FILM DOSIMETRY AND THREE-DIMENSIONAL VERIFICATION OF CONFORMAL DOSE DISTRIBUTIONS IN STEREOTACTIC RADIOSURGERY

by

James L. Robar

B.Sc. (Physics and Physiology), McGill University, 1994
M.Sc. (Medical Physics), McGill University, 1997

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We accept this thesis as conforming to the required standard

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Department of Physics and Astronomy

The University of British Columbia
Vancouver, Canada

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Abstract

The measurement of stereotactic radiosurgical dose distributions requires an integrating, high-resolution dosimeter capable of providing a spatial map of absorbed dose. Although radiographic film is an accessible detector fulfilling these criteria, its application to the dosimetry of larger photon fields in radiotherapy has been limited by dependencies of emulsion sensitivity on depth in phantom, field size and orientation relative to the beam axis. The first part of this thesis examines the applicability of radiographic film specifically for the dosimetry of 6 MV radiosurgical beams. We show that while, for large (e.g. 20 cm x 20 cm) photon beams, the error in measured dose due to a depth-dependence of emulsion reaches 15%, the corresponding maximum error for a 2.5 cm diameter radiosurgical beam is reduced to 1.5%. For radiosurgical beams this error is comparable to the measured achievable reproducibility of film dosimetry (1.1%) and the potential error incurred due to orientation dependence (1.5%). We also demonstrate that the dependence of film sensitivity on field size is negligible for beams ranging from 1.0 cm to 4.0 cm in diameter at isocentre. The marked difference between radiosurgical and larger 6 MV photon beams in the context of film dosimetry is explained using EGS4 Monte Carlo simulation. For larger fields, significant increases in the Compton-scattered photon population, particularly below 400 keV, result in dependencies on depth and field size. In contrast, the relative increase of this low-energy component is negligible for radiosurgical photon fields. Finally, the problem of volume averaging in radiosurgical film dosimetry is addressed by evaluating a new, high-resolution CCD-based transparency digitizer in terms of spatial linearity, dynamic range, signal-to-noise ratio and uniformity.

The second part of this thesis presents the design considerations and clinical evaluation of a novel phantom system facilitating the measurement of conformal radiosurgical dose distributions using one or multiple arrays of up to 20 radiographic films separated by 3.2 mm-thick tissue-equivalent spacers. Using EGS4 Monte Carlo simulation and experimental measurement, we show that this geometry preserves tissue-equivalence to within 1%. The phantom provides 0.25 mm in-plane spatial resolution, and bicubic-interpolated isodose surfaces may be interpolated with an estimated spatial
accuracy of 1.0 mm throughout the dose volume. Dedicated software has been developed to automate the process film digitization, ordering and orienting of film images, conversion of scanned pixel value to dose, interpolation within the measured volume and export of images in DICOM format for co-registration of planned and measured three-dimensional dose distributions. Benchmark tests and example conformal dose verification studies demonstrate that this technique provides a practical method of quantifying even minor errors in radiosurgical treatment delivery.
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1 Stereotactic radiosurgery: background

1.1 Introduction

Stereotactic radiosurgery (SRS) is a treatment modality using accurately directed beams of radiation for the treatment of lesions or functional disorders in the brain. Since its inception in 1951, the technique has been used to treat a variety of brain abnormalities including malignant and benign lesions, arteriovenous malformations (AVMs) and functional disorders. The aim of radiosurgery is to deliver a high, usually single-fraction radiation dose to the target volume while minimizing dose to surrounding healthy tissues. Compared to conventional radiotherapy with larger target volumes, radiosurgery demands greater accuracy in treatment delivery, and the desired spatial and numerical tolerances for dose delivery are ± 1 mm and 5%, respectively.\(^1\) This accuracy has been achieved through technological advances in both lesion localization (through high-resolution, three-dimensional imaging techniques) and in techniques for accurate dose delivery. Recently, improved sparing of adjacent tissues has been achieved through conformal radiosurgery, whereby the shape of the administered dose is tailored in three-dimensions to match the potentially irregular volume of lesion.

While these technological advances have increased the potential for accurate radiosurgical treatment, by necessity, measurement techniques for dose treatment verification have been developed in tandem. The recent capacity to deliver complex, conformal radiosurgical dose distributions has necessitated a dosimetry system providing quantitative, high-resolution dose measurement in three dimensions. This thesis describes the relevant dosimetric considerations and methodology of a novel system satisfying these requirements, and illustrates example clinical applications.
1.2 Development of stereotactic radiosurgery

1.2.1 Stereotaxis and early x-ray radiosurgery

The technique of stereotactic radiosurgery was pioneered by the Swedish neurosurgeon Lars Leksell who combined mechanical precision provided by stereotaxis with the delivery of x-rays for treating functional brain disorders. The practice of stereotaxis, as described originally by the neurologist Spiegel and neurosurgeon Wycis, initially involved the attachment of a rectilinear frame to the skull of a patient to establish a three-dimensional coordinate system in which anatomical sites could be accurately localized. Leksell developed his own stereotactic system consisting of an arc-centred frame onto which a surgical tool could be mounted. The frame was carefully positioned on the skull using a plaster mould such that the centre of the arc coincided with the intended target point, and thus the instrument could be introduced into the brain along a radial path from any point along the arc.

Leksell later replaced the invasive surgical tool with a 200 kVp x-ray tube with the intention of using a cross-fire of photon beams to deliver a high dose to the centre of the arc while minimizing the dose to peripheral tissues. Initial positive results of the technique were reported in irradiating the gasserian ganglion for treatment of trigeminal neuralgia in a patient for which other treatments had proven ineffective. It soon became clear, however, that the orthovoltage x-ray beams were incapable of producing a well-defined dose distribution, particularly for deeply-seated targets in the brain. Higher energy x-ray tubes were unavailable at the time, and therefore other types of radiation were considered.

1.2.2 Radiosurgery using energetic particles

In 1958, Leksell, with the physicist and biologist Börje Larsson, began to assess the use of proton beams at the Gustav Werner Institute of Uppsala University. Pioneering groundwork had been reported previously by Lawrence and Tobias on human treatment using beams of protons, deuterons and helium ions at the University of California synchrocyclotron. Kjellberg also reported on the use of proton beams from the 160 MeV cyclotron at Harvard. Larsson expanded upon this work with detailed radiobiological experiments on animals, and demonstrated that the overlapping Bragg
peaks from multiple proton beams could confine a lethal dose to a volume of tissue only millimetres in diameter, with a very sharp dose gradient. To date, treatment using heavy particles has proven to be effective, and beams of helium, carbon and neon ions have been used, as well as neutrons from an isocentrically-mounted cyclotron. The production of heavy particle beams is associated with high cost and technical complexity, however, and therefore heavy-particle radiosurgery has been limited to several centres worldwide.

1.2.3 Radiosurgery using cobalt-60

The need for a more practical method led to the identification of the radioisotope \(^{60}\)Co as a suitable radiation source for radiosurgery. A 1000-Ci \(^{60}\)Co radiotherapy unit had already been developed in Canada by 1952, and much of the required dosimetric data was available. In addition to its improved accessibility, \(^{60}\)Co provided suitably penetrating radiation (1.17 MeV and 1.33 MeV gamma rays) and a reasonably long half-life (5.26 years). Leksell and Larsson employed the stereotactic approach by developing a treatment unit consisting of 179 \(^{60}\)Co sources on the surface of a sphere. The sources were focussed individually by a primary collimator attached to the treatment unit and also by an interchangeable secondary collimator helmet attached to the treatment couch. Larsson demonstrated that the new device, called the Gamma Knife, was capable of producing sharply-defined lesions in animal brain without significant damage to surrounding tissues. Despite this promising result, there was considerable opposition from radiation oncologists to the new device because the treatment involved using single-fraction doses of up to 180 Gy (3 to 5 times greater than those used in conventional radiotherapy). An overruling decision by the Swedish parliament eventually permitted installation of the first Gamma Knife and treatment began at the Karolinska Institute in Stockholm in 1967. Using photon beams with diameters ranging from 4 mm to 14 mm, Leksell created small lesions in the brain for treatment of cancer patients with intractable pain. Positive outcomes of these early treatments led to radiosurgical pallidotomy and thalamotomy for treatment of Parkinson’s disease.

Early Gamma Knife treatments were limited to functional abnormalities. Malignant and benign tumours were not treated since they were not easily localized using the
imaging techniques of the time. The primary imaging modality was pneumoencephalography, which required the injection of large volumes of air into the spinal column. The patient was then strapped to a chair and tilted or inverted in order to cause the air to displace cerebrospinal fluid. High-contrast radiographs provided orthogonal views of the ventricles, and standard brain maps were consulted to estimate the target location. Thus, the target was not actually visible in the images, but this was not seen as a precluding factor since the anatomical origins of the disorders were also not known with accuracy. The introduction of computed tomography (CT) by Hounsfield in 1972 caused a revolution in radiosurgical treatment, since benign and malignant intracranial tumours could finally be visualized and treated.

The original Gamma Knife was used in treating 1361 patients in Stockholm from 1968 to 1987. The second and third models, installed at the UCLA Medical Centre in Los Angeles and in Buenos Aires, respectively, had 201 $^{60}$Co sources, produced more spherical dose distributions and could treat lesions not only on the midline but throughout the cranial volume. After 1972 the device became available commercially* and currently there are over 125 units worldwide. It is estimated that 41,100 Gamma Knife treatments were performed from 1968 to 1995.

1.2.4 Linear accelerator-based radiosurgery

Larsson et al. first suggested the isocentrically-mounted linear accelerator (linac) as a possible source of radiation for radiosurgery, and within a decade Betti and Derechinsky described the use of multiple non-coplanar, converging arcs for the production of an ellipsoidal dose distribution in the brain. In 1985, this technique was first implemented clinically by Colombo et al. in Vicenza and in Heidelberg by Hartmann et al. In North America, Harvard University in Boston was the first to adapt the multiple-arc approach, and Podgorsak et al. developed a dynamic technique at McGill University in Montréal.

The 6 MV to 10 MV x-ray photon beams used in linac-based radiosurgery are typically focussed to diameters ranging from 10 mm to 40 mm at isocentre by mounting

* produced by Elekta Instruments, A.B. since 1972
Figure 1.1 An isocentrically-mounted linear accelerator equipped with stereotactic collimation. The intracranial target is aligned to the location of the isocentre, and both the gantry and couch may be rotated about axes through this point as indicated. These degrees of freedom allow beam entry points on the superior hemisphere of the skull over a solid angle of approximately 160°x160°.

stereotactic, tertiary collimation onto the linear accelerator (Figure 1.1). This collimation can be easily installed and removed, and thus, unlike the Gamma Knife, the linear accelerator is not dedicated to radiosurgical treatment. Numerous investigations have concluded that dose distributions from linear accelerators and Gamma Knife systems are
essentially identical.\textsuperscript{24,25,26,27,28} Currently, there are over 400 installations of linear accelerator-based radiosurgical systems worldwide.\footnote{The majority of linac-based systems are currently available commercially from BrainLAB, AG; Radionics, Inc.; Medtronic Sofamor Danek; and Howmedica Leibinger GmbH.} This thesis pertains to linear accelerator-based radiosurgery in particular, and the relevant principles are described in further detail in section 1.4, below.

1.3 Medical indications

The intended goal of the radiosurgical procedure varies depending on the particular abnormality to be treated. Radiosurgery is normally an option for lesions that are either surgically inaccessible or for which surgery has a high morbidity. The variety of diseases treatable with radiosurgery has expanded as medical imaging modalities and radiosurgical treatment procedures have improved. Treatments may consist of either a large dose in a single fraction (which is termed stereotactic radiosurgery) or may be fractionated to divide the dose among several sessions (which is referred to as stereotactic radiotherapy).\textsuperscript{23}

1.3.1 Non-malignant lesions

Stereotactic radiosurgery has been used for the treatment of benign tumours including acoustic neuromas,\textsuperscript{29,30} meningiomas\textsuperscript{31,32} and craniopharyngiomas.\textsuperscript{33} In this case, the goal of the treatment is usually not the disappearance of the lesion, but a cessation of growth. The treatment of arteriovenous malformations (AVMs) has been well established\textsuperscript{34} as having a high associated obliteration rate (approximately 80%) and a low complication rate (several percent). This effectiveness may be attributed to the fact that AVMs may be accurately circumscribed in orthogonal angiograms and respond gradually to single-fraction radiosurgery (compared to excision, which may result in abrupt haemodynamic changes or hemorrhaging).\textsuperscript{34}

1.3.2 Malignancies

Malignant tumours are often treated using stereotactic radiotherapy, with the same radiobiological rationale for fractionation as for conventional radiotherapy, but with
improved accuracy in dose delivery. Identification of the clinical tumour volume is often complicated by the presence of microscopic disease, and malignancy may extend into the region beyond the contour visible in the images.\textsuperscript{35} In addition, as the tumour volume increases, the single fraction isocomplication dose decreases, indicating the possible importance of dose fractionation for larger lesions.\textsuperscript{36} Radiosurgical treatment of brain metastases is common since they are often easily delineated and are frequently detected while still sufficiently small. They may be treated using either stereotactic radiosurgery or radiotherapy.\textsuperscript{37,38}

1.3.3 Functional disorders

As mentioned previously, treatments of functional disorders such as pain, trigeminal neuralgia, movement disorders and Parkinson’s disease were common during the early development of radiosurgery.\textsuperscript{4,13} Pituitary treatments were also performed because localization of this gland by radiography was readily achieved. More than 2500 radiosurgical pituitary treatments have been conducted since 1958.\textsuperscript{39,40,41,42} These procedures include pituitary suppression in the treatment of metastatic breast cancer and diabetic retinopathy, and treatment of Cushing’s disease, Nelson’s Syndrome, acromegaly and prolactin-secreting tumours. In North America, alternative medical procedures have since diminished the role of radiosurgery in this area.

1.4 Linear accelerator-based radiosurgery

Typically, the entire radiosurgical process, including lesion localization (imaging), treatment planning and radiation delivery can be completed in a single day. The various steps of the procedure, as outlined below, are usually performed by a team composed of a neurosurgeon, a radiation oncologist and a medical physicist.

1.4.1 Lesion localization

Lesion localization refers to the establishment of a coordinate system containing the patient’s head and the subsequent specification of the volume of the target in three dimensions. At the beginning of the radiosurgical procedure, a carbon fibre or metallic head-ring is rigidly attached to the patient’s skull using four sharp pins. The head-ring serves the functions of i) immobilizing the head during both imaging and treatment and
ii) facilitating attachment of a localizer box. The localizer box defines a Cartesian coordinate system by means of Z- or V-shaped fiducial marks that are clearly visualized in CT and MRI images. Multiple image sets from different modalities may be spatially co-registered using tools commonly available in radiosurgical planning software. As shown in Figure 1.2, a single axial image provides the lateral and anterior/posterior coordinates of a particular structure relative to the fiducial marks, while the superior/inferior coordinate of the slice may be determined from the separation between marks in the image. Angiography or digital subtraction angiography (DSA) is used for imaging AVMs, and three-dimensional localization is achieved by acquiring orthogonal views. Recently, functional imaging modalities have also been employed. Positron emission tomography (PET) and single photon emission computed tomography (SPECT) images have been used to localize lesions and to differentiate between neoplastic disease and non-active, visually abnormal tissues. Functional MRI (fMRI) has been employed to localize the motor and visual cortex with the aim of limiting the dose to these regions.

* Some localizer boxes and head-rings are not compatible with MRI. In such a case a magnetic resonance image set may be co-registered to a CT image set based on anatomical landmarks.
1.4.2 Treatment planning and dose calculation

Treatment planning begins by transferring the acquired image data to the treatment planning computer. The locations of fiducial marks are detected in at least one image set (usually CT) with an automated algorithm. After delineating the volume of the lesion in one or more image sets, an arrangement of beams is established with the goals of adequate dose coverage of the lesion and sparing of viable tissues and critical structures (e.g. optic chiasm or brainstem). The position of each beam is defined by a specific combination of gantry and couch angles, and a beam arc may be administered by rotating the gantry about the head while the couch is stationary. For conventional, circular-beam radiosurgery, the treatment plan variables include: beam or arc position, arc length, fractional dose delivered by each beam (beam weighting) and the diameter of each beam at isocentre.
For a given beam arrangement, the treatment planning software calculates the planned dose distribution within the brain based on previously-measured beam characteristics. These normally include\textsuperscript{47,48} the tissue-maximum ratio (TMR), percent depth dose (PDD), total scatter factor ($S_t$) and off-axis ratio (OAR).\textsuperscript{*} The majority of treatment planning systems determine the dose from a single beam at any arbitrary point $(x,y,z)$ with respect to the isocentre using:\textsuperscript{34,35}

$$D(x,y,z) = (MU) TMR(d, A) S_t(A) OAR(A, r_{xy}) \left( \frac{SAD}{SAD - z} \right)^2$$  \hspace{1cm} [1-1]$$

where $MU$ is the number of monitor units delivered, $d$ is the depth of the point $(x,y,z)$ below the surface, $A$ is the field size at isocentre, $SAD$ is the source-to-axis distance of the linear accelerator (usually 100 cm), $r_{xy}$ is the projected radial distance of the point $(x,y,z)$ from the beam axis at isocentre and positive $z$ is toward the beam source. The total dose $D_{arc}$ from a beam arc is usually calculated by summing the dose from $N$ fixed beams along the arc trajectory (where the angular interval between beams is typically\textsuperscript{62} 10 degrees):

$$D_{arc}(x,y,z) = \sum_{i=1}^{N} D_{arc,i}(x,y,z)$$  \hspace{1cm} [1-2]$$

The dose contribution from the $i^{th}$ arc segment is determined using:

$$D_{arc,i}(x,y,z) = \left( \frac{MU_T}{N} \right) TMR(d_i, A) S_t(A) OAR(A, r_{xy,i}) \left( \frac{SAD}{SAD - z_i} \right)^2$$  \hspace{1cm} [1-3]$$

where the $MU_T$ is the total number of monitor units for the arc:

$$MU_T = \frac{D_{iso} \cdot N}{S_t(A) \sum_{i=1}^{N} TMR(d_i, A)}$$  \hspace{1cm} [1-4]$$

and $D_{iso}$ is the dose prescribed to the isocentre for the arc.

\textsuperscript{*} Appendix A provides definitions of the relevant dose functions used in this calculation.
Rice et al.\textsuperscript{48} have demonstrated that, for circularly-collimated 6 MV radiosurgical beams, the dose calculated using equation [1-1] or [1-3] is usually accurate to within 3% without corrections for surface obliquity or variations in tissue density.*

1.4.3 Treatment delivery

The treatment planning software provides the required accelerator parameters, including collimator sizes, MU settings, couch and gantry positions for each beam and a set of printed target positioner films. These films attach to the outer walls of a target positioner box, which is mounted to the headring (Figure 1.3) to facilitate alignment of the patient with the linac isocentre using orthogonal wall-mounted lasers. Once in position, the patient is treated, and the delivery of radiation (e.g. five beams or beam arcs) requires approximately 20 minutes.

* provided that the beam does not pass through air cavities below the skin surface.
1.5 Conformal stereotactic radiosurgery

The intersection of multiple circularly-collimated beams or beam arcs produces a dose distribution that is roughly spherical. While some shaping of the distribution may be achieved by using asymmetric beam arrangements, it is mainly the lower dose levels that are affected, and the high-dose region remains approximately ellipsoidal.

A spherical or ellipsoidal distribution is spatially incongruent with the volume of an irregularly-shaped lesion. Adequate dose coverage of the target therefore necessarily involves delivering a high dose to the surrounding healthy tissue. This becomes particularly important for larger lesions (i.e. > ~ 2 cm across) since the volume of viable tissue contained within the ellipsoidal distribution becomes significant. In the past, improved conformity has been achieved by using multiple-isocentre techniques, whereby several spherical distributions are overlapped to cover the irregular target volume. However, this approach introduces significant dose inhomogeneities and the presence of resultant dose hot-spots has been shown to correlate with increased frequency of complications ranging from minor neurologic deficits to severe focal deficits, in some cases leading to seizure, coma and death.\(^\text{49}\)

In order to improve conformity while preserving dose homogeneity, several techniques have been developed to produce photon beams that are customized according to the shape of the lesion. Hacker \textit{et al.}\(^\text{50,51}\) have used the four independent jaws of the linear accelerator to produce different rectangular fields for each arc of a fractionated treatment. Leavitt \textit{et al.}\(^\text{52}\) have designed cam-operated rectangular trimmer blocks upstream from conventional circular collimators to create asymmetric fields. Serago \textit{et al.}\(^\text{53}\) have used elliptically-shaped collimator openings to cover the maximum extent of the lesion throughout a beam arc. Bourland and McCollough\(^\text{54}\) and Bourque\(^\text{55}\) describe the use of static conformal beams produced using apertures matching the beam's-eye-view projection of the lesion. These collimators are similar to those used in this present work and are manufactured by pouring a low melting-point alloy into a customized mould. Recently,\(^\text{56,57,58,59}\) micro-multileaf collimators (MLCs) have become available commercially and consist of an array of narrow, opposed tungsten leaves that can be retracted independently to define an irregular field shape. Although micro-MLCs are expensive, they offer the practical advantage of computer automation of field shaping and
also may facilitate dynamic field shaping whereby the field shape is varied continuously throughout a beam arc.\textsuperscript{60}

Calculation of the dose delivered by shaped beams may be done by using Clarkson integration\textsuperscript{61} to determine irregular field beam parameters based on circular-field commissioning data. As shown in Figure 1.4, the field is divided into \(N\) segments, whereby the lengths of segments along the perimeter are equal (i.e. \(\theta\) is not fixed). Equation [1-1] may be modified\textsuperscript{62} by using an estimated value of scatter factor for the irregular field \((S_{\theta}(\text{irreg}))\), and interpolated values of both off-axis ratio \((OAR_p)\) and tissue-maximum ratio \((TMR_p)\). The dose at point \(P\) having coordinates \((x,y,z)\) is given by:

\[
D_{\text{irreg}}(x,y,z) = MU_j TMR_p S_{\theta}(\text{irreg}) OAR_p \left( \frac{SAD}{SAD - z} \right)^2
\]  

[1-5]

where the Clarkson integration is used to calculate \(S_{\theta}(\text{irreg})\):

\[
S_{\theta}(\text{irreg}) = \sum_{i=1}^{N} S_{\theta}(2r_i) \frac{A_i}{A_t}
\]  

[1-6]

Here, \(S_{\theta}(2r_i)\) is the scatter factor for a circular field of diameter \(2r_i\) and \(r_i\) is the length of the \(i^{th}\) ray from the isocentre location \((O)\) to the edge of the field. \(A_i\) is the area of the \(i^{th}\) segment, and \(A_t\) is the total area of the irregular shape.

The TMR at point \(P\) is calculated using the interpolation:

\[
TMR_p = \frac{\alpha}{\theta} \left( TMR_p - TMR_{p'} \right) + TMR_{p'}
\]  

[1-7]

where

\[
TMR_p = \left( \frac{O_P}{OA} \left( TMR_A - TMR_O \right) \right) + TMR_O
\]  

[1-8]

\[
TMR_{p'} = \left( \frac{O_{P'}}{OB} \left( TMR_B - TMR_O \right) \right) + TMR_O
\]  

[1-9]
TMRA, for example, is the TMR for a field of radius OA and the depth at the location of point A. Note that points A, B, and O may be at different depths due to surface obliquity. A similar method is used to calculate the interpolated off-axis ratio, OAR_p.

![Diagram](image)

*Figure 1.4 Partitioning of an irregular field into segments for the Clarkson integration.*

1.6 **Accuracy and sources of error in radiosurgery**

1.6.1 Suggested accuracy

According to the most recent report on stereotactic radiosurgery by the American Association of Physicists in Medicine,\(^1\) the desired spatial and numerical accuracy in radiosurgical dose delivery are ± 1 mm and 5%, respectively. Achieving this accuracy involves minimizing the contributions of several sources of uncertainty incurred in lesion localization, treatment planning and in treatment delivery.

1.6.2 Localization

In both CT and MR imaging, the accuracy of localization in radiosurgery is commonly cited as ± (in-plane pixel size) and ± (slice thickness).\(^63\) For CT, the in-plane pixel size is usually 0.7 × 0.7 mm\(^2\) and the slice separation is typically between 1.0-3.0
mm. For MRI, the pixel size ranges from 0.5 x 0.5 mm² to 2 x 2 mm² and the slice thickness ranges from 1.0 to 4.0 mm. The voxel size provides only an estimate of the optimal localization uncertainty, however, and image distortions will further limit localization accuracy. This has been particularly important for MRI, for which magnetic field inhomogeneities may increase localization error.⁶⁴,⁶⁵ Hence, MRI has been found to provide superior diagnostic information (with improved tissue discrimination) but reduced geometric fidelity in comparison with CT imaging. Recently Novotný et al.⁶⁶ have shown that uncertainty in MRI may lead to errors in the calculated volume of lesions based on the image set of up to 13.1%, but this uncertainty remains below the uncertainty in contouring of lesions. Alexander et al.⁶⁵ have routinely found systematic error in MRI stereotactic coordinates with a median of 4 mm compared to computerized tomography (CT) coordinates. The high spatial resolution of angiography and DSA has permitted localization of AVMs with an uncertainty ranging from 0.3 mm⁶⁷ to 1.0 mm.²⁷ However, even if imaging resolution does not introduce significant uncertainty, tissues in the brain may shift by up to 1.0 mm⁶⁸ between imaging and treatment. The possible uncertainties inherent in target localization are usually included as a spatial margin included in the planning target volume (PTV).

1.6.3 Treatment delivery

There are several sources of uncertainty incurred during alignment of the patient prior to treatment and in delivering the radiation. First, flexion of the head-ring contributes an uncertainty of approximately 0.6 mm.¹ Second, using the wall-mounted lasers in the treatment room, the uncertainty in aligning the patient according to the location of the isocentre is approximately 1.0 mm to 1.5 mm. Third, the isocentre is not a fixed point in space, and in fact the centre of the radiation field may vary within a sphere typically 0.2 mm⁶⁷ to 1.0 mm²³ in diameter, over the range of motion of the couch and gantry.

1.6.4 Treatment planning

While simple dose calculation algorithms (e.g. equations [1-1] and [1-5]) have been proven accurate for simple geometries and circular fields,⁶⁸ errors in analytically-calculated dose distributions may be incurred for irregular beam shapes,⁶⁹ geometries
with tissue inhomogeneities (e.g. air cavities\textsuperscript{70}) or significant surface obliquity. The nature and magnitude of these errors depend highly on the particular treatment planning system used and on the patient geometry. Where possible, these errors are quantified through comparison with direct measurement of absorbed dose.

### 1.6.5 Combination of uncertainties

It has been suggested\textsuperscript{1} that the individual sources of spatial uncertainty add in quadrature. For example, if localization is performed using both CT and angiography, the net uncertainty is given by

\[
\delta_{\text{NET}} = \left( \delta_{\text{CT}}^2 + \delta_{\text{angio}}^2 + \delta_{\text{fix}}^2 + \delta_{\text{setup}}^2 + \delta_{\text{iso}}^2 + \delta_{\text{tissue}}^2 \right)^{\frac{1}{2}}
\]

where \( \delta_{\text{CT}} \), \( \delta_{\text{angio}} \), \( \delta_{\text{fix}} \), \( \delta_{\text{setup}} \), \( \delta_{\text{iso}} \) and \( \delta_{\text{tissue}} \) are the uncertainties due to CT imaging, angiography, patient fixation, patient setup, isocentric alignment and tissue motion, respectively.

Table 1.1 gives examples of optimal, typical and worst-case combinations of uncertainties, for the ranges cited above. These examples assume MRI is not used as the reference image set for localization and no error is incurred in treatment planning (i.e. dose calculation). The “worst-case” corresponds to a 3 mm CT slice separation and use of a re-locatable fixation system such as a thermoplastic mask. Note that in this case, while the individual sources are within the tolerances cited in the literature, the combined uncertainty becomes comparable to that in conventional radiotherapy\textsuperscript{71} (i.e. without stereotactic localization).
Table 1.1 Combination of uncertainties in stereotactic radiosurgery. Values are indicated in units of ± mm.

<table>
<thead>
<tr>
<th></th>
<th>optimal case</th>
<th>typical case</th>
<th>worst case</th>
</tr>
</thead>
<tbody>
<tr>
<td>CT</td>
<td>1.3</td>
<td>1.7</td>
<td>3.3</td>
</tr>
<tr>
<td>angiography</td>
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<td>0.5</td>
<td>1</td>
</tr>
<tr>
<td>fixation</td>
<td>0.6</td>
<td>0.6</td>
<td>2</td>
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<tr>
<td>setup</td>
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<td>1</td>
<td>1.5</td>
</tr>
<tr>
<td>tissue motion</td>
<td>0</td>
<td>0.5</td>
<td>1</td>
</tr>
<tr>
<td><strong>total</strong></td>
<td><strong>1.8</strong></td>
<td><strong>2.6</strong></td>
<td><strong>4.6</strong></td>
</tr>
</tbody>
</table>

* 0.6 x 0.6 x 1.0 mm³ voxel size (value indicated is RMS uncertainty), head-ring fixation
† 1.0 x 1.0 x 1.0 mm³ voxel size, head-ring fixation
‡ 1.0 x 1.0 x 3.0 mm³ voxel size, repositionable fixation (e.g. thermoplastic mask system)
2 Dosimetry in stereotactic radiosurgery

2.1 Basic concepts

2.1.1 Dose and kerma

Absorbed dose, or simply dose, is defined as the mean energy, $dE$ deposited by ionizing radiation in a medium per unit mass, $dm$:

$$D = \frac{dE}{dm} \quad \text{[2-1]}$$

The SI unit for dose is the gray (Gy), where 1 Gy = 1 J/kg. For x-ray beams, the dose is deposited in the medium by electrons or positrons set into motion as a result of photon interactions including photoelectric absorption, Compton scattering, pair production and triplet production. For high-energy photon beams, the dose is not administered at the point of photon interaction, but along the trajectories of charged particles set into motion as they ionize and excite atomic electrons.

The sum of the initial energies of all charged particles set into motion at the point of photon interaction per unit mass is called the kerma (kinetic energy released to matter), $K$:

$$K = \frac{dE_{tr}}{dm} \quad \text{[2-2]}$$

where $dE_{tr}$ is the energy transferred per unit mass $dm$ of the medium. Note that kerma is related to the energy fluence of the photon beam $\psi$ and the mass energy transfer coefficient $\bar{\mu}_{tr} / \rho$ for the medium averaged over the photon spectrum:

$$K = \psi \left( \frac{\bar{\mu}_{tr}}{\rho} \right) \quad \text{[2-3]}$$

The energy of charged particle is expended through both collisional and radiative losses, and thus the total kerma has two components accordingly:
\[ K = K_{\text{col}} + K_{\text{rad}} \]  \[2-4\]

where

\[ K_{\text{col}} = \Psi\left(\frac{\mu_e}{\rho}\right)(1 - g) = \Psi\left(\frac{\mu_{\text{ab}}}{\rho}\right) \]  \[2-5\]

and

\[ K_{\text{rad}} = \Psi\left(\frac{\mu_e}{\rho}\right)g = \Psi\left(\frac{\mu_{\text{ab}}}{\rho}\right)\frac{g}{1 - g} \]  \[2-6\]

Here, \( g \) is the average fraction of the electron’s kinetic energy lost through bremsstrahlung radiation. For electrons set into motion in tissue by photons of energies up to 20 MeV, this fraction is small, and \( g \) does not exceed 0.073. \( \mu_{\text{ab}}/\rho \) is the mass absorption coefficient for the medium averaged over the photon spectrum.

2.1.2 Electronic equilibrium

The relationship between dose and kerma in any small volume of medium depends on the degree to which electronic equilibrium exists in that volume. The net amount of energy within a small volume depends on the difference between the energy transferred into the volume and that taken out by both photons and charged particles (i.e. electrons):

\[ E = \sum E_{\text{in}}^{\text{photons}} - \sum E_{\text{out}}^{\text{photons}} + \sum E_{\text{in}}^{\text{cp}} - \sum E_{\text{out}}^{\text{cp}} \]  \[2-7\]

Electronic equilibrium is a condition whereby the energy deposited by electrons within a volume equals the energy removed by electrons, i.e.:

\[ \sum E_{\text{in}}^{\text{cp}} = \sum E_{\text{out}}^{\text{cp}} \]  \[2-8\]
When a megavoltage photon beam is incident upon a slab of medium (water, for example), the condition of electronic equilibrium is not established at the surface due to the fact that a gradual build up of electrons occurs. For example, consider Figure 2.1a) which shows a photon beam incident upon a volume of water divided into several slabs along the depth dimension. In each slab, the photon beam sets two recoil electrons into motion, and each electron has an average range of four slabs. For high-energy beams, recoil electrons will emerge primarily in the forward direction,\(^2\) as shown. The absorbed energy in each slab is proportional to the amount of ionization and therefore the sum of the lengths of electron tracks crossing the slab. If a single electron track crossing one slab is assigned an arbitrary dose value of 1, note that in the first slab there is an absorbed dose of 1, and in the second slab the dose is 3 (i.e. two tracks + two half-tracks). As tabulated, the dose builds up over a region of 4 slabs until it is constant with depth. Collisional kerma \((K_c)\), in contrast, is constant with depth and is equal to 8 for all slabs. Also tabulated in the figure are the energies transferred into \((E_{in})\) and carried out of \((E_{out})\) each slab by electrons. Note that when complete buildup of dose has been achieved, \(E_{in}\) is equal to \(E_{out}\). This simplified example thus demonstrates three phenomena:

i) The dose builds up over a certain range of depth (called the buildup region) and reaches a maximum value at a certain depth called \(d_{max}\), where \(d_{max}\) is approximately equal to the electron range;

ii) For depths > \(d_{max}\), \(D=K_{col}\); and

iii) For depths > \(d_{max}\), \(E_{in}=E_{out}\), and electronic equilibrium is established.
Figure 2.1  a) Diagram showing buildup of dose and establishment of electronic equilibrium within a volume of water a) neglecting and b) including attenuation of the photon beam. In b), transient electronic equilibrium exists at depths > $d_{\text{max}}$. $E_{\text{in}}$ is the kinetic energy of the electron as it enters a particular slab and $E_{\text{out}}$ is the energy of an electron as it leaves a slab (which is subsequently deposited in regions downstream).
The example in Figure 2.1a) is simplistic in that it neglects the attenuation of the photon beam. If the fluence of photons decreases as a function of depth, the relationship between dose and kerma resembles that shown in Figure 2.1b). In this case, for depths greater than $d_{\text{max}}$, $D = \beta K_{\text{col}}$, where $\beta$ is a constant usually slightly greater than one. This condition is termed "transient electronic equilibrium." 

### 2.2 Radiosurgical photon beam dosimetry

#### 2.2.1 Dosimeter requirements for stereotactic radiosurgery

A dosimeter is a detector used for the measurement of dose absorbed by irradiated media. With the exception of calorimetry, no dosimeter measures absolute dose directly. Instead, other quantities such as ionization or chemical change are related to absolute dose through calibration. A suitable dosimeter, for radiotherapy applications in general, provides the capacity to measure absolute dose with an associated uncertainty significantly less than the uncertainty allowed in treatment delivery. The ideal dosimeter would also provide a wide dynamic dose range, tissue equivalence in terms of atomic number and electron density, insensitivity to environmental factors and no significant dependencies on beam parameters such as type of radiation (e.g. photons, electrons or protons), energy or dose rate. For dose verification in stereotactic radiosurgery, three additional requirements become important:

1. **very high (preferably sub-millimetre) spatial resolution in order to measure inaccuracies exceeding the tolerance specified in radiosurgery (± 1mm);**
2. **the ability to integrate dose from several beams or beam arcs to measure the distribution resulting from an entire treatment; and**
3. **the capacity to provide a spatial distribution of dose in two, or preferably three dimensions to provide a complete map of the measured dose, particularly for verification of irregular, conformal distributions.**
As described below, the high-resolution measurement of the spatial distribution of absorbed dose in radiosurgery is complicated mainly by two beam characteristics: lateral electronic disequilibrium and high dose gradients.

2.2.2 Lateral electronic disequilibrium

A unique feature of radiosurgical photon beams is the presence of lateral electronic disequilibrium across a significant proportion of the beam profile. This effect is similar to that occurring in the buildup region in Figure 2.1a), except that it results in a loss of electronic equilibrium in the lateral direction (perpendicular to the beam axis). As shown in Figure 2.2, the velocities of recoil electrons set into motion may have an initial lateral component and also tend to disperse laterally due to scattering processes. Electrons near the edge of the beam therefore deposit energy at locations beyond the geometric photon beam boundary, as shown for point A. The dose at points inside the photon beam but near the edge are also affected—compared to point C on the central axis, for example, at point B there is a lack of scatter from the right, resulting in a decrease of dose. The lateral mismatch between kerma and dose, as plotted in Figure 2.2 thus produces a broadening of the beam penumbra, and this effect becomes more pronounced for higher beam qualities.* In general, transient electronic equilibrium does not exist at points closer to the photon beam edge than the mean electron range. This means that for smaller radiosurgical beams (< 3 cm diameter for 6 MV), even the dose on the central axis may be reduced due to this effect.

In regions of electronic disequilibrium, conversion of directly-measured quantities (such as collected charge from an ionization chamber) to absolute dose becomes difficult. As explained by Khan,\textsuperscript{74} several factors are required to convert ionization to dose, and calculation of these factors (e.g. ratios of averaged stopping powers) requires knowledge of the electron spectrum,\textsuperscript{75} which is challenging in regions of electronic disequilibrium. The conversion factors are therefore typically only tabulated for transient

\* Although disequilibrium widens the penumbra, for 6-10 MV radiosurgical photon beams the dose fall-off is nevertheless very sharp, and the dose gradient in the penumbral region is greater than 20% per millimetre.
Figure 2.2 Schematic diagram of lateral electronic disequilibrium. At a single depth in phantom, the dose at point C is deposited equally by electrons from the right and left sides. Point B, close to the photon beam edge, is missing contributions from the right and therefore the dose falls below the kerma inside the geometric boundary of the photon beam. Point A is outside the beam but receives dose from electrons originating inside.

electronic equilibrium conditions. For ionization chambers, calculation of dose also requires a gradient correction* in regions where $D\neq K_{coh}$, since the electron fluence will be greater on one side of the chamber (e.g. that closer to the central axis in Figure 2.2) than on the other side. This factor has been calculated for regions of transient electronic equilibrium but becomes more significant and increasingly difficult to determine.

* In the notation of the AAPM TG-21 protocol, this factor is called $P_{rep}$. 

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accurately in regions of pronounced electronic disequilibrium and as chamber volume increases.

2.2.3 High dose gradients and volume averaging

Dose gradients (change in dose per unit length) are significant in the inherent profile of a single radiosurgical beam and also in most dose distributions created by multiple, intersecting beams. This introduces a challenge in measuring the dose across the dose fall-off region, since the detector must be sufficiently small to avoid volume averaging effects. As discussed recently by Charland\textsuperscript{77} and others,\textsuperscript{78} the measured profile, $P_{\text{measured}}$, will be blurred according to the convolution of the true dose profile $P_{\text{true}}$ and the line-spread function (LSF) of the detector:

$$P_{\text{measured}}(x) = P_{\text{true}}(x) \otimes \text{LSF}(x)$$  \[2-9\]

Therefore, if the volume of the detector is too large, it will act as a low-pass filter and the dose fall-off region will be broadened or distorted. The requirement of a high-resolution detector limits the selection of appropriate dosimeters, but several have been employed to date as discussed in the next section.

2.3 Summary of dosimeters used for radiosurgery

2.3.1 Point-dose measurement

Although ionization chambers are the most common dosimeters for conventional radiotherapy applications, most are inappropriate for beam measurement in radiosurgery because of their large active volumes. However, micro- and mini-ionization chambers\textsuperscript{79} with sensitive diameters of less than 2 mm have been used to measure beam profiles\textsuperscript{77} by sweeping them along a line perpendicular to the beam axis. Measurement of absolute dose in the presence of electronic disequilibrium remains difficult, however, since the air cavity augments lateral disequilibrium.\textsuperscript{80}

The use of silicon diodes is more common in radiosurgery, particularly for relative dose measurements, and various studies are available assessing their
The diode dosimeter consists of a p-type/n-type semiconductor junction, with an active volume typically greater than 60-130 μm in thickness and 2-3 mm in diameter. The diode itself may be encased in metal or potted in epoxy. Diodes for photon beam dosimetry may also have a backing of tungsten/epoxy to shield low-energy backscatter, thereby reducing the over-response of the detector to low energy photons. Dawson et al. have indicated that diode-measured dose profiles in radiosurgery differ from the “true” profiles by as much as 2 mm. In their model, this disparity results from the combination of two competing effects: i) broadening of the penumbra due to volume averaging effects and ii) increased attenuation electron fluence by the diode composition. Beddar et al. have demonstrated that if the diode is placed perpendicular to the beam axis (so the sub-millimetre dimension is seen by the photon beam) the shape of the measured profile is erroneously asymmetric and depends on the diode orientation relative to the direction of travel. In the parallel orientation, the profile is symmetric but volume averaging becomes significant. Diodes also suffer from temperature and dose-rate dependencies.

Diamond dosimeters are relatively novel detectors with dimensions similar to diodes, but offer the advantage of approximate tissue equivalence. Rustgi et al. have demonstrated that diamond-measured dose profiles are comparable to those measured with diodes. Diamond detectors also suffer from a dose-rate dependence, but in contrast to diodes, offer isotropic radiation sensitivity.

Plastic scintillation detectors may also be used for measuring point-doses or one-dimensional distributions since the scintillator light output varies linearly with absorbed dose. Light output may be measured by coupling the scintillator to a photomultiplier tube (PMT). The scintillator is typically 1 to 2.5 mm in diameter and several millimetres in length, and therefore in one orientation, high-resolution dosimetry is possible. Scintillators also offer approximate tissue equivalence and dose-rate independence. Although Cerenkov light produced in the coupling fibre contributes to background signal, for photon beams this effect is small and may be subtracted using a reference fibre.

Thermoluminescent dosimeters (TLDs) have been used for radiosurgical dosimetry either individually to obtain point doses or in an array. TLDs are usually composed of lithium fluoride doped with magnesium and titanium and are available as
small ribbons or rods. The thickness of TLD crystals may be as low as 0.14 mm and therefore high-resolution dose measurement is possible, at least on a point-by-point basis. TLD dosimetry is somewhat time-consuming and complicated, however, since each crystal must be calibrated individually, has its own thermal history and must be annealed separately prior to irradiation and again for dose readout. Despite the complexity of the process, Ertl et al. have found that TLD dosimetry is effective for radiosurgical profile measurements with an uncertainty of 2%.

2.3.2 Two-dimensional dose measurement

Radiographic film is a recommended method for measurement of dose profiles in radiosurgery since it has an inherent spatial resolution that is among the highest of all available dosimeters (typically up to 20 line pairs per millimetre). In practice, the spatial resolution is limited by that achievable in digitizing the film, and thus sub-millimetre resolution is possible. Therefore, film potentially eliminates the problem of volume averaging discussed in section 2.2.3. For this reason, radiosurgical dose profiles recorded using novel dosimeters are often compared to film measurements in benchmark studies. Film provides the capacity for high spatial-resolution measurement of the two-dimensional distribution of integrated dose, thus satisfying the three criteria for radiosurgical dose verification as discussed in section 2.2.1. In addition, film is inexpensive and conventional infrared or broadband densitometers or scanners may be used for readout. However, its application to large-field (i.e. non-radiosurgical) megavoltage photon beam dosimetry has been significantly limited by several factors including dependencies of the measured dose on depth in phantom, field size and on chemical processing conditions. In practice, this has made accurate dose calibration complex and time-consuming. A discussion of the details of radiographic film dosimetry and its limitations are given in section 2.4 below.

Radiochromic film is a relatively new dosimeter and consists of a mylar base coated with thin (7 μm) layers of polydiacetylene polymer emulsion. The emulsion turns deep blue when irradiated, and the recorded optical density has been shown to be linear with dose. The inherent spatial resolution of radiochromic film is better than 60 line pairs/mm, and as for radiographic film, it is ultimately limited by the densitometer
Radiochromic film is approximately tissue equivalent in terms of electron stopping power and mass absorption coefficient. No chemical processing is required, and the optical density in exposed areas stabilizes within approximately 24 hours. The film is available in a format providing a wide dose range (3-100 Gy). However, this dosimeter suffers from several drawbacks. The main reported problem is a random nonuniformity of up to 15% in the image. Secondly, to take advantage of the full dynamic dose range, it must be digitized using a laser densitometer with a wavelength closely matched to the absorption peak of the film. Helium/neon microdensitometers are suitable for this application, but are expensive and not as widely available in radiotherapy clinics as conventional densitometers with infrared-based or broadband light sources. Thirdly, the film is very expensive and available only in small sheets. Several reports are available on the use of radiochromic film for radiosurgical dosimetry. Bjarngard et al. have demonstrated that the high spatial resolution permits accurate dose measurement in the presence of electronic disequilibrium. McLaughlin et al. describe radiochromic dosimetry for Gamma Knife radiosurgery. Ramani et al. describe the verification of dynamic radiosurgical dose distributions.

2.3.3 Three-dimensional dose measurement

Reports of three-dimensional radiosurgical dosimetry are sparse, mainly due to the lack of a practical dosimeter capable of providing accurate, high-resolution, volumetric maps of integrated dose. Chan and Ayyangar have described the use of ferrous sulfate fixed in aqueous gel for the measurement of radiosurgical dose distributions. In this method, Fe$^{2+}$ ions in the gel are oxidized to Fe$^{3+}$ upon irradiation and the concentration of the latter is determined by measuring the longitudinal relaxation ($T_1$) and transverse relaxation ($T_2$) times using MRI. Although this approach provides three-dimensional dose mapping and near tissue equivalence, accuracy was limited by dose averaging within the large voxel volume ($1.2 \times 1.2 \times 7$ mm$^3$) and MRI artifacts. Moreover, the ions

* GafChromic MD-55-2, International Specialty Products, Wayne, NJ, USA.
† MD-55-2 is available in 5" x 5" sheets. At the time of writing, the cost per sheet is approximately CDN$50.
in the gel diffuse over time, precluding analysis of the irradiated gel after an hour or two.\textsuperscript{96}

Both Ibbott et al.\textsuperscript{97} and Meeks et al.\textsuperscript{98} have described the use of polyacrylamide (PAG) gels for conformal radiosurgical dose verification. This dosimeter consists of a matrix of acrylic monomers in a tissue-equivalent gelatin, and irradiation induces chain polymerization. The distribution of the polymer may be determined by MRI whereby maps of $R_2 = T_2^{-1}$ have been shown to be proportional to dose.\textsuperscript{99} Although these studies show reasonable results, the authors demonstrate that for practical radiosurgical dosimetry the technique is limited in terms of spatial resolution due to the MRI voxel size. It was also indicated\textsuperscript{100} that the signal-to-noise ratio decreases with smaller voxel sizes, necessitating spatial smoothing of the dose distribution and thus ultimate loss of resolution.

2.4 Radiographic film dosimetry

2.4.1 Film structure and image formation

As shown in Figure 2.3, dual-emulsion radiographic film consists of silver halide emulsion bonded to two sides of a polyester base substrate. The emulsion is protected by a gelatin supercoat that prevents damage of the emulsion during processing. Shown in parentheses are the thicknesses of the emulsion and base layers for Kodak X-Omat V, a typical film used in dose verification. Other relevant physical properties\textsuperscript{*} are given in Table 2.1, as provided by Yeo et al.\textsuperscript{101}

The emulsion consists of silver halide suspended in a gelatin layer and is usually 90% to 99% AgBr and 1% to 10% AgI. Ag\textsuperscript{+}, Br\textsuperscript{−} and I\textsuperscript{−} ions within a single emulsion grain form a crystal lattice. It is believed that the presence of iodine ions, in small quantities, creates point defects in the crystal, whereby Ag\textsuperscript{+} ions are displaced from their positions in the lattice. This creates a population of interstitial Ag\textsuperscript{+} ions that are mobile. The I\textsuperscript{−} ions may also create line defects, whereby a plane in the lattice is shifted or distorted. These defects serve to sensitize the crystal lattice. Additional sensitization is

\textsuperscript{*} The density of emulsion (including silver halide and gelatin) corresponds to the chemical composition listed in the Methods and Materials section on page 64.
achieved by adding a sulfur-containing compound such as allythiourea, which combines with silver halide to form silver sulfide. Crystal point defects, line defects and AgS molecules are referred to as sensitivity specks.

Figure 2.3 Cross section of a dual-emulsion radiographic film (not to scale).

Table 2.1 Physical properties of Kodak X-Omat Verification film.

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Crystal structure</td>
<td>face-centred</td>
</tr>
<tr>
<td>AgBr thickness</td>
<td>0.0010654 mm</td>
</tr>
<tr>
<td>Emulsion thickness</td>
<td>0.0127 mm</td>
</tr>
<tr>
<td>Base thickness</td>
<td>0.1778 mm</td>
</tr>
<tr>
<td>Average grain size</td>
<td>0.6 μm</td>
</tr>
<tr>
<td>AgBr density</td>
<td>6.5266 g/cm³</td>
</tr>
<tr>
<td>Emulsion density (silver halide + gelatin)</td>
<td>2.2 g/cm³</td>
</tr>
<tr>
<td>Surface density</td>
<td>0.69535 mg/cm²</td>
</tr>
</tbody>
</table>

As shown in Figure 2.4, when energy is transferred to the emulsion, for example via photoelectric absorption of a photon, an electron is liberated from the bromine ion. This electron travels through the crystal until it is trapped at the location of a sensitivity speck. With a negative charge, the sensitivity speck then attracts interstitial Ag⁺ ions, which combine with the electron to form an uncharged silver atom. The uncharged Br atoms are then taken up by the emulsion. This process repeats as more photons are absorbed by the emulsion, and a complex of multiple Ag atoms is formed at the site of
the sensitivity speck. According to the Gurney-Mott hypothesis, at least two atoms constitute a latent image formed in the emulsion.

For low-energy photons (e.g. in the visible spectrum), the latent image formation proceeds as depicted in Figure 2.4, but the situation is somewhat different for x-ray photons. In this case, a single high-energy Compton recoil electron may liberate electrons from multiple bromine ions, possibly in different emulsion grains. Only 3% to 10% of the energy transferred by the photon is expended in this fashion, however, and most is lost in ionizing or exciting atoms in the gelatin.

It is believed that the presence of the multiple Ag atom complex acts as a catalyst in the chemical development process. Development involves the reduction of additional Ag$^+$ ions in the grain to form elemental silver. More specifically:

$$2 \text{AgBr} + \text{H}_2\text{Q} + \text{NaSO}_3 \rightarrow 2 \text{Ag} + \text{HBr} + \text{HQSO}_3\text{Na} + \text{NaBr} \quad [2-10]$$

where $\text{H}_2\text{Q}$ is the developer (hydroquinone), $\text{Na}_2\text{SO}_3$ is sodium sulfite (a preservative and clarifying agent), $\text{HBr}$ is hydrobromic acid, $\text{HQSO}_3\text{Na}$ is hydroquinonemonosulfonate.
and NaBr is sodium bromide. The goal of the development process is a differential reduction between latent film grains (containing Ag complexes) and non-latent grains (which will also be reduced but at a much lower rate). Within a single grain, development is an all-or-none process, and developed grains appear black.

2.4.2 Radiographic film dosimetry

For dosimetric applications, the film is positioned within a volume of the medium (commonly a phantom of water or water-equivalent plastic) in which the dose is to be determined. The film must be shielded from both water and room light, so usually it is pressed between layers of tissue-equivalent plastic and sealed. It is also necessary to eliminate air gaps proximal to the film since these may cause significant aberrations in the recorded optical density. After developing the exposed film, the pattern of optical density is usually recorded using a scanner or scanning densitometer which measures the transmission of light through the film. As indicated in section 2.3.2, the light source used may be broadband (i.e. white light), infrared, or confined to a band of wavelengths (e.g. HeNe laser). Traditionally the measured quantity is optical density (OD), given by the logarithm of the ratio of incident \( I_0 \) and transmitted \( I_t \) light intensities:

\[
OD = \log_{10} \left( \frac{I_0}{I_t} \right) \quad [2-11]
\]

The optical density is then related to a dose by using an established calibration, called the sensitometric curve. This curve must be specific to the batch of film used and is determined by measuring optical densities resulting from a number of known doses. The relationship between OD and dose may be linear or non-linear depending on the film type and dose range used, and eventually saturates when all grains are rendered latent through the process shown in Figure 2.4. It should be noted that quantities other than true OD may be measured (e.g. gray-level pixel value from a Charge Coupled Device (CCD)-based scanner) and the sensitometric curve may also be affected by the response of the scanner itself.
2.4.3 Radiographic film: dosimetric considerations

As mentioned earlier, radiographic film offers technical advantages of high spatial resolution, two-dimensional measurement, integration of dose, acceptable uniformity and good signal-to-noise ratio even for short exposures. The practical advantages of film include permanence of the recorded optical density, accessibility, low cost (a few dollars per sheet), and the capacity to be digitized using a wide array of scanners or densitometers.

Although these advantages make film an attractive candidate particularly for high-resolution dosimetry in radiosurgery, most of the previous experience with film dosimetry has focused on measurement of dose created by larger (non-radiosurgical) photon beams. For large photon fields, several undesirable effects limit the accuracy of quantitative dose measurement. The most important of these arise from the fact that the physical properties of silver halide in emulsion differ significantly from those of water. The effective atomic number \( Z^* \) of emulsion (including both gelatin and silver halide) is approximately 34.7\(^*\) compared to 7.51 for water. This disparity results in a difference between emulsion and water in terms of both electronic stopping power (Figure 2.5) and mass attenuation coefficient \( \mu/\rho \) (shown in Figure 2.6).

Although film is not an optimal dosimeter (failing the criterion of tissue-equivalence discussed in section 2.2.1), accurate dosimetry is not precluded by a difference in the dose absorbed by film and that absorbed by water, so long as the film does not significantly perturb the dose distribution in the surrounding medium. The disparity between the dose deposited in emulsion and water can be accounted for by the sensitometric curve established during calibration. What is more crucial is that this calibration remains consistent over a range of measurement conditions, particularly beam energy. Since the electronic stopping power changes relatively gradually with energy (less than 10% over three decades of energy), accurate film dosimetry is complicated primarily by variation in \( \mu/\rho \) relative to water with changes in the photon spectrum. Spectral changes below 400 keV are especially problematic, since in this region, \( (\mu/\rho)_{\text{emulsion}} \) diverges rapidly from \( (\mu/\rho)_{\text{water}} \) due to the approximate \( Z^3 \)

\( \text{calculated using the composition given on page 64.} \)
Figure 2.5 Mass stopping power normalized to water for electrons in radiographic film emulsion and base.

Figure 2.6 Mass attenuation coefficient ($\mu/\rho$) for radiographic film emulsion and base normalized to that of water, shown as a function of photon energy.
dependence of the probability of photoelectric absorption. In this low-energy end of the spectrum \((\mu/\rho)_{\text{emulsion}}\) increases approximately as \(1/(h\nu)^3\), where \(h\nu\) is the photon energy. In the context of practical film dosimetry, this introduces several complications.

First, for larger (i.e. non-radiosurgical) photon beams, the variation of the photon spectrum with depth in phantom causes a depth-dependence of the measured sensitometric calibration curve. The total photon spectrum can be separated into two components, and each varies differently with depth. The primary component, composed of photons that have not yet interacted in the phantom, is attenuated with depth and is hardened whereby lower-energy incident photons are attenuated at shallower depths. In contrast, the photon population of the scattered component, produced from single or multiple Compton events in the phantom, increases with depth. The energy of a scattered photon, \(h\nu'\), is given by the Compton equation:

\[
h\nu' = \frac{h\nu}{1 + \frac{h\nu}{m_e c^2} (1 - \cos \theta)}\]  \[[2-12]\]

where \(m_0\) is the rest mass of the electron, \(c\) is the speed of light and \(\theta\) is the angle at which the scattered photon emerges relative to the incident photon. For example, Compton scatter of a 2 MeV photon (i.e. close to the average energy of photons in an incident 6 MV spectrum) at an angle of 90° will produce a photon of energy 0.814 MeV. Thus multiply-scattered photons reaching the film may have energies below 400 keV where film greatly over-responds to radiation relative to water. Moreover, the proportion of photons in this low-energy region will increase with depth. This causes a depth-depending shift of sensitometric curve.

This effect can be significant and becomes more pronounced for lower beam qualities. For example, for a 10 \(\times\) 10 cm\(^2\) Co-60 beam, at depths greater than 15 cm, using a depth-unspecific sensitometric curve may result in errors in measured dose of up to 30%. For a 25 \(\times\) 25 cm\(^2\) 4 MV beam, differences of 16% have been reported. At 18 MV, a 7% an increase in film sensitivity has been observed between the depths of 1.0
cm to 12.0 cm for a 10 x 10 cm\(^2\) field.\(^{108}\) For a 25 MV beam, an over-response of film of up to 5\% has been cited.\(^{109}\)

With the aim of correcting for this depth-dependence, Hale et al.\(^{108}\) have measured a series of sensitometric curves, one for each of a range of depths, and have achieved 2\% accuracy by using the appropriate curve for dosimetric studies. Williamson et al.\(^{106}\) have parameterized film sensitivity (OD per unit dose) as a function of depth in order to compensate for this dependence, and have obtained 3\% accuracy with this technique. Rather than correcting measured optical densities for the effect of phantom scatter, another approach\(^{107}\) involves removing low-energy, lateral photon scatter by placing lead foils on either side of the film plane.

The second important effect complicating practical film dosimetry is a dependence of film sensitivity on field size. As demonstrated recently by Burch et al.,\(^{107}\) as the field size increases there is a rise in the relative population of Compton-scattered photons reaching the measurement point. This effect also has significant consequences. For example, for a 4 MV beam, the measured dose may vary by up to 20\%, depending on the field size used to establish the sensitometric curve.

There is an additional possible dependence on film orientation relative to the axis of the beam, but the reported severity of this effect varies considerably in the literature. Depending on the beam quality, film type and phantom arrangement, the magnitude of this dependence ranges from negligible\(^{107}\) for a 4 MV beam and a common verification film\(*\) to 10\% for 4 MV beam and an older, industrial-type film.\(^{†}\)

Thus, with dependencies of film sensitivity on depth, field size, and possibly orientation, radiographic film dosimetry would require establishing a two- or three-dimensional array of sensitometric curves, whereby each curve corresponds to a specific depth/field size/orientation combination. Since the sensitometric curve should be re-measured for different batches of film, this calibration procedure would become lengthy and subject to error. In addition, the problem may be complicated for the dosimetry of

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* Kodak X-Omat V, Eastman Kodak Company
† Kodak RPM-2 type M, Eastman Kodak Company
conformal beams, where it may be difficult to estimate the appropriate sensitometric curve to use for irregular field shapes.

2.5 **Aim and rationale of this thesis**

This thesis consists of two separate parts with the common aim of facilitating accurate, high spatial resolution dosimetry for conformal stereotactic radiosurgery.

2.5.1 **Part one: the use of radiographic film for stereotactic radiosurgical dosimetry**

The first section is an investigation of the applicability of radiographic film specifically for quantitative dosimetry of 6 MV radiosurgical beams. This study was originally inspired by recent data reported by Burch *et al.*\(^{107}\) illustrating that the depth-dependence of film sensitivity becomes less pronounced as field size decreases. Since stereotactic radiosurgery involves the smallest field sizes commonly produced by clinical linear accelerators (1 cm to 4 cm in diameter), the depth and field-size corrections required in converting the recorded film optical density to dose should be minimized. Although there are examples in the literature illustrating the application of radiographic film for accurate dose measurement for narrow photon beams,\(^{84,93,110,111,140}\) the depth and field size dependencies of film sensitivity have not been quantified for radiosurgical dosimetry. An additional investigation into a possible dependence of sensitivity on film orientation is conducted in this section. The magnitude of errors introduced by depth, field size and orientation dependencies are compared to that incurred due to fluctuation in processing conditions and variability between films. Finally, the performance of a new CCD-based digitizer is assessed in the context of digitizing dosimetric films.

This investigation is of practical value since, with the current trend toward conformal techniques in radiosurgery, there is an increasing need for high-resolution measurement of integrated dose distributions. We have previously published a summary\(^{112}\) of the majority of the content of this first section in the literature.
2.5.2 Part two: A system for three-dimensional dose verification in radiosurgery

The second section of this thesis presents a direct application of the results of part one. This section describes the design considerations and development of a novel phantom system capable of providing quantitative measurement of three dimensional dose distributions in radiosurgery. This system includes both hardware and software components and has been designed with aim of providing a practical tool for radiosurgical commissioning, periodic quality assurance or verification of challenging patient treatments. Benchmark tests of this system are presented and clinical examples are described. We have reported previously on the design and validation of this system elsewhere in the literature.\textsuperscript{113,114}
3 Methods and Materials I: Radiographic film dosimetry for 6 MV radiosurgical beams

This chapter describes the methods used to evaluate radiographic film dosimetry specifically for radiosurgical 6 MV beams. The goal of this series of experiments is the quantification of the magnitude of dependencies of film sensitivity on depth, field-size and orientation relative to the beam axis. The potential errors caused by these effects can be compared to the measured reproducibility of the film dosimetry technique. Monte Carlo simulation is employed to examine differences between the characteristics of radiosurgical beams and larger photon beams in the context of film dosimetry. In the final section of this chapter, methods to assess the applicability of a high-resolution, CCD-based digitizer for radiosurgical film dosimetry are discussed.

3.1 Linear Accelerator

The 6 MV photon beam used for all experiments described in this work was generated by a clinical linear accelerator*. The nominal accelerating potential for this beam is 4.7 MV as defined by the AAPM Task Group 21 protocol. In its conventional (non-stereotactic) mode of operation, beam dimensions are set by adjusting tungsten jaws in the linear accelerator head to produce rectangular field sizes ranging from 5 cm x 5 cm to 40 cm x 40 cm at isocentre. For radiosurgical applications, the jaws are fixed at 5 cm x 5 cm and the tertiary, stereotactic collimator† is mounted below. The circular radiosurgical collimators used throughout this work produce fields ranging from 10 mm to 40 mm at isocentre and are constructed of lead encased within a brass sleeve. Irregularly-shaped conformal collimators are also used and are described in section 4.3.2. The collimator apertures are tapered to match the divergence of the photon beam in order to minimize geometric beam penumbra. Figure 3.1 depicts a linear accelerator equipped with radiosurgical collimation and shows the coordinate system for the couch and gantry.

* Clinac 2100C/D, Varian Associates, Palo Alto, CA, USA
† BrainLAB, AG, Munich, Germany
The theory and operation of clinical linear accelerators has been reviewed extensively in the literature.\textsuperscript{115,116,117}

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**Figure 3.1** Front view schematic diagram of the linear accelerator, showing couch and gantry coordinate system conventions.
3.2 Film dosimetry technique

The experimental setup was designed to be easily reproducible, consisting of a single sheet of radiographic film encased within water-equivalent plastic (Solid Water™*) phantom material. For each exposure, a single 10” x 12” sheet of radiographic film† from a single batch was sealed within a light-tight Solid Water™ cassette as illustrated in Figure 3.2a). The cassette consists of two 2 cm-thick slabs of Solid Water™ sealed around three edges by nylon screws and a rubber O-ring. For each exposure, a single sheet of film was removed from its envelope and paper liner and inserted into a 0.2 mm – thick recessed compartment between the slabs. For exposures with the beam central axis parallel to the film plane, the film sheet was positioned so that its edge was flush (to within approximately ± 0.5 mm) with the edge of the phantom. This ensures that edge of the film corresponds to the phantom surface. Tightening the nylon screws fixes the position of the film throughout the experiment and removes air gaps between the film and the adjacent phantom material. The unsealed edge was covered with black, opaque vinyl tape to prevent exposure by visible light.

To minimize possible variations due to film chemistry conditions, the processor‡ used for film development was one for which daily quality assurance checks are performed and film throughput is reasonably high. Developer temperature fluctuated by less than ±0.5° C between processing sessions.

* Gammex RMI, Inc. Middleton, WI, USA
† Kodak X-Omat V, batch number 194 05 2, Eastman Kodak Company
‡ Kodak X-Omat RP, Eastman Kodak Company
Figure 3.2 a) The film is inserted in a light-tight Solid Water™ cassette sealed around three edges by an o-ring. b) The parallel (A) and perpendicular (B) beam orientations are shown. Additional Solid Water™ phantom material is added above and below the cassette for both orientations.
An infrared scanning densitometer* was used to measure one-dimensional distributions of optical density on the film. The densitometer consists of an infrared pulsed LED source with a peak wavelength of 950 nm and a detector assembly with four photodiodes mounted on the inside surface of a 50 mm diameter mirrored Ulbricht sphere. This arrangement ensures detection of light transmitted through the film within a cone centred about the incident beam with an angular width of ± 80°. The source/detector assembly is mounted onto a two-dimensional translation stage facilitating scanning over an area of 45 x 48 cm². Although the precision of translation is approximately ± 0.2 mm, the spatial resolution is ultimately limited by the 0.8 mm width of the infrared beam. As reported by Charland et al.,¹¹⁸ the resolution precludes accurate digitization of the sharp profiles of radiosurgical photon beams (i.e. in the dimension perpendicular to the central axis) unless deconvolution techniques are used to correct for the blurring due to finite source size. However, this device is accurate in digitizing the spatial distribution of optical density in regions of low dose gradient, such as along the central axis of the beam at depths greater than the depth of dose maximum. The signal from the densitometer is linear with optical density over a dynamic range reaching 4.0 optical density units (OD). The optical density measurement resolution is ±0.01 OD. For all films, the base-plus-fog optical density was subtracted from scanned optical densities.

3.3 Dependence of film sensitivity on depth

In the context of dosimetry, radiographic film sensitivity refers to the relationship between the delivered absolute dose to tissue surrounding the film and the optical density recorded on the film. As discussed in section 2.4.3, any variation in film sensitivity should be manifest in a shift or change of shape of the sensitometric curve. The method used for measuring the dependence of sensitivity on depth in phantom therefore consists of measuring a set of sensitometric curves, whereby each curve corresponds to a different depth in phantom.

For this experiment, the arrangement A shown in Figure 3.2b) was used to expose the film along the depth axis (henceforth called the “parallel beam orientation”). Room

* Wellhöfer WP102, Schwarzenbruck, Germany
Lasers were used to align the central axis and plane of the film to within approximately ± 0.5 mm. All films were exposed using the median stereotactic collimator field size of 25 mm. To acquire ample data to establish sensitometric curves in sufficient detail, 14 films were exposed with different monitor unit (MU) settings resulting in maximum doses to phantom ranging between 5 and 385 cGy. Although the goal of this series of exposures was to establish the sensitometric curves over a dose range of approximately 0-140 cGy (i.e. below the saturation region of the film), the higher maximum dose settings used were required in order to obtain a sufficient range of dose at greater depths in phantom. At least three films were exposed for each dose setting in order to quantify and average variations due to processing and scanning. This procedure was then repeated using a 20 cm × 20 cm field size for comparison of small-field and large-field results.

For each exposure, a two-dimensional profile of optical density was recorded along the depth dimension of the phantom. In scanning the film, it was necessary to ensure that the direction of travel of the densitometer was parallel to the recorded location of the central axis (i.e. so that the film was not tilted relative to the direction of scanning). To align the film, the beam profiles (perpendicular to the central axis) were scanned at depths of 5 cm and 10 cm, and the geometric centers of each profile were calculated from the scanned data. Any discrepancy between the calculated location of these profile centres indicated that the film was angled relative to the scanning direction. Once aligned, the film was scanned along the central axis location and the optical density was measured along the depth axis from the surface to a depth of 20.0 cm in 0.1 cm increments.

Each scan thus produces a series of optical density values recorded at each of 200 depths. In order to relate these optical densities to absolute doses, the dose to water corresponding to each of these depths and for each MU setting was calculated from

\[
D(d, A) = \frac{(MU) \cdot PDD(d, A) \cdot S_{t}(A)}{100}
\]  

[3-1] 

where \(D(d, A)\) is the dose at depth \(d\) for field size \(A\) at the phantom surface, \(MU\) is the number of monitor units given, \(PDD(d, A)\) is the percent depth dose and \(S_{t}(A)\) is the total...
scatter factor.\textsuperscript{119} Both $S_d(A)$ and $PDD(d,A)^*$ were measured in a water phantom\textsuperscript{†} using a p-type silicon electron diode\textsuperscript{‡} (measurement of beam characteristics been discussed in the literature\textsuperscript{81}). The $PDD$ was measured in 0.1 cm increments to a maximum depth of 25.0 cm and was linearly interpolated when required to obtain values at arbitrary depths.

After scanning all 14 films the resultant set of data consisted of 14 [optical density, dose] pairs at each of 200 depths in phantom. These data were therefore sufficient to generate sensitometric curves at each of 200 different depths in phantom, with each curve defined by 14 points.

3.4 Dependence of film sensitivity on field size

Similar sets of sensitometric curves were established, as described above, for both the minimum (10 mm diameter) and maximum (40 mm diameter) field sizes. These data permit examination of the variation between the sensitometric curves, at any arbitrary depth, over the range of field sizes used routinely in radiosurgery.

3.5 Variation of film response with orientation

In order to quantify a possible dependence of film sensitivity on orientation, a series of exposures was also performed with the film plane perpendicular to the central axis, with the gantry at 0° as illustrated in arrangement $B$ of Figure 3.2b) (henceforth called the "perpendicular orientation"). For each exposure, the film was placed below a depth of 10.0 cm of Solid Water\textsuperscript{TM} phantom material. Five separate films were exposed resulting in doses to $d_{\text{max}}$ ranging from 5 to 385 cGy.\textsuperscript{§} After film processing, the optical density was measured at ten points within the spot-pattern recorded on the film, within a radial distance of 0.5 cm from the central axis and a mean value was recorded. Using equation [3-1] to calculate absolute doses for each exposure, a sensitometric curve was established for the perpendicular orientation and compared with the sensitometric curve

\* Definitions of relevant dose functions are given in Appendix A.
\* model WP 600 Wellhöfer, Schwarzenbruck, Germany
\* Scanditronix, Uppsala, Sweden
\* At a depth of 10 cm, this will result in doses within the saturation region of the film, but for this study it was useful to compare parallel and perpendicular sensitometric curves over as wide a dose range as possible.
for the parallel orientation. For both orientations, films were processed in the same batch to minimize the effect of chemical variability.

3.6 Dosimetric reproducibility

During each experimental session for the studies described in sections 3.3 to 3.5, at least three control films were exposed using a standard setup and MU setting. These exposures were performed using the 25 mm diameter collimator to deliver a maximum dose of 87 cGy to the film in the parallel orientation. The optical density curves obtained by scanning a film along the depth dimension at the location of the central axis were compared. By repeating these exposures i) during a single session and ii) over the course of approximately six months, realistic measures of both film-to-film and session-to-session variation were obtained. In addition to processing variations, these values also include contributions such as drift and fluctuation of the densitometer signal and non-uniformity of the film emulsion.

3.7 Monte Carlo study of photon spectra in a homogeneous phantom

3.7.1 Monte Carlo simulation and the Electron Gamma Shower 4 (EGS4) code

Equations [1-1], [1-3] and [1-5] give examples of analytic dose calculation methods. These methods are fast and give correct results for simple geometries and cases without tissue inhomogeneities. Even with corrections, however, they sometimes fail to accurately predict the magnitude and/or distribution of dose in more complex geometries. An alternative to the analytic approach is Monte Carlo simulation, which takes its name from the use of random numbers to sample probability distributions governing the transport of radiation in matter. The Monte Carlo method is potentially the most accurate technique for dose calculation since it relies on the first principles of radiation interaction rather than empirical measurement or analytical assumptions. In the simulation of a photon beam, for example, the trajectory of each incident photon is followed, whereby the modes of interactions (e.g. photoelectric effect, Compton scatter or pair production) and the outcomes (e.g. the scattering angle of the photon) are determined based on random sampling of probability distributions. The trajectories of each scattered photon and particles set into motion in the medium are also tracked to
record a complete record or *history* of the incident particle. As this process proceeds, various quantities can be recorded in phantom (e.g. dose or fluence). Reducing statistical uncertainty in the result requires simulating many (often millions) of histories, and thus Monte Carlo simulations are usually time consuming and computationally demanding.

A convenient, accurate and widely-used Monte Carlo code is the EGS4 package,\textsuperscript{125} which allows the user to define the geometry and composition of a phantom in three dimensions. As illustrated in the following section and in section 4.1.1, the geometry may be specified in various coordinate systems, depending on the chosen *usercode*.\textsuperscript{*} The EGS4 code then simulates the transport of a user-defined beam of radiation in the phantom.

In addition to potential accuracy over a wider range of geometries, the Monte Carlo approach offers the versatility to record (or “score”) quantities that are physically difficult or impossible to measure. The following section, for example, describes the scoring of photon spectra at various depths in phantom.

### 3.7.2 Spectral change with depth in phantom for small fields

It has been suggested in the literature\textsuperscript{107,109,121,122} that dependencies of radiographic film sensitivity with parameters such as depth in phantom or field size arise from changes in the photon spectrum incident upon the emulsion. Thus, in order to explain variation in the sensitometric curve, it is instructive to examine the behavior of the photon spectrum in phantom for both cases. Experimental determination of megavoltage photon beam spectra is difficult and usually deduced indirectly from attenuation measurements.\textsuperscript{123,124} Therefore, for this investigation a Monte Carlo photon and electron transport simulation was used to model a 6 MV beam incident upon a water phantom.

The EGS4 Monte Carlo FLURZ code\textsuperscript{125,126} was used to define the homogeneous water phantom as shown in Figure 3.3. This user code permits the scoring of primary and

\textsuperscript{*} This is the part of the EGS4 system actually written or modified by a user. In addition to the phantom geometry, this controls the initiation of each particle history, defines which quantities are recorded in phantom and where, controls the termination of histories (i.e. when a particle exits the phantom volume) and generates output.
total photon fluence within the voxels of a phantom, which is user-defined in terms of its materials and dimensions, as specified in cylindrical coordinates. As shown in Figure 3.3 the phantom for this study is 40.0 cm in diameter and 40.0 cm deep. For comparison, two separate simulations were conducted with field sizes of 2.5 cm and 25.0 cm. For both field sizes, photons in the incident beam were sampled from a 6 MV incident photon spectrum provided previously by Mohan, selected to be representative of that produced by the linear accelerator used in this work.

The full photon spectrum was scored in phantom along the depth axis at 1.0 cm increments within a 1.0 cm-diameter column along the centre of the cylinder. 15×10⁶ incident histories were recorded for each beam. The Parameter-Reduced Electron Step Algorithm (PRESTA) was used to model electron transport, and the electron cut-off (ECUT) and photon cut-off (PCUT) parameters were set to 0.521 MeV and 0.010 MeV, respectively.

3.7.3 Applicability of the 6 MV incident photon spectrum

The study described involves a comparative examination of spectra in phantom, and therefore does not require the incident spectrum to match that from a particular model of linear accelerator. However, the spectrum should be at least representative of that of incident 6 MV beams for both the large- and small-fields. This was examined by using the EGS4 user code DOSRZ to score the dose deposited by the photon beam (characterized by the published spectrum) along the central axis within a homogeneous water phantom. The phantom arrangement was similar to that shown in Figure 3.3, except that total dose was scored within a 0.5 cm diameter column along the central axis at 0.5 cm depth increments. The simulation was repeated for both 2.5 cm and 25.0 cm diameter beams using the Mohan spectrum as input. 50 million and 200 million incident histories were tracked for the small and large beams, respectively. The recorded dose values were normalized to that deposited at the depth of maximum dose. The percent-depth-dose curves thus generated were compared to those measured using a p-type silicon diode* and an IC10 ionization chamber† for the small and large fields, respectively.

* model WP 600 Wellhöfer, Schwarzenbruck, Germany
† Wellhöfer, Schwarzenbruck, Germany
Figure 3.3 Homogeneous water phantom defined using the FLURZ user code for examination of photon spectral changes in phantom.

3.8 Characteristics of a CCD-based scanner for digitization of radiographic film

As mentioned in section 3.2, the infrared scanning densitometer used in the experiments described previously has inadequate spatial resolution for accurate measurement of radiosurgical beam profiles or two-dimensional radiosurgical dose distributions. This section describes the methods used to assess the feasibility and
performance of a new CCD-based transparency digitizer* for the scanning of dosimetric films with improved spatial resolution. No reports on the use of this digitizer for film dosimetry were available, and therefore a study of its performance in this application was required. The criteria considered in this assessment were: spatial linearity, optical density dynamic range, signal-to-noise ratio, signal stability and image uniformity. Several of the methods employed here are similar to those performed recently by Mersseman and De Wagter, 130 Meeder et al. 131 and Teslow. 132

3.8.1 Digitizer specifications

The digitizer examined in this work is a relatively new device employing a cold-cathode fluorescent light source and an 8000-element linear CCD array detector. Films are positioned on a 20.32 cm × 25.4 cm glass bed, which slides into a compartment in the body of the scanner and is thus shielded from external room light. As shown schematically in Figure 3.4, the film is positioned between the light source, focusing lenses and the CCD array. The light source, CCD array and associated optics translate in the dimension perpendicular to the length of the CCD array to acquire a two-dimensional spatial distribution of optical density. Images may also be acquired from reflective media placed on the upper glass bed. In this configuration, the upper two sources are used and the rightmost mirror in Figure 3.4 pivots to alter the optical pathway.

The digitizer specifications summarized in Table 3.1 indicate that, compared to scanning densitometers like that described in section 3.2, the principal advantages are i) significantly improved spatial resolution and ii) increased speed. The hardware spatial sampling of the digitizer is configurable over the range of 0.508 to 0.025 mm/pixel in the dimension of the linear CCD array. In the direction of travel of the array, the stepper motor driving the array is capable of two thousand steps per inch, resulting in a limit on spatial resolution in this dimension of 0.013 mm/pixel. For the purposes of stereotactic radiosurgical dosimetry, a spatial resolution of 0.254 mm/pixel is sufficient and therefore was used for all digitization performed in the present work.

* Agfa Duoscan, Agfa Division, Bayer Corporation, Wilmington, MA, USA
Table 3.1 Transparency digitizer manufacturer specifications.

| Hardware spatial resolution range (mm / pixel) | 0.508 to 0.025 (horizontal)  
| 0.508 to 0.013 (vertical) |
| Dynamic range (OD) | 0.15 to 3.4 |
| Image depth (bits) | 8, 10 or 12 |
| Scanning duration per line (ms) | 10 |
| Light source | Cold cathode fluorescent |
| CCD type | Tri-linear coated, 8000 elements |

Figure 3.4 Schematic diagram of the transparency digitizer source, detector and associated optics.

It should be emphasized that the values in Table 3.1 are nominal values cited by the manufacturer only, and as will be shown in the results of Chapter 5, the actual characteristics differ somewhat from those listed. Data regarding spatial linearity, signal-to-noise ratio and uniformity were unavailable from the manufacturer.
3.8.2 Spatial linearity

For high-resolution dosimetric applications, it is essential that the pattern of optical density on the film is digitized without introduction of significant spatial distortion. To assess spatial linearity, a distortion grid phantom was scanned. This pattern consists of an accurately-printed matrix of $98 \times 250$ black (OD > 4.0) dots printed on a transparency. Each dot is 0.50 mm in diameter and the separation between vertical columns and horizontal rows is 2.00 mm and 1.00 mm, respectively. A travelling microscope* was used to verify that the tolerance of the printing of this pattern was accurate to within ±0.02 mm.

* Swift Instruments, Inc., San Jose, CA

Figure 3.5 A segment of the digitized distortion phantom used to quantify spatial linearity.

The distortion phantom was placed on the bed of the digitizer with the x-dimension of the grid (as defined in Figure 3.5) oriented parallel to the length of the CCD array. The pattern was digitized at a spatial resolution of 0.254 mm/pixel. Code was
written in MATLAB* to first extract profiles along the x-dimension for each detected row of dots in the image. A segment of one of these profiles is shown in Figure 3.6a). Next, the locations of the detected centres of all dots in each profile were identified. The presence of partial volume effect arising with the (0.25 mm/pixel)$^{-1}$ scanning resolution (evident in the magnified view of Figure 3.5 and in the profiles of Figure 3.6) precludes the use of a simple maximum-finding algorithm. Instead, for each dot in the profile, the location of the local maximum was determined, and then the centre-of-mass among three pixels centred about this maximum was calculated, i.e.:

$$x_{com} = \frac{(x_{max} - 1)p_{x_{max} - 1} + (x_{max})p_{x_{max}} + (x_{max} + 1)p_{x_{max} + 1}}{p_{x_{max} - 1} + p_{x_{max}} + p_{x_{max} + 1}}$$

[3-2]

where $x_{max}$ is the location of the local maximum and $p_{x_{max}}$ is the pixel value at that location. After locating the detected centre locations for all profiles along the x-dimension, a vector was calculated giving difference between the locations of detected centres and true centres (where the latter are determined using the known physical separation between dots shown in Figure 3.5) for all dots in the image. This vector, when presented as a frequency histogram, provides the distribution of deviation from linearity of the digitizer. By using an identical procedure for the y-dimension, separate measures of the linearity were obtained for the x- and y-dimensions. A segment of a profile along a vertical column of dots is shown in Figure 3.6b.) The results of this spatial linearity analysis are presented subsequently in section 5.6.1.

* The Mathworks, Natick, MA, USA
Figure 3.6 Profiles through a segment of the distortion phantom along the a) x-dimension and b) y-dimension.
3.8.3 Dynamic range

To measure the dynamic range of the digitizer, an optical density step pattern was produced by exposing a sheet of radiographic film* with an optical sensitometer.† This produces an array of 20 regions on the developed film (Figure 3.7) with OD values ranging from 0.21 to 3.45, as measured using a spot densitometer.‡ Routines were written in MATLAB to extract a 17 pixel × 35 pixel region of interest in each step. Plotting the mean value among the 595 pixels in each region of interest as a function of measured optical density provides a quantitative description of the dynamic range.

![Figure 3.7 The optical density step pattern used to quantify dynamic range.](image)

3.8.4 Calibration for dose measurement

The technique for establishing the sensitometric calibration curve is essentially the same as was described in section 3.3 whereby the film/Solid Water™ phantom material were set up in arrangement A of Figure 3.2b). Typically three films were exposed with different MU settings (e.g. 20, 50, 100 MU) to span the dynamic dose range of the film/digitizer combination. Prior to digitizing each film, it was verified that the location of central axis on the film was parallel (within ± 0.2°) to the direction of travel of the CCD array. Films were digitized using the 12-bit grayscale scanning mode (4096 grayscale levels) and 0.254 mm × 0.254 mm pixel size. A single one-dimensional vector of scanned pixel values corresponding to different depths in phantom (and therefore different doses) was obtained from the digitized image by extracting a column of pixels along the central axis location. The absolute dose values corresponding to the elements in this vector were then calculated as described in section 3.3, using equation [3-1].

* Kodak X-Omat V, batch number 194 05 2, Eastman Kodak Company
† X-Rite model 393, X-Rite Inc., Grandville, MI, USA
‡ X-Rite model 331, X-Rite Inc., Grandville, MI, USA
3.8.5 Noise in digitized film images

There are a number of separate contributions to the noise obtained in the signal from the digitizer when scanning radiographic film. The image recorded on the film, for example, will introduce its own noise due to structure and quantum mottle, and the magnitude of the recorded noise on the film will vary as a function of spatial frequency according to the Wiener spectrum. These components of noise in turn will combine with additional sources arising during the digitization process, including pixel-to-pixel read noise in the CCD, quantum noise, temperature-dependent reset noise and background noise (comprised of several sources including optical fat zero, dark current and internal luminescence) and readout amplifier noise. The aim of this study was not to isolate the various sources of noise, but rather to quantify the aggregate noise-to-signal ratio as a function of the optical density on the film. Moreover, this measurement was made only at a single spatial frequency (0.254 mm/pixel). To examine the variation of the noise-to-signal ratio with magnitude of the signal itself, the optical density of the step pattern of Figure 3.7 was again used and was digitized 60 times successively. For each image \( m \), in each optical density step \( OD \), the signal \( P_{OD,m} \) was defined as the mean of the pixel values in the region of interest, and the noise \( \sigma_{od,m} \) was defined as the standard deviation among these pixel values:

\[
P_{OD,m} = \frac{\sum_{j=1}^{NPIX} P_j}{NPIX}
\]

\[
\sigma_{od,m} = \left( \frac{\sum_{j=1}^{NPIX} (P_j - P_{OD,m})^2}{NPIX - 1} \right)^{1/2}
\]  \hspace{1cm} [3-3]
where $NPIX$ is the number of pixels in the region of interest. For each optical density step, the noise-to-signal ratio was then defined as the coefficient of variation 

$$\left(\frac{\sigma_{OD,m}}{P_{OD,m}}\right).$$

Finally, for each optical density step an ensemble coefficient of variation was calculated:

$$\langle CV_{OD} \rangle = \frac{1}{NI} \sum_{m=1}^{NI} \frac{\sigma_{OD,m}}{P_{OD,m}} \quad [3-4]$$

where $NI$ is the number of images (in this case, $NI=60$).

### 3.8.6 Signal stability

Signal stability, in this context, refers to the magnitude of variation over time of the signal from the digitizer. The digitizer must exhibit an acceptable degree of stability between successive scans in one session, for example, during which both calibration and dosimetric films would be digitized. To assess signal stability the digitizer was turned off and allowed to cool for 24 hours. The scanner was then turned on and 60 successive scans of the optical density step pattern were immediately acquired. For a particular optical density step, the variation of the mean pixel value within the region of interest was monitored as a function of the number of scans acquired. This test permits the determination of both the length of the warm-up period and of the magnitude of variation of the mean pixel value that can be expected for a given optical density after the digitizer has stabilized.

### 3.8.7 Dependence of signal on adjacent optical density

The majority of scanning densitometers used in film dosimetry employ a single, narrow beam of light aligned with a detector that measures the intensity transmitted through the film. In such an arrangement the source is collimated and the optical density is obtained for a single point (usually defined by the beam with or diameter of the collimator). The arrangement is different for the CCD-based digitizer used in this work, which instead uses an array of detector elements across from an elongated light source.
Optimally, this arrangement would permit simultaneous acquisition of an array of measurements, where each measurement reflects the light intensity transmitted at a single point and is completely independent of the optical density of neighboring regions of the film. However, for several reasons, this situation may not be realized. First, it is possible that light transmitted at oblique angles to the normal to the film may affect the intensity of the signal detected and therefore the signal detected at a given location would depend on the surrounding optical density. Second, as suggested in the literature, the scattering of light after transmission through the film may affect the magnitude of the detected signal. Third, cross-talk between CCD elements would also interfere with the independence of adjacent signals.

If this effect were predominant, the edges of a step function on film, when digitized, would be convolved with the spatial transfer function composed of the incident source profile and the spatial response of the detector. Although no serious artifacts of this nature had been observed in practice, a technique similar to that presented by Mersseman et al. was used to quantify the spatial independence of measurements. Two different sets of apertures were constructed whereby each aperture consisted of two black bands (with OD > 3.0) separated by a low optical density region ranging in width from 2.0 mm to 20.0 mm. The length of the apertures was fixed at 3.0 cm. For the first set of apertures, the bands were printed on an acetate film transparency (OD = 0.11). This low optical density aperture was used to exaggerate the effect by placing a very low optical density region in the middle of two very high optical density regions. The second set of apertures was constructed in a similar fashion, except the optical density of the aperture was that of the base-plus-fog background of radiographic film (OD = 0.21). This represents a more realistic, although worst-case situation for film dosimetry.

Each aperture was digitized at the same location within the field-of-view. For each image, the mean pixel value along the centre-line of the aperture was calculated. A systematic dependence of the scanner signal on the optical density of adjacent film regions should be indicated by a variation of this mean pixel value with aperture width.
3.8.8 Image Uniformity

Image uniformity was measured by digitizing an unexposed, processed radiographic film. Thus, non-uniformities in the image will arise from both spatial variation in the base-plus-fog optical density (OD=0.22) on the film and inherent non-uniformity in the scanning process. Of particular interest in this test were low-frequency spatial variations (in comparison to the pixel-by-pixel noise examined in section 3.8.5), which occur over the entire field of view of the digitizer. Therefore the scanned image was convolved with a $5 \times 5$ pixel$^2$ Gaussian filter prior to analysis.
4 Methods and Materials II: A system for three-dimensional dose verification in stereotactic radiosurgery

The previous chapter described the methods used to investigate several basic questions concerning the applicability of radiographic film for one- or two-dimensional dosimetry of 6 MV radiosurgical beams. This chapter builds upon the results of these previous studies for the development of a practical system capable of quantitative, three-dimensional dose measurement.

4.1 Multiple-film geometry

The focus of the first part of this discussion is the feasibility of using a multiple-film geometry for the simultaneous acquisition of multiple two-dimensional distributions of dose through a volume. A possible geometry is shown in Figure 4.1, whereby adjacent sheets of film, each of thickness $t_{film} \approx 0.2$ mm (i.e. see Table 2.1), are separated by water- or tissue-equivalent spacers (e.g. Solid Water™, polystyrene or lucite), each of thickness $t_{spacer}$. Although the spatial sampling frequency perpendicular to film planes would be set by $t_{spacer}$, the actual voxel size could be extremely small (e.g. 0.2 mm × 0.25 mm × 0.25 mm) as is required for radiosurgical dosimetry. It is conceivable that this arrangement could be sealed inside of a light- and water-proof cassette box and placed inside a phantom filled with water to simulate surrounding tissue. The dimensions $t_{cassette}$, $w_{cassette}$ and $h_{cassette}$ would be sufficient to acquire slices through an entire radiosurgical dose distribution, and each dimension should be greater than approximately 6.0 cm to acquire dose values outside of the dose fall-off region.
4.1.1 Monte Carlo validation of the multiple-film design

This multiple-film geometry immediately raises the question of tissue equivalence. Although only approximately 5-10% of the thickness of a verification film is composed of high atomic number emulsion (with the majority of the total 0.2 mm thickness comprised of base), the mass attenuation coefficient of the silver halide in the emulsion may differ drastically from that of tissue, depending on the photon energy. Moreover, if $t_{\text{spacer}}$ is reduced to place adjacent film planes closer together, the composition of the phantom will eventually diverge from that of a homogeneous, tissue-equivalent phantom and will approach that of a phantom composed entirely of base and emulsion.

Hence, in testing the feasibility of the multiple-film arrangement, it was necessary to determine the magnitude of dose perturbation to the tissue-equivalent material surrounding the film planes as a function of $t_{\text{spacer}}$. Also, since the radiosurgical beams may be incident at arbitrary angles relative to the film planes and the path length in emulsion depends on this angle of incidence, it was necessary to study this perturbation for the extreme cases of parallel and perpendicular beam orientations.

The EGS4 photon/electron transport code\textsuperscript{125} was used to simulate the interaction of a 6 MV radiosurgical beam with a realistic model of a multiple-film radiosurgical phantom. The aim of this simulation was to assess the feasibility of the geometry and to optimize the design parameters of a phantom prior to its construction. The DOSXYZ
user code\textsuperscript{136} was used to specify a phantom (illustrated in Figure 4.2) in Cartesian coordinates. The geometry of this model is similar to that which could be manufactured, with the exception of the shape of the water container surrounding the multiple-film cassette (which would be spherical or anthropomorphic for radiosurgical applications). The simulated phantom consists of a $20 \times 20 \times 20$ cm$^3$ water tank containing a multiple-film cassette box with inner dimensions of $7 \times 7 \times 12$ cm$^3$. The film was modeled to resemble that used regularly for dosimetry\textsuperscript{*} with two gelatin/silver bromide layers (each 0.012 mm thick) on either side of a polyester base (0.180 mm thick). Table 4.1 gives the composition of the modeled emulsion. 30 million incident photons were tracked per simulation, and the total dose was scored in the volume of each of the inter-film spacers. The dose perturbation due to the emulsion was assessed by repeating the simulation with an identical geometry, but with water replacing both the gelatin/silver bromide and film base regions. Simulations were repeated for $t_{\text{spacer}}$ values of 0.5, 1.0 and 3.2 mm. For each value of $t_{\text{spacer}}$, simulations were performed for the perpendicular and parallel beam orientations. In all cases, the beam was $2.5 \times 2.5$ cm$^2$ in cross-section, non-divergent and centered upon the multiple-film cassette box. The published incident spectrum\textsuperscript{127} used in all simulations had been validated previously as described in section 3.7.3.

Although Monte Carlo techniques have been used previously to model photon and electron transport in a single layer of radiographic film,\textsuperscript{101} obtaining accurate results requires careful selection of several of the simulation parameters. Specifically, it is crucial that the trajectories of electrons are partitioned into sufficiently small steps near the interfaces of thin regions (i.e. film gelatin/silver bromide or base layers) to prevent boundary-crossing artifacts. The PRESTA algorithm\textsuperscript{128} used in this set of simulations limits the total curved path of an electron as it approaches the boundary between two adjacent voxels. This partitioning of the total path of an electron into steps of decreasing length is achieved by ensuring that the total curved path of an electron is no longer than the perpendicular distance to the closest boundary.

\* Kodak X-Omat V, Eastman Kodak Company
Figure 4.2 Monte Carlo simulation arrangement used to verify the magnitude of dose perturbation due to the presence of an array of radiographic films. The perpendicular beam simulation is shown.
Table 4.1 Composition of the emulsion (silver bromide and gelatin) layer in the EGS4 Monte Carlo simulation.

<table>
<thead>
<tr>
<th>element</th>
<th>fraction by weight</th>
</tr>
</thead>
<tbody>
<tr>
<td>H</td>
<td>0.0305</td>
</tr>
<tr>
<td>C</td>
<td>0.2107</td>
</tr>
<tr>
<td>N</td>
<td>0.0721</td>
</tr>
<tr>
<td>O</td>
<td>0.1632</td>
</tr>
<tr>
<td>Br</td>
<td>0.2228</td>
</tr>
<tr>
<td>Ag</td>
<td>0.3007</td>
</tr>
</tbody>
</table>

However, this truncation must be stopped at some point (or else the electron would never actually cross a boundary), and the step size cannot be reduced below some lower limit, $t'_{\text{min}}$. This parameter is a function of the user-defined electron cut-off energy, $E_{\text{CUT}}$ and therefore can be set according to the specific geometry. For the phantom regions shown in Figure 4.2, a value of 514 keV (total energy) for $E_{\text{CUT}}$ ensures that for the maximum possible electron energy (6 MeV), $t'_{\text{min}}$ is a fraction of the emulsion thickness. This low value increases simulation time dramatically but is important for obtaining accurate deposited dose values near interfaces. The photon cut-off energy was also set intentionally low, to 3 keV. The absence of boundary artifacts was verified in the homogeneous phantom case.

4.1.2 Experimental validation of the multiple-film geometry

In addition to the Monte Carlo simulation, an experimental examination of possible beam perturbation due to the presence of multiple films was conducted for both the perpendicular and parallel beam orientations. For the perpendicular case, eighteen 7 cm x 12 cm films were contained within a sealed polystyrene multiple-film cassette box, separated by 3.2 mm-thick polystyrene spacers. The final spacer in the box was of a thickness to ensure that air gaps were eliminated. The cassette box was placed at the center of a 16 cm x 16 cm x 22 cm water tank and the room lasers were used to align the isocentre to the center of the last film in the array (i.e. that most distant from the source). A dose of 50 cGy was administered to the depth of this film using 25 mm-diameter
collimation. For comparison, this irradiation was repeated with only one film in the cassette box aligned to the same position as the last film in the multiple film case. The optical density at the centre of the spot-pattern was measured on both films using both the transparency digitizer and a spot densitometer. Note that the attenuation of the beam by seventeen films prior to depositing dose in the eighteenth film represents a worst-case scenario for the perpendicular orientation.

A similar single/multiple-film comparison was conducted for the parallel beam orientation. In this test, nineteen films were loaded in the cassette box (once again separated by 3.2 mm—thick spacers) with the central film aligned with the beam central axis. A dose of 50 cGy was administered to a depth corresponding to the vertical center of this central film. An identical irradiation was subsequently delivered with just one film in the cassette-box, at the same location as the central film for the multiple-film case. For both beam orientations, significant differences in the magnitude and/or spatial distribution of the optical density on the exposed films would indicate perturbation due to the increased quantity of emulsion in the multiple-film case.

4.2 Phantom hardware design

Based on the results of the studies described above, a radiosurgical head phantom was constructed, consisting of a water-filled plastic head shell containing a sealed polystyrene multiple-film cassette box. A sagittal view of the phantom design is shown in Figure 4.3. The head shell may be fixed to the CT scanner and linear accelerator couches using conventional stereotactic head-ring or mask immobilization devices. The cassette box, shown in Figure 4.4, holds up to twenty 7 cm × 12 cm films separated by polystyrene spacers. The thickness of each spacer is 3.175 mm. This value was based primarily on the results of the Monte Carlo simulations (presented subsequently in section 6.1). The last spacer in the box was milled to a thickness to eliminate air gaps.

The angular position of the cassette box, set at the base-plate of the phantom, may be selected to position the film planes reproducibly in coronal planes (as shown), in sagittal planes, or at a 45° angle in-between. This facilitates re-sampling of the dose
volume with multiple sets of films if required. The spatial dose information acquired from multiple sets of film may be merged in software.

The films and polystyrene spacers are loaded under safelight while tilting the box to ensure that film edges make contact with the inner surfaces of both the box bottom and one vertical surface (i.e. of the posterior wall). This defines two spatial "reference planes" as illustrated in Figure 4.5. Although films are cut using a special jig (with approximately 0.3 mm tolerance in the dimensions) the localization of these reference planes ensures that the subsequent co-registration of the measured and planned dose distributions (discussed below) is independent of the cut film size.

The cassette box is covered with a black veneer, except for a 1 mm-wide diagonal strip on one face. In the course of administering a treatment to the phantom, the room light is transmitted through the uncovered, translucent polystyrene strip, thus producing a small indexing mark on one edge of each film. These marks serve as a means for subsequent automatic orientation and ordering of digitized film images in software. Since a 6 mm-wide border around each film is discarded following digitization (to eliminate possible film edge artifacts), the indexing marks do not interfere with the recorded optical density.

4.3 Clinical application of phantom system

Figure 4.6 illustrates the use of the phantom system for the verification of radiosurgical dose distributions. The main role of the phantom is to simulate the entire treatment of a patient. To this end, the various steps of the radiosurgical process, including the phantom fixation (using, for example, the thermoplastic mask system shown in Figure 4.7), CT imaging, treatment planning and dose delivery are conducted almost exactly as they would be for a patient treatment. The intended endpoint of this process is a direct comparison of the planned and delivered dose distributions in terms of magnitude (i.e. maximum dose), location and shape.
Figure 4.3 Sagittal view of the multiple-film phantom, shown with films oriented in the coronal planes.
Figure 4.4 Multiple-film cassette box with diagonal strip.

Figure 4.5 The cassette box is loaded ensuring that films make contact with two reference planes used subsequently for co-registration of the measured dose volume with the CT set of the phantom.
Figure 4.6 Flowchart outlining the application of the multiple-film phantom in a clinical setting.

Figure 4.7 Radiosurgical phantom in the thermoplastic mask fixation system prior to CT imaging.
4.3.1 Phantom imaging

The phantom is imaged in the CT scanner after filling both the head shell and cassette box with distilled water. Optionally, a simulated target volume (e.g. made of paraffin, polystyrene, or Lucite) may be positioned in the cassette box volume. The CT scanning parameters are the same as are used during clinical imaging for radiosurgery (0.63 mm pixel size, 3.0 mm slice thickness, 2.0 mm slice separation, 120 kVp and 200 mAs).

4.3.2 Treatment planning and delivery

As described in section 1.4.2, treatment planning involves the specification of target volume contours, the beam arrangement, shaping and beam dose weighting. This is done using commercial treatment planning software* based on the acquired CT image set. The treatment plan will depend on the intended goal of the phantom study, but two scenarios are typical:

i) The lesion contour may be arbitrary in shape and designed to test a certain aspect of the radiosurgical system. An example of such a plan is shown in Figure 4.8 in which an arbitrary lesion contour has been defined to test the accuracy of treatment in regions of protrusions. Although this plan shows the lesion defined within the water-filled volume of the cassette box in the CT images, the contour may also be based on a physical model of a lesion positioned in the box prior to imaging.

ii) The lesion contour may be transferred in software from a patient's treatment plan to the phantom's treatment plan. This facilitates the simulation of the treatment of an actual lesion, for example, using the same conformal beam shapes as for the patient.

The maximum planned dose is limited to a value below the saturation region of the

* BrainSCAN, BrainLAB, AG, Munich, Germany
film/digitizer combination. Although this maximum is typically less than that administered clinically by a factor of approximately 25, this requires a simple re-normalization of the plan, and the shape of dose distributions are not affected. All planned isodoses are normalized relative to this maximum dose.

Figure 4.8 Example of a phantom treatment plan in which an arbitrary lesion shape has been defined within the water-filled volume of the cassette box.

For conformal treatments, the planning software produces a file defining the tool-path of a computer numerically-controlled hot-wire cutter* used for the manufacture of irregularly-shaped collimator mould inserts. The mould inserts are cut from a block of Styrofoam SM† so that the outer perimeter of the insert matches the divergence of the photon beam. As shown in Figure 4.9, for each conformal aperture, the mould contains a stereotactic collimator sleeve, which attaches directly to the linear accelerator collimator mount. The mould insert is positioned on an asymmetric key at the centre of the mould.‡

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* Par Scientific, Denmark
† Registered trademark of the Dow Chemical Company, Midland, MI, USA
‡ BrainLAB, AG, Munich, Germany
A low-melting point metal, Cerrobend, is poured around the insert. Cerrobend consists of bismuth, lead, cadmium and tin in proportions of 50%, 26.7%, 10% and 3.3% by mass, respectively. After the cerrobend has solidified and cooled (within 20 minutes, typically) the mould is separated into two halves and the finished collimator is removed.

Figure 4.9 Conformal collimators are manufactured by pouring cerrobend around the irregularly-shaped Styrofoam insert. The position and orientation of the insert are set by sliding it onto an asymmetric mould key.

Prior to treatment delivery, the cassette box is loaded with films and spacers and the phantom is positioned on the linac couch. Room-lasers are used to align the phantom to the location of the planned isocentre using the target positioning box. The treatment is delivered as is done clinically. Optionally, the treatment may be repeated with multiple sets of film at pre-set orientations about the phantom axis.
4.3.3 Film processing and digitization

Films are developed in a processor with microprocessor-control of both the transport speed of the film and of the developer temperature. The transport speed is set to the standard value of 106.7 cm/minute. The developer temperature is set to 29.2 ± 0.3°C, approximately 8°C lower than the value used in standard processing. This has the effect of shifting the sensitometric curve towards lower optical density (OD) values, permitting digitization of optical densities corresponding to a wider dose range within the dynamic range of the scanner (discussed subsequently in section 5.6.2).

Films are scanned using the transparency digitizer described in section 3.8.1. A grayscale scanning resolution of 0.254 mm/pixel was chosen as sufficient for dose verification purposes and also to limit the image size to below 512×512 for subsequent export in Digital Imaging and Communication in Medicine (DICOM) format. A Perspex guide has been constructed which sits on the transparency bed of the scanner and facilitates the positioning of four films to be digitized in a single pass. The guide also ensures that the long axis of each film is parallel to the direction of travel of the CCD array.

4.3.4 Film calibration

To obtain the optical density-to-dose sensitometric curve, a single sheet of film is located at the centre of the multiple-film cassette box and the remainder of the box is filled with polystyrene slabs. The cassette box is then placed in a 16 cm × 16 cm × 22 cm water tank. The top edge of the film is located at a known depth, and aligned to within ±0.5 mm of the isocentre using room lasers. The film is exposed using the median radiosurgical collimator diameter of 25 mm. Three exposures at different MU settings are normally sufficient to establish the sensitometric curve as described in section 3.8.4.

4.3.5 Phantom Software

A software application has been developed in C++ for the MS Windows NT platform to expedite the film calibration and the processing of the multiple phantom film

* Kodak 3000 RA, Eastman Kodak Company
images, thereby minimizing intervention required by the user. The tasks performed by this application are shown within the dashed box of Figure 4.6. Since four phantom films are digitized in a single pass of the scanner, the software first locates the edge of each film using an edge-detection algorithm and extracts individual film images. The indexing mark (produced as described in section 4.2) in each image is then located by software, and the film images are automatically oriented, ordered and displayed for inspection by the user (Figure 4.10).

One or more sets of multiple film images (where each set corresponds to a different angular orientation of the cassette box) are stored in a three-dimensional matrix, and a bicubic interpolation technique is used to re-sample this matrix at a spatial frequency specified by the user. This algorithm provides estimates of the dose at specified locations within the measurement volume by performing a local fit of a two-dimensional bicubic polynomial to measured dose values. This approach was deemed appropriate for this application since: i) the fitted surface is constrained so that it passes directly through measured points and ii) the fitted surface will be smooth and continuous across measured dose points. The details of the bicubic interpolation algorithm are provided in Appendix B.

For a single set of adjacent films, this technique yields an ordered set of images containing i) the original, directly-measured film images and ii) a series of interpolated images. While this interpolation technique does not increase the physical spatial sampling frequency of the phantom system perpendicular to film planes, it does serve as a means of merging multiple sets of films corresponding to different angular orientations and to smooth dose distributions in the planes perpendicular to the film planes. Pixel values in each image are then converted to absolute dose by inverting the equation of the curve fitted to the sensitometric curve. The global maximum dose within this set of images is determined and used to normalize each image into a series of user-specified isodose gray-levels. Figure 4.11 shows the effect of increasing the number of interpolated slices on the images corresponding to the low-resolution planes (i.e. perpendicular to the films). The accuracy of the interpolation technique is addressed in section 6.3.2.
Figure 4.10 Graphical User Interface (GUI) showing film planes following orienting and ordering of film images. The indexing marks are visible on the left edge of each image.
Each isodose image is then converted to 512x512 DICOM format and L-shaped fiducial marks are written into each image to define the edges of the film region to indicate the location of the reference planes shown in Figure 4.5. A DICOM header is written into each file to encode the image order, image pixel size, slice thickness and slice separation so that the measured dose volume is imported by the treatment planning software with the correct spatial scaling. The software processing and export of a set of 20 film images requires approximately 5 minutes using a Pentium 333 MHz processor. The majority of this duration is required by the bicubic interpolation routine.

4.3.6 Co-registration of planned and measured dose volumes

The treatment planning fusion tools are used to match the two reference planes of the dose volume (as defined in Figure 4.5) with the corresponding bottom and vertical inner surfaces of the cassette box. As illustrated in Figure 4.12, this co-registration is verified by matching the L-shaped fiducial marks in measured dose volume with a second set of marks on the inner surface of the cassette box. The latter are produced by attaching adhesive vinyl tape to the inner walls of the cassette box prior to CT imaging. Fine control of the co-registration is achieved by aligning the fiducial marks with the box corners in magnified views of multiple axial, coronal and sagittal planes.
Figure 4.11 Measured 90%, 80%, 50% and 30% isodose levels for an ellipsoidal dose distribution in planes perpendicular to the planes of the films. Shown is the effect of increasing the number of interpolated planes between two adjacent measured planes for a single set of 19 sagittal films. From top to bottom, the number of interpolated planes is 0, 1, 2 and 3, resulting in a total of 18, 37, 55 and 73 images, respectively. The dose distribution was produced by a geometric arrangement of five static, circular fields.
Figure 4.12 Co-registration of the measured volume and the planned CT set. For a set of films in sagittal planes, the L-shaped fiducials in the measured volume image set are matched to fiducial marks on the posterior and inferior inner surfaces of the cassette box. The measured volume has been deliberately offset in these images. This process is demonstrated above in a) a sagittal plane and b) an axial plane. Magnification permits co-registration to within ±0.63 mm the CT set (± pixel in the CT set), as shown for an axial plane in c).
5 Results I: Radiographic film dosimetry for 6 MV radiosurgical beams

The first part of this chapter presents the results\textsuperscript{112} of the experiments described in Chapter 3 for the quantification of dependencies of emulsion sensitivity on depth, field size and orientation. These data are interpreted using the results of a Monte Carlo study of the variation of the photon spectra with depth in phantom. Finally, with the aim of improving the technique to provide greater spatial resolution, the final section of this chapter presents the data illustrating the performance of a CCD-based digitizer.

5.1 Dependence of film sensitivity on depth

Figure 5.1a) shows the sensitometric curves obtained for the large (20 cm x 20 cm) 6 MV photon beam at depths of 1.0 cm, 10.0 cm and 20.0 cm in a Solid Water\textsuperscript{TM} phantom. For each depth a curve was fitted to the measured sensitometric data using the single-target/single-hit theory equation\textsuperscript{106}:

\[ OD = OD_{sat} (1 - 10^{-\alpha D}) \] \[ 5-1 \]

where \( OD \) and \( D \) are the measured optical density and given dose, respectively. The saturation density of the film, \( OD_{sat} \), was estimated by delivering a large dose (500 cGy) to a film in phantom and was held constant in the fitting algorithm. The fitting parameter \( \alpha \) represents emulsion sensitivity. Figure 5.1b) shows sensitometric curves corresponding to the same depths for the radiosurgical (2.5 cm-diameter) field. For this
Figure 5.1 Sensitometric curves for 1.0, 10.0 cm and 20.0 cm depths in Solid Water™ phantom for a) large (20 cm x 20 cm) and b) radiosurgical (2.5 cm diameter) fields.
small field the curves agree to within the reproducibility of film development and scanning (discussed further in section 5.4, below.)

The fitted curves using the single hit / single target equation correspond to the data closely, in all cases resulting in coefficient of correlation (R) values greater than 0.9957. By determining the value of $\alpha$ for each curve, a parameter representing film sensitivity was obtained as a function of depth as illustrated for the 20 cm x 20 cm square and 2.5 cm diameter circular fields is shown in Figure 5.2. As expected from the disparity in the sensitometric curves in Figure 5.1a), the film sensitivity for the large (20 cm x 20 cm) field increases systematically with depth. While small fluctuation of the values of $\alpha$ is apparent for the radiosurgical field, no systematic variation of sensitivity with depth is evident.

![Figure 5.2 The curve-fitting parameter $\alpha$, representing emulsion sensitivity, as a function of depth for the 20 cm x 20 cm radiotherapy field and the 2.5 cm diameter radiosurgical field.](image)
5.2 Dependence of film sensitivity on field size

Figure 5.3 shows the measured sensitometric curves for the range of field sizes used typically for linear accelerator-based stereotactic radiosurgery. These data are shown for a typical depth of 10.0 cm. The fitted curves are approximately congruent, indicating that there is no significant variation in sensitivity with field size over this range. Because the films were irradiated in the parallel orientation in this experiment, similar sensitometric curves could be generated for each of the three field sizes at any depth from the surface to 20.0 cm in phantom at 0.1 cm intervals. The congruence of sensitometric curves was observed for all depths in phantom.

Figure 5.3 Comparison of sensitometric curves for field sizes over the range used in radiosurgery. Dose was measured at a depth of 10 cm in phantom.
5.3 Variation of film response with orientation

Figure 5.4 compares the sensitometric curves obtained with the film plane positioned parallel and perpendicular to the central axis. The maximum discrepancy between these curves is 1.5%.

![Figure 5.4 Dependence of sensitometric curve on the orientation of the film plane relative to the central axis for a 2.5 cm diameter 6 MV photon beam.](image)

5.4 Dosimetric reproducibility

A subset of the film reproducibility data is shown in Figure 5.5 which compares the optical density recorded on six films exposed identically and processed in three batches of two films. Each batch was processed on a separate day over the course of approximately six months. At a depth of 10.0 cm, the maximum “intra-session” variation in optical density between films processed in one batch is 0.85% (where “variation” is defined by \([\text{OD}_{\text{max}} - \text{OD}_{\text{min}}] / \text{OD}_{\text{mean}}\)). The maximum “inter-session” variation in optical density between films processed on separate occasions is 2.3%. Using the fitted
sensitometric curve shown in Figure 4, this translates into intra-session and inter-session variations in the calculated dose of 1.1% and 2.9%, respectively. The component of this variability arising from the digitization process alone was found to depend slightly on optical density, ranging between 0.1% and 0.5% over the full dynamic range of the densitometer.

Figure 5.5 A subset of the reproducibility data showing the optical density measured along the central axis of a radiosurgical beam for six different films processed in three separate sessions. For a typical radiosurgical depth in phantom of 10.0 cm, the inter-session and intra-session variation in optical density, as quantified over the course of six months, are 2.3% and 0.85%, respectively.
5.5 Monte Carlo study of radiographic film response

5.5.1 Variation of photon energy spectra with depth in phantom

Figure 5.6a) shows the total (primary plus scattered) photon spectra for a large (25 cm-diameter) field on central axis at depths of 1.0 cm, 5.0, and 20.0 cm. Each spectrum has been normalized to its own maximum to illustrate relative spectral changes with depth. These data show a consistent increase in the lower energy (< 400 keV) component of the spectrum with depth due to relative increases of scattered photons reaching the scoring voxels. Changing only the field size in the simulation to a diameter of 2.5 cm resulted in the spectra shown in Figure 5.6b). In contrast to the large-field case, the spectra for the radiosurgical field show a negligible increase in the low-energy population of photons.

5.5.2 Applicability of the published incident spectrum

Figure 5.7 compares the simulated percent depth dose curves with those obtained by physical measurement. Note that the simulated percent depth dose for the 25.0 cm diameter field is compared to the measured percent depth dose for the equivalent square field size of 22.2 x 22.2 cm². While the Monte Carlo percent depth dose values for both beams are slightly lower than the measured data, at depths greater than the build-up region, the agreement is reasonable. The disparity in the buildup region is likely due to the absence of electron contamination in the incident beam, but will not be important for the simulations conducted in this work.
Figure 5.6 6 MV photon energy spectra shown at depths of 1.0 cm, 5.0 cm and 20.0 cm in phantom for a) the 25.0 cm diameter field and b) the 2.5 cm diameter field. The spectra were scored in 0.1 MeV energy bins below 1 MeV to observe spectral variation in this low-energy region.
Figure 5.7 Comparison of Monte Carlo simulated and measured percent-depth dose curves for validation of the incident 6 MV spectrum.

5.6 Characteristics of a CCD-based scanner for digitization of radiographic film

5.6.1 Spatial linearity

As described in 3.8.2, the analysis of the digitized distortion phantom image produces a vector whose elements give the spatial differences between the detected and true locations of dots in the distortion phantom. These deviations from spatial linearity are presented as frequency histograms in Figure 5.8 for the x- and y-dimensions. The majority of deviations from linearity in both the x- and y-dimensions of the scanner are in
fact smaller in magnitude that the scanning pixel size (0.25 mm). Table 5.1 summarizes these results.

<table>
<thead>
<tr>
<th>Absolute value of deviation from linearity (δ)</th>
<th>Percent of dots / x-dimension</th>
<th>Percent of dots / y-dimension</th>
</tr>
</thead>
<tbody>
<tr>
<td>δ &lt; 0.1 mm</td>
<td>83.1 %</td>
<td>86.9 %</td>
</tr>
<tr>
<td>0.1 mm &lt; δ &lt; 0.2 mm</td>
<td>13.1 %</td>
<td>10.6 %</td>
</tr>
<tr>
<td>δ &gt; 0.2 mm</td>
<td>3.8 %</td>
<td>2.5 %</td>
</tr>
</tbody>
</table>

### 5.6.2 Dynamic range

Figure 5.9 shows the relationship between the optical density of the film and the scanned 12-bit pixel value. The curve is characterized by a limited linear region for low optical densities (below approximately 0.4 OD) but is nonlinear throughout the remainder of the optical density range. While the slope of this curve remains positive throughout the manufacturer's cited dynamic range of 0.14 to 3.4 OD, the scanner becomes relatively insensitive to variations in optical density above 1.5 to 2.0 OD. The flattening of this curve at greater optical density values has important consequences in terms of the uncertainty in measured absolute dose, as quantified in section 5.6.4.

### 5.6.3 Calibration for absolute dosimetry

The pixel value ($P$)-to-dose ($D$) calibration curve for the transparency digitizer is given in Figure 5.10. The solid curve fitted to these data is similar in form to that used for the Wellhöfer digitizer in equation [5-1], with an additive fitting parameter, $P_{\text{offset}}$, to account for a background scanned pixel value level:

$$P = P_{\text{sat}}(1 - 10^{-\alpha D}) + P_{\text{offset}}$$  \[5-2\]
Figure 5.8 Histograms of spatial discrepancies between the detected and true locations of dots in the distortion phantom for the x-dimension (along the CCD array) and y-dimension (in the direction of travel).
Figure 5.9 Variation of 12-bit pixel value with absolute optical density for the transparency digitizer.

Figure 5.10 The measured sensitometric curve for the transparency digitizer.
5.6.4 Noise in digitized film images

As illustrated in Figure 5.11, up to an optical density of approximately 2.0, the coefficient of variation decreases rapidly with increasing optical density (and therefore with increasing mean pixel value). This noise in digitized images ultimately determines the uncertainty in the measurement of absolute dose. In the absence of noise, the relationship between the scanned pixel value and measured dose would be single-valued. However, as is evident in the measured calibration data in Figure 5.10, a given dose may in fact result in a range of scanned pixel values. Thus in applying the sensitometric curve to determine an unknown dose corresponding to a certain scanned pixel value, the uncertainty in the absolute dose may be estimated by the horizontal width of the measured sensitometric curve for a given pixel value. This uncertainty also depends on the dose itself because the slope of the curve decreases with increasing dose. The variation in the uncertainty in dose as a function of the dose is shown in Figure 5.12.

![Figure 5.11 Dependence of the coefficient of variation of the scanned pixel value with the measured optical density.](image-url)
5.6.5 Signal stability

The digitizer exhibits a brief warm-up period during which the signal varies significantly and systematically over time. Figure 5.13 shows the variation of the scanned pixel value during this transient period for three points along the central axis of the digitized percent depth dose film. These results indicate that, depending on the optical density of the film being scanned, failure to warm up the digitizer may result in underestimation of the pixel value by as much as 7%. After approximately four scans the signal stabilizes (where approximately 15 seconds are required per scan). Analysis of the 60 sequential scans of the OD step pattern (after the warm-up period) indicated that the coefficients of variation of the mean values of each step ($\sigma / \mu \times 100\%$) do not exceed 0.7%.

Figure 5.12 Dependence of the uncertainty in measured dose obtained from the sensitometric curve as a function of the magnitude of the dose.
Figure 5.13 The warm-up effect of the digitizer over the first several scans. Scanned pixel values are shown for three nominal optical densities on the film and have been normalized to their stable values.

5.6.6 Effect of scattered light

Figure 5.14 illustrates that for the apertures consisting of acetate transparency (OD=0.11), there is a systematic variation of the average pixel value scanned at the centre of the aperture width. The pixel value decreases (i.e. towards lighter shades of gray) by approximately 0.9% per millimetre increase in aperture width. In contrast, for the base+fog apertures (OD=0.21), no significant dependence of the scanned pixel value on the aperture width is apparent. The half-width of the error bars in Figure 5.14 correspond to the standard deviation of the pixel values along the centre-line of the aperture.
Figure 5.14 Dependence of the scanned pixel value on the width of the aperture for apertures made of acetate transparency (OD 0.11) and film (base+fog, OD=0.21). Scanned pixel values for both aperture types have been normalized to those measured for the smallest aperture (1 mm wide in both cases.)

5.6.7 Uniformity

The low-pass filtered image of the “uniform” film is shown in Figure 5.15. The gray-level window was set to emphasize non-uniformities in the digitized image. Clearly, there is a systematic variation in scanned pixel value around the edges of the scanner. The magnitude of non-uniformity was reasonably low, however, with the global coefficient of variation of pixel ($\sigma/\mu \times 100\%$) equal to 2.1%. A vertical profile through the centre of this image is shown in Figure 5.16. Re-scanning of the film after rotating it through 180° shifted the majority of non-uniform features in the image accordingly—indicating that these variations were in fact contributed largely by the film itself.
Figure 5.15 Gaussian-filtered image of the unexposed, processed film, shown with a gray-level window setting to emphasize the low-frequency spatial non-uniformities. The coefficient of variation for the entire field of view was 2.1%.

Figure 5.16 Vertical profile through the centre of the image in Figure 5.15.
6 Results II:

A system for three-dimensional dose verification in stereotactic radiosurgery

6.1 Monte Carlo validation of the multiple-film geometry

6.1.1 Perpendicular beam orientation

Figure 6.1 compares the homogeneous and multiple-film phantoms in terms of dose to the water slabs prior to and within the multiple-film cassette, for the perpendicular beam orientation and the smallest inter-film spacing (0.5 mm). As expected, the presence of this large number of closely-spaced films results in a significant increase in attenuation by the phantom compared with the homogeneous case, as reflected in the steeper curve in the region of the cassette box. Figure 6.2 illustrates that the maximum dose perturbation is reduced from 2.5% for the 0.5 mm inter-film spacing to 1.3% and 0.8% for the 1.0 mm and 3.2 mm spacing, respectively.

6.1.2 Parallel beam orientation

For the parallel beam orientation, the perturbation caused by the presence of the films is most pronounced across the central 2 cm of the profile, which is shown in Figure 6.3 for the three different values of inter-film spacing. As for the perpendicular beam orientation, the perturbation of dose is greatest for the 0.5 mm inter-film spacing and reaches 2.1% for the spacers closest to the central axis of the beam. This perturbation is markedly diminished for the 1.0 mm spacing. For the 3.2 mm spacing, any disparity between the multiple-film and homogeneous cases is below the statistical uncertainty of the simulation (< approx. 0.6%).
Figure 6.1 Comparison between the homogeneous and multiple-film phantoms in terms of Monte Carlo calculated absolute dose absorbed by the water slabs prior to and within the cassette-box for the 0.5 mm inter-film spacing and the perpendicular beam orientation.
Figure 6.2 Comparison between the homogeneous and multiple-film phantoms in terms of Monte Carlo calculated dose to the inter-film water spacers for perpendicular beam orientation. For the 0.5 mm, 1.0 mm and 3.2 mm inter-film spacing, the absolute dose differs by a maximum of 2.5%, 1.3% and 0.8%, respectively.
Figure 6.3 Comparison between the homogeneous and multiple-film phantoms in terms of Monte Carlo calculated dose to the inter-film water spacers for parallel beam orientation. For the 0.5 mm inter-film spacing, the absolute dose differs by as much as 3.2%. For the 3.2 mm spacing, no statistically-significant dose perturbation is apparent.
6.2 Experimental validation of the multiple-film geometry

6.2.1 Perpendicular beam orientation

For the perpendicular beam orientation, the optical density produced on a film located below 17 films and 18 polystyrene spacers was compared to that produced on a film placed at the same depth in a homogeneous phantom in which the layers of film had been replaced with polystyrene. The scanned pixel values within a 1 cm × 1 cm region at the centre of the spot pattern showed no change due to the presence of the film outside of the uncertainty due to digitization. This result was also checked by using a sensitive densitometer* to compare the optical densities at several locations on the spot-pattern. This second measurement indicated a small change due to the presence of the multiple layers of film, and the optical density corresponding to the multiple-film case was lower by 0.9%. Although this disparity is small, it was observed to be systematic throughout the central region of the spot-pattern recorded on the film.

6.2.2 Parallel beam orientation

The Monte Carlo results in Figure 6.3 for the perpendicular beam orientation indicate that the dose perturbation is most pronounced near the central axis. Therefore, measuring the perturbation of optical density recorded on a film at the centre of an array of 19 films and aligned with the central axis should represent a worst-case arrangement.

For both the single- and multiple-film cases, a two-dimensional pattern through the central-axis of the beam was obtained (i.e. in the x-z plane Figure 4.2) and profiles were digitized across the beams at each of 120 depths ranging from 5.5 cm to 17.5 cm. In Figure 6.4, these profiles are compared at example depths of 7.0 cm, 11.5 cm and 15.5 cm in phantom. No significant disparity between the single and multiple film optical density patterns were observed for the entire range of depth studied.

* Model 331, X-Rite, X-Rite Inc., Grandville, MI, USA
Figure 6.4 Comparison between profiles of optical density recorded on a film aligned with the central axis for the single-film (solid line) and multiple-film (dashed line) cases. Shown above are profiles at depths of a) 7.0 cm, b) 11.5 cm and c) 15.5 cm in phantom.
6.3 Evaluation of the phantom system

6.3.1 Circular-field treatment verification

The accuracy of circular-beam treatment is well-established (compared to conformal radiosurgical treatments, for example) and circular beam arcs are used regularly for clinical treatment. During the commissioning of the radiosurgery program at the Vancouver Cancer Centre, the absolute dose profile of circular beam treatment was verified to within 2% of treatment planning dose calculations using thermoluminescent dosimeters (TLDs). TLDs were also used to verify the absolute dose distribution at several locations in the dose distribution for a circular-beam five-arc treatment. Thus, a conventional circular-beam treatment serves as a suitable benchmark for the phantom system and a simple circular beam test case was established in the treatment planning system. The plan was comprised of five 30 mm diameter fields arranged symmetrically by using the couch and gantry combinations listed in Table 6.1 (see Figure 3.1 for coordinate definitions).

<table>
<thead>
<tr>
<th>Beam #</th>
<th>Couch angle (°)</th>
<th>Gantry angle (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>90</td>
<td>90</td>
</tr>
<tr>
<td>2</td>
<td>45</td>
<td>45</td>
</tr>
<tr>
<td>3</td>
<td>45</td>
<td>135</td>
</tr>
<tr>
<td>4</td>
<td>315</td>
<td>225</td>
</tr>
<tr>
<td>5</td>
<td>315</td>
<td>315</td>
</tr>
</tbody>
</table>

Prior to treatment planning, the phantom was imaged using CT without a phantom lesion (i.e. the cassette box was filled with water only). In the treatment planning software, a 30 mm diameter spherical planning target volume (PTV) was contoured at the approximate centre of the cassette-box volume and the isocentre was located at the center-of-mass of this simulated lesion. The isodoses in the treatment plan were...
normalized to the maximum dose of 0.48 Gy. This arrangement results in a range of delivered absolute dose values which are below the saturation region of the sensitometric curve in Figure 5.10. The treatment was administered to the phantom with a single set of 19 films in the sagittal planes of the phantom. The films and scanned images were analyzed as described in Chapter 4, and the measured isodose surfaces were exported to the treatment planning system as a series of 73 DICOM images (i.e. with three interpolated slices between each measured plane).

The measured maximum dose for this treatment was 0.49 ± 0.01 Gy. Figure 6.5 gives the comparison of the planned isodoses (white lines, calculated for each beam using equation [1-1]) and measured isodoses (grayscale levels) following co-registration of the measured dose volume. This three-dimensional co-registration permits the user to browse through the entire dose volume to view the comparison in arbitrary axial, sagittal or coronal planes (Figure 6.5 shows example images taken directly from the treatment planning display). Throughout the dose volume, the planned and measured 90%, 80% and 70% isodose surfaces agree to within 1.5 mm. While the 50% isodoses agree throughout most of the volume, disagreement of up to 2.5 mm was observed in some regions (for example, see the central panel of Figure 6.5). This mismatch at the lower isodose level has been observed by other investigators for the same treatment planning system and may arise from the fact that in regions of lower dose gradient, a small mismatch of measured absolute dose results in a significant spatial shift between planned and measured isodoses. Another explanation may be the lack of modeling of scatter in the dose calculation of the treatment planning system.140
Figure 6.5 Comparison between the 50%, 70%, 80% and 90% isodoses calculated by the treatment planning system (shown as white lines) and the corresponding measured isodoses (shown as grayscale levels) for the circular-field test case.
6.3.2 Evaluation of the interpolation algorithm for typical dose distributions

As mentioned in section 4.3.5, the phantom system facilitates re-sampling of the dose volume with sets of films in orthogonal planes, and the subsequent merging of these planes to form a three-dimensional map of administered dose. The motivation for this would be to improve the spatial sampling of the dose perpendicular to the films. However, since the exposure of two sets of films nearly doubles the time required for the procedure, it was of interest to evaluate the accuracy of the bicubic interpolation algorithm for the calculation of the dose values between the measured planes resulting from a single set of films.

This assessment was done by scanning a number of one-dimensional profiles through a typical two-dimensional conformal dose distribution recorded on film at a "low" spatial frequency of \((3.2 \text{ mm/pixel})^{-1}\) (i.e. the spatial frequency of measurement perpendicular to the film planes in the phantom system.) This "under-sampled" data set was then provided as input to the cubic interpolation algorithm to generate interpolated profiles at a higher spatial frequency of \((0.254 \text{ mm/pixel})^{-1}\). Next, the same profiles were re-digitized at a "high" spatial frequency of \((0.254 \text{ mm/pixel})^{-1}\) (i.e. the in-plane spatial frequency for the phantom system.) Figure 6.6 shows a comparison of the high-frequency \((0.254 \text{ mm/pixel})^{-1}\) digitized profiles and the interpolated profiles. The spatial agreement is within 1.0 mm throughout the volume, even in regions where abrupt changes in gradient occur.
Figure 6.6 Estimation of the accuracy of the bicubic interpolation for a typical dose distribution. The interpolated profiles (dashed line) are compared to "true" profiles scanned at a spatial frequency of 1/0.25 mm\(^{-1}\) (solid line) at locations indicated by a), b) and c).
6.4 Clinical application of the phantom system

To date, the phantom system described above has been used for several treatment verification studies at the Vancouver Cancer Centre. This section highlights two example applications that are representative of the possible roles of the technique. The first is a verification of a clinical conformal treatment, and the second describes the testing of a new commercial system for head and neck radiosurgery.

6.4.1 Verification of a conformal radiosurgical treatment

The results of the benchmark test in section 6.3.1 indicate that the planned and measured dose distributions agree to within the expected numerical and spatial tolerances. Based on these findings, the phantom technique was extended to conformal radiosurgical treatment verification. The accuracy of conformal treatment is less established since it depends on i) a different treatment planning dose calculation algorithm and ii) a new method for administering the treatment (i.e. with poured conformal collimators).

A three-dimensional verification was performed for the first single-fraction conformal radiosurgery at the Vancouver Cancer Centre. The lesion was an acoustic neuroma approximately 4.0 cm across. In replicating the patient’s treatment on the phantom, the beam apertures and orientations were copied from the patient’s treatment plan and transferred to the phantom treatment plan (Figure 6.7). The isocentre was located at the approximate centre of the cassette-box. The planned isodoses were normalized to a maximum dose of 0.49 Gy. The measured maximum dose for this treatment was 0.50 ± 0.01 Gy. Figure 6.8 gives the comparison of planned and measured isodoses in axial, coronal and sagittal planes for the conformal treatment of the lesion. The planned dose delivered by each beam was calculated by the treatment planning software using equation [1-5].
Figure 6.7 The arrangement of beams for the conformal treatment a) shown as a three-dimensional diagram and b) as viewed from the superior of the head.
The planned and measured 80% prescription isodose surfaces agree to within 1.5 mm throughout the treatment volume. However, in several regions of the dose volume the mismatch was more significant, reaching 3.5 mm.

To isolate whether this mismatch was due primarily to mechanical or algorithmic (i.e. dose calculation) errors, the manufactured beam aperture shapes were compared to the planned aperture shapes to highlight differences due to manufacturing uncertainty (e.g. limited accuracy in cutting the mould insert or positioning the insert on the mould key). The manufactured aperture shapes were obtained by two-dimensional film measurements at isocentre. For each field the position of the room lasers was marked on the film to permit alignment of the measured and planned isocentre locations. Most collimator shapes agree within the expected tolerance in hot-wire cutting of Styrofoam (approximately 1.5-2 mm at isocentre). However, one Styrofoam mould insert (corresponding to collimator number two in Figure 6.9) had apparently suffered a rotation on the mould key prior to pouring the cerrobend, producing spatial mismatches in the projected light field of up to 4 mm at isocentre*. By replacing all six of the planned shapes in the treatment planning system with the actual manufactured shapes, a new planned dose distribution was calculated. The new dose distribution reduced, but did not eliminate, the majority of the largest (i.e > 3 mm) mismatches of the 80% and 90% isodose lines. Figure 6.10 illustrates this effect for an axial plane corresponding to the beam's-eye-view of aperture number two. For this plane, re-entering the manufactured aperture shapes has reduced the largest mismatch between planned and measured 80% and 90% isodoses by approximately 2 mm. In general, spatial shifts in the isodose lines were generally smaller in magnitude than the physical modifications to the collimator aperture in the treatment plan. This is to be expected since the total distribution results from the sum of distributions from six beams and not all of the six aperture shapes in the plan were modified significantly. The remaining mismatch (below 1.5 mm) throughout

* In fact this error had been observed just prior to the patient treatment by comparing the conformal light field with the field perimeter printed on the target positioner box film. The phantom study was performed to quantify the effect of this error on the overall treatment accuracy.
the volume is within the expected mechanical tolerance of treatment delivery and phantom repositioning.

Figure 6.8 Comparison between the 70%, 80% and 90% isodoses calculated by the treatment planning system (shown as white lines) and the corresponding measured isodoses (shown as grayscale levels) for the six-field conformal treatment.
Figure 6.9 Planned aperture shapes (white outlines) and manufactured aperture shapes (dark outlines) at isocentre.
Figure 6.10 a) The maximum mismatch observed between planned and measured isodoses for the conformal treatment was 3.2 mm. b) After correcting the planned field shapes to match the shapes of the manufactured collimators, this mismatch was significantly diminished to 1.2 mm, indicating i) the significant effect of error in collimator manufacture and ii) the sensitivity of the phantom technique to inaccuracy in beam-shaping. Note that this set of isodoses corresponds to an axial plane, which was perpendicular to the planes of the films.
6.4.2 Verification of a system for head and neck radiosurgery

The phantom system was used in the evaluation of a prototype system* for stereotactic radiosurgery or radiotherapy treatment of lesions in the head and neck. The system consists of a carbon fibre tray and a moulded vacuum cushion for fixation of the torso of the patient. The patient’s head is fixed during imaging and treatment using a thermoplastic mask system. This fixation is similar to that shown in Figure 4.7, except additional support is provided to restrict motion of the shoulders. Stereotactic localization is facilitated by fiducial marks on a localizer box similar to that used in regular radiosurgery, except that it is elongated to span a greater distance in the cranial-caudal dimension. As in the conventional (head only) radiosurgical system, printouts generated by the treatment planning system are attached to this same box for alignment of the patient on the linear accelerator couch.

Because this system involves both untested hardware (e.g. the fixation system and new localization box) and software (i.e. the dose calculation algorithm), the phantom system was used to verify the following:

i) Correct transfer of stereotactic coordinates from treatment planning system to the linear accelerator;
ii) Accuracy of the isodose surfaces; and
iii) Accuracy of the absolute dose delivered.

After forming a thermoplastic mask for the phantom, it was imaged using the standard CT scanner settings (see section 4.3.1). The treatment plan consisted of an arrangement of five circular beams since the intention was to test the head and neck system without incurring error introduced by conformal collimation. The beams were incident as shown in Figure 6.11 in order to produce an asymmetric dose distribution. This asymmetry would permit identification of possible mirroring errors between the planned and delivered dose distributions.

* BrainLAB, AG, Munich, Germany
As shown in example axial, sagittal and coronal planes in Figure 6.12, throughout the majority of the dose volume, the 70%, 80% and 90% planned isodose surfaces are in good spatial agreement with the measured isodose surfaces. The prescription (80%) isodose surfaces agreed to within 1.5 mm throughout the majority of the volume and mismatches did not exceed 2.4 mm. As discussed for the circular field treatment in section 6.3.1, the spatial mismatch increases in regions where the dose gradient is low. The planned maximum absolute dose of 0.45 Gy (occurring at the location indicated by the asterix in the central coronal plane in Figure 6.12) was in agreement with the measured absolute dose of 0.46 ± 0.01 Gy.

Figure 6.11 The beam arrangement for the verification of the head and neck radiosurgical system a) shown as a three-dimensional diagram and b) as viewed from the superior of the head.
Figure 6.12 Comparison between the 30%, 70%, 80% and 90% isodoses calculated by the treatment planning system (shown as white lines) and the corresponding measured isodoses (shown as grayscale levels) for the verification of the head and neck system.
7 Discussion

7.1 Radiographic film dosimetry for small 6 MV photon beams

Although film dosimetry is most problematic for lower-energy photon beams (e.g. cobalt-60 and 4 MV) due to their lower primary-to-scatter ratio, Figure 5.1a) emphasizes the importance of corrections in converting scanned optical density to dose even for the 6 MV beam. For the 20 cm x 20 cm field, using a depth-unspecific curve may introduce an error in dose of up to 15% for the dose range examined in this study. This value is in close agreement with both analytically-modeled and experimentally-measured results reported recently in the literature for a large (i.e. non non-radiosurgical) 6 MV photon beam. Note that this error is reduced at lower doses but is confined to below 10% only for doses below 80 cGy. In contrast, for the 2.5 cm-diameter radiosurgical field, using a sensitometric curve established for a depth of 1.0 cm to calculate a dose at 20.0 cm depth would result in a maximum error of approximately 1.5%. Thus the potential error due to depth-dependence of sensitivity is minor and in fact comparable in magnitude to the reproducibility of the technique. Although the depth dependence was examined for the parallel beam orientation only, as indicated recently by Suchowerska et al. this dependence should be reduced for the perpendicular orientation, for which, compared to the parallel case, less of the dose to the film is due to photon interactions in the emulsion itself.

For the radiographic film dosimetry of 6 MV radiosurgical beams, the practical benefit of this near-invariance of the sensitometric curve is twofold. First, corrections to compensate for a depth-dependence of the film emulsion sensitivity are not required, and therefore the possibility for error in determining these corrections is removed. Second, since a single calibration curve can be used over a range of depths, the quantity of data required in calibrating the film is greatly reduced.

This measured difference between small and large photon fields in terms of the change in sensitivity with depth can be explained with reference to the variation of the photon spectrum in phantom. As depicted by the data in Figure 5.6, the photon spectra
for the large (25.0 cm diameter) and small (2.5 cm diameter) fields both exhibit variation with depth in phantom, but the nature of this variation differs markedly. For the large field, a systematic increase in the lower-energy photon population is seen, while for the radiosurgical-sized field, a gradual increase in the average photon energy is apparent. These results are supported by the simulation results by Heydarian et al.\textsuperscript{80} which indicate that the mean photon energy of a 6 MV radiosurgical beam increases with depth due to beam hardening, but for a large beam (20 cm diameter) the mean photon energy decreases with depth up to 15 cm due to an increase of phantom scatter.

In terms of the potential over-response of film emulsion, however, spectral changes below approximately 400 keV are particularly crucial. Figure 7.1 shows the 6 MV incident spectrum on the same energy scale as the mass attenuation coefficients for water, film base and film emulsion. Note that a very small component of this spectrum (< 2% of the total photon population) exists in the energy range below 400 keV. As indicated by the Monte Carlo results in Figure 5.6b) for radiosurgical fields, this low-energy tail does not increase significantly with depth in phantom. Over the range of depths of interest in radiosurgery (to 20 cm) the vast majority of photons in the spectrum exist in the energy range between 0.4 to 6.0 MeV. Thus, it is apparent that film dosimetry is particularly applicable to 6 MV radiosurgical photon beams, and may not be as well suited to either larger 6 MV beams or beams of higher or lower quality. For spectra shifted towards lower energies, predominance of the photoelectric effect would complicate or preclude the use of film due to the divergence from tissue-equivalence and a dependence of sensitivity on depth. Similar complications would be encountered for higher-energy beams, for which the approximate $Z^2$ dependence of the pair production atomic cross section\textsuperscript{142} would become important.

Due to the time-consuming nature of the Monte Carlo simulation, the study of the spectral changes in phantom was performed for the median radiosurgical collimator size of 2.5 cm only, since this beam was selected as typical for radiosurgery. Potential increased softening of the spectrum in phantom for the larger radiosurgical collimators was not investigated. However, the recent Monte Carlo simulation of linac-based
Figure 7.1 Mass attenuation coefficients of film components shown on the same energy scale as the 6 MV incident photon spectrum.

radiosurgery unit by Verhaegen et al.\textsuperscript{143} suggests that the mean photon energy at depth of dose maximum decreases by only 0.45 MeV when the aperture diameter is increased from 0.5 cm to 5.0 cm. Moreover, the experimental results in this thesis (Figure 5.3) indicate that even for the larger collimator size, the sensitometric curve does not change significantly.

For the median collimator diameter of 25 mm, the sensitometric curve varies by less than 1.5% as a function of film orientation relative to the beam axis. This is marginally greater than the intra-batch variability in optical density of 0.85%. Reports on the dependence of film response on orientation vary significantly in the literature and are somewhat difficult to compare due to differences in experimental parameters (e.g. film
type, beam quality, field size and depth in phantom). For example, Williamson observed an orientation-dependence for an older industrial-type film, reporting an increased sensitivity in the perpendicular orientation for depths in phantom below approximately 19 cm, beyond which the dependence reverses. Cadman observed a slight increase in response for CEA TVS film for a 10×10 cm² field and Co-60, 6 MV and 18 MV photon beams. A similar observation was made by Oldham for a 10×10 cm², 6 MV beam and X Omat V film. Cheng and Das have reported a slight increase (<2%) in the perpendicular orientation sensitivity for a 10 cm × 10 cm² 18 MV beam. It should be noted that all of these results were obtained with photon field sizes considerably larger than those used in stereotactic radiosurgery. In contrast, the results found in this thesis most closely agree with the recent work by Burch et al. who noted no significant orientation dependence for a smaller photon field (6 cm × 6 cm) for Kodak XV-2 film. It may be the case that for larger fields, the greater relative population of low-energy scatter in the beam augments the orientation dependence.

One of the common reservations concerning the practical use of radiographic film in dosimetry is that considerable time and attention is necessary to establish the sensitometric curve. Figure 5.1 and Figure 5.3 illustrate that the curve is approximately linear for doses below 40 cGy. As shown by Evans and Schreiner, restricting the dose to this region would offer the advantage of direct conversion of optical density to dose for relative dosimetry. The sensitometric curves for the large field in this thesis (Figure 5.1a) indicate that this would also minimize the error caused by the variation of sensitivity with depth, and thus would alleviate the need to acquire a family of calibration curves over a range of depths. However, in measuring radiosurgical dose distributions, often it may be difficult to confine the total dose to the linear region. For example, for multiple convergent-arc techniques, a minimum achievable monitor unit-per-degree rate is imposed by the linear accelerator, and thus the total dose may necessarily exceed 40 cGy. Similarly, treatments involving multiple, static conformal fields would necessitate a very low dose per field, and thus the uncertainty inherent in the dose per beam may become

* Kodak RPM-2 type M
appreciable. Therefore, to facilitate dosimetry over a larger dose range, we generally acquire sufficient calibration data to extend the curve-fitting over the non-linear region.

It should also be noted, however, that the non-linear shape of the sensitometric curve is specific to Kodak X Omat-V film and may be significantly affected by the digitizer used (evident by comparing Figure 5.1 and Figure 5.10). Several investigators\textsuperscript{103,145} have recently reported on a relatively new type of silver-halide radiographic film* characterized by a linear sensitometric curve up to 90 cGy, followed by abrupt saturation. For this film, Cheng and Das\textsuperscript{103} have shown that the relationship between optical density and dose does not vary for 4-18 MV photon spectra, but possible variation of sensitivity with depth in phantom has not been examined to date.

The low variation of emulsion sensitivity with depth and field size for radiosurgical 6 MV beams will offer an advantage in terms of dosimetric accuracy only if the variation due to processing conditions does not prevail. In this study the developer temperature was held constant to within ±0.5° C and the replenishment rate was kept as continuous as possible. However, the data in Figure 5.5 suggest that in order to achieve a variation of < 2% in measured dose, it is necessary to process calibration and dosimetric films in the same batch.

The transparency digitizer examined in this work has not been characterized in the literature to date, and provides the advantages of speed, cost-efficiency and the flexibility to scan even small films on the glass bed. The spatial resolution exceeds that required for radiosurgical dosimetry and thus provides a solution to the volume averaging problem introduced by the scanning infrared densitometer\textsuperscript{118} described in Chapter 3 as well as common dosimeters such as micro-ionization chambers,\textsuperscript{75,78,147} diodes\textsuperscript{81,82,83} and diamond detectors.\textsuperscript{148} Figure 5.8 indicates that, at a spatial sampling frequency of (0.25 mm/pixel)$^{-1}$, the magnitude of spatial distortion in the digitized image is below ±0.1 mm over the majority of the field of view. As shown in Figure 5.13, the signal is consistent between scans to within 0.7%, following a warm-up period of four scans. This variation is below variability between processed films as indicated in section 5.4. It is apparent in Figure 5.14 that for extreme gradients in optical density, (i.e. 3.0 OD to 0.11 OD, as for

* CEA TVS, CEA America, Ltd., Houston, TX, USA
the acetate film apertures) the scanned pixel value will depend on the optical density of adjacent regions of the film. This parasitic effect may be caused by peripherally-scattered light which is detected after being transmitted through the aperture. However, this dependence will not be present when digitizing radiographic film because this effect vanishes when the minimum aperture optical density is increased to that of the film base-plus-fog level. The disappearance of this dependence may result from attenuation of the intensity of scattered light to a level within the noise of the CCD array.

The principal disadvantage of the digitizer is its narrower dynamic optical density range, as compared to the infrared densitometer used for the film sensitivity studies described in Chapter 3. The effect of this limitation is a significant reduction of the range of measurable dose. As is evident in Figure 5.9 and Figure 5.10, the shape of the sensitometric curve is significantly affected by the digitizer itself, rather than by the response of the film to photon irradiation. This drawback is further emphasized by comparing the sensitometric curves for the infrared densitometer (Figure 5.1) with the transparency digitizer (Figure 5.10).

As was indicated by the manufacturer (Table 3.1), the digitizer remains at least somewhat sensitive to optical densities reaching 3.4 OD. However, the flattening of the digitizer response results in prohibitively high uncertainty in absolute dose above optical densities of 1.5-2.0 OD. This is summarized by Figure 5.12, which shows that in order to limit the uncertainty in the dose measurement to below 2%, it is necessary to confine the dose to below 50 cGy. Even for relative dosimetry, the increase in uncertainty is important because the location of generated isodoses are sensitive to the absolute dose to which the distribution is normalized, particularly in regions of low dose gradient.

In summary, to date the use of radiographic film for dosimetry of non-radiosurgical radiotherapy photon beams has been limited by dependencies of emulsion sensitivity on depth in phantom, field size and orientation. In contrast, this work shows that the inherent high primary-to-scatter ratio of 6 MV radiosurgical beams, particularly in the energy range below 400 keV, results in a minimization of these dependencies. Variation in film sensitivity with depth (from 0 to 20 cm), with field size (from 10 mm to 40 mm diameter) and with film orientation introduces potential errors comparable in magnitude to that incurred due to processing variability (1-2%). Thus, with the provision
of a high spatial-resolution digitizer for the readout of the spatial distribution of optical density, radiographic film provides an effective solution for the dosimetry of 6 MV radiosurgical beams.

7.2 A system for three-dimensional dose verification in radiosurgery

The simulation results for the perpendicular beam orientation in Figure 6.2 indicate that for the 6 MV radiosurgical beam there is a small yet systematic perturbation of dose due to the presence of the multiple layers of film. Note that estimating the distribution of the dose at the interface of even a single inhomogeneity using analytical methods is complicated due to the combined effects of multiple-photon and electron scatter. It is for this reason that the Monte Carlo method is an invaluable tool. However, a simplified explanation for the initial rise in dose to the inter-film spacers can be understood by considering the effect of replacing a thin layer of water with a thin layer of emulsion. First, suppose that adjacent to the first inter-film spacer there is a layer of water with linear attenuation coefficient \( \mu_{\text{water}} \) and thickness \( dx \). Since this layer is very thin, for a beam of \( N_0 \) incident photons incident normally, the change in the number of photons in the beam will be \( dN_{\text{water}} = -N_0 \mu_{\text{water}} dx \) and thus the number of interactions in the layer is \( |dN_{\text{water}}| \). For the 6 MV spectrum the majority of these interactions will be Compton scattering events, each of which will set a recoil electron into motion. The majority of recoil electrons will be forward-directed (i.e. towards the adjacent inter-film water spacer down-stream). If the thin water layer is then replaced by emulsion, the number of interactions will be \( |dN_{\text{emulsion}}| \) where \( dN_{\text{emulsion}} = -N_0 \mu_{\text{emulsion}} dx \). Note that the ratio of \( \mu_{\text{emulsion}} / \mu_{\text{water}} \) and therefore \( dN_{\text{emulsion}} / dN_{\text{water}} \) depends on the photon energy, but as can be seen in Figure 2.6, over the majority of the 6 MV spectrum, the difference between the mass attenuation coefficient \( (\mu/\rho)_{\text{emulsion}} \) and \( (\mu/\rho)_{\text{water}} \) is minimized. Thus, considering the ratio of the density of emulsion to that of water (Table 2.1), \( \mu_{\text{emulsion}} \approx 2.2 \mu_{\text{tissue}} \). Therefore, by replacing the thin layer of water with emulsion, the number of interactions in the thin layer, and thus the number of electrons ejected from this layer into

\[ (*) \text{Assuming } dN << N, \text{ the solution to this differential equation gives the familiar exponential attenuation relationship, } N = N_0 e^{-\mu x}. \]
the adjacent inter-film water spacer, will increase by a factor of approximately 2.2. This
will cause an increase of the dose to the inter-film spacer downstream, but the fractional
increase of dose (i.e. of the total) will be small since the emulsion is very thin (12 μm)
and the majority of electrons in motion in the spacer will in fact originate from the water
spacer or film base rather than from emulsion.

This reasoning also points to the competing effect of increased attenuation of the
photon beam by the presence of multiple film layers. This is reflected in the steeper
to of the multiple-film dose curves in Figure 6.2 and also in the eventual crossing of
the homogeneous and multiple-film dose curves.∗

The useful result of this set of simulations, however, is that for an inter-film
spacing as low as 1.0 mm, the perturbation of the dose is confined to below 2%. For the
actual value of $t_{\text{spacer}}$ used in manufacturing the phantom cassette box, the dose
perturbation is below 1% for both the parallel and perpendicular beam orientations. In
addition to preserving tissue equivalence, this spacing was chosen based on practical
considerations since it results in a maximum of 20 films for the 7 cm box width, and thus
the amount of time required for processing, scanning and data analysis is reasonable.

The minimal dose perturbation facilitates accurate measurement of absolute dose
in verifying clinical radiosurgical treatments. This is evident in the results of the circular-
beam benchmark test of the phantom system described in section 6.3.1 in which the
planned and measured maximum doses in the distribution agreed to within 2%. This is
within the dose uncertainty of the film/transparency digitizer combination as quantified in
section 5.6.4. The measured prescription isodose surface (in this example, 80% of the
maximum dose) also agreed to the corresponding planned surface within the required
tolerance specified for stereotactic radiosurgery1 throughout the volume of the simulated
spherical lesion.

The reasonable accuracy of the interpolation technique may be attributed to the
fact that the dose distributions are smooth and do not contain many high frequency details
that would be aliased by the low spatial-frequency (3.2 mm/pixel)∗1 sampling. This is

∗ Although this explanation serves to explain the general shape of the dose perturbation, it is greatly
simplified compared to the realistic case. In reality the dose will also be affected by both multiple photon
scattering and electron backscatter.
typical for most clinically relevant distributions. Even sharp features (e.g. spikes or protrusions) in the physical collimator shape result in much lower spatial-frequency variations in isodoses at depth in phantom. Hence, while the user has the option of resampling the dose volume with film sets at different orientations, for typical dose distributions a single set of films will facilitate accurate verification, even in interpolated dose images.

The clinical applications presented in Chapter 6 illustrate that the phantom system provides a means of comparing planned and measured isodoses and thus gives an indication of the aggregate spatial error arising from some or all of a number of sources including the finite CT scanning resolution, algorithmic limitations in dose calculation and beam shaping. As illustrated by conformal verification example, the technique appears to be sensitive to small spatial inaccuracies of the delivered dose.

Although the endpoint of the verification technique provides a measurement of the cumulative spatial error, in some cases it may be possible to retrospectively determine the dominant sources of error incurred in the process. This was illustrated in the example of the conformal treatment for which the effect of collimator manufacturing inaccuracy was quantified in terms of the spatial change in the isodose surfaces. Table 7.1 suggests the methods that could be followed to isolate the effect of other sources of error.
### Table 7.1 Techniques to isolate specific sources of error in radiosurgery using the phantom system.

<table>
<thead>
<tr>
<th>Source of error</th>
<th>Suggested technique</th>
</tr>
</thead>
</table>
| **Localization error** (i.e. discrepancy between the planned isocentre location and the centre of the radiation field) | 1. Use head-ring fixation to minimize positioning error  
2. CT image phantom with a small target (e.g. 5 mm diameter sphere)  
3. Locate sphere in treatment plan, place isocentre at centre  
4. Irradiate using a single circular beam with  
i) gantry at 90° and film in sagittal plane  
ii) gantry at 0° and film in coronal plane  
5. Co-register and compare centre of measured dose with isocentre |
| **Effect of inaccuracy in conformal dose calculation**   | 1. Verify the localization error as above  
2. Use head-ring to minimize positioning error  
3. Verify manufactured conformal beam apertures using 2-D film measurements  
4. Administer conformal treatment, co-register to observe inaccuracy in isodose surfaces |
| **Effect of conformal aperture manufacturing inaccuracy** | 1. Verify localization error and accuracy of conformal dose calculation as above  
2. Use head-ring fixation to minimize positioning error  
3. Administer conformal treatment, co-register to observe inaccuracy in isodose surfaces |

The alternative methods presently available for three-dimensional radiosurgical dose verification are those employing ferrous sulfate or polyacrylamide (PAG) gel dosimeters. As mentioned in Chapter 2, several recent studies are available describing the use of gels for the measurement of conformal radiosurgical dose distributions. In comparing the current work to these techniques, it is important to consider differences in the dimensions and spatial frequency of measurement voxels. To date, gel dosimeter verification studies of radiosurgical dose distributions have employed MRI for the quantitative measurement of dose, and therefore the spatial resolution is ultimately limited by the imaging voxel size. Figure 7.2 compares the geometry of the gel/MRI technique with that of the current film phantom. The relevant dimensions of the voxels and voxel spacing are compared in Table 7.2. In all cases, the spatial sampling of the dose volume for the gel studies is limited by the voxel separation in the axial dimension of the scanner. In fact, the measured dose is averaged over the volume of one measurement voxel and therefore the measurement suffers from the same volume

* to within ~1.0 mm
The smallest voxel used to date was in the recent study by Meeks et al.\textsuperscript{100} (1.56 mm $\times$ 1.56 mm $\times$ 3 mm). This group reported that the image slice separation ($t_{sep} = 6$ mm) necessarily restricted the dose comparison to two dimensions, and that the 1.56 mm $\times$ 1.56 mm pixel size ultimately limited the accuracy of the comparison in the region of the high dose gradient of the radiosurgical distribution. Moreover, for the voxel size used, the imaging noise was significant, resulting in jagged isodose lines. This necessitated spatial smoothing of dose images prior to comparison with the planned distribution. Possible improvement of the signal-to-noise ratio by image averaging was precluded by the lengthy scanning time required and the resultant high associated cost.

Table 7.2 Voxel dimensions (as defined in Figure 7.2) for studies of gel verification of conformal radiosurgery compared to the multiple-film phantom system.

<table>
<thead>
<tr>
<th>Study</th>
<th>$t_{sep}$</th>
<th>$t_{thick}$</th>
<th>$t_{pixel}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Meeks et al.\textsuperscript{100} (1999)</td>
<td>6 mm</td>
<td>3 mm</td>
<td>1.56 mm</td>
</tr>
<tr>
<td>Ibbott et al.\textsuperscript{97} (1997)</td>
<td>unspecified</td>
<td>3 mm</td>
<td>1.9 mm$^*$</td>
</tr>
<tr>
<td>Chan et al.\textsuperscript{95} (1995)</td>
<td>10 mm</td>
<td>7 mm</td>
<td>1.2 mm</td>
</tr>
<tr>
<td>Current work (single film set, no interpolation)</td>
<td>3.2 mm</td>
<td>0.20 mm</td>
<td>0.25 mm</td>
</tr>
<tr>
<td>Current work (single film set, with interpolation)</td>
<td>0.85 mm$^+$</td>
<td>0.20 mm</td>
<td>0.25 mm</td>
</tr>
</tbody>
</table>

* based on field of view of 24 cm and 128 pixel smallest image dimension.

$^+$ With three interpolated images between measured planes. Note, however, that the estimated accuracy of the interpolation algorithm in dose fall-off region was estimated to be approximately 1.0 mm (Figure 6.6).
Figure 7.2 Comparison of the Fricke and PAG gel + MRI dose readout geometry (top) with the film phantom technique (bottom) in terms of the size and spatial frequency of measurement voxels. The values of $t_{sep}$, $t_{thick}$ and $t_{pixel}$ for the reported PAG gel dose verification studies are summarized in Table 7.2.
In comparison to the results of these previous studies, the film phantom technique provides significantly smaller measurement voxels (0.25 mm × 0.25 mm × 0.20 mm). Without interpolation, the separation between measured voxels \((t_{sep})\) is smaller almost by a factor of two. Moreover, the fact that the dose distribution is sampled by an array of almost point-measurements (0.01 mm\(^3\) voxels) rather than volume averages within comparatively large voxels facilitates accurate interpolation between planes.

The PAG gel study by Ibbott indicated that the uncertainty in absolute dose measured using MRI readout is 3.4%. For the film phantom system, this uncertainty increases with the magnitude of the dose itself. Since this effect is mainly caused by the non-linearity of the CCD-based scanner, it may be remedied by using a laser scanning densitometer, for example. Nevertheless, for the applications presented in Chapter 6, it was possible to limit the uncertainty to 2%.

It should be emphasized that while the current technique offers superior spatial resolution and measurement precision for the measurement of three-dimensional radiosurgical distributions, the presence of radiographic film limits the technique to beams with high primary-to-scatter characteristics (particularly below 400 keV). Gel dosimeters, are inherently more versatile and can also be used for large photon beam dosimetry.

Finally, one of the intended features of the film phantom system was time-efficiency. The capacity to perform a dose verification within a single evening, for example, would be useful to verify a conformal stereotactic radiotherapy treatment between fractions or for periodic quality assurance. This has been achieved mainly through the development of software for automating the film calibration, image digitization, image processing and export of the measured dose volume to the treatment planning system. The typical time required for a complete dose verification (including all of the steps in Figure 4.6) is approximately four hours (Table 7.3).

In summary, we have found the multiple-film phantom technique to be accurate, versatile, cost-effective and practical. As has been illustrated by verifications of radiosurgical dose distributions performed to date, the improved spatial resolution of the technique permits detection of even minor spatial inaccuracies in treatment delivery.
Table 7.3 Approximate time required for dose verification using the film phantom technique.

<table>
<thead>
<tr>
<th>Time (minutes)</th>
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</thead>
<tbody>
<tr>
<td><strong>Phantom setup / CT imaging of phantom</strong></td>
</tr>
<tr>
<td><strong>Treatment planning (typical)</strong></td>
</tr>
<tr>
<td><strong>Film loading, phantom setup, treatment delivery</strong></td>
</tr>
<tr>
<td><strong>Processing of films</strong></td>
</tr>
<tr>
<td><strong>Digitization of films</strong></td>
</tr>
<tr>
<td><strong>Software analysis of films / export of DICOM images</strong></td>
</tr>
<tr>
<td><strong>Software co-registration</strong></td>
</tr>
<tr>
<td><strong>Total</strong></td>
</tr>
</tbody>
</table>

* This time is highly variable and depends on the complexity of the treatment plan.
† In practice, the majority of this time is spent setting up the phantom on the linear accelerator (i.e. aligning the isocentre position using room lasers) and the actual duration depends on the patient positioning system used. If a supply of pre-cut film is kept on hand, the loading of the cassette box requires only 10 minutes.
8  Conclusions and future directions

8.1.1 Conclusions

Stereotactic radiosurgical dosimetry requires an integrating detector with the capability of high-resolution measurement in the presence of high dose gradients and lateral electronic disequilibrium. With the emergence of conformal radiosurgical techniques, the additional requirement of three-dimensional dose measurement is becoming increasingly important.

Radiographic film dosimetry provides the advantages of very-high spatial resolution, low cost, good signal-to-noise characteristics and accessibility, but its application to the dosimetry of large (non-radiosurgical) photon beams is complicated by a variation of film sensitivity with changes in the low-energy region of the photon spectrum. This thesis has presented results illustrating that depth- and field size-dependencies of film sensitivity are minimal over the range of field sizes used in radiosurgery. For large (e.g. 20 cm × 20 cm) 6 MV beams, errors of up to 15% may be encountered by using a depth-unspecific sensitometric curve. Although a minor systematic depth-dependence also is seen for radiosurgical beams, the potential error incurred due to this effect is below 1.5% for field sizes ranging from 1.0 cm to 4.0 cm in diameter. Monte Carlo simulation has been used to demonstrate that this reduced dependence can be attributed to a smaller increase of scattered photon population below 400 keV with depth for radiosurgical beams. The sensitometric curve varies negligibly over the range of radiosurgical field sizes (1-4 cm diameter) and is consistent to within 1.5% as a function of film orientation. Therefore, this analysis has shown that radiographic film dosimetry is particularly suited to 6 MV radiosurgical beams since a single calibration applies to a wide range of measurement conditions. In the context of radiosurgical dosimetry, this is a useful finding, because, in comparison to other radiotherapy treatments, radiosurgery demands the capability for accurate, high spatial resolution measurement that is uniquely provided by radiographic film.
Spatial resolution in film dosimetry may be compromised by densitometers with large aperture diameters (e.g. > 1 mm). This thesis has evaluated a relatively new CCD-based transparency digitizer providing sub-millimetre spatial-resolution for radiosurgical film dosimetry. This investigation has indicated that the spatial distortion (below ±0.1 mm over >80% of the field of view) signal-to-noise (σ/μ ~ 1-2%) and dynamic range (0.2 – 2.0 OD) characteristics are suitable for radiosurgical dosimetric applications.

The second part of this thesis described the development, relevant dosimetric considerations and applications of a novel phantom system for the measurement of three-dimensional radiosurgical dose distributions. Although the multiple-film geometry is a relatively simple design, it has not been widely employed in radiotherapy, because the presence of low-energy phantom scatter in larger photon fields would result in a significant perturbation of the deposited dose. The Monte Carlo study performed in this work indicates that, for the 6 MV radiosurgical beam, the dose perturbation is below 1% for the 3.2 mm inter-film spacing and increases to a maximum of 3.2% for the 0.5 mm spacing. This low perturbation was confirmed through experimental measurement, which demonstrates that the increased perturbation by 18 films (compared to one film) was below 1% for both parallel and perpendicular beam orientations. The 3.2 mm spacing was chosen in fabricating the phantom system since it results in i) acceptable dose perturbation, ii) reasonable spatial resolution and iii) a practical number of films for rapid dose verification.

With dedicated software, this system provides a relatively rapid and practical means of directly comparing planned and measured three-dimensional dose distributions. Currently, the automated software performs the dose calibration (i.e. to establish the sensitometric curve from calibration films), orients and orders film images and generates a volume of isodose surfaces by performing an interpolation based on the user’s specifications. Thus, the output includes i) the maximum measured dose (in Gy) in the volume (which can be compared with the corresponding planned value) and ii) a series of CT-format DICOM isodose images to be exported to the treatment planning system. It is also conceivable, however, that instead of isodose levels, absolute measured dose (i.e. in Gy) could be encoded within these DICOM images. Tools in the treatment planning software allow display of the planned dose value and CT Hounsfield unit at arbitrary
locations, and therefore direct comparisons of absolute dose could be made throughout the volume.

8.1.2 Future directions

As a practical means for verifying three-dimensional dose distributions in radiosurgery, the phantom system is novel and significant in several respects. First, the technique provides a relatively rapid and automated method of measuring a complex dose distribution, and immediate co-registration with the planned distributions. Second, as discussed in the previous chapter, this method provides improved spatial resolution compared with alternative techniques discussed in the literature. Third, it is very cost-efficient and accessible in terms of both the dosimeter used and the method used to read out the dose.

We are currently developing a second prototype of the phantom hardware and will implement several changes to the design. The phantom will extend further toward the inferior for simulation of head and neck radiosurgical treatment. A smaller cassette box volume (7 cm × 7 cm × 9.5 cm) will be used to allow improved mobility of the box inside the phantom volume. The cassette box will also be locatable off-axis (by means of a locking pivot) for simulation of radiosurgical treatment throughout the cranial volume. The new cassette box and spacers will be manufactured using 3 mm-thick Solid Water™ material instead of polystyrene.

The means of detecting the dose distribution, with the use of an array of radiographic verification films, will remain unchanged. It should be emphasized that, in this work, the focus was placed on radiographic film rather than radiochromic film because of the high cost and significant image non-uniformity of the latter. The manufacturing challenges in applying a uniform monomer emulsion are being addressed, however, and therefore from a technical standpoint radiochromic film may become an increasingly attractive dosimeter for stereotactic radiosurgery. The use of radiochromic film in the phantom would lift the constraint of small field sizes due to the

* Because Solid Water™ is opaque to visible light (whereas polystyrene is translucent), for the fiducial-marking of film edges, a narrow (0.5 mm-wide) channel is cut into one vertical wall of the cassette box. This channel is then filled with a tissue-equivalent, translucent white epoxy.
near tissue equivalence of the film. The technique could also be employed for non-stereotactic applications with minimal modification to the hardware and software components. The cost of radiochromic media remains high, however, and currently the film is only commonly available in 12.7 cm × 12.7 cm sheets. Although multiple-film phantom dose verification with radiochromic film was not described in this thesis, we have conducted a feasibility study. To lower the total cost of the film, we used smaller inner cassette-box dimensions (6 cm × 6 cm× 6 cm) so that four film pieces could be cut from a single sheet. Initial investigations have indicated that that radiochromic film can also be read out using the transparency digitizer described in this work, but that the dynamic range is reduced in comparison that obtained using a He/Ne laser densitometer, as suggested by published models. A precedent for digitization of radiochromic film using a broadband light source and CCD detector is available in the literature, and the device used in this previous study offered a lower dynamic range and image depth than the digitizer described in this thesis.

The phantom system has been used over the past several months in our radiotherapy clinic and example applications have been described in Chapter 6. Based on initial benchmark results and more realistic, conformal applications, this technique was shown to facilitate the detection of small discrepancies between planned and measured distributions. It is likely that this technique will find continued application in three roles in the clinical setting: i) commissioning radiosurgical systems or additions to hardware (e.g. beam shaping devices) and software (e.g. dose calculation algorithms); ii) periodic quality assurance, particularly for conformal radiosurgery; and iii) quantitative dose verification of individual, conformal patient treatments, especially in cases where target volumes are proximal to organs of risk (e.g. brainstem or optic pathway).

In addition to these more routine clinical tests, the phantom system described in this thesis may prove to be valuable in the evaluation and development of new, advanced radiosurgical techniques. At the time of writing, the field of radiosurgery is expanding rapidly on several fronts. As was mentioned in Chapter 1, a currently active area of research focuses on improved dose conformity with the use of micro-multileaf dynamic beam shaping. With the goals of improved dose homogeneity across the target volume, better sparing of adjacent tissues, conformal treatment of larger lesions and improved
automation of treatment delivery, this approach promises to receive continued attention. The development of such a technique usually requires significant advances in both hardware and software (i.e. for treatment planning). The phantom system would provide a means for rapid and accurate evaluation of both of these components, and is sufficiently practical to permit fast and direct comparison of results as the technique is established.

Also of current interest is the expansion of the range of treatable indications using radiosurgery by extending the technique to extracranial sites. For example, the millimetre-accuracy of radiosurgery could be harnessed in the treatment of lesions in the head and neck region, lung, or at sites proximal to the spinal column. However, establishing extracranial radiosurgery as a feasible technique poses an array of future challenges. These include, for example, patient immobilization or tracking during treatment, linear accelerator gating (e.g. to reduce the effect of patient motion), localization of organs inside the body (which may be more mobile than the brain) and dose delivery proximal to air cavities. If the method of radiation delivery for extracranial treatment is similar to that used currently in linac-based radiosurgery (i.e. small 6-10 MV photon beams), the multiple-film technique could be used in this context as well, for comparison of planned and delivered dose distributions in terms of both location and shape.

Radiographic film has been used by several investigators to measure one-dimensional absorbed dose profiles proximal to phantom inhomogeneities, and air cavities can produce significant perturbation of the dose deposited by radiosurgical photon fields, in orientations both parallel and perpendicular to the beam axis.154 Although various analytic inhomogeneity-corrections are available for dose calculation, the accuracy of such methods is often limited, particularly when the beam and air cavity are comparable in size.70 The potential accuracy of radiographic film for small photon beam dosimetry, combined with the phantom system described in this work, suggests the possibility of high-resolution, three-dimensional measurement of the dose perturbation caused by air cavities. This investigation would become increasingly relevant as conformal radiosurgical treatment becomes feasible in the head and neck region (e.g. for treatment of nasopharyngeal carcinomas and lesions on the pharyngeal wall or proximal to sinuses).
9 Appendix A: Definition of dose functions

9.1 Total scatter factor

The total scatter factor\textsuperscript{119}, $S_t(A)$, is the ratio of the dose on central axis at a reference depth, $d_{ref}$ in phantom for a field size $A$ to the dose at the same location in phantom for a reference field size, $A_{ref}$, as illustrated in Figure 9.1. Thus,

$$S_t(A) = \frac{D(A)}{D(A_{ref})}$$ \hspace{1cm} [9-1]

\textbf{Figure 9.1 Phantom geometry for measurement of total scatter factor.}
The dose distributions shown in Chapter 6 were generated by a treatment planning system using equation [1-5], for which circular-field total scatter factors were measured with $d_{ref} = d_{max} = 1.5$ cm and $A_{ref} = 10 \text{ cm} \times 10 \text{ cm}$. The total scatter factor rises with field size due to the increase in scatter from the linear accelerator collimators and in phantom scatter reaching the point of measurement.

9.2 Percent depth dose

The percent depth dose\(^{155}\) gives the ratio of dose at depth $d$ on axis in phantom to the dose at depth $d_{max}$ also on axis in phantom. As shown in Figure 9.2. Both dose measurements are made with the same field size $A$ at the surface of the phantom source to surface, SSD. The ratio is expressed as a percentage:

$$P(d, A, SSD) = \frac{D(d, A, SSD)}{D(d_{max}, A, SSD)} \times 100\%$$

![Figure 9.2 Phantom arrangement for measurement of percent depth dose.](image)

[9-2]
9.3 Tissue-Maximum Ratio

As illustrated in Figure 9.3, the tissue-maximum ratio is a special case of the tissue-phantom ratio,\(^{156}\) where the reference depth is \(d_{\text{max}}\). \(TMR(d,A)\), is the ratio of dose at a depth \(d\) in phantom at isocentre level for a field size \(A\) to the dose at depth \(d_{\text{max}}\) in phantom, also at isocentre level and for the same field size:

\[
TMR(d,A) = \frac{D_{\text{iso}}(d,A)}{D_{\text{iso}}(d_{\text{max}},A)} \quad [9-3]
\]

![Diagram of tissue-maximum ratio](image)

Figure 9.3 Phantom geometry for measurement of tissue-maximum ratio.

9.4 Off-axis Ratio (OAR)

The off-axis ratio\(^{157}\) for a field size \(A\), \(OAR(r',A)\), gives the ratio of the dose at a reference depth \(d_{\text{ref}}\) and at a radial distance \(r=r'\) from the central axis (point \(P\) in Figure 9.4) to the dose at the same depth on central axis \((r=0, \text{point } Q)\):
\[ OAR(A, r') = \frac{D(d_{\text{ref}}, A, r = r')}{D(d_{\text{ref}}, A, r = 0)} \] 

Figure 9.4 Arrangement for measurement of off-axis ratio.
Appendix B: Bicubic interpolation algorithm

As described in section 4.3.5, a bicubic interpolation technique was employed for the determination of the dose at any point within the volume of the cassette box, based on measured dose values in multiple planes. The algorithm has been adapted from that presented by Press et al.\textsuperscript{138}

One or more sets of films (i.e. where each set consists of multiple films in sagittal or coronal planes) are used to measure the absolute dose at locations in a three-dimensional Cartesian grid. Suppose that, in an axial plane, the dose has been measured at each of the points marked with "•", in Figure 10.1, below.

\begin{figure}
\centering
\begin{tikzpicture}
\draw[->,thick] (0,0) -- (8,0); \node at (4,-0.5) {$\hat{x}$};
\draw[->,thick] (0,0) -- (0,8); \node at (-0.5,4) {$\hat{y}$};
\draw[thick] (0,0) -- (8,0) -- (8,8) -- (0,8) -- cycle;
\filldraw (0,0) circle (2pt); \node at (0,0) {P1};
\filldraw (8,0) circle (2pt); \node at (8,0) {P2};
\filldraw (0,8) circle (2pt); \node at (0,8) {P3};
\filldraw (8,8) circle (2pt); \node at (8,8) {P4};
\node at (0.5,4) {x_0}; \node at (0.5,0) {x_0};
\node at (4.5,0) {x_1}; \node at (0,4.5) {y_0};
\node at (7,4.5) {x_2}; \node at (4,7) {y_1};
\node at (7,7) {x_3}; \node at (8,7) {x_3};
\draw[thick] (0,0) -- (8,0) -- (8,8) -- (0,8) -- cycle;
\end{tikzpicture}
\caption{Grid of measured dose points (black dots) used as input to the bicubic interpolation algorithm.}
\end{figure}
The algorithm estimates the value of the dose \( D(x_p, y_p) \) at an arbitrarily-specified point \( P \) based on the values and the curvature of the dose surface at the four surrounding points. For each point, the partial derivatives and cross derivative with respect to \( x \) and \( y \) are computed and stored. For example, at point \( P1 \),

\[
\begin{align*}
D\big|_{P1} &= D(x_i, y_i) \\
D_x\big|_{P1} &= \frac{D(x_2, y_i) - D(x_0, y_i)}{2sx} \\
D_y\big|_{P1} &= \frac{D(x_i, y_2) - D(x_i, y_0)}{2sy} \\
D_{xy}\big|_{P1} &= \frac{D(x_2, y_2) + D(x_0, y_0) - D(x_2, y_0) - D(x_0, y_2)}{4sxsy} 
\end{align*}
\]

The spatial distribution of the dose is estimated as by an interpolation function \( f(t,u) \) of the scaled coordinates \( t \) and \( u \) where

\[
\begin{align*}
t &= \frac{x_p - x_i}{sx} \\
u &= \frac{y_p - y_i}{sy}
\end{align*}
\]

such that \( f(t,u) \) is a third-order polynomial in both \( t \) and \( u \):

\[
f(t,u) = c_0 + c_1t_0 + c_2u_0 + c_3t_0u_0 + c_4t_0^2u_0 + c_5t_0u_0^2 + c_6t_0^2u_0^2 + c_7t_0^3u_0^2 + c_8t_0u_0^3 + c_9t_0^2u_0^3 + c_{10}t_0^3u_0^3 + c_{11}t_0u_0^4 + c_{12}t_0^2u_0^4 + c_{13}t_0^3u_0^4 + c_{14}t_0u_0^5 + c_{15}t_0^2u_0^5
\]

The coefficients \( c_0 \) to \( c_{15} \) of this function can be determined and will describe a function which i) is continuous across the measured points and ii) with derivatives exactly equal to those at measured points on the grid. Determining the values of the coefficients requires the solution to the following system of sixteen equations:
This computation is expedited by recognizing that the leftmost quantity is a 16x16-element Vandermonde matrix for which a fast solution technique exists (the solution is provided in detail by Press et al.\textsuperscript{158}) After the coefficients have been determined, the dose value \( D(x_p, y_p) \) is determined by substituting them back into the bicubic formula, i.e.:

\[
D_{p} = \begin{bmatrix}
1 & x_p & y_p & x_p^2 & y_p^2 & x_p y_p & x_p^3 & y_p^3
\end{bmatrix}
\begin{bmatrix}
c_0 \\
c_1 \\
c_2 \\
c_3 \\
c_4 \\
c_5 \\
c_6 \\
c_7 \\
c_{15}
\end{bmatrix}
\]

Note that for points on the edge of the measured volume (e.g. point \( Q \) in the figure), it will not be possible to determine all of the required derivatives in equations [10-1]. For such points, the required dose value outside the volume is taken to be equal to that at the edge. Points on the edge of the volume are usually well outside the high dose-gradient region and therefore errors resulting from this assumption are likely.
minimal. Artifacts in interpolated dose points proximal to the edge of the measured volume have not been observed in practice.

It would also be possible to extend the bicubic interpolation technique to three dimensions to estimate an unknown dose value surrounded by measured dose values on the eight corners of a parallelepiped. However, this would require a significant increase in computation time.
References


62 BrainLAB AG, Dose algorithm manual, revision. 3.5, 1999.


