INVESTIGATIONS ON DYNAMIC INTENSITY MODULATED PHOTON FIELDS IN RADIATION THERAPY

By
HOMAYON PARS 

B.Sc., Concordia University, Montreal, Canada, 1993
M.Sc., University of British Columbia, Vancouver, Canada, 1997

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Department of Physics And Astronomy
The University of British Columbia
Vancouver, Canada

Date Sept. 24, 2001
Abstract

The objective of this thesis is to investigate the validity of a class of intensity modulated radiation beam delivery (IMRT) with the view towards its clinical implementation.

Intensity modulated photon beam (IMB) delivery is a new radiation delivery technique in external beam radiation therapy. The technique is complex in planning, delivery and verification and frequently requires a large number of beams to achieve a desired dose distribution. Therefore, an important consideration in exploiting the advantages offered by this new technology is to study the degree of accuracy with which highly conformal dose distributions can be achieved.

A methodology appropriate to dosimetry and quality assurance (QA) of dynamic fields is proposed in this thesis and the theoretical modeling was improved based on the results of this work.

Dynamic fields utilizing the motion of multileaf collimators (MLCs) and backup diaphragms were used to produce 1 and 2D intensity maps arbitrary in orientation with respect to collimator axis. Dynamic wedge dose profiles produced by moving diaphragms are accurate to within ±1% of those produced with conventional techniques. The accuracy of the omni wedge with arbitrary field orientations are found within ±2° of calculations.

A dosimetric verification technique used to monitor the dynamic beam delivery for IMRT plans is introduced. The design considerations and clinical evaluation of a QA phantom, facilitating the measurement of IMRT dose distribution and conversion of photon fluence to machine deliverable monitor units are described. Benchmark tests and clinical examples for IMRT dose verification techniques are carried out to demonstrate the accuracy of the technique. in-vitro and in-vivo dose measurements of dynamic IMB's were in good agreements with the calculation model.
A theoretical error analysis on the influence of systematic and random field perturbations in highly conformal beam deliveries with emphasis on its dosimetric effects is also presented. It is shown that field inaccuracies in dynamic deliveries in the order of ±1.0 mm could lead to dose errors of 20% or more. Based on the findings of our study, we have proposed a QA procedure unique to the delivery of dynamic beams.
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Any remaining deficiencies are most likely lost in translation and are entirely my fault.

I would like to dedicate this work to the memory of my brother, Mehran.

1H. Parsai, Spring 2001

“Struggle or calm
broke or a king
life’s but wind and a dream
perhaps describing some other thing;
and with the end
all will be the same again
and all will be well”

...Khayam

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1This thesis is type-set and compiled in \LaTeX. Copies in postscript, PDF, or HTML formats are available from the author.
This chapter presents an introductory background, the literature review and the rationale for this study. A summary of the thesis is also outlined.

1.1 Background: Radiotherapy

The objective in external beam radiation therapy in the treatment of cancer is the eradication of tumor cells with ionizing radiation either as a single mode of treatment or in combination with other forms of treatment such as surgery, chemotherapy and/or brachytherapy. The goal is to conform the spatial distribution of the prescribed radiation dose to the precise three dimensional configuration of tumor while keeping the dose to the surrounding healthy tissues below their tolerance.

Traditionally, automated beam shaping devices such as lead blocks, physical wedge filters, and compensators have been used to minimize the dose to surrounding critical structures and to modify the incident radiation intensity to the desired 2D projection of the target. They are meant to improve the homogeneity of the dose in the target volume (target volume is the volume containing the tumor plus some additional margin
to account for the spread of any microscopic diseases) by altering both the distribution of the primary photon fluence and the Compton scatter in the medium [1, 2].

However, conventional techniques may not be adequate as they are often unable to protect the dose-limiting tissues (tissues which have low tolerance to ionizing radiation) in complex treatment set-ups. This is due to inaccuracies in tumor coverage and the inability to avoid healthy organs that are in the path of the radiation [3, 4, 5, 6].

1.2 Literature Review

The introduction of multileaf collimators (MLC) into the design of medical linear accelerators has increased the efficiency of external radiation beam delivery. Figure 1.1 is a schematic diagram of MLC used to conform to the radiation field to the 2D projection of a target. MLCs were first introduced by Takahashi [7] and later were used in the development of 3D conformal techniques. Integration of MLCs into radiation delivery techniques, along with advances in computer treatment planning software, have allowed a more accurate conformation of the dose distribution to the target volume. This has led to capabilities where relatively complex geometries with multiple beams can conform to the 3D shape of the target volume.

However, despite the realization of 3D conformal techniques in clinical settings, there are cases where the degree of conformality achievable with these techniques have been found to vary greatly as a function of tumor site, extent of the disease, and the location of tumor in relation with critical structures [8]. Therefore, 3D techniques with uniform field intensities alone cannot conform well enough to arbitrary 3D geometries. This is especially true in case of concave tumors, particularly when dose-limiting tissues lie in the concavity [9]. These limitations suggest the need for further refinement in improving the current techniques.
Figure 1.1: A block diagram of MLC conforming to the shape of the target volume.

The most promising of the new approaches in 3D conformal techniques is the use of target-specific intensity-modulated beams that can achieve the simultaneous goals of uniform irradiation of the target volume and sparing of the sensitive structures at risk [10, 11, 12, 13, 14, 15, 16, 17, 18, 19].

Intensity modulated radiation therapy (IMRT) is an elegant approach in planning and delivery of external radiation beams [20, 21]. In its most advanced form, it utilizes MLCs and other beam defining parameters to produce fluence modulation across the radiation field [22, 23, 24, 25, 26, 27, 28, 29]. This enables one to fine-tune the dose distribution on an individual beam ray (or pencil beam) such that the intensity of the pencil beams can vary from 0 to 100% independent of all other pencil beams [30, 31, 32].

In its simplest form, IMRT can be realized using rather straightforward methods such as mechanical compensators that can be designed and planned with existing three
dimensional treatment planning techniques and iterative optimization [1, 9, 33, 13, 34, 35]. The drawback of such an approach is the time it takes to manually change compensators between fields and to accurately model radiation transport that include scatter and beam hardening effects (Beam hardening is a filtering effect in which the low energy end of the photon spectrum is preferentially absorbed in the material). In Chapter 2 of this thesis two methods will be described in which the intensity modulated beams (IMBs) can be generated using MLCs and backup diaphragms: the step and shoot technique and the sliding window technique. For these methods, calculations of weights of optimized fluences are performed by means of inverse-optimization techniques [36, 37, 38, 39, 40, 41, 42, 43, 44, 45, 46]. The next chapter will briefly discuss inverse and forward optimization techniques.

Figure 1.2 illustrates the major differences in methods of radiation delivery techniques between conventional and 3D conformal field shaping.
Figure 1.2: Advances in target conformation techniques in external beam radiation therapy. Two dimensional projections of the target volume are shown with a) conventional b) conformal but uniform intensity, and c) conformal intensity-modulated field shaping. Best conformation results with MLC shaping; however, the best dose distribution and conformation result from modulating the incident beam as well.
1.3 Clinical Rationale

The clinical rationale of IMRT is that it allows the escalation of dose, potentially leading to local control and survival for the same or lower probability of normal tissue complications [47, 48, 49, 50, 51]. Its ability to manipulate the intensities of individual rays permits a greatly increased control over target dose. Furthermore, intensity modulated treatments are fully computer-controlled deliveries, therefore, they lend themselves to automation to a greater degree than standard conventional treatments. Thus, in the future, intensity modulated treatments may be preferable not only from the point of view of improved outcome for the patient but also because of the increased efficiency in planning and delivery.

1.4 Objective of This Project

While there has been much progress in the development of the theory of the delivery of IMBs in the past decade, at the present time there are no in vivo models proposed to verify the accuracy with which treatment plans match the delivered doses or to determine if dose distributions in the target volume are spatially accurate and sufficiently localized to allow significant dose escalation.

The objective of this thesis is to investigate the validity of a class of IMRT, namely dynamic IMBs, with the view toward its clinical implementation. The degree to which IMBs can deliver highly accurate dose distributions to regions in close proximity to critical structures is studied. Furthermore, the need to verify the precise positioning of the MLC and other beam defining parameters in dynamic deliveries becomes important to ensure accurate and safe delivery to the patient. Therefore the weaknesses and strengths of such a delivery scheme is studied in detail in this thesis.
1.5 Thesis Summary

The objective of the thesis, the literature review and the rationale for this research project has been described in this chapter. Basic terminologies and definitions important in this thesis are discussed in Chapter 2. Chapter 3 describes the relevant materials used in the experimental portion of our work.

The first part in our experimental work described in Chapter 4 is the implementation of a dynamic delivery utilizing MLCs and backup diaphragms for a commercially available medical electron linear accelerator. 1D dynamic fields produced with motion of the backup diaphragms are used in combination with an orthogonal wedge field and one un-wedged field to create wedged intensity maps with orientation independent of collimator rotation axis (omni wedges).

Chapter 5 describes the dosimetry and verification techniques used in the delivery of IMBs. Our approach is to use beagle dogs as the subjects of IMRT irradiation to a target volume adjacent to the spinal cord. Physical measurements of the dose distributions are made in phantoms, in deceased dogs, purposely bred subject dogs, and dogs with spontaneous head and neck tumors. An evaluation of known normal tissue response from IMRT treatments with those seen in conventional radiation therapy is carried out.

Chapter 6 studies the effect and influence of field perturbation on dose distribution due to positional inaccuracies in the beam defining parameters. Correlation of simulations of IMRT beam delivery with planned dose distribution obtained using the available treatment planning algorithms is presented to illustrate the significance of leaf positioning errors in dynamic IMRT. Finally Chapter 7 summarizes the results of this study and concluding remarks are presented.
This chapter is a brief overview of some of the relevant terminology and definitions useful in this study. Concepts of forward planning, inverse planning, different techniques in delivering IMBs using MLCs are discussed. Types of interactions of ionizing radiation with matter important in radiotherapy are also presented.

2.1 Treatment Planning: An Optimization Technique

Treatment planning is an optimization problem. The objective is to find the treatment plan that best meets the goals of delivering a tumorcidal dose and maintaining low rates of complications.

2.1.1 Forward Planning

Forward planning technique is a trial-and-error procedure where beam parameters, such as beam angle, beamweights and other indices defining a treatment plan are selected manually to yield an optimum plan with the goal of delivering the maximum allowable dose to the tumor. The technique is to try a number of different possible approaches that
best achieve the objectives of the plan, compute the dose distribution using available treatment planning algorithms, evaluate the plan, then repeat the task until the plan meets the prescribed criteria. This is known as forward treatment planning. It is used in conventional methods of beam optimization and is achieved chiefly by human decision making.

However, it is difficult to create treatment plans by forward planning techniques for IMBs. This is because there are too many possibilities to explore in order to arrive at an optimum treatment. IMRT requires the variation of beamweights of hundreds or even thousands of beam rays to achieve an optimum dose distribution. This is not practical with conventional means. For this inverse techniques are developed.

### 2.1.2 Inverse Planning

The inverse treatment planning technique starts from the desired dose distribution and employs mathematical inversion to derive a set of optimal intensity modulated profiles for beams with predefined directions. The technique can be thought of in terms similar to
Computed Tomography (CT), in that it models a conformal dose around the target and backprojects through different tissues to find a non-uniform fluence that must be delivered to give the uniform dose. Therefore, to achieve optimal uniform dose distributions with IMRT, conventional forward planning must be replaced with inverse planning techniques in which the beams are designed from specifications of the required dose distribution, created by the computerized optimization of weights of beam rays rather than by trial-and-error of homogeneous beam profiles. Figure 2.1 is the schematic representation of beam delivery generated with forward and inverse techniques.

Methods by which IMRT is delivered can be realized by MLCs where multi-segmented or fully dynamic leaf movements are used in producing the modulated beam intensities.

## 2.2 Field Modulation

### 2.2.1 Internal Wedge

One method of producing field modulation to achieve uniform dose distributions is the insertion of beam modifying devices in the radiation field. External or motorized wedges are examples of such devices that are capable of varying the incident fluence intensity by means of differential attenuation of the beam (see Figure 2.2). The beam characteristics of such fields are described by parameters such as transmission factor, wedge angles, and the wedge factor. They are defined as follows:

- The wedge angle is a measure of how much an isodose line is tilted by the presence of a physical wedge and, while not quite descriptive of its full effect, it is used to quantify the effect of each wedge filter and is loosely defined as the compliment of the angle between the central axis line and the line tangent to an isodose curve at 10.0 cm depth in water equivalent medium. Scattered radiation causes the angle
of the isodose curve to decrease with increasing depth.

- Wedge factor is the ratio of beam output with and without the wedge in the field measured at the centre of the beam, at a given depth.

2.2.2 MLCs

Although, traditionally fluence modulation has been performed by wedges or cerrobend blocks, the process would be highly labour-intensive considering the large number of beams that may be needed for optimum intensity modulated treatments. With recent advances in computer-controlled beam shaping systems in commercial medical linear accelerators (linac) and the development of MLCs, it is possible to modulate the intensity of the beam through multi-segmented narrow slit-fields or by virtue of fully dynamic MLC movement.

Figure 2.2: Cross sectional view of linac head (left) showing various components. A round-end design MLC (right) illustrates the leakage through the leaf ends of two opposing leaves.

Because MLCs are primarily an efficiency device with original design intent to replace lead-blocks for field shaping, its use as a field modulator in IMRT demands that they
must be sufficiently thick to provide the necessary attenuation and sufficiently narrow to provide the required spatial resolution in the direction normal to the leaf motion. The MLC used in this study is mounted in the treatment head of an Elekta linac (Elekta Oncology Systems, Crawley, UK) and is shown schematically in Figure 2.2. They have round-end design with two banks of tungsten leaves. There are 40 leaf-pairs, each projecting to 1.0 cm in thickness at 100 cm distance from the radiation source. The round-end design can keep the penumbra spatially invariant. This design type ensures uniform dose distributions across the field and eliminates spatial dependence of dose gradient near the field edges. A disadvantage of such a design however, is the partial transmission through the round-end of the leaf which may cause differences in calculation and measurements of dose distribution and needs to be accounted for [52]. Figure 2.2 (left) shows the transmission through the leaf ends schematically. To account for this effect, leaf transmission is approximated by a function describing the radiation transport through the MLC.

\[\text{This is the geometric penumbra caused by the fact that the x-ray source is not point-like but rather finite in size.}\]
Chapter 2. Basic Concepts in IMRT

Table 2.1 summarizes the specifications of the type of MLC used in our study.

Table 2.1: Specifications of Elekta MLCs. Over-travel is the distance leaves can travel pass the mid-line of beam axis.

<table>
<thead>
<tr>
<th>MLC Design</th>
<th># of leaf pairs</th>
<th>Physical thickness at iso.</th>
<th>%</th>
<th>Max. leaf velocity</th>
<th>Over-travel at iso.</th>
<th>Max F.S</th>
</tr>
</thead>
<tbody>
<tr>
<td>round-end</td>
<td>40</td>
<td>7.5 cm</td>
<td>1 cm</td>
<td>1.8%</td>
<td>2 cm/sec</td>
<td>12.5 cm</td>
</tr>
</tbody>
</table>

The following two sections will describe the methods with which radiation intensity is modulated with MLCs.

2.2.3 Static Field Modulation

While the research work described here focuses on dynamic fields, we will briefly discuss the beam modulation by means of step-and-shoot or multi-segmented technique.

The modulation of the beam intensity in static mode is realized by a few irregularly shaped MLC subfields. In contrast to the fully dynamic delivery technique (see Section 2.2.4), in the step-and-shoot technique, the MLC leaves are moved through a sequence of discrete steps and dose increments. The beam is only turned on to deliver a fraction of the total dose while the leaves maintain their position in the field. With the beam turned off, the leaves are driven to the next predefined positions and the next segment of delivery is carried out. This pattern in delivery is repeated for a number of steps until all the required subfields are completed so that the superposition of the subfields combined will yield the total dose distribution desired by the planning algorithm [53]. Figure 2.3 is the block diagram representation of a step-and-shoot delivery using MLCs.
Chapter 2. Basic Concepts in IMRT

2.2.4 Dynamic Field Modulation

Dynamic methods of intensity modulation include any technique in which the MLCs or backup diaphragms are moving while the radiation beam is being delivered. The intensity modulation of the beam is achieved by varying the horizontal separation of the opposing
Figure 2.4: Schematic diagram of a dynamic delivery where the beam modulation is realized by sweeping a window of MLCs with varying widths and velocities across the field.

Leaf-pairs. Figure 2.4 is the block diagram of dynamic delivery of the MLC in producing field modulations of various intensities as a function of time. Since it is the moving leaves that generate the intensity modulation, we will briefly discuss the equations that describe the motion of the leaf-pairs. The intensity profile $I(x)$ (expressed in monitor units\(^2\) (or time)) making up the IMB as a function of position in the radiation field may be described as follows:

$$I(x) = t_t(x) - t_l(x)$$  \hspace{1cm} (2.1)

where $t_t$ and $t_l$ are the cumulative times at which the trailing and leading leaves are at position $x$ respectively while the radiation beam is being delivered. Differentiating Equation 2.1 yields:

$$\frac{dI}{dx} = \frac{dt_t(x)}{dx} - \frac{dt_l(x)}{dx}$$  \hspace{1cm} (2.2)

\(^2\)Monitor units is a cumulative radiation dose which increases monotonically with total elapsed time while the beam is on. If the dose rate is constant, they are directly proportional.
Chapter 2. Basic Concepts in IMRT

and if \( v \) represents the velocity of the \( i \)th leaf at location \( x \) in the field, then the intensity profile in MU as a function of distance is:

\[
\frac{dI}{dx} = \frac{1}{v_i(x)} - \frac{1}{v_l(x)}
\] (2.3)

Spirou et al [16] have shown that for cases where \( dI/dx \) is positive (the gradient of intensity profile is positive), the optimal solution is always to move the leading leaf at maximum speed, \( v_{max} \). This will make the negative part of Equation 2.3 minimal and opens the field as fast as possible to obtain positive slope. That is, from Equation 2.3 we have:

\[
\begin{align*}
  v_l &= v_{max} \\
  v_t &= \frac{v_{max}}{1 + v_{max} \left(\frac{dI}{dx}\right)}
\end{align*}
\] when \( \frac{dI}{dx} \geq 0 \) (2.4)

Similarly when the gradient of the intensity profile \( I(x) \) is negative, the trailing leaf is moved at its maximum allowable velocity to make the positive term of Equation 2.3 minimal:

\[
\begin{align*}
  v_l &= v_{max} \\
  v_t &= \frac{v_{max}}{1 - v_{max} \left(\frac{dI}{dx}\right)}
\end{align*}
\] when \( \frac{dI}{dx} < 0 \) (2.5)

Therefore the above equations describe the velocity sequence needed to modulate the intensity profile during a continuous, unidirectional sweep of the independently controlled MLC leaves while delivering the dose at a constant rate.

The dynamic modulation with MLC has an advantage in that any number of intensity levels can be delivered without a significant increase in treatment time. Furthermore, in comparison to static intensity modulation, optimized intensity modulated fields can often be realized more precisely in dynamic mode, especially when steep dose gradients are required in the field. Because of the continuous uninterrupted leaf motion in dynamic deliveries, they are capable of producing "smoother" dose profiles than static deliveries (see Chapter 4). The problem of underdosage and tongue-and-groove effects are also
avoided to some extent in dynamic delivery by means of leaf synchronization\(^3\) [54].

However, a disadvantage of the method is the potential increase in dosimetric errors because of the large number of MLC segments and the complexity of the intensity pattern in combination with possible intratreatment motion of the target [31]. In Chapter 6 we will further study the potential dosimetric consequences of field errors in dynamic beam delivery.

### 2.3 Charged Particle Equilibrium

Another topic that needs to be discussed briefly in this chapter is the concept of charged particle equilibrium (CPE) in radiation therapy as it has practical importance in conversion of transfer of energy into a medium (kerma) to deposition of energy in the medium (absorbed dose).

The idea of CPE is that there should be as many electrons set in motion as are brought to rest in a given volume, \(V\). That is, the energy carried in and that carried out are balanced for both directly and indirectly ionizing radiations [55]. Figure 2.5 illustrates the required conditions for CPE to exist. In the context of dose distribution with ionizing radiation in biological tissues, CPE is required to ensure uniform dose distribution to the boundaries and across the volume of interest. In dynamic IMRT, however, because of the narrow sliding field-widths which are often comparable to the size of the penumbra, CPE is difficult to achieve. To account for loss of CPE at the boundaries of the volume of interest, the intensity may be increased.

\(^3\)The tongue and groove effect is due to protruding elements on one side of the leaf where their physical thickness is increased and therefore cast thin shadow strips in the field that causes the overall dose to decrease.
2.4 Basic Interactions of Ionizing Radiation

Interactions of electromagnetic radiation with matter may be described by four major processes. Coherent scattering (Rayleigh scattering), photoelectric effect, Compton or incoherent scattering, and pair production [55, 56, 57]. Any of these interactions can be characterized by (a) the attenuating characteristics and (b) the attenuation coefficient of the beam in the absorbing material. Reduction in the number of photons \( d\psi \) in the medium is proportional to the number of incident photons \( \psi_0 \) and the thickness of the material \( dx \):

\[
d\psi = -\mu \psi dx
\]

integrating the above equation yields:

\[
\psi = \psi_0 \exp(-\int_0^x \mu_i dx)
\]

or

\[
\psi = \psi_0 e^{-\mu x}
\]
where $\mu_t$ is the linear attenuation coefficient, proportional to the incident radiation energy and has units of $1/\text{cm}$ if $x$ is expressed in cm. The total attenuation coefficient of the incident beam is the sum of the individual coefficients:

$$(\mu_t/\rho)_{\text{total}} = \frac{\sigma_{\text{coh}}/\rho + \tau_{\text{ph}}/\rho + \sigma_{\text{incoh}}/\rho + \pi_p/\rho}{\text{therapeutic range}}$$

$$(\mu_t/\rho)_{\text{total}} = \frac{\sigma_{\text{coh}}/\rho + \tau_{\text{ph}}/\rho + \sigma_{\text{incoh}}/\rho + \pi_p/\rho}{\text{diagnostic range}}$$

where $\sigma_{\text{coh}}$, $\tau_{\text{ph}}$, $\sigma_{\text{incoh}}$, and $\pi_p$ are attenuation coefficients for coherent scattering, photoelectric effect, Compton scattering, and pair production respectively. Table 2.2 shows the relative importance of various interactions of photons with matter.

Table 2.2: Various interactions of electromagnetic radiation with matter showing energy range and their relative importance in megavoltage radiation therapy.

<table>
<thead>
<tr>
<th>Photon interaction</th>
<th>Energy range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coherent scattering</td>
<td>$\leq 10$ keV</td>
</tr>
<tr>
<td>Photoelectric</td>
<td>$10 - 100$ keV</td>
</tr>
<tr>
<td>Compton</td>
<td>$0.1 - 3$ MeV</td>
</tr>
<tr>
<td>Pair production</td>
<td>$\geq 1.02$ MeV</td>
</tr>
</tbody>
</table>

Of all the above processes mentioned in Table 2.2, the Compton process is the most important form of interaction in biological tissues for both therapeutic and diagnostic energies and so it will described in more detail.

Let us consider collisions of a photon with a free electron (Strictly speaking, no electron is unbound since even the outer electrons are bound by a few electron volts, however if the binding energy of the electron is much less than the energy of the photon, it can be considered as free electron). The electron receives a portion of the incident photon energy and as result is ejected at an angle relative to the incident photon direction. The energies of the ejected electron and the scattered photon can be calculated from the principles of conservation of momentum and energy and are given as [56, 58]:

$$\begin{align*}
\text{E_{incident}} &= \text{E_{ejected}} + \text{E_{scattered}} \\
\text{P_{incident}} &= \text{P_{ejected}} + \text{P_{scattered}}
\end{align*}$$
\[ E_c = h\nu \frac{\alpha(1 - \cos \phi)}{1 + \alpha(1 - \cos \phi)} \]  
(2.10)

\[ h\nu' = h\nu_0 \frac{1}{1 + \alpha(1 - \cos \phi)} \]  
(2.11)

where \( h\nu_0, h\nu', \) and \( E_c \) are the energies of the incident photon, the scattered photon and the ejected electron, respectively, \( \alpha = h\nu_0 / m_e c^2 \), where \( m_e c^2 = 0.511 \text{ MeV} \) is the rest energy of the electron and \( \phi \) is the angle of the scattered photon.

Since the Compton interaction is a collision of a photon with a free electron, the energy of the photon must be much larger than the binding energy of the electron. This is in contrast to the photoelectric effect which is more probable when the energy of the incident photon is equal to or slightly greater than the binding energy of the electron. Thus as the photon energy increases beyond the binding energy of the K electrons, the photoelectric effect decreases with energy and the Compton effect becomes more important. However, the probability of the Compton interaction also decreases with increasing energy after a certain point [57].

When Compton electrons are produced in biological tissues, they undergo multiple scattering and lose energy predominantly by excitation and ionization, creating secondary electrons and free radicals that cause damage to DNA. Approximately 2/3 of damage caused by x-ray in tissue is mediated by free radicals. This results in deposition of energy in the medium which is quantized by the concept of absorbed dose (1 Gy = 1 J/Kg). Also, if the Compton electron interacts with the electromagnetic field of a nucleus it suffers a sudden deceleration by the strong attraction of positive nucleus and negative electron. The electron recedes from the collision with a loss of energy which appears as a photon with energy \( h\nu \) (Bremsstrahlung radiation).
This chapter describes the experimental materials used in this study. In addition to the description of the linear accelerator, the radiation detectors used in measuring the dose distribution such as films, diodes and TLDs are also discussed.

3.1 Linear Accelerator: The Linac

A commercial linac (Elekta Oncology Systems, Crawley, UK) capable of generating high frequency electromagnetic waves to accelerate charged particles to energies ranging between 4 – 25 million electron volts (MeV) was used in our study. Figure 3.1 is a block diagram of the major components of a linac common in most modern radiotherapy machines. The source of the electrons is a heated cathode in an electron gun. The energetic electrons produced are directed to collide with a high-$Z$ (normally tungsten) target material where they are stopped with the emission of bremsstrahlung x-rays. To control the beam uniformity and flatness, cone shaped flattening filters are used to reduce the dose rate at the beam centre by means of differential attenuation of photons and they produce a symmetric beam across the the field. Finally to shape the beam to the desired
Fig. 3.1: Block diagram of a typical electron linac showing its major components.

projection of the target volume, the radiation beam is collimated through components such as primary, and secondary collimators, MLC, and custom made blocks.

3.2 Dosimetry Equipment

Dosimetry and quality assurance in radiotherapy is an important step towards a successful treatment outcome. Existing methods in dosimetry of high-energy beams include the use of dosimeters such as ionization chambers, thermoluminescent dosimeters (TLDs), solid state detectors, and films. While the first three detectors are generally used in one dimensional point-like dose measurements, radiographic films can be used in measuring
the dose distribution in 2D where spatial resolution and simultaneous integration of dose at all points may be desired. Although not used in this study, other detectors such as electronic portal imaging devices (EPID) are emerging as both verification and dosimeters in quality assurance of treatments [59, 60, 61].

3.2.1 Ionization Chamber

Ionization chambers have calibration factors that are traceable to standard labs which are used in a calculation of the absorbed dose. Figure 3.2 illustrates the schematic representation of a 0.6 cm\(^3\) ionization chamber used in our study. The material from which the wall of the chamber is made of is matched in atomic number to that of air\(^1\) in the cavity. The reason for this is to ensure that the energy spectrum of the electrons liberated in the wall of the chamber is similar to that in air or the medium in question.

The chamber is connected to an electrometer and the charge liberated in the chamber upon irradiation is collected by the central electrode and converted to absorbed dose by applying appropriate correction factors.

3.2.2 Thermoluminescent Dosimeters

Certain crystalline materials, when exposed to ionizing radiation, emit light in the form of fluorescence which is proportional to the amount of radiation to which they are exposed. When exposed to high temperatures, electrons are liberated to recombine with holes and result in the emission of light photons. Figure 3.3 shows the energy bands of the crystalline material and possible transmission modes corresponding to the emission of thermoluminescence photons.

Lithium fluoride (LiF) was the material used in our study. It has been shown to be

\(^{1}\)Effective atomic number of air: 7.78
stable with almost negligible fading at room temperature. It has a low effective atomic number\(^2\) which does not differ greatly from that of tissue\(^3\) and therefore the absorbed doses are comparable in both media.

Since these material are artificially doped with impurities, the electron-traps produced in crystals have different physical characteristics which will change the sensitivity and response in each material. Therefore, TLDs need to be calibrated in a known radiation field prior to conversion of their response to dose. The sensitivity factor for a TLD is calculated as follows:

\[
EC_{C_i} = \frac{<TL>}{T_{Li} - T_{bkg}}
\]  

\(^2\)Effective atomic number of LiF: 8.31
\(^3\)Effective atomic number of tissue: 7.42
Figure 3.3: The left diagram represents the formation of an electron-hole pair leading to the population of electron-hole traps in the material. The right diagram illustrates the two possible modes of recombination when the temperature of the material is raised and the emission of TL photon occurs.

where

\[ < TL > = \frac{1}{N} \sum_{i=1}^{N} (TL_i - TL_{bkg}) \]  \hspace{1cm} (3.2)

\( ECC_i \) is the electron correction coefficient of the \( ith \) TLD. \( < TL > \) is the mean TLD response and \( TL_{bkg} \) is the response due to background radiation taken from several unexposed TLDs, \( TL_i \) is the response due to the \( ith \) TLD and \( N \) is the total number of TLDs in a batch of TLDs.

The dose deposited in an individual TLD is calculated from their response measured using a photomultiplier tube and the correction factors described above as:

\[ Dose_i = \frac{ECC_i \cdot TL_i}{\Gamma} \]  \hspace{1cm} (3.3)

where

\[ \Gamma = \frac{ECC_{i_{sample}} \cdot < TL_{sample}>}{D_{cal}} \]  \hspace{1cm} (3.4)

\( \Gamma \) is the correction factor of a sample TLD exposed in parallel to a known dose of radiation, \( TL_{sample} \) is charge in nC, and \( D_{cal} \) is the known dose given to the sample TLDs.
Chapter 3. Experimental Materials

TLDs are annealed prior to irradiation according to the manufacturer’s recommendations at 400°C for 1 hr followed by 100°C for 2 hrs to ensure electron-free traps. The samples are then read and appropriate calibration factors are incorporated in order to obtain the correct response in units of dose, cGy.

Another advantage of using TLDs is their useful range, going from $10^{-3}$ cGy to $10^3$ cGy. TLDs are dose rate independent and reproducible to within ±3%. Furthermore, they are reusable, when appropriate annealing procedures are performed. However, after substantial use, eventually due to extensive radiation damage the traps will be permanently damaged and the response may be biased.

A drawback of using TLDs is lack of uniformity in their response. Also at high temperatures migration of trapping centres may occur, leading to the instability of the response.

3.2.3 Diodes

Semiconductor detectors used in this work are $p$ type detectors. In applications of radiation dosimetry, the electron-hole pairs generated due to irradiating the diode are measured as charge collected per unit radiation dose (sensitivity of the detector). The sensitivity coefficient of the semiconductor detectors in terms of absorbed dose is [62]:

$$\epsilon = C \sqrt{\zeta \left( \frac{1}{\tau_0} + \eta D \right)^{-1/2}} \quad (3.5)$$

where $C$ and $\eta$ are constants, $\zeta$ is the minority carrier diffusion coefficient, $\tau_0$ is the minority carrier life-time of the un-irradiated diode and D is the absorbed dose.

3.2.4 Films

Films may be regarded as an attractive alternative to TLDs and diodes for measuring dose distribution with high spatial resolution in two dimensions. They have a limited
Figure 3.4: Measured characteristic curve for Kodak XV films. Data were fitted to a fourth-order polynomial and the linear portion of the curve was used to convert optical density to absolute dose in cGy.

Useful range over which the optical density of the film is linearly proportional to the dose. See Figure 3.4.

Radiographic films (Eastman Kodak Co, Rochester, NY.) were used to measure the 2D planar dose distributions in planes parallel and normal to the beam axis. Conversion of optical density-to-absolute dose was made by calibrating the films in a known field. The concentration of exposed silver halide grains on the film was measured using a densitometer (Wellhöfer Dosimetrie, GmbH and Co, Schwarzenbuck, Germany) with a spot size of 0.1 mm. The relation between optical density and intensity is:

\[ OD = \log\left(\frac{I_o}{I}\right) \]  

(3.6)

where \( OD \) is the optical density of the exposed film, \( I_o \) is the intensity in the absence of the film. Sensitivity of the film with depth and field size was measured with a set of
films exposed at various doses (2 – 200 cGy). Un-exposed films were also processed at the same time as calibration films to measure the fog density due to the film emulsion. Density due to fog was subtracted from the density on each of the calibration films. It is important to remeasure the film sensitometric curve for each new batch of films as the chemical composition of films may be slightly different and consequently different dose-density relation may be observed.

A fourth-order polynomial was used to fit data and conversion of optical density-to-dose. In general:

\[
Dose = [c_n(OD)^n + c_{n-1}(OD)^{n-1} + \cdots + c(OD)]
\]

The values of \(c_i\) are determined from the initial data set using polynomial fit to the data. A measured graph of dose-versus-density is shown in Figure 3.4.

With all the advantages of film dosimetry in the megavoltage energy range, the use of film as dosimeter is still limited due to various difficulties associated with films such as energy dependence, film orientation and its nonlinearity with dose beyond certain dose.

The film measurements were benchmarked against other dosimeters measured in a water phantom. Measurements using films and diodes were made with solid water slabs used for buildup and backscatter.
Chapter 4

DYNAMIC AND OMNI WEDGED FIELDS

Adapted from the original article published in: Medical Physics, Vol 27 1623–1634, July 2000.

This chapter describes the implementation of dynamic wedge on an Elekta linac, produced through the use of computer-controlled motion of MLCs and/or backup diaphragms. The notion of mixed fields to produce 2D intensity maps arbitrary in orientation with respect to collimator axis is also studied in development of omni fields. The mathematical formulation and theoretical considerations needed to realize these goals are discussed and results are presented.

4.1 Theoretical Considerations

The use of non-uniform photon beams or a combination of uniform and non-uniform radiation beams (see Figure 1.2) to produce 1D field gradients has been shown to increase the dose uniformity to the target volume and as a result reduces the probability of toxicity to surrounding healthy tissues [16].
The work described in this chapter uses an Elekta SLi linac equipped with a single steep-angle motorized wedge, known as the internal wedge, placed in the head of the linac and moved in and out of the beam by computer control (see Chapter 2). When in the beam, it is designed to produce a wedged field with maximum attainable slope of 60° at 10 cm depth in water. To produce isodose curves with wedge angles smaller than 60°, a weighted combination of open and wedged field irradiations is used to modulate the beam to arbitrary angles between $0 \leq \theta_{\text{eff}} \leq 60°$. The final wedged field has an effective wedge angle $\theta_{\text{eff}}$. This technique is called the universal wedge technique. Shakford et al [63] have presented the effective wedge angle for the universal wedge as:

$$\theta_{\text{eff}} = \tan^{-1}\left[\frac{\omega_w}{\omega_o + \omega_w}\tan \theta_w\right]$$

where $\omega_o$ and $\omega_w$ are the relative weights for the open and wedged fields, respectively. In section 4.1.2 a similar generalization for the dynamic wedge is given to produce fields with arbitrary wedge angles and orientations independent of collimator angle.

### 4.1.1 Dynamic Wedge: 1D Intensity Modulation

By moving the MLCs or the backup diaphragms continuously while the beam is being delivered, non uniform intensity profiles can be produced. Figure 4.1 is a block diagram, illustrating the 1D intensity modulation of the photon fluence produced by continuous motion of a backup diaphragm.

The first goal in this study was to produce dynamic wedged fields with wedge angles identical to those used in conventional treatments. The objective was to provide a dynamic wedge dose distribution that when combined with open fields in different proportions can change the wedge angle at will. In dynamic mode, the relative fluence is determined by the number of MUs delivered to each segment. That is, unlike conventional wedged fields where beam modulation is achieved by differential attenuation of
photons in the material, in dynamic mode the incident photon intensity is changed by controlling the amount of MU each element in the field is exposed to.

Figure 4.1: Schematic illustration of wedged dose distribution created by sweeping the left diaphragm during an uninterrupted exposure. The movement of the diaphragm is defined by a set of control points (Ctl. Pt. 0 to Ctl. Pt. N) describing the position of the diaphragm at any point in the delivery of the total number of MUs.

To produce wedged dose distributions of clinically relevant field sizes, the moving diaphragms must travel past the beam axis. With the current design of the linac in this study, the backup diaphragms and the MLCs can over-travel to only 12.5 cm beyond the central axis of the beam. This will allow a dynamic field modulation that is 25 cm wide in the transverse direction by 40 cm long in the longitudinal direction.
To produce a dynamic wedged field, diaphragms or MLC leaves were moved by monitoring the fraction of the total dose (number of MU) at each control point during exposure. For example, the first segment was given a cumulative percentage MU of 0% and the final segment had a value of 100%. The procedure followed in designing the dynamic wedged fields is outlined below [64]:

- Diaphragms parallel to MLCs (Y-diaphragms), were used in beam modulation. One of the diaphragms was kept stationary while the other moved.

- The motion of a diaphragm was defined with a set of control points, moving by 1.0 cm. The speed of the diaphragm was controlled so that the motion between two control points was linear.

- The initial control point describes the 0 MU delivery, corresponding to the open field, and the diaphragm motion was such that the field delivery ended with the moving diaphragm 1.0 cm away from the adjacent diaphragm to avoid possible collision.

- The portal shape was set by MLC leaves; diaphragms either were set to ±12 cm or to the edge of MLC portal.

- Due to physical constraints (over-travel of the diaphragms past the mid-line of the field), the maximum symmetric field size achieved was $25 \times 40 \text{ cm}^2$.

### 4.1.2 The Omni Wedge: 2D Intensity Modulation

Wedged fields are routinely used in radiation beam deliveries to optimize the treatment outcome. However, in some clinical cases there are situations in which simultaneous optimal MLC conformation of the target volume and desired dose distribution using an
internal wedge may not be achievable. This is mainly due to mechanical limitations of the hardware design where the internal wedge direction is fixed with respect to the orientation of MLC leaves. This is shown in Figure 4.2.

Figure 4.2: Schematic diagram illustrating the direction of the internal wedge, dynamic wedge, and the optimal MLC configuration to the target volume and the desired field orientation, the wedge orientation $\alpha$ and the transformation coordinates, $x'$, $y'$.

A possible solution to this problem is the development of the omni wedge, that is, the combination of mixed fields (orthogonal wedged fields and an open field) in different proportions to produce arbitrary changes to the angle and orientation of the field. The usefulness of this technique and the need for its clinical implementation arise from the fact that the wedge angle and wedge orientations may be changed without the need to rotate the collimators; hence, it is possible to achieve the best conformation to the target while obtaining the desired dose distribution in the field. Figure 4.3 illustrates
Chapter 4. Dynamic and Omni Wedged Fields

Figure 4.3: Schematic diagrams illustrating the principle of the omni wedge technique. Dynamic and internal wedges are orthogonal in direction (right). Coordinate transformations from $xyz$ to $z'y'z'$ are made to determine the orientation, $\alpha$ and the new effective wedge angle, $\theta_{eff}$ (left), respectively.

this principle schematically with appropriate coordinate transformation.

Milliken et al [65] have presented the general expression for the omni wedge angle, $\theta_{eff}$ (the angle the isodose lines make with central axis of the beam at a certain depth in water), and the angle of the final wedged field, $\alpha$ (the orientation of the wedge). The total dose due to the sum of the wedged and open fields is:

$$D_{tot} = \omega_o D_o + \omega_d D_x + \omega_i D_y$$  \hspace{1cm} (4.2)

where $\omega_o D_o$, $\omega_d D_x$, and $\omega_i D_y$ are weighted doses due to open, dynamic and internal wedged fields along the $x$ and $y$ axis of the linac respectively such that:

$$\omega_o + \omega_d + \omega_i = 1$$  \hspace{1cm} (4.3)

In general, the omni wedge is assumed to produce dose gradients in orthogonal directions. Also by definition, the tangent of the wedge angle is equal to the ratio of the gradients along the wedge axis to the gradient in depth. For wedged fields in the $x$-direction we
have:

\[
\tan \theta_x = \frac{\partial D_x}{\partial z} / \frac{\partial D_x}{\partial z}
\]

for \( \frac{\partial D_x}{\partial y} = 0 \)  

(4.4)

and similarly the gradient along the y-direction is:

\[
\tan \theta_y = \frac{\partial D_y}{\partial y} / \frac{\partial D_y}{\partial z}
\]

for \( \frac{\partial D_y}{\partial x} = 0 \)

(4.5)

Because of the \( x \) - and \( y \)-dependence of the wedge angle, when these two wedged fields are combined the new wedge angle varies with both \( x' \) and \( y' \), the transformation coordinates of the final field (see Figure 4.3). The final wedged field if rotated from the \( x \)-axis would have an effective wedge angle \( \theta_{eff} \) such that:

\[
\tan \theta_{eff} = \frac{\partial D_{tot}}{\partial x'} / \frac{\partial D_{tot}}{\partial z'}
\]

for \( \frac{\partial D_{tot}}{\partial y''} = 0 \)

(4.6)

With transformation coordinates:

\[
x' = x \cos \alpha + y \sin \alpha \text{ and } y' = -x \sin \alpha + y \cos \alpha \text{ and } z' = z
\]

(4.7)

Combining Equations 4.2 - 4.6 by direct substitution and appropriate coordinate transformation from Equation 4.7, and noting that in open field, the isodose curves are symmetric about the central axis and therefore, the dose gradient along \( x \) and \( y \) axes is zero \( (\partial D_0/\partial x = \partial D_0/\partial y = 0) \), we obtain:

\[
\tan \theta_{eff} = \frac{\omega_d \cos \alpha \tan \theta_x (\partial D_x/\partial z) + \omega_i \sin \alpha \tan \theta_y (\partial D_y/\partial z)}{\omega_d (\partial D_x/\partial z) + \omega_i (\partial D_y/\partial z) + \omega_o (\partial D_o/\partial z)}
\]

(4.8)

rearranging Equation 4.8 yields:

\[
\tan \theta_{eff} = \frac{\kappa \omega_d \cos \alpha \tan \theta_x + \omega_i \sin \alpha \tan \theta_y}{\kappa \omega_d + \omega_i + \kappa_o \omega_o}
\]

(4.9)

Equation 4.9 is the general expression describing the relation between the omni wedge angle as a function of \( \alpha \). \( \kappa \) is the ratio of depth-dose curves measured at the central axis
of the beam for two orthogonal wedged fields, that is:

\[ \kappa = \frac{\partial D_x}{\partial z} / \frac{\partial D_y}{\partial z} \quad (4.10) \]

Equation 4.10 is a function of field size, wedge angle, and energy. Its value is determined experimentally from the ratio of depth doses [65].

\( \kappa_0 \) is the ratio of the slopes of central axis depth dose curves to the combined wedge fields, defined as:

\[ \kappa_0 = \frac{\partial D_0}{\partial z} / \frac{\partial D_y}{\partial z} = \frac{\partial D_0}{\partial z} / \frac{\partial D_x}{\partial z} \quad (4.11) \]

If modulation is done with one wedge only, that is, \( \omega_i = 0 \), then Equation 4.9 reduces to:

\[ \tan \theta_{eff} = \frac{k \omega_d \cos \alpha \tan \theta_x}{\kappa \omega_d + \kappa_0 \omega_0} \quad (4.12) \]

A limiting case arises when we let \( \alpha = 0 \) and \( \kappa_0 = 1 \). Furthermore, first-order approximations yield \( \kappa \) to be close to unity, that is, if the effects of beam hardening and scatter contributions are ignored, then Equation 4.12 is reduced to:

\[ \tan \theta_{eff} = \frac{\omega_d \tan \theta_x}{\omega_d + \omega_0} \quad (4.13) \]

This is Equation 4.1 with \( \theta_x \) replaced with \( \theta_w \). We shall show that this approximation is in fact sufficient to be used in some clinical cases.

The above equations were used to solve for \( \alpha \) and \( \theta_{eff} \), given the beam weights for the dynamic, internal and open fields, or for the normalized beam weights of the dynamic, internal and open fields, given the desired values of the wedge angle and the wedge orientation.

### 4.2 Experimental Methods

A number of studies were performed to determine the dosimetric characteristics of the dynamic and omni wedges. In some cases, these were compared with mechanical wedges.
The stability and robustness of these wedges were also measured.

Output factors for various fields and depths were measured to study the dependence of the wedged fields on these parameters. To identify any discrepancy in energy spectra between the designed dynamic beam and those produced with the internal wedge, we compared depth doses at the central axis (CAX) of the beam and at off-axis points.

The effect of gravitational forces on the motion of the leaves in dynamic delivery and its influence on dose distribution was also studied by measuring diaphragm motion at various gantry positions.

Two delivery techniques; step-and-shoot and fully dynamic were used to compare with the conventional methods with emphasis on the optimal dose distribution in the volume of interest and efficiency in irradiation time.

In-air fluence measurements were made using an array of 46 p-type silicon diode detectors (see Chapter 3) with a spatial resolution of 0.5 cm and an inherent buildup of 0.84 g/cm² of water equivalent material.

To study the dose distribution as a function of depth in a dynamic field, a scanning ion chamber array system (24 ion-chambers with 1.0 cm spacing between each detector) installed in a water tank (Wellhöfer Dosimetrie, BmbH and Co, Schwarzenbuck, Germany) was used to measure the incident radiation intensity at various depths and field sizes in water. The ionization was sampled in each detector in 5 mm increments across the field and 1.00 ± 0.25 cm in depth.

Films sandwiched tightly between solid-water phantoms were also used in the dosimetry of dynamic and omni fields. A conversion table from optical density to absolute dose was generated by measuring the sensitometric curves of the films for a given energy and film orientation (see Chapter 2). In order to be in the linear region of characteristic curves and not saturate the films, only 1/2 of the total MU was delivered to the films. Optical density of the films were then adjusted for data analysis.
Chapter 4. Dynamic and Omni Wedged Fields

4.2.1 Clinical Applications

A clinically relevant field with concavities, suitable for applications of dynamic beam deliveries was constructed. Both dynamic and static intensity modulated beams were used to produce isodose distributions with arbitrary angles and with arbitrary orientations in field gradient. The goal was to achieve the optimal leaf settings (optimal conformation to the target volume) while delivering a wedged dose distribution to the region of interest at any desired orientation, as occurs in clinical set-ups. For this particular application, the optimum MLC configuration was achieved such that the collimator rotation required a $-60^\circ$ rotation. To assess the principal methodology of the omni wedge, we imposed a constraint such that a $40^\circ$ wedged dose distribution be delivered in the lateral direction, thereby ensuring that the implementation of a single internal wedge or a single dynamic wedge by itself would not be a sufficient solution for this case. An omni wedge with three beam arrangements, one open, one with internal, and one generated in the dynamic mode with appropriate weightings was used to create the required dose distribution given the MLC settings.
4.3 Results and Discussion

Table 4.1 is the tabulated wedge factors in solid water for various field sizes and two different delivery modes (dynamic wedge (DWF), and internal wedge (IWF)) for photons with nominal energy of 6 MV.

Table 4.1: Wedge factors for 6 MV photons measured as a function of field size at depth of 10 cm in water with source to detector distance of 100 cm.

<table>
<thead>
<tr>
<th>Field size (cm(^2))</th>
<th>Dynamic wedge factor (DWF)</th>
<th>Internal wedge factor (IWF)</th>
</tr>
</thead>
<tbody>
<tr>
<td>5</td>
<td>0.373</td>
<td>0.271</td>
</tr>
<tr>
<td>10</td>
<td>0.387</td>
<td>0.275</td>
</tr>
<tr>
<td>15</td>
<td>0.398</td>
<td>0.278</td>
</tr>
<tr>
<td>20</td>
<td>0.412</td>
<td>0.282</td>
</tr>
<tr>
<td>25</td>
<td>0.425</td>
<td>0.285</td>
</tr>
</tbody>
</table>

Figure 4.4 shows the dependence of wedge factors for both the dynamic and the internal wedged fields, measured as a function of depth (Fig. 4.4a) and field size (Fig. 4.4b).

Table 4.2: Measured transmission factors (TF) for Elekta SL20 linac MLCs and backup diaphragms for two nominal photon energies. Transmission factors are the ratio of closed to open fields measured in air.

<table>
<thead>
<tr>
<th>Energy (MV)</th>
<th>X-Jaws</th>
<th>Y-Jaws</th>
<th>MLCs</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>0.004</td>
<td>0.104</td>
<td>0.018</td>
</tr>
<tr>
<td>18</td>
<td>0.005</td>
<td>0.119</td>
<td>0.018</td>
</tr>
</tbody>
</table>

Field sizes ranged from 5 to 25 cm\(^2\) and depths were varied from 1.5 to 20 cm. The difference in wedge factors between those produced with an internal wedge and in dynamic mode may be due to the high transmission in dynamic wedge delivery and
Figure 4.4: Wedge factors measured for dynamic and internal wedged field techniques for 6 MV photons. a) is the measured data as a function of depth for a $25 \times 38 \text{ cm}^2$ field size, and b) is the graph of wedge factors as a function of field size measured at 10 cm depth and a source to detector distance of 100 cm. Diaphragm velocity was 1.0 cm/sec for dynamic wedge delivery.

high absorption in the internal wedge. The Y diaphragm (thickness: 3.0 cm of lead) has a nominal transmission (percentage of radiation transmitted through diaphragms) of approximately 10% for 6 MV photons, and its primary design purpose is to be used as backup to the MLCs to prevent leakage through the leaves. The measured transmission factors through X-, Y-Diaphragms, and the MLCs are shown in Table 4.2.

Figure 4.5 shows the profiles measured for 6 MV photons at a depth of 10 cm in solid-water for various fields. Figure 4.5a was produced using the dynamic motion of the diaphragms and normalized to their corresponding open field at central axis. The variation in wedged fields in amplitude of the dose is due to contribution from scatter radiation as it varies as a function of diaphragm settings. Figure 4.5b is the comparison of dynamic beam profiles with those measured using an internal wedge. To compare the
Figure 4.5: a) Dynamic beam profiles measured at 10 cm depth in water normalized to their corresponding open fields. Total dose given to each field was 200 MU with a dose rate of 400 cGy/min. b) Internal (dashed lines) and dynamic dose profiles (solid lines) measured at 10 cm depth normalized to 1. b) is the comparison of the shape of dynamic versus internal wedge profiles.

shape of profiles, both dynamic and internal wedge profiles were normalized to 1.0 at central axis of the beam.

To facilitate the use of dynamic beam delivery, both diaphragms, Y1 and Y2, (see Fig. 3.1) were used to produce 1D wedged fields. If beam modulation is identical with either diaphragm, this will reduce the need for having to rotate the collimator for beam modulations reflected in the plane of the wedged field. The results are shown in Figure 4.6a which illustrates the excellent agreements for both Y1 and Y2 diaphragms.

The effect of gravitational forces on the dynamic motion of the diaphragms and MLCs and its influence on final dose distribution was studied by moving the diaphragms at various gantry angles. Figure 4.6b is a plot of dynamic profiles obtained by sweeping the Y2 diaphragm across the field for gantry angles of 0, 90, and 270 degrees. The results
show the robustness of the system to compensate the effect of gravity on the motion of the diaphragms.

Depth dose curves were measured with a single-detector scanning system and compared to those measured with internal wedge. Figure 4.7 shows percent depth dose (%DD) curves for $10 \times 10 \text{ cm}^2$ field (Fig. 4.7a) and for $25 \times 25 \text{ cm}^2$ field (Fig. 4.7b), measured on CAX of the beam. Measurements for $25 \times 25 \text{ cm}^2$ field at 7.0 cm off the central axis of the beam (Fig. 4.7c) is also shown. The profiles show the dependence of the dose as a function of depth, both on central axis and off axis cases. The agreement for all measurements was excellent for both internal wedge and the dynamic delivery. Data points are shown with horizontal error bars to indicate the positional uncertainty of the detector array in each scan.
Figure 4.6: a) Y1-Y2 motion of the diaphragms showing the beam modulation with either diaphragm. The Y2 profile is reflected about the central axis of the beam to provide better comparison of the two beams. b) Shows the comparison of beam profiles measured at various gantry angles to determine the robustness of the system for different clinical gantry angles.
Figure 4.7: a) Percent depth dose for $10 \times 10 \text{ cm}^2$ field measured along the beam axis. b) The corresponding percent depth dose measurements for $25 \times 25 \text{ cm}^2$ on the central axis. c) Percent depth dose measured in the heel region of the wedge, 7.0 cm off the central axis for $25 \times 25 \text{ cm}^2$ field. Horizontal error bars ($\pm 1 \text{ mm}$) show the measurement uncertainty in spatial positioning of the detectors.
Figure 4.8: Comparison of dynamic (solid lines) and internal wedge (dashed lines) isodose lines for a 10 x 10, 15 x 15, 20 x 20, and 10 x 20 cm² half-beam-blocked fields. Depth in water is plotted along the ordinate and distance from central axis along the abscissa. The measurements were made with source to surface distance of 100 cm.

Wedged fields as a function of depth and lateral distance are shown in Figure 4.8 for 10, 15, and 20 cm² field sizes, and a 10 x 20 cm² half-beam-blocked field. Measurements were carried out using a scanning water tank system with 24 linear array chambers. Data were normalized to the depth of maximum dose. Little difference was seen between the dose distribution produced by the dynamic and internal wedges.

The omni field measurements shown in Figure 4.9 as a function of depth was used to compare the measured dose distributions obtained from the universal wedge technique and the omni wedge technique. An omni wedge with 45° orientation from the linac's
principal axis and a gradient in field intensity such that the effective angle at a given depth would be 30° was produced. Equal weightings were assigned to each segment ($\omega_0(0.33)$, $\omega_d(0.33)$, and $\omega_i(0.33)$). Measured wedge factors from Table 4.1 were used to determine the relative MU for each segment (open 33 MU, dynamic 87 MU, and internal 120 MU). Figure 4.9a show the isodose plots measured with films exposed to 50% of the total dose for a $10 \times 10$ cm² field size placed normal to the beam axis. Figure 4.9b is the corresponding field measured with film parallel to the beam axis. The wedge orientation and wedge angle were measured from the tangent of the isodose line through the central axis of the beam. The measured data were within ±2° of the expected values. This example demonstrates the accuracy of the technique for obtaining a desired wedge angle and wedge orientation.
Figure 4.10: Comparison of isodose plots (depth vs distance) for an omni wedge and a universal wedge. The dashed lines represent the universal wedge isodose contours. The solid lines represent the omni wedge isodose lines obtained using the omni wedge equations. The measured wedge angles were within ±1° for both methods. The field orientation for the omni field was rotated to measure 45° from the axis of the linac. The deviation was ±1°

The universal wedge concept described in Section 4.1 was used to produce wedged isodose distributions for a 10 × 10 cm² field measured at 100 cm from the source. Measurements were carried out using films positioned parallel to the beam axis. The universal wedge used a combination of internal and open fields with appropriate weightings to produce isodose lines with 30° angle. The fraction of the dose for the open field was 52 MU and 174 MU for the internal wedged field. Figure 4.10 demonstrates the agreement of the two methods and emphasizes that the omni wedge approach can be used to produce wedged dose distributions made by the internal wedge. The measured and expected
Figure 4.11: Comparison of a 45° external wedge with internal, and dynamic wedged fields produced using the universal wedge concept. Dose profiles measured using a linear array of diodes. Internal and dynamic dose profiles were scaled to the dose on the central axis of the external wedge profile, and the external wedge profile itself was normalized to 1.0 along the beam axis.

values of the wedge angles were within ±1° and that of the wedge orientation was ±1°.

Figure 4.11 presents a comparison of wedge profiles for three different fields with 45° wedges. A conventional external wedge, the combination of internal and open field (universal internal wedge), and dynamic wedged fields (universal dynamic wedge) were used to compare dose distribution along the lateral axis of the linac. The external wedge was designed so that it yields a wedge angle of 45° at a depth of 10 cm. The universal wedge equation was used to produce a field with wedge angle similar to that of the external mechanical wedge. The fraction of the dose for each case was determined by normalizing the weight of each segment to the wedge factors for a 15 × 15 field at 10 cm
depth such that the dose delivered by each of the wedge techniques would be equivalent on the central axis. The calculated MU for the open and wedged fields of the universal wedge were 30 and 252 MU, respectively. Figure 4.11 shows the dose profiles measured using a linear array of diodes. Internal and dynamic dose profiles were scaled to the dose on the central axis of the external wedge profile, and the external wedge profile itself was normalized to 1.0.

The difference in universal wedged profiles from the external wedged field is of no intrinsic consequence. This is because the mechanical external wedge was not designed to match universal wedge profiles. However, the good agreement at the centre of the beam shows the adequacy of the approximations made with the universal wedge technique to
yield clinically acceptable results.

Figure 4.12 is a plot of dose profiles generated in a dynamic field and step-and-shoot methods. The results demonstrate the robustness of both techniques in producing identical dose profiles. The plots are taken from densitometer scans of films.

Finally, Figure 4.13 shows the result for the clinical example presented in this chapter. Final field orientation was $-60 \pm 2^\circ$ from the axis of the linac and the measured wedge angle shown as the plot of isodose curves versus lateral depth was designed to be $40^\circ$. The measured value was $40 \pm 2^\circ$. The relative weights for each segment were calculated to be 0.04, 0.35, and 0.61 for the open, dynamic, and internal wedge segments, respectively. The measured orientation angle was $-60 \pm 2^\circ$. The clinical implication of these results is better conformation to the target volume with optimum dose distribution.
Figure 4.13: Plot of the isodose lines of a clinical example of an omni wedge obtained from exposure of a film positioned normal to the beam at 10 cm below solid water phantom (top) and parallel to the beam axis (bottom) sandwiched between water phantom material. The measured field orientation was $-60 \pm 2^\circ$ and wedge angle was measured at $40 \pm 2^\circ$. 
4.4 Summary

The objective of the study described in this chapter was to provide a better understanding of limitations and capabilities in the delivery and dosimetry of dynamic beams using both the backup diaphragms and MLCs.

In general, a dynamic wedge has the versatility to be implemented in many ways, ranging from a fixed wedge angle to a number of wedge angles by varying the percent MU between each control point. In our implementation, a fixed wedge angle was produced (and hence a fixed set of MU settings per diaphragm positions); it was shown that shallower wedge angles can be produced by the universal wedge concept. An important advantage of the dynamic wedge technique is that it requires fewer photons to deliver the same dose distribution than in conventional techniques. In this way, the implementation in a planning system is the same as for the fixed wedge. The omni field becomes useful when the optimal MLC orientation differs from the wedge direction. As was illustrated, this technique is capable of creating fields independent of collimator orientation, giving the freedom to use MLCs to obtain optimal target conformation without compromising the dose constraints to the healthy tissues. This was achieved by sequential irradiations of dynamic, universal, and open fields.

Based partly on the findings and results of our research study, a proposal has been put forth to the linear accelerator manufacturer to implement dynamic delivery on this type of linac and will be included in the next release of their application software.
In this chapter experiments to study the precision and degree of accuracy with which dynamic IMB can deliver highly localized dose distributions to tumors near critical structures are presented. This was achieved by dynamically driven MLCs (sliding window technique). Measurements of dose distribution were performed \textit{in vivo} and \textit{in-vitro} using TLDs, films, and diodes. Results and discussion are presented.

5.1 General Considerations

IMRT has the ability to arbitrarily modulate the weights of individual rays (intensity distribution) of a radiation beam to conform to the three dimensional shape of a target volume. However, owing to the dynamic nature of the beam delivery and the novel to sophisticated algorithms used to generate the beams, the dosimetry of dynamic MLC beams is extremely crucial. Therefore, transition from conventional delivery to a full
3D-dynamic intensity modulated beam (IMB) delivery is not trivial, and it requires the introduction of procedures for dosimetry different than those used in conventional techniques [4, 49].

There are a number of reports in the literature describing the importance of dose verification in IMRT and the significance of different approaches required for each technique [47, 52, 66, 67, 68, 69, 70, 71, 72].

5.1.1 Calculation of The Dose Matrix

The dose computation engine uses a pencil beam superposition approach, the Macro Pencil Beam (MPB) model [73]. In summary, the MPB divides the radiation field into a finite number of beamlets, each with equal cross sections (1 × 1 cm² at isocentre plane). The model considers each 1 × 1 cm² bixel in the phantom to be the source of Compton scattered radiation which is transported to the dose point. Head scatter (head scatter is the contribution of scattered radiation originating in the head of the linac), using a dual source model, is also calculated as arising from bixels matched to the width of MLC leaves. In short, a dual source model utilizes the flattening filter as the primary source of Compton scattered fluence coming from the head and the target as the source of primary fluence. In both cases, the intensity and distribution of the scatter sources are obtained by an iterative fit to measured data.

5.1.2 Fluence Optimization

A seven equi-angular approach in designing the beam configuration was adopted for this study. The weight of each pencil-beam calculated by the MPB model is optimized by means of an inverse planning algorithm based on projections onto convex sets (POCS) [45, 74]. The method is different from other iterative optimization techniques in that it
does not utilize an objective function in its minimization process but rather assumes that the constraints are expressed as convex sets and that the intersection of these sets presents a solution that satisfies the imposed constraints. The constraints are usually target dose uniformity, upper limit on maximum and integral dose to the organs at risk, and permitting only fluences with positive or zero intensity. The relation between the optimized fluences and the prescribed dose is described by Cho et al [46]. The optimized profiles for each beam were input into a fluence-to-leaf trajectory translator to sequence the beam shapes defined by the MLC [75]. Both MPB and POCS algorithms have been integrated into the in-house treatment planning software, Prism [76, 77]. The pencil beam computation of dose array and the optimization of fluences were carried out on an Intel computer (700 MHz Pentium III processor) with 512 MB of memory running under Linux operating system. Calculation times were dependent on the resolution of the dose sampling grid.

5.2 Experimental System

The approach to testing the accuracy of the designed IMRT beams is to use live animals as subjects and to treat a target adjacent to the spinal cord in the head and neck region. In addition to dosimetric measurements, the patterns of radiation damage to spinal cord and soft tissues in control subjects receiving conventional conformal therapy were compared with those receiving IMB delivery.

Steps necessary in planning, dose computation, fluence optimization, MU verification, delivery, and dosimetry of the dynamic beams carried out in this study are illustrated in Figure 5.1.

An IMRT option was installed on the SL/MLCi system of an Elekta linac for research purposes and is currently under assessment by an international consortium of research
The method by which IMBs were produced in this chapter involves the continuous motion of leaves and backup diaphragms. Chapter 2 gives a more general overview of the realization of dynamic beam deliveries using MLCs. For the Elekta linacs, the position of the leaves is controlled in real time from a video signal, fed into a CCD camera via an optical system placed in the head of the linac as a function of delivered MU. Values of these variables at given points in the delivery were defined as “control points”. Leaf velocities were constant between each control point.

In this study, for dosimetry of the IMRT plans, two types of phantoms were used:
(a) **Phantom 1:** a semi-cylindrical multi-slice phantom with dimensions shown in Figure 5.3a, designed for individual beam dosimetry and MU calibration;

(b) **Phantom 2:** a cylindrical multi-slice phantom (see Figure 5.3b) designed to accommodate various dosimeters was used for verification of composite plan delivery.

Physical measurements of the dose distributions using TLDs, films, ion chamber and diodes were made in phantoms, in deceased dogs, and in subject canines. Cylindrical rods of TLDs (lithium fluoride, LiF, with dimensions of $1 \times 1 \times 2$ mm) were used to measure the dose at selected points in subject canines and in phantom. To reduce statistical fluctuations in data analysis, each TLD was calibrated and assigned a sensitivity factor, $\Xi$, determined in a known field. Calibration was performed by irradiating a batch of TLDs to a known dose, and the response of the irradiated dosimeters was read using a TLD-reader (Harshaw-Bicron, Solon, OH). The dose imparted to each TLD was then calculated as follows:

\[ D_{TLD} = \frac{\Xi_{sample} TL}{\Gamma} \]  

(5.1)

where

- $\Xi_{sample}$ is the sensitivity or the correction coefficient of a TL element.
- $\Gamma$ is the calibration factor in units of nC/cGy.
- and $TL$ is the signal in nC for a given TL element in the field.

To obtain 2D intensity maps with higher spatial resolution and better dose distribution information, films were used in planes perpendicular and parallel to the beam axis. The dosimetry of films in IMRT fields, and its advantages and limitations have been described in the literature [52, 69, 78, 79] and briefly in Chapter 3. The relationship between the optical density of the film and the incident radiation beam intensity was
Determined by studying the sensitometric curves for a given batch of film and fitting the data to an \textit{nth} order polynomial. That is,

\begin{equation}
D_{\text{film}} = \sum_{i=1}^{n} c_i (OD)^i
\end{equation}

and

- \( c_i \) is the \textit{ith} coefficient of the \textit{nth} order polynomial.
- \( OD \) is the measured optical density of the film at a given depth in phantom and for a given dose.

The optical density of the irradiated films was measured using both the Wellhöfer densitometer with an aperture size of 0.1 mm and an accuracy of ±0.01 and range of 0 – 3.2 density units, and a Lumisys 50 laser film scanner (Lumisys Inc, Sunnyvale, CA).

A linear array of 46 P-type silicon diodes (Profiler, Sun Nuclear Inc., Melbourne, FL), each with an effective area of \( 1.65 \times 1.65 \) mm, and with 5 mm separation between diodes was used to measure the dose profiles along different tracks of MLC leaves. The total inherent buildup to the detector junction was 0.84 g/cm\(^2\) with 2.8 g/cm\(^2\) of Acrylic as backscatter material.

Measurements of all the detectors were first benchmarked against known dose with a calibrated ionization chamber and a solid water phantom.

### 5.2.1 Fluence to Monitor-Unit Conversion

IMRT treatment plans were developed for volumes designed to simulate a paraspinal tumor. A series of experiments were conducted on each plan before it was applied to live or cadaver dogs. Measurements consisted of determining MU-to-dose accuracy and dose distributions for individual beams and the composite plans. The experiments are described below.
Individual Beam Dosimetry in Phantom 1

Two independent methods were used to obtain the MUs required to deliver the prescribed dose for each beam of each plan. The need for this approach stemmed from (a) the incomplete knowledge of delivery in the conversion of fluence-to-dose in the optimization algorithm, (b) the translation of fluence-to-leaf and diaphragm motions, and (c) the need to evaluate the applicability of the MPB model in dynamic IMRT.

The first method was to use a measurement in phantom 1 (Fig. 5.3a) to determine the dose for an estimate of the number of MUs and then scale the MUs based on the dose measurement. To convert the inverse plan output, expressed in relative fluence units to machine deliverable MUs, Equation 5.3 was used as a first-order approximation to calculate the dose due to contribution from a primary ray intersecting the normalization point (see Figure 5.2) in the field. That is:

\[
\mathcal{F} = \frac{TD}{\left( \sum_{i} C \cdot TPR_i \cdot ROF_i \cdot \omega_i \cdot ISF_i \right)}
\]  

where,

- \(\mathcal{F}\) is the fluence-to-MU conversion factor. It was used to rescale the fluence profiles from each beam to dose profiles in units of MU.

- TD is the calculated composite tumor dose in cGy to the normalization point due to all beams.

- \(C\) is the machine calibration, equal to 1.0 cGy/MU.

- TPR is the tissue-phantom-ratio for the smallest measurable field size with ion-chamber and at the depth of the normalization point.
• ROF is the relative output factor. A value from a $1 \times 5$ cm$^2$ field was used based on the average field size for all control points.

• $\omega_i$ is the weight of the $i$th beam element of the primary ray for a given beam.

• ISF is the inverse square correction.

• $N$ is the total number of beams.

The phantom was designed so that the gantry remained upright and the depth to the detectors along the delivered ray was set approximately equal to the depth of the normalization point in the patient anatomy for each beam in the plan.

Figure 5.2 shows the intersection of the primary rays with the normalization point along an MLC track that was used to measure the dose distribution in the phantom geometry which was used to simulate the depth of measurements to the calculation point with respect to the beam angle and primary rays intersecting the point.

In order to reduce statistical fluctuations in measurement uncertainties, several points were compared along different tracks from which a correction factor was obtained to correct the dose profiles obtained. The average of the scaling factors for the dose to the norm points along different tracks was used to obtain the final MU for each beam:

$$\Lambda_{\text{avg}} = \frac{1}{N} \sum_{i}^{N} \left[ \frac{D_{m}}{D_{e}} \right]_{i}$$  \hspace{1cm} (5.4)

where $\Lambda_{\text{avg}}$ is the average correction factor, $D_{m}$ is the measured dose on a given track, $D_{e}$ is the expected dose and $N$ is the number of tracks.

The second method was to use the dose calculation model (MPB) to calculate MUs. The translated beams were imported into the same phantom geometry as above (Fig. 5.3a) in which measurements were made. If good agreement between measurements and calculations was obtained the calculations were repeated in the subject anatomy to determine
Figure 5.2: Schematic illustration of the method introduced for verification of calculated MU of the IMBs. Parameters, $s$, $d$, $d'$, $\alpha$, $\beta$, and $l$ were found from the geometry of the primary ray intersecting the normalization points which were then used in calculating the dose to the norm point. The dose to multiple points along an MLC track was measured with a linear array of diodes.

the appropriate MU for patient geometry. However, if there were discrepancies, MUs for subject irradiation were calculated from the phantom measurements described as above. This methodology was adapted to validate the MPB calculations as well as to gain confidence in the quality assurance (QA) and dosimetry of IM beams.

**Verification of Plan Delivery in Phantom 2**

To verify the 3D doses resulting from the output of the inverse planning process, a cylindrical QA phantom appropriate for the dosimetry of IMBs was fabricated (see Fig. 5.3b). The advantage of these designs lie in their versatility to position various detectors in them and change phantom geometry as needed. CT images of the IMRT phantom (phantom
Figure 5.3: Schematic diagrams of IMRT QA phantoms for MU rescaling (a), phantom 1, and dosimetry verification of patient beams (b), phantom 2.

2) with 2 mm axial resolution were taken and transferred to the treatment planning computer where the target and critical structures of the patient were imported onto the phantom and used for in-phantom dosimetry verification. The cross-sectional view of the phantom is shown in Figure 5.8. The bright spots indicate metallic spacers in the catheter used as position indicators of the 2 mm long TLD's located between the spacers. This method allowed the position of each TLD to be located to within ±1 mm.

The MPB algorithm was used to calculate point doses and planar dose distributions in regions of interest. TLD's were inserted into the phantom for absolute dose measurements. 2D dose measurements were carried out using films. The optical density of the exposed films were converted to 2D dose maps.

2D dose measurements were also carried out in phantom using films. Measurements were made of dose distributions from individual beams, as well as of the composite dose from all seven beams. The measured and calculated dose distributions were then compared for any discrepancies between the actual delivery and simulations.
5.2.2 Delivery Verification

To further investigate the true dose distribution in subject canines due to inhomogeneities in the path of the beam, we performed dose measurements in deceased dogs with catheters placed in the spine and in the tissues simulating the target volume. TLDs were irradiated by either single beams or due to composite beams from a given plan.

The correlation of the measured and the planned dose distribution in subjects was accomplished using the image fusion technique [80]. The procedure was as follows: since planning, dose calculations, optimization and MU calibrations were done based on the treatment planning CT of the subject canines, fusion of treatment day CTs (taken immediately prior to irradiation) was performed to locate the position of the field TLDs in planning coordinates of the CT images, thus allowing the calculated planned beams to be correlated with the measured doses from TLDs.

Furthermore, using image correlation, a quantitative study was done to assess the daily variation in setup accuracy of the patients during the course of treatment by means of correlating anatomical structures and identifying any variation in positioning of the TLDs.

5.2.3 In vivo Measurements

A population of 15 beagle dogs was selected for this study from which 9 were chosen to be treated in IMRT mode and 6 were used as control subjects, treated with conventional conformal techniques [81]. The target in each dog was a paraspinal volume with a concavity encompassing the spinal cord (see Fig. 5.9). The irradiation and handling of the canines was made possible by collaborations with the Department of Clinical Veterinary Sciences at Washington State University (WSU). Both institutions shared common equipment for the delivery of IMRT beams.
The treatment plan for each dog was designed so that the same target dose was given an equal number of fractions to each dog regardless of whether it was IMRT or conventional delivery. The prescribed dose to the target volume was 4.0 Gy per fraction for a total of 21 fractions. In the optimization procedure, a constraint of 2 Gy/day to the cord was enforced. All IMRT subjects received the same optimization scheme, as well as similar expansion of margins in the definition of planning target volume (PTV), number of beams (7), equiangular gantry angles (53, 106, 159, 212, 265, 318), and beam energy (6 MV). Control subjects were irradiated with 4 isocentric beams, one dorsal, one
ventral and two parallel opposed lateral fields with MLC shaping of the port to conform to the target volume. The field size in each case was approximately $6 \times 8 \text{ cm}^2$.

In preparation for treatment, each dog was fitted with an alpha cradle immobilization device, indexed to fit reproducibly onto both the CT bed and the linac treatment couch (see Fig. 5.4). In addition, a mold of the upper dental arcade was made to attach to a stereotactic head immobilizer with fiducial markers for the purpose of localization and more reproducible positioning [82, 81]. The transfer of subject canines from the CT bed to the treatment bed was made with minimal setup errors. Port films were taken during the course of treatment to determine the positioning accuracy prior to the actual treatment. In IMRT mode, beams were delivered at 100 MU/min; in control subjects 400 MU/min was used. Dogs were anesthetized in all procedures.

Prior to the delivery of the beams, for both control subjects and those irradiated in IMRT mode, each dog was fitted with a catheter filled with cylindrical TLDs into the air passage and an \textit{in vivo} dosimetry was performed to determine the dose to the esophagus.

A one year post-irradiation was allowed to study patterns of radiation damage and pathological responses to radiation in both group of dogs.
5.3 Results

5.3.1 MU Calibration

Figures 5.5 and 5.6 present a set of absolute dose profiles after MUs were rescaled. The profiles are along a specific MLC track and are compared with MPB calculations and the desired fluence. Fluence profiles from POCS were normalized to the measured dose at the normalization point at isocentre for each beam and they are only plotted for the purpose of comparison of the shape of the desired and measured profiles. Discrepancies seen between POCS and those from MPB and measurements are due to several factors including the lack of phantom scatter, penumbra effects and leaf transmission in the POCS algorithm. Consequently, the POCS output is the relative in-air fluences and not absolute dose. The vertical error bars on the measured points show the inherent sensitivity in TLD’s. The maximum deviations from in-phantom measurements and those calculated with MPB was as low 2% for beams with normalization points at shallow depths. However, the response of the diodes was found to be slightly dependent on energy spectrum with depth. Sensitivity-response curves were generated to account for the detector response with depth.

Table 5.1 shows the results for calculated MUs using Equation 5.3 and measured doses with rescaling coefficients for each beam. As shown, the final delivered doses are in good agreement with expected doses on central axis of the beams.

As a quality assurance measure, prior to the electronic-transfer and delivery of the beams, a virtual delivery was carried out for each IMB file on a computer program to visually inspect the position of leaves at each control point. Figure 5.7 is a snapshot of the simulation output, showing the position of the leaves, the expected intensity map, the fluence profile along a given track, and the position of two opposing leaves as a function
Figure 5.5: Dose profiles for various beams and depths. Measurements were made in solid-water with a linear array of diodes. The comparison between the desired fluence profiles (dotted lines), calculated (solid lines) and measured (points with error bars) data are shown here. Data presented are for track 20, 0.5 cm superior to isocentre. Measured depths for beams 1 through 7 are: 2.0, 3.0, 5.0, 7.0, 7.0, 3.5, and 2.5 cm, respectively.
Figure 5.6: Dose profiles for various beams and depths cont’d. Measurements were made in solid-water with a linear array of diodes. The comparison between the desired fluence profiles (dotted lines), calculated (solid lines) and measured (points with error bars) data are shown. Data presented are for track 20, 0.5 cm superior to isocentre. Measured depths for beams 1 through 7 are; 2.0, 3.0, 5.0, 7.0, 7.0, 3.5, and 2.5 cm, respectively.
Chapter 5. Dosimetry of Dynamic IMB

Table 5.1: Calculated and measured doses, rescaled to deliver 400 cGy to the normalization points in the target volume. Measurement was performed for 6 MV photons.

<table>
<thead>
<tr>
<th>Field #</th>
<th>Calculated MUs</th>
<th>Expected Dose [cGy]</th>
<th>Rescaling factor</th>
<th>Measured Dose [cGy]</th>
<th>%Δ</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>205</td>
<td>44.4</td>
<td>0.961</td>
<td>44.0</td>
<td>-0.9</td>
</tr>
<tr>
<td>2</td>
<td>188</td>
<td>69.6</td>
<td>0.818</td>
<td>71.0</td>
<td>+2.0</td>
</tr>
<tr>
<td>3</td>
<td>223</td>
<td>64.3</td>
<td>0.802</td>
<td>64.3</td>
<td>0.0</td>
</tr>
<tr>
<td>4</td>
<td>237</td>
<td>20.7</td>
<td>0.645</td>
<td>20.7</td>
<td>0.0</td>
</tr>
<tr>
<td>5</td>
<td>236</td>
<td>64.8</td>
<td>0.853</td>
<td>65.9</td>
<td>+1.7</td>
</tr>
<tr>
<td>6</td>
<td>178</td>
<td>67.2</td>
<td>0.796</td>
<td>67.4</td>
<td>+0.3</td>
</tr>
<tr>
<td>7</td>
<td>199</td>
<td>69.0</td>
<td>0.872</td>
<td>68.8</td>
<td>-0.3</td>
</tr>
</tbody>
</table>

of time across the field.

5.3.2 In vitro Results

Dose distributions in planes parallel and normal to MLC-leaf motion were measured in 1 and 2D with subject canine beams first transferred onto a phantom geometry for verification and assessment of measured doses versus the calculated doses. A CT image of the IMRT phantom with the the target cross-section and spinal cord used for in-phantom dosimetry verification is shown in Figure 5.8. TLDs were positioned in the phantom with beam parameters (beam isocentre, and couch indices) adjusted to correlate the position of the TLDs in phantom with the intended anatomical structures in the subject canine coordinates.

Figure 5.10a shows the in-phantom results, acquired using TLDs in the transverse plane of the beam. The measured composite doses from all seven beams are due to beam modulation from a given MLC track, 0.5 cm superior to the isocentre, parallel to the motion of the MLCs. There are small discrepancies of two TLDs in the superior end of
Figure 5.7: A computer-generated output of leaf position (top-left), intensity map (top-right), fluence profile (bottom-left) along a given track in the field, and leaf-pair motion (bottom-right) for two opposing leaves for a given beam angle of one clinical case studied.

The in-phantom measurements in Fig. 5.10a when compared with calculations. This was due to damaged TLDs. Owing to the spatial resolution of locating the TLD coordinates on the CT images (resolution of typically 2.0 mm) and in the image fusion process, the horizontal error bars are shown to indicate the expected positional uncertainties in locating the TLDs. The vertical error bars are the inherent sensitivities associated in TLD readings. The variation in dose correlation is within the limits of our protocol which is ±3%. Figures 5.13a and 5.14a are 1D correlations between measurements and dose computations in the phantom geometry in longitudinal directions, normal to the motion of the leaves. Figure 5.13a represents the dose to a set of points centrally located
in the target volume, while Figure 5.14a shows the results for the dose received to points in the critical structure.

Calculated and measured composite isodose curves are also presented for track 20, 0.5 cm superior to isocentre line in Figure 5.11. The measurement was carried out with films placed in the phantom. To avoid saturation of the film, only 1/4 of the total dose was delivered. Figure 5.12 shows that fluence modulation scales linearly with MU.
Figure 5.9: Lateral view of subject canine CT with calculated isodose lines overlaid on the image. The position of TLDs both in longitudinal and transverse directions are also shown.

5.3.3 In Subject Results

An IMRT plan similar to those used for the trial canine subjects was delivered to a cadaver and the results are presented here. The correlation of the measured and the planned dose distribution was accomplished using the image fusion technique. The pre- and post-treatment CT images of the subjects were resampled in all 3D axes to locate the position of the field TLDs in planning coordinates of the CT images. Figure 5.9 is the fused planning and treatment CT image of the subject canine showing a lateral view of the target cross section, critical structures, and location of the catheters with TLDs both in longitudinal and transverse directions. The image is overlaid with the
Figure 5.10: Measured (points with error bars) and the calculated (solid lines) composite dose distributions. The measured data with TLDs both in phantom (a) and in subject canine (b) were due to beam modulation produced along MLC track 20, 0.5 cm superior to isocentre in the transverse plane.

calculated isodose curves, showing the dose in the volumes of interest and fall-offs near the boundaries of the target and other structures.

In vivo dose measurements in the 9 IMRT subjects showed doses to TLDs, placed in the airways close to the target volume of less than 200 cGy which is less than 50% of the target dose. This was in agreement with the constraints imposed by the optimization of the beams.

The preliminary results of the irradiation of the 15 canines indicate that the treatment goal of reduced spinal cord toxicity was met. The 6 control subjects irradiated with conventional treatment techniques developed severe radiation induced neurologic deficits. The neurologic deficits included partial or complete loss of sensory and motor function to front and hind legs, and neck pain. Due to the severity of the neurologic deficits, the animals were euthanized within 2.5 months post-irradiation. The 9 canines irradiated with
Figure 5.11: Comparison of measured (dashed lines) and calculated (solid lines) isodose curves along a plane 0.5 cm superior to beam axis.

IMRT were euthanized one year post-irradiation with no evidence of neurologic disease. No adverse early radiation effects were noted in these animals. The musculature in the target area in these 9 dogs showed evidence of radiation induced myositis (inflammation of muscles). The myositis was expected due to the large radiation dose to the target area. A more detailed description of the biological results is presented elsewhere [81].

Figure 5.10b shows the results for the subject dogs measured with TLDs 0.5 cm superior to isocentre in the transverse plane of the beam. TLD results shown in Fig. 5.10b show results of two TLDs with lower doses than expected (inferior position). This may be due to portions of one of the beams (gantry angle: 120°) going through the metal
Figure 5.12: Measured data showing linearity of fluence (dose) with MU. a) is fluence modulation for track 20 (0.5 cm superior to beam axis) with doses varied in 25% increments. b) is the dose Vs MU graph showing the dose linearly scales with output of the machine. Different regions (1st peak, the valley and the 2nd peak) of the dose profiles were fit to a first order polynomial.

C-arms of the couch, causing underdosage to that area. Figures 5.13b and 5.14b are measurements done in the cadaver, correlating the dose in the target area and the spinal cord with the MPB calculations. One of the constraints selected in our protocol was to restrict the dose to the spine to 50% or less of the target dose. The agreement between measurements and calculations is better than ±3%. However, any small discrepancies between measured TLDs and point dose calculations may be due to the position of some TLDs that are near field edges and consequently in regions of sharp dose gradients. The dashed lines in Figure 5.14 emphasizes this point. A 2 mm shift of TLDs in anterior direction (2 mm shift in couch motion in posterior direction) can result in errors greater than 20%.

The increase in dose to the spine in the inferior part of the phantom measurements in Figure 5.14 is due to the curvature of the cord which is not present in the phantom.
Table 5.2: Tabulated treatment parameters for the 9 IMRT dogs used in this study. The dose-rate used in IMRT mode was 100 cGy/min. The control subjects were irradiated at a dose-rate of 400 cGy/min dose-rate with a 4-field conventional treatment regimen.

<table>
<thead>
<tr>
<th>IMRT study</th>
<th># of fields per frac.</th>
<th>Avg # of segts. per field</th>
<th>Avg # of MU per beam</th>
<th>Avg delivered time per frac. (min)</th>
<th>Avg depth to isocentre (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>7</td>
<td>133±6</td>
<td>508±67</td>
<td>35±5.0</td>
<td>6.4±1.3</td>
</tr>
<tr>
<td>2</td>
<td>7</td>
<td>115±15</td>
<td>427±31</td>
<td>29±2.0</td>
<td>4.5±1.0</td>
</tr>
<tr>
<td>3</td>
<td>7</td>
<td>110±15</td>
<td>450±64</td>
<td>30±4.5</td>
<td>4.5±1.0</td>
</tr>
<tr>
<td>4</td>
<td>7</td>
<td>104±13</td>
<td>445±41</td>
<td>31±3.0</td>
<td>4.0±1.0</td>
</tr>
<tr>
<td>5</td>
<td>7</td>
<td>98±11</td>
<td>303±17</td>
<td>21±2.0</td>
<td>5.5±1.4</td>
</tr>
<tr>
<td>6</td>
<td>7</td>
<td>95±14</td>
<td>294±49</td>
<td>20±3.4</td>
<td>6.0±1.8</td>
</tr>
<tr>
<td>7</td>
<td>7</td>
<td>119±7</td>
<td>397±29</td>
<td>28±2.0</td>
<td>4.5±1.0</td>
</tr>
<tr>
<td>8</td>
<td>7</td>
<td>85±24</td>
<td>299±55</td>
<td>20±4.0</td>
<td>5.4±1.0</td>
</tr>
<tr>
<td>9</td>
<td>7</td>
<td>98±14</td>
<td>340±94</td>
<td>24±6.0</td>
<td>4.9±1.3</td>
</tr>
</tbody>
</table>

The dose plot in Figure 5.14b is truncated in the superior direction because of difficulties in placing the catheter in spine.

5.4 Discussion

The results of the irradiation of the 15 canines indicate that the goal of using IMRT to irradiate target volume with sparing a critical normal structure was met. Unfortunately, the measure of physiological response used was incapable of providing a more accurate measure of the dose delivered. Nevertheless, valuable information was gained by this as well as by the other methods used. As Figure 5.14 indicates, the dose in the spinal cord was quite sensitive to position accuracy. In addition the cord was in a steep dose “valley”. These two factors coupled with the uniform response of the control dogs, certainly provide some level of confidence in the spatial and dosimetric accuracy of the system.

However, the significance of the work reported here also includes the results from the
Figure 5.13: Measured (points with error bars) and calculated (solid lines) composite dose distributions. Measurements were carried out in the target volume with TLDs in phantom (a) and in subject canine (b) in the longitudinal plane.

other experimental methods. These were designed to validate critical steps in the entire IMRT procedure. In this case the procedure included the CT imaging, the beam optimization and dose calculation verification, spatial and dosimetric accuracy in phantom and biological materials and the successful delivery of a fractionated course of IMRT treatment.

As with any treatment planning system, it is important to validate the results of the IMRT planning, including optimization. Within the limits of measurement accuracy, we have validated the algorithm used. In general, the sources of error in our measurements are attributed to the following parameters:

- The physical resolution of the dosimeters (TLDs, diodes, and ion-chamber) was a limiting factor in defining the normalization points in patient space. To reduce the effect of random errors, several points along different tracks were taken for measurement.
Figure 5.14: Measured (points with error bars) and calculated (solid line) composite dose distributions measured in the spine with TLDs in phantom (a) and in subject canine (b). The dashed line in b) shows the effect on dose distribution in spine due to 2.0 mm shift.

- Problems associated with film dosimetry in high energy fields such as conversion of optical density (OD) to absolute dose due to OD saturation.

- The inherent limitation on TLD dose measurement accuracy (±3%).

- The imaging resolution of the pre-treatment and post-treatment CT images, and consequently the accuracy in image fusion and dose correlation.

- In regions of steep dose gradients of 10%/mm or in beam profiles, positional uncertainties greater than ±1.0 mm could lead to errors of 20%.

A number of studies in the literature have described various methodologies appropriate to the dosimetry of IMBs. Pasma et al [19] have presented results on treatment dose verification of IMRT beams using portal imaging systems. Their portal dose measurements agreed to within 2% of those calculated. In the study presented here, calculated
and measured dose profiles agreed to within 2-3% of each other. Low et al [79, 78] have also presented quantitative results on dose verification using in-phantom measurements. And Wang et al [52] have tested the accuracy of dose delivery in IMRT and found discrepancies in excess of 5% of central axis dose. Our in-phantom measurements on central axis of the beam shows discrepancies of less than 2% from calculated doses (see Table 5.1).

Other investigators have also worked on measurements and verification of IMRT beams which has led to understanding the capabilities, limitations and implementation of this new technique at clinical level [47, 68, 69, 72]. However, most of the results presented in the literature on dosimetry of IMBs are mainly carried out to obtain spatial dose information in phantom geometry. One advantage of the work described in this study is the assessment of anatomic dose distribution in biologically relevant tissues and fractionated dose delivery in vivo. It can be argued that the in vivo verification technique presented here is not practical for routine clinical use. Nonetheless, it provides important information that in-phantom measurements is incapable of providing; that is, how well the dose distribution generated in a dynamic field can match the planned dose not only spatially but also anatomically.
5.5 Summary

An *in vivo* study for dosimetric verification of intensity modulated beams for a dynamic MLC technique was presented in this work. A methodology for verification and calibration of MU and absolute dose profiles was presented and tested for validity. In general, due to the absence of anatomical structures with patient-specific density, external contours, and the presence of inhomogeneities, phantom measurements may not be an accurate reflection of anatomical dose distribution in patients. A more accurate method in verification of dose distributions, although not trivial to achieve, is the delivery of beams in phantoms with density and inhomogeneities similar to those in real patients.

The 2D dosimetric verification of the beams was performed with film. However, due to the limitations of film dosimetry in high-energy fields, there are drawbacks to using film as a routine quality assurance tool. The 1D profiles were measured using TLDs and a linear array of diodes. Both TLDs and diodes were calibrated in known fields to convert their signal to absolute dose.

The results of our one-year clinical trial study have allowed us to have a better understanding of IMRT and to present a more accurate assessment of the technique. The canine study included all variables present in a clinically relevant situation. The work described here offers a unique approach in assessing the potential outcome of IMRT and its clinical implementation. The method of dose verification described in this study provides both spatial and anatomic dose information that is important in evaluating a new dose delivery technique.
This chapter focuses on the effects and influences of errors in dose distributions due to spatial inaccuracies of moving MLC leaves and backup diaphragms during delivery of dynamic IMB. Quantitative assessments of random and systematic errors introduced in delivery of dynamic beams due to field defining parameters have been carried out. Results and discussion of our findings are presented.

6.1 General Considerations

In dynamic fields utilizing MLCs, the dose delivered in the treatment field is directly affected by the width of the field in a sweeping window technique, therefore, any error
in spatial accuracy of diaphragm and/or leaf settings will manifest itself not only to the boundaries of the target volume but will propagate throughout the entire volume irradiated. In fact there are many factors that effect the IMB deliveries, but none is more important than accuracy required to deliver these beams. Therefore, given the intent of IMRT, where sharp fluence gradients are produced near the boundaries of the field with critical structures in the vicinity of the target, there is the inevitable question of how accurate and safe these deliveries are. That is, uncertainties in the performance of the MLC during dynamic delivery raise the question as to whether the dose delivered in the dynamic treatment accurately agrees with the dose predicted by the optimization algorithm. These uncertainties may be attributed to some degree to the tolerance levels or mechanical limitations of the beam defining parameters that include positional and velocity inaccuracies.

In dynamic mode, a source of error in positional accuracy of the field defining parameters is the variation in the velocity of the moving parts. This is due to the required slowing down and speeding up of the leaves and the moving diaphragms between control points to produce the necessary gradients in the field. In Chapter 2 we described the mathematical derivation needed for dynamic deliveries. Factors that may cause such errors in positional accuracy include:

- delay time in response of motors pushing and pulling the leaves;
- inconsistent torque in the motors for the same leaf motion;
- limitations such as overheating in optical system with which leaves are tracked in field.

Furthermore, the current protocols on medical linear accelerator tolerance settings for parameters defining the radiation field are only made such that they meet the standard
quality assurance (QA) specifications for conventional beam deliveries [83]. For example, discrepancies of ±2 mm in MLC or collimator settings are accepted limits. However, due to sharp gradients over small regions in dynamic deliveries, these limits may not be adequate for IMRT.

A number of investigators have reported on various errors and their influence on the dosimetric outcome of the delivery of these new modalities [84, 85, 86]. These have led to new strategies for reducing the error margin in comparison to more conventional techniques [66]. However, in the area of dosimetric analysis more work needs to be done before accurate implementation of these new beam delivery techniques can be carried out.

We have studied two sources of uncertainties deemed important in dynamic therapy of photon beams, namely, errors caused by uncertainties in velocity and in position of MLC leaves and backup diaphragms. Various limiting cases are modelled in our study and recommendations and discussions based on our results are presented.

### 6.2 Methods in Error Analysis

This study was carried out so that the methods and results of our data analysis maintained its generality and may be applied to any linac capable of delivering dynamic IMBs and equipped with an MLC as described here.

An IMRT prescription file designed for this study was set up so that the intensity modulated fields produced were the result of movement of the leaves and diaphragms throughout delivery.

Figure 6.1 is a diagram of a typical MLC system with rounded-end leaf design shown in the inset of the Fig. 6.1.

The motion of MLC was indexed to the appropriate number of MU rather than the
time so that the number of photons received by a point in the field was proportional to some percentage of the total dose during exposure. The speed of the leaves and backup diaphragms were allowed to vary between 0 to ±2 cm/sec.

### 6.2.1 Field Perturbation

Radiation fields delivered in dynamic mode were perturbed both randomly as well as systematically by adding or subtracting positional offsets to each of the MLC leaves and/or diaphragms at each control point in the IMRT delivery sequence. The systematic and random offsets were applied to:

- the position of the leaves alone;
- the position of the diaphragms alone;
• positional offsets in both leaves and diaphragms simultaneously;

• velocity offsets to leaves alone;

• velocity offsets to diaphragms alone;

• velocity offsets to both leaves and diaphragms.

Perturbation in leaf motion in each segment of the field was accomplished by inputting the optimized fluence files into a noise-generator and varying the position of the leaves from their nominal settings. Appendix A shows an example of the simulations carried out in this study with the output of the perturbed files when a ±2 mm random offset was applied to the motion of the leaves. Figure 6.2 is a schematic illustration in perturbation

![Diagram](image)

**Figure 6.2:** Leaf perturbation using a Gaussian distribution with standard deviation $\sigma$ and mean $\mu$. $\delta$ is the perturbation in cm.

of MLCs by an amount $\delta$ introduced in the field. The probability of finding a leaf perturbed by an amount $\delta$ obeyed the following normal distribution:

$$\frac{1}{\sigma \sqrt{2\pi}} \int_{-\infty}^{+\infty} exp\left[-\frac{1}{2}\left(\frac{\delta - \mu}{\sigma}\right)^2\right]d\delta$$  \hspace{1cm} (6.1)
where $\mu$, and $\sigma$ are the mean and standard deviation of the distribution, respectively, with $\sigma$ as:

$$
\sigma = \sqrt{\frac{\sum_{i=1}^{N} (\delta_i - \mu)^2}{N - 1}}
$$

(6.2)

where $N$ is the total number of runs.

As a safety precaution, to avoid mechanical collisions of the moving parts, some linac control systems also impose additional constraints on their delivery scheme. For example, the Elekta linac control system requires a 1-cm minimum separation at the plane of the isocentre, 100 cm from the radiation source. The constraint holds true for the opposing leaves plus the adjacent leaves on either side to avoid interdigitation between the leaves (see Figure 6.1). This mechanical limitation was modelled in our study by introducing a 1 cm gap constraint on the values selected from the Gaussian distribution. Therefore, for example, to simulate the delivery of an Elekta linac, any violation due to field perturbations that caused the gap between leaves to be less than 1 cm in distance was corrected for by calculating the difference between the minimum gap requirement and position of the two opposing leaves as follows:

$$
\text{If } |J^L + J^T| \begin{cases} 
< 1 \text{ cm} & \text{then, } \frac{1-(|J^L+J^T|)}{2} + J^{vt} \\
\geq 1 \text{ cm} & \text{no violation}
\end{cases}
$$

(6.3)

where $J$ indicate the position of the leaves. The superscripts $J^L$ and $J^T$ describe the leading and trailing leaves, respectively. The term $J^{vt}$ indicates the addition of the correction term to both leading and trailing leaves. The presence of the absolute value sign makes the calculation coordinate invariant with respect to the midline of the field. This constraint was not applied to all simulations presented here in order to model delivery by linacs that do not have this limit.

Mathematically the perturbation in velocity of the leaves was accomplished by first calculating the rate of change in position of the two adjacent control points normalized to
the fractional change in time, \( \Delta t \), and next multiplying the value by some scaling factor, \( f \) (see Equation 6.9). We define the velocity of a leaf prior to perturbation as:

\[
v = \frac{J_n - J_{n-1}}{\Delta t}
\]  

(6.4)

\( v \) is the initial unperturbed velocity term input to the control algorithm, \( J_n, J_{n-1} \) are the respective positions of the leaf at \( n \), and \( n - 1 \) control points. Similarly, the velocity of the perturbed leaves, \( v' \) resulting in final perturbed position, \( J' \) was defined as:

\[
v' = \frac{J'_n - J'_{n-1}}{\Delta t}
\]

(6.5)

with boundary conditions satisfying:

\[
v' = \begin{cases} 
0 & \text{if } v' \leq v_{\text{min}} \\
 v_{\text{max}} & \text{if } v' \geq v_{\text{max}} 
\end{cases}
\]  

(6.6)

\( \Delta t \) is the time differential each point is exposed to in the field and is given by:

\[
\Delta t = \frac{MU_{\text{tot}}(\%MU_n - \%MU_{n-1})}{D}
\]

(6.7)

where \( D \) is output of the machine in MU/sec, \( MU_{\text{tot}} \) is total the monitor units delivered in a beam and \( \%MU_n \) is the percent of the total MU in segment \( n \).

Although possible in step-and-shoot delivery, the leaf-sequencing process for fully dynamic deliveries generally does not allow backward motion of the leaves. There are two reasons for this; one is reducing the overall treatment time, and second is that any backward motion in a unidirectionally-moving leaf/diaphragm results in over or underdosage. Therefore, in our study, leaf or diaphragm perturbations were made such that negative velocities were rejected. This was done by comparing the respective unperturbed and perturbed positions of the leaves or diaphragms. If the noise added resulted in the final position of the perturbed parameter to be smaller than its initial position, it implied
negative motion and therefore the position was clamped to the original unperturbed position (zero motion). The statistical implications of this selective perturbation and the clamping of the noise outside of the constrained window was such that the noise distribution was a clipped Gaussian rather than a true Gaussian distribution. The same model was implemented if any variation due to velocity perturbation of the leaves caused an overshoot or resulted in backward motion.

The random noise generator had two components that described different states of the system independently; a velocity component, and an uncertainty in position. Therefore, the final perturbed position of the leaves was such that the magnitude of the noise could vary either as a function of the leaf speed or position, or both, simultaneously. This model of the variation in dynamic noise was chosen to represent a realistic simulation of the actual delivery since the velocities of the leaves/diaphragms vary from one control point to the next and therefore the noise associated with it should reflect the appropriate variations in speed.

In addition to the random noise, systematic shifts to the position of either the MLCs, or the backup diaphragms (see Eqn. 6.8) were introduced to further investigate the influence of errors due to the miscalibration of beam defining parameters. The static perturbation was such that the leaves, X- or the Y-diaphragms were shifted by a constant amount in or out of the field independent of each other, allowing one to simulate different cases. Thus, the gap between beam defining elements could result in a systematic shift along an axis, or an increase/decrease in the field size.

The final positions of the perturbed parameters in the output file were then checked to ensure no constraints were violated. For example, if the perturbed leaf velocity was outside a range defined by minimum and or maximum values, the output velocity was “clamped” to meet the boundary constraints (see Eqns. 6.5 and 6.6). An advantage of placing such limits on our random distribution was that, it ensured the outlier noise
values were clipped off, therefore, only errors of realistic magnitudes were considered. Similarly if the gap constraint placed on the separation between two opposing leaves was violated, Equation 6.3 was used to recalculate and meet the required distance. The final position of a perturbed leaf or diaphragm is then described as:

\[ J' = \sigma G + P + J \]  

(6.8)

where \( P \) is the systematic shift term assigned to the leaves or diaphragms, or both, \( J \) is the value of the unperturbed leaf, defined in Equation 6.4. \( G \) is a unit variance distribution whose statistics obey the normal distribution described in Eqn. 6.1 scaled by an arbitrary coefficient \( \sigma \) defined as an approximation to the width of the distribution as the root mean square (rms) sum of the standard deviations of the distributions for each perturbed parameter:

\[
\sigma = \sqrt{\sigma_v^2 + \sigma_p^2}
\]

(6.9)

with \( \sigma_v = f v \)

\( \sigma_v \) is the standard deviation in velocity component of the noise, varied by the scaling factor \( f \), and \( \sigma_p \) is the standard deviation in random positioning. Equations 6.8 and 6.9 were both applicable to MLCs and the backup diaphragms.

Simulations for the entire course of a treatment (10 fractions) with random errors introduced in delivery were done to study the effect of cumulated dose errors. The average dose error summed over all fractions was calculated as:

\[
Err_{avg} = \frac{1}{N} \sum_{i=1}^{N} D'_i
\]

(6.10)

where \( D'_i \) is the integral dose per fraction from the contribution of all the dose points in a given structure when perturbed. \( N \) is the total number of fractions. The percent error in each fraction was also calculated as:

\[
\%Err = \frac{D'_i - D_i}{D_i} \cdot 100
\]

(6.11)
6.2.2 Simulations

Simulations were carried out on an Intel-based Linux platform. The dose calculation routine and inverse optimization of beam rays are described in more detail in Chapter 5. In short, doses are the result of 3D dose calculations utilizing the macro pencil beam model, with the primary beam fluence convolved with an empirically determined, depth-dependent kernel. In order to decouple measurement errors from those caused due to moving parts during beam delivery, only computer simulations were performed, and errors reported are the difference between calculated dose without leaf/diaphragm perturbations and calculated dose with perturbations.

6.2.3 Clinical Applications

We studied two clinical sites for which dynamic IMB was a desired mode of delivery. These were a T1 stage adenocarcinoma of the prostate and a clival meningioma (brain tumor). Plans generated for both sites used a 9-coplanar, equiangular beams. To provide geometrical information of the target volumes and organs at risk, CT images were imported into the treatment planning computer and contours of interest were drawn for dose calculation purposes. In the first case, beam optimization was performed with intentions of avoiding the optic chiasm and brain stem which were next to the target volume (see Fig. 6.3). The bladder and rectum were considered as the critical structures in optimization for the second case (see Fig. 6.14).

Calculations were done with fine resolution grids, allowing resolution of errors due to small shifts in the field. However, this was at the expense of longer calculation times. For the clinical cases studied here a 1.0 mm grid resolution was used. For prostate, the volume of interest was approximately $9 \times 17 \times 6$ cm which led to calculation times of approximately 7 hours per plan. For the brain tumor simulations, the volume of
interest was $5 \times 6 \times 1.5$ cm, resulting in approximately 1.5 hours of calculation time per plan. It should be noted that in clinical cases to reduce calculation times, the grid resolution is typically 5 mm. Furthermore, the criteria in introducing errors was based on the manufacturer's guidelines for leaf and diaphragm calibration tolerances. In most cases, the spatial accuracy required for each leaf or backup diaphragm, depending on field size, was at best $\pm 1.0$ mm. In virtual delivery of the beams the positional accuracy of diaphragms for small fields and large offsets was also studied.
6.3 Results and Discussion

Simulation results of 16 different cases (9 for brain and 7 for prostate) for each clinical example are presented in several forms: in differential and integral dose volume histograms, and 2D planar dose distributions for different planes in the field. Dose volume histograms (DVH), both integral and differential, are presented in % of maximum dose.

6.3.1 Simulations for Brain Study

Table 6.1 shows the variation in the field defining parameters we introduced for various cases in the brain study.

Table 6.1: Tabulated parameters used in random and systematic perturbations of MLCs and backup diaphragms. Leaf and diaphragm velocities were clamped between 0 and 2 cm/sec. The standard deviations, $\sigma$ of the perturbed parameters are in units of centimeters.

<table>
<thead>
<tr>
<th>Case study</th>
<th>Random Perturbation</th>
<th>Systematic Perturbation</th>
<th>Leaf-Gap constraint (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Multi-leaves</td>
<td>Backup Jaws</td>
<td>Multi-leaves</td>
</tr>
<tr>
<td></td>
<td>$\sigma_l$</td>
<td>$\sigma_d$</td>
<td>$P_l$</td>
</tr>
<tr>
<td>1</td>
<td>±0.15</td>
<td>±0.15</td>
<td>0.0</td>
</tr>
<tr>
<td>2</td>
<td>±0.10</td>
<td>±0.15</td>
<td>0.0</td>
</tr>
<tr>
<td>3</td>
<td>0.0</td>
<td>0.0</td>
<td>+0.1</td>
</tr>
<tr>
<td>4</td>
<td>±0.15</td>
<td>0.0</td>
<td>+0.1</td>
</tr>
<tr>
<td>5</td>
<td>±[0.15 - 0.20]</td>
<td>±[0.15 - 0.20]</td>
<td>0.0</td>
</tr>
<tr>
<td>6</td>
<td>0.0</td>
<td>±[0.15 - 0.25]</td>
<td>0.0</td>
</tr>
<tr>
<td>7</td>
<td>0.0</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>8</td>
<td>0.0</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>9</td>
<td>±0.15</td>
<td>±0.2</td>
<td>0.0</td>
</tr>
</tbody>
</table>

Different values were given to the terms $\sigma_v$ and $\sigma_p$ in Equation 6.9 to vary the width of the distribution due to velocity and position uncertainties and their rms sum is presented in the above table as standard deviations, $\sigma_l$ and $\sigma_d$. However, the results of perturbation...
Figure 6.3: Transverse view of the brain with contours of the target cross-section and the critical structures outlined. Calculated isodose lines are also shown for the case with nominal settings. The dashed lines represent the direction of the incoming beams.

due to velocity term of Equation 6.9 is implicitly defined in the final perturbed positions (see Appendix A). The parameters $P_t$ and $P_d$ (in cm) are also defined in Equation 6.8.

Figure 6.3 is a transverse view of the brain with structures of interest drawn. The calculated isodose lines are also superimposed to show degree of conformity with ideal dose distributions.

A plot of leaf-trajectory (MLC position as a function of time) for two opposing leaf-tracks in the field is shown in Figure 6.4a. The subsequent perturbed leaf trajectories were overlaid on the corresponding nominal leaf settings to illustrate the type and magnitude of random noise introduced in our study. Figure 6.4b show the fluence profiles for the
Figure 6.4: a) Leaf trajectory from two opposing leaves in a dynamic field. b) is the corresponding fluence profiles. The grey lines in figures (a) and (b) represent the state of the leaves/diaphragms when perturbed randomly with a standard deviation of ±1.0 mm. The black lines are ideal deliveries.

The amplitude of the noise introduced for a single leaf in dynamic delivery is shown in Figure 6.5. The results shown in Figure 6.6 illustrate the effect in changing velocities of the leaves. The position of the perturbed leaf was interpolated linearly in time with zero slope to account for the overshoot whenever the velocity violation was triggered. Figure 6.6 shows the nominal and perturbed leaf velocity as a function of control points for two opposing leaves.

Figure 6.7a shows the changes to integral DVHs of random field perturbation introduced to each subfield of the nine beams in the brain study. The leaves were allowed to close-in, simulating zero distance separation. The maximum allowed velocity of the leaves or the diaphragms was set to 2 cm/sec. Figure 6.7b is the comparison of integral dose histograms when the backup diaphragms were also allowed random perturbation.
See cases 1 and 2 in Table 6.1 for the magnitude of the standard deviations.

Figure 6.8 illustrates the effect on integral DVHs of systematic and systematic plus random errors combined. Figure 6.8a shows a +1 mm static shift to leaf motion, while Figure 6.8b is the static shift in (a) plus a random noise added to the MLC motion with standard deviation of ±1.0 mm (see cases 3 and 4 in Table 6.1).

Figures 6.9a through 6.9d compares the differential DVHs of nominal and perturbed fields when a dynamic-noise with standard deviations ranging between ±0.15 to ±0.2 mm was added to the leaf/diaphragm motion.

Differential histograms for an ideal delivery could be interpreted as a delta function, indicating uniform energy deposition throughout the target and any variation from that
Figure 6.6: Velocities for right (a) and left (b) leaves for a given leaf-pair as a function of control points. The corresponding perturbed velocities (grey lines) are shown superimposed on the nominal velocity (black lines). $\sigma$, the standard deviation of the random perturbation was ±1.0 mm.

indicates nonuniformity of dose to the structure of interest and more importantly the spread of dose to adjacent organs. The results from Figure 6.9 show decrease in the overall dose to the target and other structures at risk as the dynamic noise to the leaves was increased.

One caveat in presenting our data in this form (integral or differential dose volume histograms) is that they do not reveal where specific hot or cold spots in the volume of interest occurs but show the overall decrease or increase in dose. Since many inverse planning algorithms utilize DVHs as constraints, this analysis can shed light on the clinical importance of different optimization results given the possibility of delivery errors.

Figure 6.10 show the simulation results for random perturbations in the diaphragms only. The standard deviations were between ±0.15 to ±0.25 cm. The perturbed parameters are tabulated in case 6 of table 6.1. The results of Figure 6.10 show the measure of dose conformity which is severely affected in the target volume. This is seen from the
Figure 6.7: Random field perturbation introduced in leaf and backup diaphragm motion. a) is the simulation results when zero gap constraint was allowed and b) with a 1 cm minimum separation between opposing leaves. Dashed lines are the perturbed cases and the solid lines are the nominal doses for gross tumor volume (GTV), brain stem and optic chiasm. The magnitude of the noise was varied from one segment to the next with standard deviations for leaves and diaphragms in a) ±2.0 mm and in b) ±1.5 mm. The parameters are also shown in cases 1 and 2 of Table 6.1 respectively.

The width of differential histograms.

Figure 6.11 is the results of integral dose distributions in shifting the diaphragms systematically by ±1.0 mm. The perturbed parameters are shown in cases 7 and 8 in Table 6.1.

Results from Figure 6.11 show that deviations in dose distributions are not uniform about the mean. This is understood by the following argument. For the specific example studied, both MLCs and the backup diaphragms were used simultaneously in producing the optimized intensity maps. The probability of having segments in which leaves are trailing or leading their corresponding backup diaphragms or vice versa is purely dependent on the intensity gradients required at each subfield. For example, at times where small fields are required (typically smaller than 1 × 1 cm), the backup diaphragms were
Figure 6.8: a) Dose volume histograms comparing the effects of field perturbation between the nominal dose distributions (solid lines) with a +1.0 mm systematic shift to the leaves (dashed lines) in each subfield. b) DVHs when systematic (+1.0 mm) and random errors in leaf (±1.5 mm) motion are combined. Dose histograms are shown for gross tumor volume, optic chiasm and brain stem. Standard deviations of the random distribution are shown in Table 6.1, cases 3 and 4, respectively.

required to pull-in and form the desired field with the opposing leaves, especially if the MLCs were prevented from closing in to zero cm separation. Now, in the above example, when the diaphragms have been shifted by +1 mm, the leaves would still provide adequate blockage of radiation (ignoring the low transmission and leakage through the leaves), and the increase in dose is not as significant as is shown in Figure 6.11b. However, in Figure 6.11a where the diaphragms are shifted so that the subfields in each beam are closed in, there will be additional blockage thus reducing the portal and the fluence reaching the field.

To study the effect of random errors over the entire course of fractionated treatment, the velocity component and position of MLCs and backup diaphragms were perturbed randomly in 10 different runs to simulate a normal treatment regimen for the brain case.
See case 9 in Table 6.1.

Figure 6.12 is a comparison between the average percent dose difference of 10 simulations with the nominal dose distributions. The results in Figure 6.12 show a difference in the 2D planar dose distribution to the target volume and other critical structures by as much as 20%. The accumulated dose volume histogram of both the nominal dose distribution and the average of DVHs is also shown in Figure 6.12.

Figures 6.13a through 6.13d are the 1D profiles of percent dose difference calculated for Figure 6.12 plotted in different planes. The profiles are a better indication of the magnitude of dose nonuniformity to each organ due to the random errors introduced. Figure 6.13a is the in-plane profile through the optic chiasm, target volume, and the brain stem, showing the brain stem receiving hot spots where the dose is higher by as much as +7% while in the same plane the target volume is being under dosed by as much as −16% of the prescribed dose. Figures 6.13b through 6.13d are cross-plane profiles through the brain stem, target volume, and the optic chiasm, respectively, showing variation in dose distribution. The extent in which each organ is covered in the dose profile is shown.
Figure 6.9: Differential dose volume histograms illustrating dose distributions to gross tumor volume, brain stem, and the optic chiasm. The perturbed doses are shown in grey colors and nominal doses are the black lines. Total $\sigma$ for figure a) was $\pm 2.0$, for b) $\pm 2.2$, for c) $\pm 2.5$, and for d) $\pm 2.8$ mm. The results show the degradation of dose distribution and the measure of dose inhomogeneity as $\sigma$ increases. Simulations refer to case 5 in Table 6.1.
Figure 6.10: Histograms showing the variation from the norm and the effect on dose distribution in gross tumor volume, brain stem and optic chiasm due to random perturbations in moving diaphragms only. The perturbed doses are shown in grey lines and nominal doses are presented with black lines. Total $\sigma$ due to random errors in diaphragm motion in figure a) is $\pm 1.5$ mm, in b) is $\pm 1.8$, in c) is $\pm 2.1$, and d) is $\pm 2.5$ mm (see case 6 in Table 6.1). The results show the gradual degradation of dose homogeneity as $\sigma$ increases.
Figure 6.11: Integral dose distributions in gross tumor volume, optic chiasm and brain stem for nominal (solid lines) simulations and those with systematic shifts (dashed lines) introduced to each subfield in moving diaphragms only. a) is due to $-1.0$ mm shift and b) is due to $+1.0$ mm shift to each segment. Perturbed parameters are also given in cases 7 (figure b) and 8 (figure a) in Table 6.1

Figure 6.12: 2D surface plot of percent dose difference (left) due to random perturbation of both MLCs and diaphragms averaged over 10 fractions. Average integral dose volume histograms for each organ (right) is also shown. The standard deviation of the random noise is $\pm 2.5$ mm and are given in case 9 of Table 6.1. Errors of up to 20% are observed in this case.
Figure 6.13: One dimensional profiles of percent dose difference plotted in-plane (a) showing the percent error in dose to the target volume, optic chiasm, and the brain stem. Figures (b), (c), and (d) are cross plane plots of the brain stem, target volume, and the optic chiasm, respectively. They show the level of dose non-uniformity to the organs due to errors produced by field perturbation in case 9 shown in Table 6.1.
6.3.2 Simulations for Prostate Study

Perturbations similar to those described in the previous section were also used for the second clinical example given in this chapter, a tumor of the prostate. Reasons such as different field geometries, different energy required, organ motion, and different tolerance levels for critical structures justified the repeat of some of the above simulations. Furthermore, we wanted to examine the dependence of the magnitude of errors seen on treatment site. Table 6.2 shows the tabulated parameters used for the prostate simulations. The definition of parameters are similar to those in Table 6.1.

Table 6.2: Parameters used in simulation of random and systematic perturbation of MLCs and backup diaphragms in the prostate study. $\sigma$, is in units of centimeters.

<table>
<thead>
<tr>
<th>Case Study</th>
<th>Random Perturbation Multi-leaves $\sigma_l$</th>
<th>Random Perturbation Backup Jaws $\sigma_d$</th>
<th>Systematic Perturbation Multi-leaves $P_l$</th>
<th>Systematic Perturbation Backup Jaws $P_d$</th>
<th>Leaf-gap constraint (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>$\pm 0.1$</td>
<td>$\pm 0.14$</td>
<td>$+0.2$</td>
<td>$+0.2$</td>
<td>1.0</td>
</tr>
<tr>
<td>2</td>
<td>$\pm 0.25$</td>
<td>$\pm 0.25$</td>
<td>$-0.2$</td>
<td>$-0.2$</td>
<td>1.0</td>
</tr>
<tr>
<td>3</td>
<td>$\pm 0.1$</td>
<td>$\pm 0.14$</td>
<td>0.0</td>
<td>0.0</td>
<td>1.0</td>
</tr>
<tr>
<td>4</td>
<td>$\pm 0.2$</td>
<td>$\pm 0.25$</td>
<td>0.0</td>
<td>0.0</td>
<td>1.0</td>
</tr>
<tr>
<td>5</td>
<td>0.0</td>
<td>$\pm [0.2 - 0.3]$</td>
<td>0.0</td>
<td>0.0</td>
<td>1.0</td>
</tr>
<tr>
<td>6</td>
<td>0.0</td>
<td>0.0</td>
<td>$+0.1$</td>
<td>$+0.1$</td>
<td>1.0</td>
</tr>
<tr>
<td>7</td>
<td>0.0</td>
<td>0.0</td>
<td>$-0.1$</td>
<td>$-0.1$</td>
<td>1.0</td>
</tr>
</tbody>
</table>

Figure 6.14 is a cross sectional CT image of the prostate. The calculated isodose lines and the contoured critical structures and the target volume are superimposed on the image. The field size for each beam was on average $6 \times 8$ cm.

Figure 6.15 are histograms for nominal and perturbed fields for cases 1 and 2 of Table 6.2. The magnitude of the random parameters in Figures 6.15a and 6.15b are identical except in that in (a), $+2.0$ mm shift was added and in (b), a $-2.0$ mm systematic shift was added to the fields. Cases 3 and 4 of Table 6.2 are random perturbations in leaf and
Figure 6.14: Transverse view of the prostate with the contours of the target cross-section and the critical structures drawn. The calculated isodose lines are also shown.

the diaphragm motion. The calculated dose distributions are shown in Figure 6.16. The results of the above two cases are interesting in that, random perturbations with standard deviations shown in Table 6.2 do not affect the dose distribution in the prostate as much as it varied the dose distribution in the brain simulations. One plausible explanation is that in this case, the opening of the portal in each subfield is wider than the brain subfields. Therefore, when the leaves or the diaphragms are perturbed, the effect is not as pronounced. For the brain study, leaf separations on average were approximately 0.5 cm (for cases where 0 gap constraint was allowed) versus the 1.4 cm average separation of the leaves for the prostate study. This also will make the calculation of output factor more difficult as it is a rapidly changing function of field size for small fields.
Figure 6.15: Dose volume histograms for gross tumor volume, bladder, and the rectum. a) The variation in dose when both systematic errors of $\pm 2$ mm and random errors of $\pm 1.5$ mm were added to leaf and diaphragm motion. b) The effect of error on dose distribution when systematic errors of $-2.0$ mm and random errors of $\pm 3.0$ mm were introduced in leaf and diaphragm motion. Dashed lines represent the perturbed cases and the solid lines are ideal doses. Doses are rescaled to 100% of the maximum dose. Standard deviations of distributions are also given in cases 1 and 2 of Table 6.2.

Figure 6.17 shows the effect of positional inaccuracies in moving diaphragms during irradiation. The velocity component of each diaphragm was varied from its nominal speed in each segment and is shown in terms of standard deviation in position in Table 6.2, case 5. The results show decreases in doses as the variation in velocity component of the diaphragms increases; however, the dose degradation is not as severe as that shown in Figure 6.12.

Figure 6.18 shows the effect of systematic shifts to the leaves and diaphragms on DVHs. Each beam segment was shifted by $\pm 1.0$ mm, simulating errors due to mis-calibration of the diaphragms. As expected, both Figures 6.18a and 6.18b are nearly symmetric about the ideal doses. This is because when subfields are closed in, the diaphragms dictate the rate of fluence reaching the field and when opened by $+1.0$ mm,
Figure 6.16: Dose volume histograms for gross tumor volume, bladder and rectum. a) is the variation of dose distribution from nominal doses (solid lines) when random perturbations (dashed lines) of ±1.5 mm was introduced, and b) is the comparison when random errors of ±3.0 mm was simulated (see case studies 3 (a) and 4 (b) of Table 6.2). Doses are rescaled to 100% of the maximum dose.

the leaves determine the field size but in this case the leaves were not perturbed.
Figure 6.17: Dose volume histograms illustrating comparison of volumetric dose distributions to gross tumor volume, bladder and the rectum for nominal (solid lines) doses with those randomly perturbed (dashed lines) with diaphragm motion. a) is random perturbation of ±2.0 mm, b) ±2.2 mm, c) ±2.5 mm and d) ±3.0 mm (case 5 of Table 6.2). Note as the random error in the diaphragm motion increases so does the error in dose distribution.
Figure 6.18: Comparison of integral dose distributions to the gross tumor volume and other critical structures due to nominal (solid lines) and systematic (dashed lines) errors. a) is a shift of +1.0 mm, and b) is a shift of −1.0 mm to each segment. See cases 6 and 7 in Table 6.2.
6.3.3 Quantitative Analysis

Prescribed doses to the target volume in the planning stage was such that 50% of the volume received at least 95% or more of the delivered dose. Table 6.3 below shows the desired doses to each organ and the percent doses received to 50% of the volume in each organ under ideal delivery settings.

Table 6.3: Desired doses to 95% to each organ and percentage of prescribed dose given to 50% of the total volume in each organ in reference simulations.

<table>
<thead>
<tr>
<th>Desired Doses (cGy)</th>
<th>Brain Target</th>
<th>Brain Stem</th>
<th>Optic Chiasm</th>
<th>Brain Target</th>
<th>Bladder</th>
<th>Rectum</th>
</tr>
</thead>
<tbody>
<tr>
<td>%Dose to 50% of volume</td>
<td>150</td>
<td>40</td>
<td>60</td>
<td>180</td>
<td>35</td>
<td>2</td>
</tr>
</tbody>
</table>

Doses from Table 6.3 were used to determine the corresponding percent volume receiving the same dose when field parameters were perturbed. Table 6.4 shows the variation in partial volume for each study case with numbers in brackets as the percent deviation from nominal doses.

One interesting observation that was noted from the results of the simulations is that even though the noise added in each case was purely random with a probability density described in Equation 6.1, the integral dose distributions calculated from our simulations always was less than the nominal dose distributions. Intuitively, one would expect that if the distribution were purely random, then for many repetitions of the simulations the mean of the perturbed cases should converge to the norm doses. This proved not to be the case here. The following arguments are given to explain this.

There are two reasons why cumulative dose in a randomly perturbed field tend to decrease when compared to the nominal doses. The first is that the constraints placed on
Table 6.4: Percent volume receiving the prescribed doses shown in Table 6.3 after they were perturbed. Values in brackets are percent errors.

<table>
<thead>
<tr>
<th>Case study</th>
<th>Brain</th>
<th>Brain Stem</th>
<th>Optic Chiasm</th>
<th>Prostate</th>
<th>Bladder</th>
<th>Rectum</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2.0 [-96]</td>
<td>45.0 [-10]</td>
<td>40.0 [-20]</td>
<td>97.0 [+94]</td>
<td>65.0 [+30]</td>
<td>56.0 [+12]</td>
</tr>
<tr>
<td>2</td>
<td>1.0 [-98]</td>
<td>42.0 [-16]</td>
<td>38.0 [-24]</td>
<td>0.2 [-100]</td>
<td>38.0 [-24]</td>
<td>44.0 [-12]</td>
</tr>
<tr>
<td>3</td>
<td>85.0 [+70]</td>
<td>53.0 [+6]</td>
<td>75.0 [+50]</td>
<td>26.0 [-48]</td>
<td>50.0 [0.0]</td>
<td>52.0 [+4]</td>
</tr>
<tr>
<td>6</td>
<td>0.0 [-100]</td>
<td>40-41 [-20]</td>
<td>18-21 [-64]</td>
<td>80.0 [+60]</td>
<td>61.0 [+22]</td>
<td>55.0 [+10]</td>
</tr>
<tr>
<td>7</td>
<td>0.0 [-100]</td>
<td>40.0 [-20]</td>
<td>18.0 [-64]</td>
<td>0.2 [+100]</td>
<td>41.0 [-18]</td>
<td>48.0 [-4]</td>
</tr>
<tr>
<td>8</td>
<td>84.0 [+68]</td>
<td>53.0 [+6]</td>
<td>76.0 [+52]</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>9</td>
<td>0.2 [+100]</td>
<td>41.0 [-20]</td>
<td>36.0 [-28]</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

The algorithm producing the perturbations, namely, the limits on velocity of the leaves and/or diaphragms and the separation between the leaves, cause the random distribution to be asymmetric. That is, the errors do not superpose linearly. When subfields in a dynamic delivery are perturbed randomly with the leaves and backup diaphragms restricted only to move forward (no backward velocity), there are instances where the perturbed position of the leaves will surpass the position of opposing Y-diaphragms such that they overlap. This is shown schematically in Figure 6.19 where the perturbed leaf covers the portal thus reducing the dose to only transmission through the leaves and backup diaphragms. This is because the diaphragms are providing the additional blockage. There is no way for these constraints to produce a larger field opening, thus leading to an asymmetry in the delivered dose compared to the ideal values.

The second reason which may be the more dominant effect, is due to the number of degrees of freedom with which a field can randomly be perturbed. Consider only two opposing leaves and the two opposing diaphragms as an example to justify our
Figure 6.19: Schematic diagram illustrating the blocking of the beam portal due to field perturbation when diaphragms and the leaves are used in field modulation.

argument (see Fig. 6.19). Because the fluence reaching the field is dependent on the portal shape, and because the diaphragms and leaves can move independently, there are 4 possible permutations due to two effects (an increase and a decrease in the final field) with only one of them resulting in the increase of total dose. That is, because each of these permutations is equally probable, the probability for any choice is 1/4 or 25%.

To find the probability for obtaining any mixture of beam portal without differentiating between the increase or decrease in field size, we must add the probabilities corresponding to each possible resultant field:

\[
P_< = P_<^l P_<^d + P_<^d P_<^l + P_>^l P_>^d = 75% \
\text{decrease in field size}
\]

with all \( P_s = 50\% \)

The subscripts \(<\) and \(>\) stand for probability of having smaller and larger fields, respectively and superscripts \(l\), and \(d\) stand for fields produced by the leaf and diaphragm.
openings. We note that the sum of probabilities for all possibilities is 1 because something is bound to happen after perturbation. Therefore, only 25% of the time there may be instances where the perturbed subfield is larger than the nominal segment but the dominant effect results in reducing the integral dose.

This phenomena is especially true and becomes more important when the synchronization process of the leaves and diaphragms are in place to avoid the tongue and groove underdosage effect and also to minimize transmission through the leaves.

A flat-top fluence profile was produced with a dynamic prescription utilizing the motion of the leaves and diaphragms to test the validity of our argument. The results are shown in Figures 6.20 and 6.21. In Figure 6.20 diaphragms were deliberately parked outside of the field, leaving the leaves to march across the field using a sliding window technique with a constant velocity of 1 cm/sec and a portal size of 1 cm$^2$ per segment.
Figure 6.21: Flat-top fluence profile utilizing the motion of the leaves and diaphragms. a) only leaves are perturbed by ±1.5 mm and in b) both leaf and diaphragms are perturbed by ±1.5 mm. The velocity constraint imposed on the leaf motion in a) was $0 \leq v_t \leq 2$ cm/sec, however, in b) dose distributions with and without constraints are shown (see case 2 of Table 6.5).

The solid line represents the motion of the leaves with no errors, and the dotted line shows perturbations introduced to each subfield with a positional standard deviation of ±1.5 mm. See Table 6.5 for the magnitude of noise and the parameters perturbed.

The effect on dose distribution due to velocity constraints imposed on the leaves is clearly shown in Figures 6.20a, with the asymmetric distribution in the profile where the average perturbed fluence is higher than the norm. This is due to rectification of velocity perturbations (no backward motion constraint). When the leaves could move backward and forward, simulating negative and positive velocity errors, an evenly distributed dose about the nominal profile is produced as shown in Figure 6.20b.

On the other hand when the diaphragms are also used in field blocking (see Fig. 6.21),
Table 6.5: Magnitude of the noise added to both MLCs and backup diaphragms of the flat-top fluence profile. \( \sigma \), the standard deviations of the perturbed parameters are in units of centimeters.

<table>
<thead>
<tr>
<th>Study</th>
<th>Random Perturbation</th>
<th>Systematic Perturbation</th>
<th>Leaf-Gap constraint (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Multi-leaves</td>
<td>Backup Jaws</td>
<td>Multi-leaves</td>
</tr>
<tr>
<td>1</td>
<td>( \pm 0.15 )</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>2</td>
<td>( \pm 0.15 )</td>
<td>( \pm 0.1 )</td>
<td>0.0</td>
</tr>
</tbody>
</table>

almost always the resultant perturbed doses are decreased. Figure 6.21a is the perturbation of MLCs with diaphragms unperturbed but only following the leaves with no perturbation, and Figure 6.21b is the result of field perturbations due to both MLCs and the backup diaphragms. A case of no velocity constraint is also shown in Figure 6.21b for the purpose of comparison.

6.3.4 Routine QA in Dynamic IMBs

Dosimetric QA of dynamic beams both in-patient (*in vivo* measurements) and in-phantom were presented in the previous chapters. However, to a large extent the method used depends on one’s assessment of degree of complexity of the beam modulations and accuracy required. Methods to verify the accuracy and stability of leaves proposed in this study are perhaps sufficient to detect errors that may be dosimetrically significant. As we have demonstrated in this chapter, the flat-top fluence profile is sensitive to both dosimetric errors as well as positional inaccuracies in MLCs and diaphragms. These tests can be done either using a linear array of detectors placed along a MLC track or with film dosimetry techniques, or perhaps portal imaging systems, if available.

More general dosimetric and mechanical QA tests may be carried out to investigate the stability of the leaves and diaphragms on routine basis. For example, leaf-pairs (typically
Table 6.6: Quality Assurance tests, frequencies, dosimetric and mechanical tolerances required for MLCs and backup diaphragms to ensure accurate delivery of dynamic IMBs are shown. The term $L_t/F_w$ is the ratio of total number of leaves in a beam to the number of fractions in one week.

<table>
<thead>
<tr>
<th>Frequency</th>
<th>Dosimetry (%/mm)</th>
<th>Mechanical (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>MLC</td>
<td>JAWS</td>
</tr>
<tr>
<td>Daily Tests:</td>
<td>$L_t/F_w$</td>
<td>Both Jaws</td>
</tr>
<tr>
<td>Tolerances:</td>
<td>$\leq \pm(1.0-3.0)$</td>
<td>$\leq \pm(1.0-3.0)$</td>
</tr>
<tr>
<td>Mech+Dosi:</td>
<td>$\leq(1.0-3.0)%/mm$</td>
<td></td>
</tr>
<tr>
<td>Weekly Tests:</td>
<td>All Leaves</td>
<td>Both Jaws</td>
</tr>
<tr>
<td>Tolerances:</td>
<td>$\leq \pm(1.0-2.0)$</td>
<td>$\leq \pm(1.0-2.0)$</td>
</tr>
<tr>
<td>Mech+Dosi:</td>
<td>$\leq \pm(1.0-2.0)%/mm$</td>
<td></td>
</tr>
<tr>
<td>Monthly Tests:</td>
<td>All Leaves</td>
<td>Both Jaws</td>
</tr>
<tr>
<td>Tolerances:</td>
<td>$\leq \pm1$</td>
<td>$\leq \pm1$</td>
</tr>
<tr>
<td>Mech+Dosi:</td>
<td>$\leq(1.0-1.5)%/mm$</td>
<td></td>
</tr>
</tbody>
</table>

between 4 to 8 pairs) may be instructed to move across the field with constant speed but different intensities (different number of MU is assigned in each segment for each leaf pair) to study the effect of tongue-and-groove underdosage as well as leaf stability. To test the positional accuracy of the leaves due to the changing velocity between subfields, explicit pauses in leaf motion and beam delivery of the flat-top profile can be inserted to determine the errors due to acceleration and or deceleration of the leaves from the difference between the commanded positions and the actual positions. If these effects are significant, fluctuations in intensity profiles (dose) will be observed. The positional inaccuracies can be tested by inserting explicit pauses while the beam is on, and resuming the leaf motion after a time interval. If the leaves over-travel or under-travel from their
defined position, the error will manifest itself in cold or hot spots at regions in field where the leaves were instructed to stop. The width of the cold or hot spots is a measure of positional inaccuracy in leaf motion.

Table 6.6 is derived from analysis of the results of this study. It provides guidelines for tolerances allowed when performing patient specific QA for dynamic IMBs. The tests are done to ensure dosimetric and mechanical stability of the machine throughout the course of treatment.

The above tests for comparison of the measured intensity modulated fields to expected values may be done in terms of relative shape of the dose distribution when films are used to compare the 2D intensity maps and in absolute terms when diodes are used to measure the absolute dose profiles. Although, our QA procedure outlined is designed for dynamic IMBs, the limits and tests may equally be applied for step-and-shoot technique as well.
6.4 Summary

In this study we have demonstrated that the quality of treatment plans and their dosimetric outcome when using highly modulated photon beams strongly depends on accurate execution of beam deliveries throughout the course of treatment. We have shown that failure of precise positioning of beam-defining parameters such as backup diaphragms and multileaf collimators, two major components in delivery of dynamic IMBs, can have significant dosimetric consequences.

To ensure safe delivery in dynamic IMBs mechanical tolerances should be less than 0.2 mm in magnitude.

Furthermore, we have developed a methodology to test the stability and dosimetric validity of dynamic IMBs that can be used at clinical settings to ensure safe and accurate beam deliveries.
This chapter summarizes the work described in this study. Conclusions are drawn with emphasis on outlining the clinical strengths and weaknesses of dynamic IMB deliveries.

7.1 Summary of The Thesis

Dynamic IMRT is a technique in which MLC leaves are instructed to move to a sequence of predefined locations during irradiation. Figure 7.1 is an example of a complex target geometry and the degree with which dynamic IMRT can achieve highly conformal dose distributions. The potential advantages of such modality include:

- better conformation to the target volume with probability of dose escalation;
- higher dose uniformity in the region of interest;
- lower toxicity to healthy surrounding tissues;
- the ability to conform to concave targets (see Figure 7.1) that otherwise is impossible to achieve with conventional modalities;
Figure 7.1: Measured 2D doses shown as surface (left) and as dose isocontours (right) along with profiles for a MLC track of a concave target geometry produced using dynamic IMBs.

- computer-control beam optimization has significantly reduced human intervention thus reducing probability of introducing operator errors;

- and its ability to reducing the overall treatment time.

However, all of this comes at a cost. IMRT has introduced new concepts that require in-depth investigations prior to its implementation. The development of inverse planning models, dose computation engines, interpreters to convert 2D IM fluences into appropriate machine deliverable leaf sequences, and new methodologies in ensuring their accurate delivery are examples of the major areas that this new modality of radiation delivery has demanded.

The goal of this research project was to investigate the validity of some of these concerns. Various areas of dynamic IMBs were studied with emphasis on its limitations, and appropriate methodologies in its implementation were developed. The project was divided in three areas.
7.1.1 Part I. Dynamic and Omni Fields

The first part of the project involved the implementation of two clinical applications, the realization of dynamic intensity modulated fields by moving backup diaphragms, and the coupling of orthogonal wedged fields (omni wedge) to produce wedged fields independent of collimator rotation. The rationale for this work was to overcome the restriction that current hardware design has placed on beam delivery techniques utilizing the omni wedge technique.

It was shown that the dynamic wedge technique utilizing the backup diaphragms meets the required criteria needed for clinical implementation and is capable of providing intensity modulated fields comparable to those produced with internal or external mechanical wedges. Dynamic wedged fields require fewer photons to deliver the same dose distribution which reduces the treatment time. One drawback of the dynamic wedge technique however, is that the methods for calculating the MU to give the appropriate dose can be complicated. However, the method used in this work was designed to allow a treatment planning system to treat mechanical and dynamic wedges in the same manner.

MLC conformation to the target volume was optimized without compromising the dose distributions in wedged fields. This was the implementation of the omni wedge on the Elekta linac. We showed this technique is capable of creating fields independent of collimator orientation, giving the freedom to use MLCs to best fit the target volume. The equations derived in Chapter 4 give an accurate prediction of the wedge angle and wedge orientation.

Based on the results of this study a proposal was put forth to the linear accelerator manufacturer to implement the dynamic wedge and omni wedge on this type of linac on the upcoming release of the control system software.
Chapter 7. CONCLUSIONS

7.1.2 Part II. Dosimetry of Dynamic IMBs

The precision and degree of accuracy with which dynamic IMB could deliver highly localized dose distributions to targets near critical structures was studied. Measurements of dose distributions were performed both in vivo and in vitro using TLDs, films, and diodes. These measurements were used to improve and validate dose and MU calculations for highly modulated dynamic beams. They provided a test of the spatial accuracy of the delivery method. They also provided a validation of the inverse planning algorithm (POCS).

In this study a unique approach in evaluating the potential outcome of IMRT and its clinical implementation was presented. Results of dosimetric verification of IMBs delivered to subject canines presented here included all variables present in a clinically relevant situation. The method of dose verification described provided both spatial and anatomic dose information that is important in evaluation of a new dose delivery technique.

7.1.3 Part III. Field Perturbation in Dynamic IMBs

Finally, a theoretical analysis of the influence and effects of field perturbations in highly modulated beams was presented. The impact of field uncertainties due to spatial inaccuracies of moving MLC and backup diaphragms was studied by introducing random and systematic errors in delivery.

We have shown the dosimetric consequences that may result from perturbation of these parameters. The magnitude of the errors were assessed by means of cumulative and differential dose volume histograms, planar dose distributions, and point dose calculations carried out for nominal and perturbed fields.

Quantitative analysis of our data presented for two different clinical cases showed that, any inaccuracy resulting from random leaf and diaphragm positioning errors in dynamic
delivery must be less than 1.0 mm and systematic inaccuracies in field modulations must be less than 0.2 mm to avoid large dosimetric errors.

7.2 Quality Assurance in IMRT

A major impediment to the development of dynamic IMBs is the problem of verification and accuracy in beam delivery. In the area of dosimetric analysis, factors that impose limitations on accurate delivery of IMRT have received less attention to date. Indeed, most of the concerns addressed here are similar to those in conventional conformal therapy, and while most of the issues are relevant to both techniques, the quality assurance aspects of the two are not the same. Because of the nature of delivery of dynamic IMRT and the current restrictions placed on beam-defining parameters, the demand for rigorous quality assurance that requires higher accuracy is the key factor in successful delivery of this new technique.

Deviations of ±2.0 mm in spatial accuracy of MLCs from its specified position may be an acceptable limit for fixed field treatments (conventional treatments), but the same does not hold true if dynamically produced highly modulated fluences are used, as the impact on the final dose distribution may be significantly different. Therefore, from analyses based on our error modeling, it is recommended that random inaccuracies in leaf and diaphragm positioning in dynamic beams be less than 1.0 mm to avoid large errors in regions with high dose gradients. Table 6.6 is a summary of the the tests and tolerances required on a daily and weekly basis in delivering dynamic IMBs.
7.3 Future of IMRT

It is very likely that IMRT will be the driving force to better control the treatment of cancer with external beam therapy, minimize acute treatment-related morbidity, making dose escalation feasible and ultimately improve local tumor control.

In the past few years much attention has been devoted to development of inverse planning algorithms and dose calculation models that can provide intensity maps capable of conforming to complex geometries that were not possible with conventional techniques. Now the trend has changed. It is equally well-understood that development and introduction of new techniques in radiation therapy requires detailed quality assurance that is unique to this type of delivery scheme and that ensures its desirable outcome.

7.4 Concluding Remarks

What can we conclude about the IMRT as we come to the end of this study? In other words, can we confidently state that IMRT is an accurate tool that will become a practical reality with promising potential? –The answer is a cautionary yes. Only if there are accurate dosimetry techniques in place.

We demonstrated that the dosimetry and quality assurance of IMRT and conventional techniques are markedly different and errors in each technique can have significant different outcome. The most obvious dosimetry verifications of dynamic IMBs are tests such as: long term stability of moving parts, sensitivity of dynamic MLC and backup diaphragms as a function of gantry angle and detection of variations in dose distributions either by means of in vivo dosimetry, portal imaging, or in vitro measurements should be in place to ensure successful beam delivery.

Furthermore, from the experience we gained in this project, it is recommended that
initial implementation of simpler techniques such as step-and-shoot, dynamic omni fields, and 1D dynamic wedged fields may be a better approach in understanding the nature of beam delivery as they are less complex in both dosimetry and QA in comparison to fully dynamic delivery techniques with continuously varying intensities.

An investigation on delivery and dosimetry aspects of dynamic IMBs in photon fields was presented in this thesis. We have developed a methodology in dosimetry of dynamic IMBs and have shown both the advantages and potential dosimetric consequences in delivery of IMBs. Our objectives have been met.


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of intensity modulated 3d conformal treatment plans based on biological indices.

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1999.


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[22] T R Bortfeld, D L Kahler, T J Waldron, and A L Boyer. X–ray field compensation

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[83] L Xing, Z-X Lin, S S Donaldson, Q T Le, D Tate, D R Goffinet, S Wolden, L Ma, and A L Boyer. Dosimetric effects of patient displacement and collimator and gantry


The appendix summarizes an example of simulation files of nominal and perturbed positions of MLCs and diaphragms for a given beam, specified by the percent of total MU. Only two segments (first and last subfields) in each case is shown. A random positional offset of \( \sigma = \pm 2 \) mm was introduced to leaf motion in each subfield.

<table>
<thead>
<tr>
<th>Beam Identification</th>
<th>Leaf/diaphragm positions (input file)</th>
<th>Leaf/diaphragm positions (output file)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Patient Name</td>
<td>Brain</td>
<td>Brain</td>
</tr>
<tr>
<td>Treatment ID</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Treatment Name</td>
<td>IMRT</td>
<td>IMRT</td>
</tr>
<tr>
<td>Beam Name</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Beam Number</td>
<td>1/9</td>
<td>1/9</td>
</tr>
<tr>
<td>Beam MU</td>
<td>77</td>
<td>77</td>
</tr>
<tr>
<td>Number of Control Points</td>
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<td>29</td>
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<tr>
<td>Energy</td>
<td>6 MV</td>
<td>6 MV</td>
</tr>
<tr>
<td>Modality</td>
<td>XRAYS</td>
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### Appendix A. Input/Output of Simulation Files

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<th>0</th>
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<tbody>
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<td>0.0</td>
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<td>1.000</td>
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<td>X2 Diaphragm</td>
<td>1.0</td>
<td>1.000</td>
</tr>
<tr>
<td>Y1 Diaphragm</td>
<td>-1.1</td>
<td>-0.943</td>
</tr>
<tr>
<td>Y2 Diaphragm</td>
<td>2.1</td>
<td>2.196</td>
</tr>
<tr>
<td>trk, L2, L1</td>
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<td>3.166 -2.139</td>
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<tr>
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<tr>
<td>trk, L2, L1</td>
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<td>3.089 -2.225</td>
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<table>
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</tr>
<tr>
<td>X1 Diaphragm</td>
<td>1.0</td>
<td>1.000</td>
</tr>
<tr>
<td>X2 Diaphragm</td>
<td>1.0</td>
<td>1.000</td>
</tr>
<tr>
<td>Y1 Diaphragm</td>
<td>3.7</td>
<td>3.911</td>
</tr>
<tr>
<td>Y2 Diaphragm</td>
<td>-2.7</td>
<td>-2.814</td>
</tr>
<tr>
<td>trk, L2, L1</td>
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<td>-1.842  2.779</td>
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<td>21  -1.7  2.7</td>
<td>-1.872  2.841</td>
</tr>
<tr>
<td>trk, L2, L1</td>
<td>22  -1.7  2.7</td>
<td>-1.844  2.744</td>
</tr>
</tbody>
</table>

---

*End Of Beam*

trk\(^1\), L2\(^2\), L1\(^3\)

---

\(^1\)trk: track number.

\(^2\)L2: position of left leaf.

\(^3\)L1: position of right leaf.
Appendix B

GLOSSARY

List of Abbreviations

BEV  beam's eye view
CAX  central axis of the beam
CPE  charged particle equilibrium
CT   computed tomography
CTV  clinical target volume
$d_{\text{max}}$ depth of maximum dose
dDVH differential dose volume histogram
DVH  dose volume histogram
DWF  dynamic wedge factor: ratio of wedged to open field, measured on the central axis of the beam at a given depth in water.
EPID electronic portal imaging device
GTV  gross tumor volume
IM   intensity modulation
IMB  intensity modulated beams
Appendix B. Glossary

IMRT intensity modulated radiation therapy
IWF internal wedge factor: see definition of DWF.
MLC multileaf collimator
MPB macro pencil beam: see reference [72].
MU monitor units: cumulative radiation dose which increases monotonically with total elapsed time while the beam is on.
NTCP normal tissue complication probability
OD optical density
POCS projections onto convex sets: see reference [46].
PTV planning target volume: the volume encompassing the tumor with additional margin to include microscopic disease, lymphatic drainage and margin for patient movement and setup uncertainties.
rms root mean square
QA quality assurance
ROF relative output factor: ratio of doses for a given field size to a $10 \times 10$ field measured at maximum dose depth.
SD standard deviation
TCP tissue complication probability
TF transmission factor: ratio of blocked to open field measured in air.
TLD thermoluminescence dosimetry
TPR tissue phantom ratio: ratio of dose at a given point in phantom to the dose at the same point at a fixed reference depth, usually 5 cm.
1D one dimensional
2D two dimensional
3D three dimensional
**List of Units**

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Unit Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>°C</td>
<td>degree celsius</td>
</tr>
<tr>
<td>cGy</td>
<td>centigray</td>
</tr>
<tr>
<td>cm/sec</td>
<td>centimeter per second</td>
</tr>
<tr>
<td>cm²/g</td>
<td>centimeter square per gram</td>
</tr>
<tr>
<td>g</td>
<td>gram</td>
</tr>
<tr>
<td>keV</td>
<td>kilo electron volt</td>
</tr>
<tr>
<td>MeV</td>
<td>million electron volt</td>
</tr>
<tr>
<td>mm</td>
<td>millimeter</td>
</tr>
<tr>
<td>MU/sec</td>
<td>monitor unit per second</td>
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<tr>
<td>MV</td>
<td>megavoltage</td>
</tr>
<tr>
<td>s</td>
<td>second</td>
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<tr>
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<td>joule per degree kelvin</td>
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<tr>
<td>J/kg</td>
<td>joule per kilogram</td>
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