linac collimator jaw settings.

Graphs for the comparison of the percentage depth dose curves for square fields of side length ranging from 2.4 cm to 9.1 cm defined by either the jaws alone or the micro-MLC are plotted in Fig. 6.15 for the 6 MV photon beam. When the micro-MLC was used to define the field size, the jaws were maintained at a fixed setting of 9.8x9.8 cm². Small differences between the PDDs of identical field size incident on the phantom surface were observed when the field size was defined by the jaws versus the micro-MLC. The average difference beyond the build-up region for the 9 curves was 0.7%. However, in this case, the jaws were found to generate a slightly higher PDD than the micro-MLC. Measurement of the PDDs for fields defined by the jaws were obtained with an ion chamber while the PDDs for fields defined by the micro-MLC were obtained with a diode, explaining the reason for the large variation in the build-up region.

Similarly to 6 MV photon beams, the PDDs for the 18 MV photon beam again showed comparable results for open fields and MLC-defined irregular fields. Graphs for the comparison of the percentage depth doses for square fields of side length ranging from 2.4 cm to 9.1 cm defined by either the linac jaws alone or the micro-MLC are
plotted in Fig. B.9 for an 18 MV photon beam. The results for our 6 MV photon beam agree well with previously published data for the same machine; however, no data has yet been published for the 18 MV x-ray beam.

The percentage depth doses for the micro-MLC were found to be no more than 2% different from the percentage depth doses generated for the same size fields formed by the conventional photon jaws. Thus, the addition of an micro-MLC does not significantly affect the depth dose characteristics of the treatment machine. The PDD data for fields defined by the conventional linac jaws can be used in the treatment planning process even when the MLC or micro-MLC is used to shape the field.
Fig. 6.15: Percentage depth dose curves for the 6 MV photon beam in water when the jaws and the micro-MLC are used to define the various square field sizes of side length (a) 2.4 cm, (b) 3.0 cm, (c) 3.6 cm, (d) 4.2 cm, (e) 5.1 cm, (f) 6.0 cm, (g) 6.9 cm, (h) 8.0 cm, and (i) 9.1 cm. In the case of the micro-MLC, the jaws were fixed at 9.8 x 9.8 cm².
Chapter 6

Results and Discussion

The percent depth dose data for circular fields defined by the micro-MLC of diameter 0.6, 1.2, 1.8, 2.4, 3.0, 3.6, 4.2, 5.1, 6.0, 6.9, 8.0, and 9.1 cm were obtained and are displayed in Fig. 6.16. The series of curves, all containing a fixed jaw setting of 9.8×9.8 cm², demonstrate the dependence of \( d_{\text{max}} \) on field size. For instance, the 0.6 cm diameter field showed a \( d_{\text{max}} \) at a depth of 1.2 cm while the 5.1 cm diameter field showed a \( d_{\text{max}} \) at a depth of 1.5 cm. The series of curves for the 18 MV beam are displayed in Fig. B.10. The \( d_{\text{max}} \) shift for the 12 fields is indicated in Table 6.1. The \( d_{\text{max}} \) shift toward the surface for fields smaller than 5×5 cm², as stated in Section 2.4, has been studied in detail before⁹,¹⁰ and is attributed to changes in phantom scatter.

![Figure 6.16](image.png)

**Fig. 6.16**: Percentage depth dose curves for the 6 MV photon beam in water when the jaws and the micro-MLC are used to define the various circular field sizes of diameter 0.6 cm, 1.2 cm, 1.8 cm, 2.4 cm, 3.0 cm, 3.6 cm, 4.2 cm, 5.1 cm, 6.0 cm, 6.9 cm, 8.0 cm, 9.1 cm, and 10.0 cm. The jaws were fixed at 9.8×9.8 cm².

**Table 6.1**: Positions of \( d_{\text{max}} \) for the 6 MV photon beam in water for various field size diameters.

<table>
<thead>
<tr>
<th>Field diameter (cm)</th>
<th>( d_{\text{max}} ) (cm)</th>
<th>Field diameter (cm)</th>
<th>( d_{\text{max}} ) (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.6</td>
<td>1.2</td>
<td>4.2</td>
<td>1.5</td>
</tr>
<tr>
<td>1.2</td>
<td>1.4</td>
<td>5.1</td>
<td>1.5</td>
</tr>
<tr>
<td>1.8</td>
<td>1.6</td>
<td>6.0</td>
<td>1.5</td>
</tr>
<tr>
<td>2.4</td>
<td>1.6</td>
<td>6.9</td>
<td>1.5</td>
</tr>
<tr>
<td>3.0</td>
<td>1.6</td>
<td>8.0</td>
<td>1.5</td>
</tr>
<tr>
<td>3.6</td>
<td>1.5</td>
<td>9.1</td>
<td>1.5</td>
</tr>
</tbody>
</table>
A comparison of the percent depth dose curves for a 10x10 cm\(^2\) field defined by the linac jaws alone, the MLC with a fixed linac jaw setting of 25x25 cm\(^2\), and the micro-MLC with a fixed linac jaw setting of 9.8x9.8 cm\(^2\) is given in Fig. 6.17 and Fig. B.11 for the 6 MV and 18 MV x-ray beams, respectively. It is evident that there is essentially no difference in the PDD when the field size is defined by either the jaws alone, the MLC, or the micro-MLC, confirming that the PDD data for fields defined by the conventional linac jaws can be used in the treatment planning process even when the MLC or micro-MLC is used to shape the field.

Fig. 6.17: Comparison of the percentage depth doses for 6 MV photons in water with the collimator jaws alone set to 10x10 cm\(^2\), with the collimator jaws set to 25x25 cm\(^2\) and the MLC leaves set to 10x10 cm\(^2\), and with the collimator jaws set to 9.8x9.8 cm\(^2\) and the micro-MLC leaves set to 10x10 cm\(^2\).

6.6 Beam Profiles

Fig. 6.18 and Fig. 6.19 show the beam profiles for a 6 MV photon beam of field size 10x10 cm\(^2\) defined by the MLC in both the direction of leaf motion and
perpendicular to the direction of leaf motion. In both cases, the leaves were positioned to abut on the beam central axis. However, the scan in Fig. 6.19 was taken along the beam central axis where the leaves abut. The average peripheral dose of approximately 16% is a result of the leakage due to the minimum separation that must be maintained between abutting leaves to avoid collision. Evidently, this high relative dose is unacceptable in certain circumstances, and therefore, the leaves must be made to abut off the beam central axis to reduce the peripheral dose.

Fig. 6.18: Beam profile in the direction parallel to the leaf motion for a 6 MV photon beam with jaws set to 25x25 cm$^2$ and the MLC set to 10x10 cm$^2$.

6.7 Penumbra

Penumbra widths as a function of field size were evaluated for the linac jaws alone, the MLC, and the micro-MLC for square fields centered on the central axis. The penumbra for the leaf sides is somewhat less than for the leaf ends, as seen in Fig. 6.20 and Fig. 6.21 for the 6 MV photon beam and Fig. B.12 and Fig. B.13 for the 18 MV photon beam, since the leaves follow beam divergence in this direction. For the 6 MV photon beam at $d_{max}$, the penumbras for leaf ends were found to have a mean difference of 0.32 mm and 0.6 mm greater than the penumbra for the leaf sides for the MLC and micro-MLC, respectively. At a depth of 10 cm, the mean difference was 0.94 mm and 0.6 mm
between the leaf ends and the leaf sides for the MLC and the micro-MLC, respectively. Since the micro-MLC has a variable leaf width, it generally has a stronger field size dependence. Previous measurements have shown that the effective penumbra widths for the leaf sides are 0.5 to 1.5 mm narrower than the width for the leaf ends and the edge functions of the sides of the leaves are very close to the edge function obtained from a divergent system. However, due to the finite size of the leaf width, the scalloping effect is inevitable with MLC fields. Measurements show that the effective penumbra width is 3 to 5 mm wider than the width for the focused or divergent system.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{fig619.png}
\caption{Beam profile perpendicular to the direction of leaf motion for a 6 MV photon beam with jaws set to 25x25 cm$^2$ and the MLC set to 10x10 cm$^2$.}
\end{figure}

The penumbras for the MLC leaf sides were, on average, 0.62 mm greater than the penumbra for the lower X jaws while the penumbras for the MLC leaf ends were, on average, 0.25 mm greater than the upper Y jaws. On the other hand, the penumbras for the micro-MLC leaf sides were, on average, 0.04 mm less than the penumbra for the lower X jaws while the penumbras for the micro-MLC leaf ends were, on average, 0.36 mm less than the upper Y jaws. Data for the 6 MV photon beam agreed well with previously published data.
A. Irregular fields defined with micro-MLC

The penumbra width for fields shaped by the micro-MLC were observed to be small and did not vary significantly as a function of leaf position. When a leaf is retracted away from the central axis, the field edge is determined by the lower part of the curved end, yet when the leaf position is beyond the central axis, the field edge is produced by the upper part of the curved leaf end. From the geometric point of view, it is expected that the penumbra width shall be larger when the field edge is created by the upper part of the curved leaf end than when the field edge is created by the lower part of the curved leaf end. However, the average penumbra width for the 3x3 cm\(^2\) field was 2.0±0.6 mm for the leaf sides and 2.5±0.2 mm for the leaf ends for the 6 MV photon beam. Similarly, the penumbra width did not visibly change from the mean value of 3.6±0.2 mm for the leaf sides and 0.38±0.1 mm for the leaf ends for the 18 MV photon beam. This has also been previously verified.\(^{13}\) The smaller penumbra of the micro-MLC compared to the conventional MLC is expected since the micro-MLC is located farther from the x-ray source, therefore reducing the geometric penumbra. However, penumbras at deep depths are expected to be even broader, similar to a conventional MLC system. Furthermore, because the maximum field size that can be set by the jaws is only 9.8x9.8 cm\(^2\) when using the micro-MLC, there is a geometric limit to the variation in penumbra that can occur as a result of beam divergence, scatter, and transmission through the leaf ends.

The uniformity of the beam penumbra for asymmetric and off-axis field shapes has obvious advantageous. No special calculation of dose distributions with a treatment planning computer or beam data measurements are required as a function of leaf position. However, undulations in the beam penumbra which occur when the leaves are made to define angled straight edges is something which has yet to be investigated.
Chapter 6  
Results and Discussion

**Fig. 6.20**: Comparison of the 80% to 20% penumbras for a 6 MV photon beam in the crossplane direction for the lower X jaws or perpendicular to the direction of leaf motion (leaf side penumbra) for the MLC and micro-MLC at SSD = 100 cm and depth $d_{\text{max}}$. In the case of the MLC and micro-MLC, the jaws were maintained at $9.8 \times 9.8$ cm$^2$.

**Fig. 6.21**: Comparison of the 80% to 20% penumbras for a 6 MV photon beam in the inplane direction for the upper Y jaws or parallel to the direction of leaf motion (leaf end penumbra) for the MLC and micro-MLC at SSD = 100 cm and depth $d_{\text{max}}$. In the case of the MLC and micro-MLC, the jaws were maintained at $9.8 \times 9.8$ cm$^2$. 

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6.8 Interleaf Transmission and Leakage

When two opposing leaves meet, the curvature of the leaf ends result in radiation leakage. In addition, to prevent opposing leaf pairs from colliding with each other, a minimum separation is maintained between two closed leaves. This causes radiation leakage at the junction. Another source of radiation leakage is the small nominal gap that is maintained between two adjacent leaves to reduce friction during leaf movement. Radiation transmission through this gap is minimized by the tongue and groove design described earlier.

A. Irregular fields defined with MLC

The leakage between the leaves of the MLC was measured with the leaves completely closed and relative to that of a 10x10 cm\(^2\) field defined by the jaws. The minimum was found directly under the leaves while the maximum was found between the leaves. The results are presented in Fig. 6.22. The average interleaf transmission at SSD = 100 cm and depth \(d_{\text{max}}\) for the 6 MV photon beam defined by the MLC was found to be 12.8\% \pm 0.8\% when opposing leaves abut along the y-axis (see Fig. 5.2(a)). However, when the leaves are made to abut at 2.5 cm (see Fig. 5.2(b)) and 5 cm (see Fig. 5.2(c)) off the y-axis, the average interleaf transmission along the y-axis decreased to 0.57\% \pm 0.4\% and 0.50\% \pm 0.4\%, respectively. Although, from Fig. 6.2, the transmission at the position where the leaves abut was 17.6\%, 17.4\%, and 8.3\%, when the leaves were made to abut on the y-axis, at 2.5 cm off the y-axis, and 5 cm off the y-axis, respectively.

Similarly, for the 18 MV photon beam, the average interleaf transmission was found to be 10.4\% \pm 0.6\% when opposing leaves abut along the y-axis. However, when the leaves are made to abut at 2.5 cm and 5 cm off the y-axis, the average interleaf transmission along the y-axis decreased to 0.61\% \pm 0.03\% and 0.53\% \pm 0.03\%, respectively. Fig. B-14 depicts the transmission values for the 18 MV photons through the MLC leaves. However, from Fig. B.15, the transmission at the position where the leaves abut was 13.8\%, 13.9\%, and 6.6\%, when the leaves were made to abut on the y-axis, at 2.5 cm off the y-axis, and at 5 cm off the y-axis, respectively.
These results agreed with previously measured transmission between the leaves of between 1.5 and 2.0% for 6 MV photon beams and 1.5 to 2.5% for 18 MV beams.\textsuperscript{14,15} Leakage between the leaves was roughly 0.25% to 0.75% higher than transmission through the leaves. In addition, the attachment screws increased the transmission still further to a maximum of about 3%. Transmission through abutted (closed) leaf pairs was as high as 28% for 18 MV photons on central axis. Off axis, the abutment transmission decreased as a function of off-axis distance to as low as 12%.

B. Irregular fields defined with micro-MLC

The leakage between the leaves of the micro-MLC was measured with the leaves completely closed and relative to that of a 9.8×9.8 cm\textsuperscript{2} field defined by the jaws. The results are presented in Fig. 6.24. The average interleaf transmission at isocenter at a depth of \(d_{max}\) was found to be 13% ± 2% when opposing leaves abut along the y-axis. However, slight leaf misalignment has been observed to raise this value to 23%.\textsuperscript{16} When the leaves are made to abut at 2.5 cm and 5 cm off the y-axis, the average interleaf transmission along the y-axis decreased to 2.4% ± 0.5%. However, from Fig. 6.25, the transmission at the position where the leaves abut was 16.0%, 10.2%, and 3.0% when the leaves were made to abut on the y-axis, at 2.5 cm off the y-axis, and at 5 cm off the y-axis, respectively. Furthermore, if leaves are made to abut at 4.5 cm off the y-axis but the points where the leaves abut are staggered so that adjacent leaves were alternately fully extended and fully retracted, the transmission at the points where the leaves abut is reduced to a mean of 4.5%. This reduction is a result of the shape of the leaf ends since there is now no direct path between opposing leaves for the divergent x-rays to transverse.

Similarly, for the 18 MV photon beam, the average interleaf transmission was found to be 6.1% ± 0.8% when opposing leaves abut along the y-axis. However, when the leaves are made to abut at 2.5 cm and 5 cm off the y-axis, the average interleaf transmission along the y-axis decreased to 1.8% ± 0.3% and 1.2% ± 0.3%, respectively. Fig. B-16 depicts the transmission values for the 18 MV photons through the micro-MLC leaves. However, from Fig. B.17, the transmission at the position where the leaves abut
was 6.3%, 3.3%, and 1.3%, when the leaves were made to abut on the y-axis, at 2.5 cm off the y-axis, and at 5 cm off the y-axis, respectively. Staggering of the micro-MLC leaves for the 18 MV photon beam showed little reduction in transmission.

This agrees with previously reported values of transmission through and between the leaves for the 6 MV beam at isocenter of 1.9% and 2.4% respectively. Others have reported interleaf leakage between the micro-MLC leaves of 2% and intraleaf leakage through the depth of the leaves of 1.3%.

Although the transmission through and between leaves is slightly higher than those reported for the conventional MLC, the values are still lower than those for standard lead-alloy shielding blocks which have typical transmission values of around 3 to 3.5%. Transmission where the leaves abut on the central axis was relatively high. Maximum reductions in transmission can be achieved when the leaves overtravel by a maximum of 5 cm. This can be advantageous in clinical situations when the jaws can not be made to shield the points where opposing leaves meet.

![Graph](image)

**Fig. 6.22**: Transmission at the depth of dose maximum through closed leaf pairs for 6 MV photons when the MLC leaf ends were positioned to abut on the central axis, at 2.5 cm off central axis, and at 5.0 cm off central axis. [scan taken in direction perpendicular to the direction of motion of the leaves]
Fig. 6.23: Leakage through the closed MLC leaf ends for 6 MV photons when the MLC leaf ends were positioned at 0 cm, 2.5 cm, and 5.0 cm. [scan taken in the direction of motion of the leaves]

Fig. 6.24: Transmission at the depth of dose maximum through closed leaf pairs for 6 MV photons when the micro-MLC leaf ends were positioned to abut on the central axis, at 2.5 cm off central axis, and at 5.0 cm off central axis. [scan taken in direction perpendicular to the direction of motion of the leaves]
Fig. 6.25: Leakage through the closed micro-MLC leaf ends for 6 MV photons when the micro-MLC leaf ends were positioned at 0 cm, 2.5 cm, and 5.0 cm. [scan taken in the direction of motion of the leaves]

6.9 Comparison of Treatment Plans

The MLC treatment plan was done on CadPlan, while the micro-MLC treatment plan was done on BrainSCAN. Therefore, the two different systems introduce a variable also; however, it was assumed that CadPlan models the MLC well and BrainSCAN models the micro-MLC well. The calculated isodose distributions for the MLC and micro-MLC systems are displayed in Fig. 6.26 and Fig. 6.27. The films obtained from the treatment delivery are displayed in Fig. 6.28 and Fig. 6.29. It is evident from simple observation of these films that the micro-MLC-based plan was only slightly superior in its dose conformity to the conventional MLC-based plan. This is an indication that the effect of leaf width becomes less crucial when more beams are used in the treatment. The measured isodose distribution from the films is plotted in Fig. 6.30 and Fig. 6.31. The results suggest that the dosimetries of the MLC and micro-MLC systems are comparable in multiple field 3D conformal treatments. This has also been observed by other investigators. However, the smaller penumbra of the micro-MLC system may be advantageous in cases where the tumor is highly irregular or extends near the brain stem and spinal cord.
Comparison of the single isocenter, 9-field static micro-MLC and MLC treatments show similar coverage of the target volume. However, measurement of the 100% isodose contour is smaller in the case of the MLC compared to the calculated isodose contour whereas the opposite is true for the case of the micro-MLC.

In comparison with standard open radiation fields, those produced with the MLC or micro-MLC system give a more uniform dose to the target volume and reduce the dose to surrounding brain tissue. The treatment planning for intracranial targets is straightforward with the micro-MLC compared to circular collimation where multiple isocenters would be required. Consequently, a reduction in time and labour to the treatment planning process and treatment delivery are achieved. This would be even more so for fractionated stereotactic radiosurgery.

Guidelines established by the Physics Committee of the Radiotherapy Oncology Group (RTOG)\textsuperscript{19} can be used to assess radiosurgical treatment plans. The RTOG recommend that dose homogeneity within the target volume be evaluated based on the ratio of the maximum target dose to the prescription dose (MDPD). An MDPD ratio between 1.0 and 2.0 is desired; however, an MDPD ratio greater than 2.5 is considered unacceptable. The RTOG also recommend that conformity of the prescription isodose volume to the target volume be evaluated based on the ratio of the prescription isodose volume to the target volume (PITV). A PITV ratio between 1.0 and 2.0 is desired; however, a PITV ratio greater than 2.5 is considered unacceptable. Previous evaluation of a series of targets for stereotactic radiosurgery showed that most lesions treated with a micro-MLC system using between 3 and 5 static fields meet with RTOG guidelines.\textsuperscript{20} However, very irregular targets are best treated with intensity modulated radiotherapy to achieve an acceptable PITV ratio of between 1.0 and 2.0.

\textbf{6.10 Summary}

The in-phantom dosimetric parameters, such as the relative dose factor (RDF), scatter factor (SF), percent depth dose (PDD), tissue-air ratio (TAR), and tissue maximum ratio (TMR), are determined by their field shape created by the tertiary blocking. On the contrary, the in-air dosimetric parameter, namely the collimator factor (CF), is determined
by the square or rectangular field shaped by the secondary collimator jaws of the linac and is independent of any tertiary blocking. Therefore, the method of MU calculation is the same for MLC fields as the conventional method using Cerrobend blocks. However, as the MLC field becomes much smaller than the collimator jaw opening or is extremely irregular so that part of the blocked area is close to the central axis, the measured in-air output factor becomes significantly lower than that obtained by the collimator jaws alone. Furthermore, the surface of the jaws closest to the source provides the limit on the amount of scattered radiation contributing to the output.

The micro-MLC device allows minimal transmission through and between the leaves. Because of this minimal transmission, the linac jaws can remain fixed while the micro-MLC field size is changed. However, transmission between leaf ends is highly dependent on leaf alignment and it is critical that the leaves are properly aligned. Dosimetry data for single isocenter treatments using computer controlled field shaping with micro-MLC demonstrate the ability to conform the dose distribution to an irregularly shaped target volume.

![Calculated isodose distribution generated from the treatment plan using CadPlan for the MLC conformal fields.](image)

Fig. 6.26: Calculated isodose distribution generated from the treatment plan using CadPlan for the MLC conformal fields.
Fig. 6.27: Calculated isodose distribution generated from the treatment plan using BrainSCAN for the micro-MLC conformal fields.

Fig. 6.28: Image of the film exposed to 9 MLC shaped treatment fields at the midplane of the target volume.
Fig. 6.29: Image of the film exposed to 9 micro-MLC shaped treatment fields at the midplane of the target volume.

Fig. 6.30: Measured isodose distribution from the treatment delivery using the MLC to shape the target volume.
Results and Discussion

Fig. 6.31: Measured isodose distribution from the treatment delivery using the micro-MLC to shape the target volume.

6.11 References


Reducing the dose to normal tissues and, where advantageous, escalating the dose to the target volume, has been made possible by the increasingly sophisticated radiation therapy planning and treatment. This is a result of the advancements in technology, CT and MR imaging, 3D planning software, and more powerful computer hardware. Consequently, an increase in the use of non-coplanar beams, stereotactic techniques, and inverse planning has been observed.

Computer-controlled multileaf collimators (MLCs) are a powerful tool with which optimized conformal therapy dose distributions may be delivered. They are becoming a standard feature for beam shaping of modern megavoltage radiotherapy machines and consist of a large number of narrow, closely-abutting leaves, each independently controlled and positioned through the MLC’s controller computer. The physical characteristics of a tertiary multileaf collimator must ensure that its dosimetric characteristics resemble those of a beam block. The main limitation in adjusting the MLC to conform to the shaped field is the discrete step size. Optimization of the MLC field involves placing the largest number of leaf ends tangent to the field edges, while maintaining the same internal area, as originally prescribed. An MLC, due to its larger effective penumbra, may be unsuitable for use in cases when the tumor volume extends very close to critical normal structures. More recently, the advent of a micro-multileaf collimator (micro-MLC) with a smaller leaf width can better conform to the smaller targets of regions such as the brain and the head and neck. The usefulness of micro-MLCs, on the other hand, is limited by their relatively small maximum field size.

The method for the collection of beam data for two beam energies and various field size combinations was presented for the linac jaws alone (open fields), the conventional MLC, and the micro-MLC. Beam profiles and depth dose data were required for entry into the BrainLAB treatment planning system, before treatment plans could be generated for the micro-MLC.
The in-phantom dosimetric parameters, such as the relative dose factor (RDF), scatter factor (SF), percent depth dose (PDD), tissue-air ratio (TAR), and tissue maximum ratio (TMR), are determined by their field shape created by the tertiary blocking. On the contrary, the in-air dosimetric parameter, namely the collimator factor (CF), is determined by the square or rectangular field shaped by the secondary collimator jaws and is considered independent of any tertiary blocking. Therefore, the method of MU calculation is the same for MLC fields as the conventional method using Cerrobend blocks. As the MLC field becomes much smaller than the collimator jaw opening or is extremely irregular so that part of the blocked area is close to the central axis, the measured in-air output factor becomes significantly lower than that obtained by the collimator jaws alone. Furthermore, the surface of the jaws closest to the source provides the limit on the amount of scattered radiation contributing to the output.

The micro-MLC device allows minimal transmission through and between the leaves. Because of this minimal transmission, the linac jaws can remain fixed while the micro-MLC field size is changed. However, transmission between leaf ends is highly dependent on leaf alignment and it is critical that the leaves are properly aligned. Dosimetry data for single isocenter treatments using computer controlled field shaping with micro-MLC demonstrate the ability to conform the dose distribution to an irregularly shaped target volume. Careful dose verification, however, is necessary before clinical implementation. Since the housing of the micro-MLC is larger than standard circular collimators, clearance between the patient, gantry, and treatment couch is reduced, making some treatments no longer possible.
Section:

A.1 Quality Assurance

Following clinical implementation of any device, a routine quality assurance program must be implemented to ensure continual proper functioning of the device. The range and extent of use of the MLC in the clinic is a determining factor as to the type and frequency of tests performed. Checks related to both the data transfer over the network of the MLC files as well as the mechanical movement of the leaves must be considered. A quality assurance program for the Varian MLC in static mode has been suggested by Mubata et al.\(^1\) It addresses patient specific and periodic checks related to day-to-day reproducibility of the MLC files, leaf positional reproducibility, leaf calibration, virtue of the software and accuracy of the digitizer used to design the MLC field, alignment of the MLC carriage with respect to the secondary collimators, as well as light field and mechanical alignment. Again, like for the acceptance testing, rotational checks are performed to ensure that the added weight due to the MLC does not distort the isocenter position over time due to the effects of gravity. The following is a summary of the suggested quality assurance program for the Varian MLC. The pre-treatment and daily checks are carried out by the technologists, while the more detailed monthly and quarterly checks are performed by the physicists.

A. Pre-treatment

**QA Test #1**: Prior to the initial treatment of each patient, illumination of the MLC shaped fields onto the original simulation film or DRR should correspond within 2 mm at all boundaries.
Appendix A

B. Daily

**QA Test #2**: Visual verification of the leaf positions with respect to the outlined aperture shape should be inspected on the screen of the MLC workstation before each treatment to ensure that each leaf has moved to the proper placement.

C. Weekly

**QA Test #3**: The printed template of a stored MLC file involving all the leaves is compared with the light field projection of the MLC leaves. With the collimator and gantry angles set to 0°, the projection of the light field at 100 cm SSD is expected to be within 0.5 mm of the template.

D. Monthly

**QA Test #4**: The accuracy and reproducibility of leaf positions should be checked over the entire range of travel of the leaves. Calibration of the leaves is carried out using a stored MLC file in which all of the leaves, one carriage at a time, move from -16 cm (i.e., 16 cm across central axis) to +16 cm (i.e., 16 cm away from central axis) in 4 cm increments. With the collimator and gantry angles set to 0°, the difference between the projection of the light field at 100 cm SSD and the actual leaf settings should not exceed 1 mm for each increment.

**QA Test #5**: The veracity of the "shaper software" and the accuracy of the digitizer used to design the shape of the MLC aperture should be verified. The coordinates of a single point that is repeatedly digitized should all be within 0.5 mm of one another. In addition, placing a calibrated grid over the digitizer light box and digitizing a set of points should give readings within 1 mm of the actual coordinates of each point. These points should be chosen from within the clinically used area.

E. Quarterly

**QA Test #6**: The skew of the leaf carriage is checked with respect to the light field in order to verify that the MLC leaf carriage is properly aligned with the opposing X collimator jaw. Each of the MLC leaves from carriage A is positioned at -1.0 cm (i.e., 1.0
cm across the central axis) while the opposing X1 jaw is set to 1.0 cm. The leaves of carriage B and the X2 jaw are retracted. At an SSD of 100 cm only light transmitted from between the curved ends of the leaves should be visible, if the carriage is properly aligned with the opposing collimator jaw. This is depicted in Fig. A.1(b). On the other hand, if the carriage is skewed compared to the opposing collimator jaw, a triangle of light will be visible, as illustrated in Fig. A.1(a). The same check is performed for carriage B in conjunction with the X2 jaw.

![Diagram](image)

**Fig. A.1**: Diagram depicting the position of the MLC carriage and the opposing collimator jaw in the determination of the skew of the MLC carriage. The expected pattern from a properly aligned carriage is shown in (a), while the triangular view of the light field due to a misaligned carriage is shown in (b).

**QA Test #7**: A software check of the leaf carriage skew may be verified by examining the secondary feedback values for each leaf. The visual display unit of the MLC controller displays each leaf setting as well as each leaf's current position. When the MLC leaves are completely retracted, the actual position of all the leaves from one carriage should be within 5 counts. If the MLC carriage is skewed with respect to the collimator jaws, some of the leaves from a carriage will then have to travel greater distances in order to reach their rest positions. Therefore, if the secondary feedback values for leaves A1 to A26 or B1 to B26 vary by more than 5 counts, a radiation check of the leaf carriage skew should be performed.

**QA Test #8**: The skew of the leaf carriage may be checked with respect to the radiation field by setting the position of the leaves and jaws, as illustrated in Fig. A.2. All leaves
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are set to 7.0 cm away from central axis, except for leaves A4 and B23 which are both set to -6.5 cm. The X and Y collimator jaws are set to 15 cm and 26 cm, respectively. The collimator and gantry angles are set to 0°, and a film is placed at an SSD of 100 cm at depth $d_{\text{max}}$. The position of both carriages, the Y jaws, and the light field center are marked on the film and the film is exposed to 50 MU, 6 MV photons. The skew of carriage A is obtained from measuring the distance between the top of leaf A4 and the Y2 jaw in two places on the developed film. This corresponds to lines $P_AQ_A$ and $P_BQ_B$. The difference in length between these two lines should not be more than 1.0 mm. Similarly, the skew of carriage B is obtained from measuring the distances between the bottom of leaf B23 and the Y1 jaw.

Fig. A.2: Diagram of the set-up for the determination of the skew of the leaf carriage with respect to the radiation field edge.

The A and B carriages were found to be aligned with the Y1 and Y2 jaws. The film used to detect the proper alignment of the leaf carriages is shown in Fig. A.3. The line $P_AQ_A$ and $P_BQ_B$ were 22.4 cm and 22.5 cm in length, respectively. The difference of 1 mm is just within the tolerance meaning that the A carriage and Y2 jaw were aligned. Due to scattered radiation detected on the film, it is somewhat difficult to judge exactly where the edge of the leaf and the collimator jaws are with respect to the radiation field. For this reason, part of the variation could be a result of measurement error. Similarly, lines $R_AS_A$ and $R_BS_B$ were both 22.4 cm in length.
Fig. A.3: The pattern of the exposed film used for measuring the skew of the MLC leaf carriages with respect to the radiation field.

**QA Test #9:** The MLC field radiation center is checked by first setting the gantry at $0^\circ$. By rotating the collimator from $270^\circ$ to $90^\circ$ and noting the position of the cross hairs for each rotation, verification that the cross hairs define a circle of radius less than 1.0 mm about the isocenter is attained. Next, a film is placed at SSD 100 cm and depth $d_{\text{max}}$ in a solid water phantom. The jaws are set to $20\times20$ cm$^2$ and all leaves are closed, except for leaves 13 and 14 from both carriages which are set to 10 cm, as illustrated in Fig. A.4. The film is exposed to 50 MU for each of the collimator rotations of $315^\circ$, $0^\circ$, and $90^\circ$. On the developed film, the center of the spokes should intersect within a circle of radius 1.0 mm.

![Diagram of the set-up for the determination of the MLC field radiation center.](image)

**Fig. A.4:** Diagram of the set-up for the determination of the MLC field radiation center.
On the developed film, the interleaf transmission as well as the leakage radiation where the leaves abut is evident in Fig. A.5. By drawing a line through the center of each spoke, a triangle is formed. The circle that encompasses this triangle has a radius of roughly 0.5 mm and therefore is within tolerance.

Fig. A.5: The spoke pattern of the exposed film used for the determination of the MLC field radiation center.

QA Test #10: Over time, gravity may affect the positions of the MLC leaves due to wear and tear of the pieces on which the MLC carriage is mounted. This may lead to disagreement between the position of the carriage at gantry angles of 90° versus 270° due to backlash. The MLC carriage sag may be determined by first setting the collimator at 0°. By rotating the gantry from 270° through 0° to 180° and noting the position of the cross hairs for each rotation, verification that the cross hairs define a circle of radius less than 1.0 mm about the isocenter is obtained. Next, a film is placed on its edge, perpendicular to the length of the couch and the center of the film is placed roughly at the
isocenter. The MLC leaves are all set to 0.5 cm and the X jaws are set to 1.0 cm behind the leaves, while the Y jaws are each set to 13.0 cm, as illustrated in Fig. A.6. The film is exposed to 50 MU for each gantry position of -120°, 0°, and 120°. On the developed film, the center of the spokes should intersect within a circle of radius 1.0 mm.

Fig. A.6: Diagram of the set-up for the determination of the MLC carriage sag.

The film was exposed for the gantry at 240°, 0°, and 120°. From the exposed film, shown in Fig. A.7, the attenuation of the beam in each spoke is observed by the lightening of each spoke across the film. Again the center of each spoke was determined and a line drawn so that the intersection of these lines could be observed. The maximum variation between any two lines was found to be 5 mm implying the variation would define a circle of radius 2.5 mm. This value exceeds the maximum tolerance.

Fig. A.7: The spoke pattern of the exposed film used for the determination of the MLC carriage sag.
In addition, the removal of inactive patient files should be performed periodically. A check that all active interlocks are operational as well as the functionality of the network system with respect to the stability and accuracy of the data transfer should be performed. Film scans to evaluate that the inter-leaf transmission is less than 3% and the abutted leaf transmission is less than 25% are all required.

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B.1 Introduction

This appendix presents dosimetric data for the 18 MV photon beam and complements data of Chapter 6 which dealt with the 6 MV photon beam.
B.2 Collimator factor

Fig. B.1: CF measurements for an 18 MV photon beam defined by the jaws. The "collimator exchange effect" for rectangular fields with a side of equivalent square equal to 10 cm and elongation ratios ranging from 2 to 7 are represented.

Fig. B.2: Comparison of the conventional collimator factor for the 18 MV photon beam with jaws fixed at 25×25 cm² with the measured collimator factor for fields defined by the MLC.
B.3 Relative Dose Factor

Fig. B.3: RDF measurements for an 18 MV photon beam defined by the jaws.

Fig. B.4: Relative dose factor for 18 MV photons, 200 MU, MLC leaves in the "cross-field" position.
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B.4 Scatter Factor

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Fig. B.10: Percentage depth dose curves for the 18 MV photon beam in water when the jaws and the micro-MLC are used to define the various circular field sizes of diameter 0.6 cm, 1.2 cm, 1.8 cm, 2.4 cm, 3.0 cm, 3.6 cm, 4.2 cm, 5.1 cm, 6.0 cm, 6.9 cm, 8.0 cm, 9.1 cm, and 10.0 cm. The jaws were fixed at 9.8×9.8 cm².
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B.6 Penumbra

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![Graph showing leakage through the closed micro-MLC leaf ends for 18 MV photons when the micro-MLC leaf ends were positioned at 0 cm, 2.5 cm, and 5.0 cm.]

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