PORTAL SCATTER TO PRIMARY DOSE RATIO OF 4 TO 18 MV PHOTON SPECTRA INCIDENT ON HETEROGENEOUS PHANTOMS

By

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ABSTRACT

Electronic portal imagers designed and used to verify the positioning of a cancer patient undergoing radiation treatment can also be employed to measure the in vivo dose received by the patient. This thesis investigates the ratio of the dose from patient-scattered particles to the dose from primary (unscattered) photons at the imaging plane, called the scatter to primary dose ratio (SPR). The composition of the SPR according to the origin of scatter is analyzed more thoroughly than in previous studies. A new analytical method for calculating the SPR is developed and experimentally verified for heterogeneous phantoms. A novel technique that applies the analytical SPR method for in vivo dosimetry with a portal imager is evaluated.

Monte Carlo simulation was used to determine the imager dose from patient-generated electrons and photons that scatter one or more times within the object. The database of SPRs reported from this investigation is new since the contribution from patient-generated electrons was neglected by previous Monte Carlo studies. The SPR from patient-generated electrons was found here to be as large as 0.03.

The analytical SPR method relies on the established result that the scatter dose is uniform for an air gap between the patient and the imager that is greater than 50 cm. This method also applies the hypothesis that first-order Compton scatter only, is sufficient for scatter estimation. A comparison of analytical and measured SPRs for neck, thorax, and pelvis phantoms showed that the maximum difference was within ±0.03, and the mean difference was less than ±0.01 for most cases. This accuracy was comparable to similar analytical approaches that are limited to homogeneous phantoms. The analytical SPR method could replace lookup tables of measured scatter doses that can require significant
In vivo doses were calculated by combining our analytical SPR method and the convolution/superposition algorithm. Our calculated in vivo doses agreed within ±3% with the doses measured in the phantom. The present in vivo method was faster compared to other techniques that use convolution/superposition. Our method is a feasible and satisfactory approach that contributes to on-line patient dose monitoring.
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CHAPTER 1

INTRODUCTION

1.1 AIM OF THIS WORK

Radiotherapy portal images are analogous to planar X-ray films since the patient is positioned between the radiation source and the flat imager. The megavoltage treatment beam serves as the photon source. Portal imaging was originally developed to verify the position of a cancer patient relative to the external photon beam used for treatment. Portal images could also be used for patient dose verification immediately after treatment. Furthermore, the treatment accuracy increases when deviations between planned and measured doses are identified and corrected.

The majority of the imager signal is from primary photons that pass through the patient without interacting. The remainder of the imager signal comes from scatter. Scatter includes photons that interact one or more times within the patient as well as patient-generated electrons. Scatter is a problem when verifying in vivo patient doses from portal images since novel algorithms are required to estimate the radiation dose from scatter. The goal of the current work is to develop, validate, and apply a method to calculate absorbed dose in the patient from absorbed dose measured at the portal imaging plane.
1.2 Structure of This Thesis

This thesis contains eight chapters having the common theme, scatter dose in radiotherapy portal images. Chapter 1 provides background information on radiation therapy with photon beams, portal imaging, and sources of scatter dose in portal images. As well, the clinical linear accelerators that produce the photon beams are described and an example of the resulting dose distribution in water is presented.

A literature review on portal scatter dose estimation methods is given in chapter 2. These methods are divided into three broad categories: (i) those that use the same calculation algorithms as for dose calculation within the patient, (ii) theoretical approaches, and (iii) empirical techniques. This review focuses on the advantages and disadvantages of each method, accuracy of the resulting dose data, and ways in which the data have been used within the radiotherapy process.

Chapter 3 describes the operation and calibration of the imager used in this thesis for measurement of portal dose images. The imager is a two-dimensional array of $256 \times 256$ liquid ionization chambers and has a total imaging area of $32.5 \times 32.5 \text{ cm}^2$.

Chapter 4 presents the analytical method developed here for calculating the dose from scatter in the portal image. The analytical scatter calculation is a theoretical approach and applies the equations for photon attenuation, divergence, and Compton scatter within tissue. To illustrate the technique for portal scatter dose calculation, several examples are included to show the effect of photon energy and heterogeneous scattering objects on the scatter dose.

Validation of the analytical method for calculation of the dose from scatter is provided in two separate chapters. Chapter 5 documents comparison of the analytical results and Monte Carlo simulation data for homogeneous and anthropomorphic scattering objects. Chapter 6 gives the details of an experimental validation of the analytical method where
measurements of the scatter dose were carried out for homogeneous and heterogeneous scatterers over a wide range of radiation beam energies and beam areas. Statistical analysis of the results are included in both chapters 5 and 6.

In chapter 7 the analytical scatter method is applied to the problem of extracting the phantom dose using a pair of measured and calculated portal dose images. The method described to compute the portal dose images does not require measured data to predict the total dose at the imager. This technique is advantageous compared to empirical methods for imager dose calculation that can require a significant amount of time to measure the data needed for the scatter computation algorithm.

Chapter 8 concludes the thesis. It provides a summary of the major results and potential avenues for future work.

1.3 THESIS HIGHLIGHTS

- Monte Carlo simulation results and experimental measurements of the scatter to primary\(^1\) dose ratio (SPR) at the portal imager are reported for a wide range of clinically relevant cases.

- The equations for photon attenuation and Compton scattering were applied to analytically calculate the SPR in radiotherapy portal images.

- Analytically calculated SPRs show good agreement with Monte Carlo simulation results and experimental measurements.

- The analytical method has the following advantages:
  - since the imager dose from scatter is approximated by a uniform distribution, the calculation is only performed at the centre of the field, which reduces the

\(^1\)Primary photons pass through the object (patient) without interacting or scattering.
computation time

- the method is based on an X-ray computed tomography scan of the patient, and thus is suitable for heterogeneous cases such as the thorax
- the method does not rely on a database of scatter dose measurements
- the method can account for the detector response as a function of photon energy.

- Application of the analytical method is illustrated:

  - calculation of portal dose images normalized to the dose at the centre of the phantom. Quantitative comparison of calculated and measured portal dose images can provide a quality assessment of the treatment.
  - extraction of the in vivo phantom dose. The method for extracting the phantom dose is faster than several similar techniques.

1.4 **Accuracy Required in Radiotherapy**

Normal tissue can become nonfunctional after exposure to radiation and irradiation of normal tissue inevitably occurs during external beam radiotherapy. The probability of normal tissue complication (NTCP) versus dose follows a sigmoidal curve. This probability can also be considered as the percentage of patients who experience treatment complication versus the total dose to the normal tissue. In clinical practice, the prescribed radiation dose is chosen to correspond to an NTCP value of 5%. Since the slope of the NTCP curve versus dose is steep, accurate dose delivery is important to avoid increasing the possibility of normal tissue injury. The steepness of the NTCP curve means that changes in delivered dose of either ±10% can give marked changes in the probability of normal tissue damage.
The tumour control probability (TCP) as a function of tumour dose (that is, the percent of patients whose tumour was controlled) has also been experimentally determined to be sigmoidal. The slopes of the TCP curves are shallower than for the NTCP curves because of the radiobiological variation in the tumours for different patients (see for example reference [87]). Practical experience with radiation therapy has shown that the dose to control microscopic disease ranges from 40 to 50 Gy, while gross tumours are controlled by doses from 60 to 65 Gy.

The recommended accuracy in absorbed dose for radiotherapy varies. Mijnheer et al. [78] propose that the accuracy should be 3.5% of the combined uncertainty for random and systematic errors, given as one standard deviation. In many cases, larger values are accepted and in a few cases an even smaller value is desired. The International Commission on Radiation Units and Measurements [88] concluded that an accuracy of ±5% is needed. The ability to fulfill such accuracy requirements depends partly on the control of random and systematic errors; human, software, and hardware mistakes; and tumour dose homogeneity [28, 88]. The reproducibility of patient setup for each fraction of the irradiation is one possible source of inaccuracy. Portal imaging is used to verify the accuracy of daily setup and has the potential to verify the dose delivered to the patient.

1.5 CLINICAL LINEAR ACCELERATOR

The components of a radiotherapy linear accelerator are shown in figure 1.1 (a review of medical electron accelerators is given by Karzmark et al. [50]). In a medical electron accelerator, an electron beam is accelerated to megavoltage energies. This electron beam is then converted via the bremsstrahlung process to a photon beam in the X-ray target. The target is made of a high-atomic number metal such as tungsten to improve the

\(^2\)The unit of radiation dose is the gray, Gy, and 1 Gy = 1 J kg\(^{-1}\).
efficiency of bremsstrahlung photon production. The maximum energy of the photon energy spectrum is equal to the energy of the incident electron beam. If the energy of the electrons striking the target is 6 MeV, then the nomenclature for the resulting photon spectrum is 6 MV. Sample photon energy spectra for 6 and 18 MV beams are given in figure 1.2: these spectra were taken from [80] and [127], respectively. Movable collimator jaws define the size of the square or rectangular photon beam. The collimators, which are usually made of tungsten, attenuate the X-ray beam by several orders of magnitude and are designed to cleanly cut-off the radiation beam with minimal penumbra.

1.6 PORTAL IMAGER

A portal image is analogous to an X-ray image used for diagnosis of bone fractures. Portal images are obtained with the same photon source as for the radiation treatment: a sample image is given in figure 1.3. The main purpose of portal imaging is to guide the repositioning of the patient between treatment fractions and to ensure that the patient is positioned as intended. Portal images can be recorded on X-ray film or with a wide range of two-dimensional, electronic detectors for immediate on-line display. Compared to electronic detectors, film has several disadvantages for portal imaging. For example, the optimal film exposure dose is lower than the typical dose received by the film while a patient is treated, and therefore the treatment has to be interrupted to remove the film. As well, the time to develop and read the film prevents intervention to correct patient set-up before treatment. Finally, the X-ray image must be digitized first if image enhancement is required, while a portal image is already digitized so that image enhancement can be carried out on-line. Munro [81] and Boyer et al. [18] review electronic portal

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3The probability of normal tissue damage is decreased by delivering the radiation dose in small fractions rather than in one large dose. The total radiation dose is typically delivered to the patient in approximately one minute exposures given five days a week over five to seven weeks. Each exposure is termed a fraction, and the dose per fraction is \( \approx 1.8 \) to 2 Gy.
Figure 1.1. Cross-sectional diagram of a radiotherapy linear accelerator. Electrons, created by thermionic emission, are injected into the accelerator guide by the electron gun. Microwaves from the klystron provide power to accelerate the electrons, and the bending magnet rotates the electron beam to hit the bremsstrahlung target. The resulting photon beam is made uniform in intensity across the beam area by the flattening filter, and then collimated by primary and secondary collimators.
Figure 1.2. Photon energy spectra at the centre of the beam for linear accelerators with nominal accelerating potentials of 6 MV (—) and 18 MV (---).

Figure 1.4 shows where the portal imagers are located with respect to the patient treatment couch. This figure also defines the air gap between the patient and the imager that will be discussed throughout this work. Figure 1.5 is a photograph of the liquid matrix portal imager (this imager was used for measurement of the portal dose images discussed in chapter 7). The detector is mounted on a mechanical arm so that the imager can be retracted out of the way of the radiotherapists who set-up and treat the patient.

Electronic portal imagers were designed for imaging, rather than the measurement of radiation dose. However, several groups have applied these imagers for measurement of the dose to the patient, or in vivo dosimetry. The application of electronic portal imagers for dosimetry is reviewed in chapter 2.
Figure 1.3. A sample portal image of a head phantom taken with a photon beam for radiation therapy. The contrast in the image is lower than that for diagnostic X-ray images. To better visualize the patient anatomy the image was digitally processed to enhance the contrast. (Image reprinted from page 373 of [124], with permission from Elsevier Science.)

1.7 Patient Density Data

1.7.1 X-ray Computed Tomography

In radiation therapy, the patient’s physical density is measured in a set of axial images using X-ray computed tomography (CT). Figure 1.6 shows an example of an axial CT image of the chest. Each area element (or pixel) of the image corresponds to a volume element (voxel) within the patient. The third dimension corresponds to the slice thickness. This data is used when planning the course of radiation therapy for the patient. In this section, the method for measuring CT images and the interpretation of the measured data are briefly reviewed. An introduction to CT technology and image reconstruction methods can be found in [19].

For CT the internal structure of the patient is reconstructed from multiple projections obtained with a kilovoltage X-ray tube. During each axial slice acquisition, the X-ray
Figure 1.4. Schematic diagram of a clinical linear accelerator with a portal imager. The position of some portal imagers is fixed, while for other imagers the distance between the patient and the imager can be varied. In the current work, the size of the air gap between the patient and the imager is a key parameter for calculating the portal dose images.

tube circles the patient and X-rays that pass through the patient are absorbed by a ring of detectors surrounding the patient. The intensity of X-rays reaching the detector depends on absorption of the beam by the tissues, which the beam passes through. Image reconstruction methods are then used to analyze the multiple projections to obtain the density and position of the different structures contained within each slice.

The CT reconstruction process results in a two-dimensional matrix of numbers in the range from near 0.0 up to values near 1.0. These numbers correspond to the average linear attenuation coefficient of the tissue contained in each voxel. To display these numbers on a computer screen, they are scaled to a larger range, and also normalized to the attenuation coefficient of water:

$$CT \text{ Number} = 1000 \times \frac{\mu_{\text{pixel}} - \mu_{\text{water}}}{\mu_{\text{water}}}. \quad (1.1)$$

Water, therefore, has a CT number of zero; lung CT numbers are $\approx -800$ and bone CT numbers range from +300 to +1000. These CT numbers can be converted to phys-
Figure 1.5. The liquid matrix portal imager mounted on a Varian linear accelerator. In (a), the imager is retracted into the gantry, while (b) is a multi-exposure photograph and shows several different positions of the imager behind the patient treatment couch. An advantage of this imager for portal dosimetry applications is that large air gaps between the patient and the imager can be used, which simplifies the calculation of the imager dose from patient scatter. (Images reprinted from page 123 of [81], with permission from the W. B. Saunders Company.)

1.7.2 PHANTOMS

To calibrate radiation therapy devices and verify dose calculation algorithms, the radiation doses must be measured in settings that mimic actual treatments. This is achieved by placing the dosimeter in a volume of material that is radiologically tissue-equivalent. Such a volume of material is called a phantom. The important radiological properties for tissue equivalence are scattering and absorption of ionizing radiation.

A tissue-equivalent material should have the same density and number of electrons per gram as the tissue it replaces [112, 113]. Tissue-equivalence between a given tissue and tissue substitute may be verified by comparing the density, mass energy absorption
coefficients, $\mu_{ab}/\rho$, and restricted collision stopping powers, $L/\rho$, of the two media for all energies of interest. Water is readily available, inexpensive and is an ideal phantom material to replace muscle [3, 54, 120]. Since water is fluid, computer controlled movement of the dosimeter within the phantom is possible, which saves considerable time when measuring the doses at more than one point. When verifying or commissioning dose algorithms, the doses are usually measured in a 40x40x40 cm$^3$ water phantom, called a watertank or standard phantom.

$^4$Stopping power quantifies the average energy loss of an incident charged particle per unit track length and is given the symbol $dT/dx$ (units MeV/cm). Energy loss by electrons occurs from bremsstrahlung (a radiative process) as well as ionization and excitation (collisional losses). Consequently, the total stopping power is the sum of the radiative and collisional parts:

$$\frac{dT}{dx} = \left(\frac{dT}{dx}\right)_{\text{rad}} + \left(\frac{dT}{dx}\right)_{\text{coll}}. \quad (1.2)$$

The restricted collisional stopping power, $L$, only includes collisional energy losses by the electron along its path that are below a given threshold energy. Above this threshold, the electrons that are knocked out as a result of the collision create tracks of their own called delta rays. The mass restricted collisional stopping power is the ratio of the restricted collisional stopping power $L$ and the mass density, $\rho$. 

---

Figure 1.6. X-ray computed tomography images of the mediastinum and pulmonary parenchyma. The images in (a) and (b) are the same but use different gray-scale ranges to view the various tissues. In (a), the fat, lymph nodes, and vascular structures in the mediastinum can be differentiated. In (b), the fine details of the pulmonary parenchyma can be seen.
Suitable tissue substitutes for muscle, lung, bone, bone marrow, and other tissues have been designed and verified [21]. A heterogeneous phantom can be as simple as a stack of rectangular slabs of different tissue substitutes, known as a slab phantom. Anthropomorphic phantoms are designed to conform more closely to the shape and size of the standard human body [22].

For this thesis, phantom materials are of interest to ensure that the scatter model correctly accounts for the scattering and absorption in different tissues. The Monte Carlo simulation software package EGS4, which will be used later for the Monte Carlo validation of the scatter model, comes complete with files specifying the restricted mass stopping powers and mass attenuation coefficients for a variety of tissues, common phantom materials, and materials found in dosimeters [83]. For the experimental validation, tissue substitutes will be chosen from those listed in the International Commission on Radiation Units and Measurements Report 44 [21] based on their degree of tissue equivalence, availability, cost, and ease of use.

1.8 SOURCES OF SCATTER DOSE IN PORTAL IMAGES

Medical linear accelerators produce a spectrum of photon energies as described in section 1.5. The maximum photon energy in the spectrum is approximately equal to the energy of the electrons exiting the accelerator waveguide. Typical spectra for treating subcutaneous tumours have a maximum photon energy between 4 to 24 MeV. Within tissues, the most important mode of photon interaction over this energy range is by Compton scattering. At the higher photon energies, pair production also becomes important. Figure 1.7 shows the relative importance of these two interaction modes [49]. At radiotherapy energies, photoelectric and photonuclear events can be ignored [8]. Compton scattering, pair production, bremsstrahlung, annihilation, and Raleigh scattering are
Figure 1.7. Percentage of the energy transferred to water by Compton scattering (—) and by pair production (— - —) versus the energy of the incident photon. The range of incident photon energies encompasses the energies from medical linear accelerators for treating tumours located beneath the skin.

described briefly in this section as they are important for portal scatter dose studies.

1.8.1 COMPTON SCATTERING AND PAIR PRODUCTION

Compton scattering is the most important interaction between photons and soft tissue for radiotherapy. For a 2 MeV photon interacting in water, 99.3% of the total cross-section is accounted for by this type of scatter. Figure 1.8 illustrates both Compton scattering and pair production. In Compton scattering, the incident photon scatters off an atomic electron and changes direction. Some of the energy of the incident photon is transferred to the electron, ejecting the electron from the shell. The interaction cross-section per unit mass (or Compton mass attenuation coefficient) is independent of the atomic number, Z, of the scattering medium [8].
The Klein-Nishina cross-section for Compton scattering is given by:

\[
\frac{d\sigma}{d\Omega}(\theta) = \frac{r_o^2}{2} (1 + \cos^2 \theta) \left\{ \frac{1}{1 + \alpha(1 - \cos \theta)} \right\}^2 \left\{ 1 + \frac{\alpha^2(1 - \cos \theta)^2}{[1 + \alpha(1 - \cos \theta)](1 + \cos^2 \theta)} \right\} \quad (1.3)
\]

where

- $\Omega$ is the solid angle of the scattering cone
- $r_o$ is the classical electron radius
- $\theta$ is the photon scattering angle (see figure 1.8)
- $\alpha$ is the initial energy of the photon divided by the rest mass energy of the electron.

This cross-section is graphed as a function of photon scattering angle $\theta$ in figure 1.9. Equation (1.3) assumes that the electrons are free and stationary. To correct for the electron binding energy and motion of the electron, the Klein-Nishina cross-section is multiplied by the incoherent scattering function, $S$ [49].

In pair production, a photon is absorbed within the electromagnetic field of a nucleus and gives rise to an electron/positron pair. The photon requires a minimum energy
Figure 1.9. Differential Klein-Nishina cross-section versus angle of the scattered photon ($\theta$), for initial photon energies of 0.5 MeV (--), 2 MeV (- - -), and 6 MeV (•••). This shows that Compton scattered photons, at radiotherapy energies, are preferentially scattered in the forward direction.

of 1.022 MeV. At values that are well above this minimum threshold energy, the electron/positron pair are strongly forward directed. The energy of the incident photon minus the threshold energy is not necessarily split equally between the kinetic energy of the electron and positron. The mass attenuation coefficient for pair production is approximately proportional to the atomic number of the medium, $Z$ [8].

Electrons generated from Compton and pair production events within the patient can also deposit dose at the imaging plane. Figure 1.10 is a graph of the energy spectra for Compton electrons set in motion by monoenergetic photons. From figure 1.10, one can see that electrons are set in motion with energies almost as great as the maximum photon energy present in the beam. Electrons and positrons from pair production events can have energies between 0 and $(E_\gamma-1.022)$ MeV, where $E_\gamma$ is the energy of the incident photon. Electrons lose energy at the rate of $\approx 2$ MeV cm$^{-1}$ in water. Thus the highest
energy electrons will pass through up to $\approx 12$ cm of water. This was the reason for tracking the patient-generated electrons in the current study.

1.8.2 BREMSSTRAHLUNG

The electrons set in motion by Compton scattering or pair production can give rise to scattered photons through bremsstrahlung radiation. In this process, the electrons are scattered and therefore accelerated by the electrically charged nuclei and consequently radiate energy in the form of bremsstrahlung photons.

The fraction of the incident electron energy that is radiated as bremsstrahlung, called the bremsstrahlung yield, depends on the initial energy of the electron and the atomic number of the material [8]. The bremsstrahlung yield for water, which is radiologically comparable to soft tissue such as muscle, is plotted in figure 1.11. The yield for cortical bone is also shown in the graph since cortical bone has a higher effective atomic number.
Figure 1.11. *Comparison of the bremsstrahlung yield for water (—) and bone (−−−−) for incident electron energies from 0.01 to 25 MeV. Over the energy range for radiation therapy, the bremsstrahlung yield for bone is approximately 1.5 times that for water. (Data from [8]).*

than water. It can be concluded from this graph that the bremsstrahlung yield for bone is $\approx 1.5$ times the yield for water.

1.8.3 Annihilation

Positrons formed through pair production lose kinetic energy as they ionize and excite molecules along their trajectory. At the end of their track, the positron slows down enough so that electron capture can occur [8, 49]. The two particles annihilate each other, their charges are neutralized, and their masses are converted into 1.022 MeV of energy. This energy is radiated in the form of two 0.511 MeV photons that leave the annihilation site in opposite directions. If the positron annihilates in flight while it still has kinetic energy, the kinetic energy is passed on to the annihilation photons. According to Berger [9], as cited in Attix [8], the average fraction of the positron’s kinetic
energy that is converted to annihilation radiation is comparable to the fraction going into bremsstrahlung radiation.

1.8.4 RAYLEIGH SCATTERING

In Rayleigh or coherent scattering the scattered photons are forward peaked and no energy is lost by the incident photon. For low-Z materials like water, Rayleigh scatter is negligible. For example, the ratio of the coherent-to-incoherent cross-section for water at 2 MeV is only 0.0003.

Nevertheless, since Rayleigh scattering was a significant problem when validating the portal scatter model in Spies et al. [114], a brief review of the physics of this interaction is included here. The differential cross-section per unit solid angle is given by [49]

\[ \frac{d\sigma}{d\Omega} = \frac{r_0^2}{2} \left( 1 + \cos^2 \theta \right) [F(x, Z)]^2 \]  

(1.4)

where \( F(x, Z) \) is the atomic form factor, \( x = \sin(\theta/2)/\lambda \), and \( Z \) is the atomic number of the material. For small values of \( \theta \), \( F(x, Z) \approx Z \), while for large values of \( \theta \), \( F(x, Z) \) approaches zero.

The portal scatter study in [114] used copper phantoms (\( Z=29 \)) and found that coherent scatter contributed approximately 16% of the total dose on the central beam axis. In the current work, coherent scatter was neglected for the following reasons. First, the total scatter dose here is small. Second, the materials used in the current study that have high atomic numbers [bone and aluminum (\( Z=13 \))], are not as extreme as copper (\( Z=29 \)). Third, the volume of the high atomic number materials exposed to the photon beam is much less than the total phantom volume, unlike the study by Spies et al. [114] where the entire exposed volume was composed of copper.
1.9 QUANTITIES TO DESCRIBE A RADIATION BEAM

Throughout this thesis, comparison is often carried out between different methods for estimating the dose. For example, the measured dose could be compared to the dose calculated analytically or through Monte Carlo simulation. This section briefly describes two methods to present the doses for this work, as well as the normalization of this data.

1.9.1 DEPTH DOSE CURVES

If the dose is measured in a tank of water versus depth for photon beams, curves as shown in figure 1.12 result. These depth dose curves are characterized by a steep dose buildup region from the surface to the maximum dose, and then show an exponential fall-off past the depth of maximum dose. The exponential decrease occurs due to the attenuation of the primary photon beam. The reason for the rapid buildup of dose within the first few centimeters of the water surface can be explained by examining the range of the secondary electrons. Consider first the case where the electrons have a very short range, as for a very low-energy photon beam. In this situation, the dose is deposited very near the photon interaction site. For each centimeter layer in the water, the same number of photons interact (neglecting attenuation of the photon beam and photon backscatter), and all the electrons lose all their kinetic energy in the layer from where they originated. In this case, if photon attenuation is included, the maximum dose occurs at the surface. For higher energy beams, the electrons traverse several centimeters of water before coming to rest. If the same number of photons interact at each depth, within the first few centimeters, the number of electrons in motion per centimeter will increase with depth. Consequently, the dose rises. At some depth, due to the attenuation of the photon beam, equilibrium is reached between the number of electrons starting and
Figure 1.12. Graph of the relative dose measured in water as a function of depth for 6 (—) and 10 MV (- - -) photon beams. The depth of maximum dose is 1.5 cm for the 6 MV beam and 2.5 cm for 10 MV. The data is for a 10×10 cm² radiation field.

the number stopping, and the maximum dose will occur at this depth. This depth is known as the depth of maximum dose.

In figure 1.12, the curves were normalized at a depth of 10 cm. The dose from contaminant electrons created by photons interacting with components in the treatment head is no longer clinically significant at a depth of 10 cm for high energy photon beams [111], with the exception of the 50 MV racetrack microtron [40]. In our work, this normalization facilitates comparison of depth dose curves measured experimentally and calculated from Monte Carlo simulation, since the simulation omitted the contaminant electrons.

1.9.2 **Dose Profiles**

The plot of the dose as a function of distance from the centre of the beam is termed the dose profile, and a sample is shown in figure 1.13. In this case, the normalization
Figure 1.13. Relative dose versus off-axis distance from the centre of the beam in a water phantom. The data is for a 6 MV beam and a 10×10 cm² field at a depth of 10 cm.

dose is the dose on the central beam axis (zero off-axis distance) at this depth. Since the profile was calculated at the isocentre,⁵ where the beam size is defined, the 50% dose points occur at the geometric beam edge at -5 and +5 cm. The dose just outside the geometric field edges is nonzero since the electrons scatter laterally along their path and some photon transmission occurs through the collimators.

1.10 Motivation to Calculate the Imager Dose from Scatter

The slopes of the normal tissue complication probability curves are steep. Precise dose delivery for each fraction is therefore important. Portal imagers may be calibrated

⁵The isocentre is a point located at the intersection of the central beam axis and the axis of rotation of the gantry. For the linear accelerators in the current work, the isocentre is at 100 cm from the bremsstrahlung target in the treatment head.
Chapter 1. Introduction

to record the dose at the imaging plane, and this information may be used as part of the quality assurance of the treatment.

The total dose at the imager is the sum of the dose from primary (unscattered) photons and scattered particles. Chapter 2 reviews current methods for calculating the imager dose. While some centres have developed their own algorithms for computing the imager dose from scatter, solutions to this problem are still of interest since existing solutions suffer from limitations that will be discussed in the next chapter.
CHAPTER 2

LITERATURE REVIEW

An important challenge to the widespread use of on-line in vivo dosimetry with portal imagers has been the development of accurate portal scatter dose calculation methods for patients. Portal scatter estimation methods have evolved from using the same algorithms as those used for patient dose calculation [76, 130], to advanced Monte Carlo methods [26]. The physicists' time investment to implement the scatter calculation algorithm is still a significant problem, for example, using approaches that depend on a database of measured scatter doses for predicting the portal scatter dose.

In this chapter, existing scatter estimation methods are described and classified into three categories. First are those approaches that use exactly the same calculation algorithm as used for dose calculation within the patient. The second group contains those techniques based on theoretical treatment of the transport and scatter of the photons through the patient, air gap, and portal imager. The third class includes methods that are based on experimental measurements of the portal scatter dose. Within each category the review is chronological. The areas of interest include the limits, accuracy, uses, and theory. A brief background is provided at the start of the chapter to provide general information on the physical characteristics of portal scatter.
Chapter 2. Literature Review

2.1 BACKGROUND: PORTAL SCATTER DOSE

Estimates of the scatter from patients are also important when considering the construction of radiotherapy facilities, since the thickness of the concrete to shield staff and the public from unintentional irradiation depends on these scatter estimates. Reports summarizing the methods and results of scatter estimates are available and provide background information on patient scatter. When calculating shielding requirements, the patient scatter is estimated by measuring a quantity defined as the scatter fraction of dose (SF). A definition of SF is required for this discussion, however, the exact definition for SF in shielding purposes is irrelevant here. Therefore, for this study, a definition of SF is chosen that is more relevant to portal dosimetry. In portal dosimetry, the SF of dose is the ratio of the scatter dose to the total dose, both measured at the imaging plane.

The works by Taylor et al. [122] and Shobe et al. [110] are complementary Monte Carlo and experimental studies for shielding calculation that provide comprehensive SF data for a wide range of scatter angles. Although the SFs were not studied at zero degrees,¹ which would have been of interest here, several features of the author's research are worth mentioning. The maximum value of the scatter fraction occurs with the minimum amount of buildup material over the ionization chamber or scoring voxel. This occurs because of the low energy of the scattered photons as well as the presence of electrons originating from the phantom. The SF decreases with increasing thickness of material over the detection layer. This decrease is rapid at first because of the higher attenuation of the low energy scattered photons. The SF then decreases more slowly due to the attenuation of the primary and scattered photons. The variation of the SF versus buildup thickness was also reported by Droge and Bjarnard [27].

¹Zero degrees was defined as the direction of the central axis of the beam. Non-zero angles were defined by the beam central axis, the isocentre, and the (off-axis) position of the dose scoring voxel (or ionization chamber).
Chapter 2. Literature Review

Taylor *et al.* [122] computed SF data with the Integrated Tiger Series Version 3.0 Monte Carlo simulation code. This code required several hours to several days to obtain an accuracy of 4% or better when using a Hewlett Packard series 9000/model 735 UNIX workstation. They chose cutoff energies\(^2\) of 0.1 MeV for electrons and 0.01 MeV for photons. Agreement between measured and Monte Carlo results for the SF data was on average 0.83 (expressed as a ratio) with a standard deviation of 40%, which was satisfactory for shielding calculation.

Reports on Monte Carlo estimates of scatter for portal dosimetry include those by Jaffray *et al.* [47], McCurdy and Pistorius [70], Swindell and Evans [117], and Partridge and Evans [95]. Jaffray *et al.* [47] provide data for a single phantom thickness (17 cm thick polymethacrylate) and show that the SFs decrease with increasing beam energy (from 6 to 24 MV) on the central beam axis, and also show good agreement between measured and calculated SF data at 6 MV. McCurdy and Pistorius [70] reported the scatter fraction for a photon counting detector\(^3\) for singly and multiply scattered photons over a wide range of air gaps, field sizes, phantom thicknesses, and beam energies.

Jaffray *et al.* [47] showed that with no air gap, the first-order Compton scatter fluence dominated the SF of photon fluence for monoenergetic photon beams from 2 to 20 MeV. McCurdy and Pistorius [70] reported the SF of photon fluence for singly and multiply scattered photons for 6 and 24 MV photon beams and showed that first-order Compton scatter continues to dominate the SF as the size of the air gap increases.

Swindell and Evans [117] provide extensive Monte Carlo results for the portal scatter to primary dose ratio\(^4\) (SPR) for a 6 MV beam. They reported SPRs on the central axis

\(^2\)When a particle's energy falls below the cutoff energy, the particle is no longer tracked and the history is terminated.

\(^3\)For a photon counting detector, the detector response is independent of energy. The detector signal is the sum of the number of incident photons and the photon energy has no influence on the response.

\(^4\)The portal scatter to primary dose ratio is the ratio of the dose from scatter radiation to the dose from primary photons, both measured at the imaging plane.
for circular beams with areas up to 320 cm$^2$, homogeneous water phantoms ranging from 5 to 35 cm thick, and isocentre to detector distances from 10 to 100 cm. The magnitude of the SPR varied from less than 0.005 to over 0.30, and the number of photon histories required per SPR varied from several million to several hundred million. They showed good agreement between measured and Monte Carlo results, with root mean squared absolute differences of less than 0.01 (for example, the absolute difference between an SPR of 0.21 and 0.25 is 0.04). This work was extended by Partridge and Evans [95] who reported SPRs for a beam energy of 10 MV.

2.2 SCATTER ESTIMATES FROM TREATMENT PLANNING SYSTEMS

Two portal scatter calculation methods that use patient treatment planning systems$^5$ are described in this section. The first is the Delta Volume algorithm, which is based on ray-tracing the primary and scattered photon paths. The second is the convolution/superposition method that uses dose kernels computed with Monte Carlo simulation.

2.2.1 DELTA VOLUME DOSE ALGORITHM

Wong et al. [130] described the use of the Delta Volume dose calculation algorithm for computing portal dose images for heterogeneous phantoms and patients. The calculated portal doses agreed within 3% with film and ionization chamber measurements. Although only a $^{60}$Co beam was investigated, extension of the method to higher photon energies was expected to be feasible.

In the Delta Volume method [104, 129, 131], the dose computation space was divided into voxels, each of which was assigned a physical and electron density equal to the

$^5$ A treatment planning system consists of computer software that calculates the dose to the patient from a limited amount of data on the radiation beams as well as the patient contour and density data.
average for that voxel. For a $^{60}$Co beam, most of the dose deposited in a voxel comes from electrons set in motion by photons that interact (or Compton scatter) for the first time within that voxel: this component of the dose is termed primary dose. The remainder of the dose is mainly from photons that scatter first outside of the dose deposition voxel, and then interact again within the dose deposition voxel: this part of the dose is called scatter dose. The Delta Volume algorithm accounted for the dose from primary, first and second order Compton scatter, as well as photons scattered more than twice (which was classified as multiple scatter). The dose from primary, first, and second order scatter was calculated by ray-tracing through the volume and computing the photon attenuation along the ray paths. The Klein-Nishina coefficient was used to calculate the probability of Compton scattering. The multiple scatter dose component was computed with an empirical formula.

The Delta Volume method assumed that the photon source was a point source, which was a limitation of their approach since real photon sources have a finite extent. This mostly affects the primary dose near regions of major density changes (for example, near bone), and will tend to give a sharper transition between the two areas than will exist in reality. It was expected that an improved model for the photon source could be designed from an array of point sources, and that it would be sufficient to consider the extended source for the primary calculation only.

Ying et al. [133] calculated portal images using the planning CT data and the Delta Volume method. Discrepancies between measured and predicted portal dose images can arise from several sources, including: (i) changes in patient anatomy during treatment, (ii) errors in beam delivery, and (iii) inaccuracies in the treatment planning algorithm. Ying et al. [133] assumed that changes in patient anatomy during treatment caused the differences between the predicted and measured images. They proposed correcting the planning CT data using the measured portal dose image and an iterative algorithm.
to modify the CT data. Convergence was obtained when the measured and calculated portal dose images agreed sufficiently well. The patient dose was subsequently calculated from the altered CT data. This approach was demonstrated using computer simulation for a chest phantom, where the density and size of the lungs was allowed to vary.

McCurdy and Pistorius [72] investigated an analytical approach for portal scatter dose calculation that is similar to the Delta Volume Algorithm. They used Compton kinematics and ray tracing to determine the fluence of first-order Compton scattered photons at the portal imaging plane. Results for homogeneous and heterogeneous phantoms showed good agreement with predictions of the first scatter fluence calculated using Monte Carlo simulation.

2.2.2 CONVOLUTION/SUPERPOSITION METHOD

Another method that estimates the patient scatter dose with a treatment planning system is the approach developed by McNutt et al. [73, 74, 75, 76] based on earlier work by Papanikolaou [91]. This technique applies the convolution/superposition algorithm [67, 68, 106, 107, 108] to calculate the dose within the patient and the portal imager. A discrete, voxelized description of the phantom is used with uniform physical and electron densities in each voxel. In the convolution/superposition method, the dose is calculated by convolving the total energy released per unit mass (TERMA) in each voxel with pre-calculated three-dimensional dose deposition kernels. The total dose at each voxel is computed by superimposing the dose contributed by interactions within each voxel in the phantom that is irradiated. Dose deposition kernels have been calculated in a spherical water phantom by scoring the dose deposited in spherical coordinates from a forced photon interaction at the centre of the phantom.

\[^{6}\text{McNutt et al. [73] worked with the ADAC Pinnacle treatment planning system (ADAC, Milpitas, CA) which is now available from Philips (Philips, Amsterdam, Netherlands).}\]
The extended phantom concept. A photon interaction at \( r' \) will generate a shower of photons and electrons. Some of these shower particles may lead to dose deposited at \( r \). The dose at the imager is calculated by stretching the dose deposition kernel \( A \) across the air gap between the phantom and imager. (Adapted from [73] and [76]).

The air gap between the patient and the portal imager is a very large, low density heterogeneity. In heterogeneous regions the dose deposition kernels are scaled by the electron density of the medium relative to the electron density of water. This scaling is known as radiological pathlength scaling. Figure 2.1 illustrates the extended phantom concept where the dose deposition kernels are scaled between the patient and the imager. Figure 2.2 illustrates primary and scatter images calculated with the convolution/superposition algorithm. McNutt et al. [76] found that the imager dose profiles calculated with the convolution/superposition algorithm agreed within 4% with measured dose profiles. For their comparison they normalized the profiles to the dose at the central axis of the imaging plane. Therefore, only the relative amplitudes of the measured and calculated profiles were compared.
Figure 2.2. Portal images computed with convolution/superposition. (a) The neck phantom for this example showing the left lateral field (indicated by the solid diverging lines) and the central axis of the field (dashed line). The images provide a qualitative illustration of the (b) primary dose and the (c) scatter dose. The bright band in the primary image corresponds to the photon path through the trachea, while the dark band (far left) in this image aligns with the photons that have traversed the spinal column.

McNutt et al. [74, 75] reconstructed the doses within heterogeneous phantoms using a measured portal image. A drawback of their dose reconstruction method is the time required to compute the patient dose using a measured image. The most computationally intensive part of their technique was the convolution of the reconstructed primary energy fluence within the phantom with the dose deposition kernels. The number of computations for this step was approximately proportional to

\[ \mathcal{O} \propto N_F N_D N_V \]  

where \( N_F \) is the number of fields for the treatment, \( N_D \) the number of points within the phantom at which the \textit{in vivo} dose is to be calculated, and \( N_V \) is the number of voxels exposed to the primary beam. The symbol \( \mathcal{O} \) in equation (2.1) stands for ‘order of’.\(^7\)

\(^7\)The estimate of the order assumes that the dose in the patient is calculated from the dose deposition point of view, rather than with the dose interaction method (see for example [106]).
2.3 Theoretical Calculation of the Scatter Dose

The theory of photon and electron transport is well established and accurate estimates of the portal scatter dose are possible with Monte Carlo simulation. To reduce the sometimes prohibitive time required for theoretical calculation, simplified cases are studied to derive straightforward rules for predicting the scatter dose. The theoretical approaches reviewed here vary dramatically in complexity, from a powerful hand calculation (simple scatter to primary dose ratio model of Swindell and Evans [117]) to full Monte Carlo simulation of the treatment that requires parallel computer processors.

2.3.1 Scatter to Primary Dose Ratio (SPR) Model

Swindell and Evans [117] derived a simple model from first principles for the portal scatter to primary dose ratio (SPR) on the central axis for homogeneous scattering objects. This group used Monte Carlo simulation to show that the scatter to primary dose ratio is uniform across the imager for large air gaps between a homogeneous scattering object and the portal imager. This result was applied in their SPR model. At their institution the portal imager was located at a fixed source to detector distance of 200 cm, which satisfied the requirement of a large air gap.

The physical model of the SPR was developed by examining the photon fluence at the portal imager from primary photons as well as first and second order Compton scatter. The model predicts that for a cylindrical, homogeneous slab of water of thickness $t$, placed symmetrically about the isocentre and irradiated by a circular radiation field of area $A$ at the isocentre,

$$ \text{SPR} = k_0 A t (1 + k_1 t)(1 + k_2 A) \quad (2.2) $$

where

$$ k_0 = 0.0266 \frac{(L_1 + L_2)^2}{(L_1 L_2)^2}, \quad (2.3) $$

$$ L_1 = L_2 = \frac{D}{R} $$

and $D$ is the radiation energy and $R$ is the radiation pathlength.
Figure 2.3. Phantom geometry used in the SPR model. A circular beam of area $A$ is incident upon a circular water phantom of thickness $t$ located symmetrically about the isocentre. The isocentre is at distance $L_1$ from the photon source, and the detector is at a distance $L_2$ from the isocentre. This figure is based on a similar figure in [117].

$$k_2 = -\frac{1}{2\pi} \left[ \frac{1}{L_1^2} + \frac{1}{L_2^2} + \left( \frac{1}{L_1} + \frac{1}{L_2} \right)^2 \left( \frac{2}{3} + \frac{3\kappa}{2} \right) \right],$$

(2.4)

$L_1$ is the source-to-isocentre distance, $L_2$ is the isocentre-to-detector distance, and $\kappa$ is the mean energy of the 6 MV photon beam expressed in units of 0.511 MeV (that is, 1.81 MeV/0.511 MeV). Figure 2.3 illustrates the phantom geometry used for their model. Since there is no simple expression for $k_1$, this constant was optimized by comparing equation (2.2) to an extensive set of Monte Carlo data. Constant $k_1$ depends weakly on $L_2$ and for $60 \leq L_2 \leq 100$ cm, $k_1 \approx 0.002$ cm$^{-1}$. As $L_2$ decreases below 50 cm, $k_1$ increases, which is understandable since $k_1$ was interpreted to be proportional to the ratio of the number of twice-scattered detected photons to the number of once-scattered detected photons. The first term in equation (2.2), $k_0At$, states that as a first approximation, the SPR is proportional to the irradiated volume (for example, the SPR is $\approx 1\%$ per liter of irradiated scatterer for $L_1 = L_2 = 100$ cm). The agreement was better than 0.01 between their Monte Carlo data and their experimental SPR measurements, as well as between the model and the Monte Carlo data.

Their Monte Carlo code included Compton scattering and pair production, but not
coherent scattering and bremsstrahlung production. The scored quantities at the detector were the energy $E$ and status (denoted by superscripts $P$ for primary and $S$ for scattered) of every photon that arrived in a particular area. The SPR was calculated from

$$\text{SPR} = \frac{\sum_{k=1}^{K_S} R(E^S_k)}{\sum_{k=1}^{K_P} R(E^P_k)}$$

(2.5)

where $R(E)$ is the detector response, which describes the efficiency with which detected photons are converted into the detector output signal. Three types of detector responses (photopeak, photon counter, and Compton) were studied and the corresponding response functions are:

- photon counter $R(E) = \text{constant}$
- Compton detector $R(E) = \text{electron recoil energy}$
- photopeak detector $R(E) = E$.

Most detectors used in portal imaging devices are Compton detectors, since the dominant interaction is a single Compton scattering, for which the signal is proportional to the average energy imparted to the recoil electron. Thus $R(E)$ can be calculated from the appropriate Klein-Nishina cross-section.

The SPR model for a 6 MV photon beam is in use in a clinical trial testing missing tissue compensators\(^8\) for tangential breast radiotherapy. Compensators are designed from portal images to determine the thickness of the breast along the source-ray lines. This model has been used with two detectors, a purpose-built detector consisting of a linear array of scintillating crystals [36], and the liquid matrix ionization chamber electronic portal imaging device [35].

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\(^8\)A missing tissue compensator is designed to attenuate the photon beam so that the resulting dose within the tumour volume is more homogeneous. The compensator is constructed from high density material (for example, lead shot) and is inserted between the patient and the photon source. The thickness of the lead shot at each point in the two-dimensional grid depends inversely, to a first approximation, on the thickness of the tissue along the ray line between the photon source and the dose computation point.
Chapter 2. Literature Review

Since a more uniform dose to the breast for large patients was observed with a 10 MV beam than with a 6 MV beam, Partridge and Evans [95] validated the SPR model at 10 MV, again using Monte Carlo data. They showed that the model was accurate at 10 MV for \( L_2 \) greater than 60 cm and for all field areas \( A \) up to 625 cm\(^2\). Moreover, they concluded that the model was sufficiently accurate to use in the design of missing tissue compensators for breast radiotherapy with a 10 MV beam.

Hansen et al. [42] incorporated an approximate form of the SPR model of Swindell and Evans [117] into a method to calculate the \textit{in vivo} dose within the patient. \textit{In vivo} doses in an anthropomorphic phantom agreed with measurements within 3%. \textit{In vivo} doses were calculated by deriving the primary energy fluence (PEF) within the phantom using a measured portal image, and then convolving this PEF with the convolution/superposition dose deposition kernels. The PEF at the imaging plane was computed by dividing the total dose by \( \text{SPR}/(1+\text{SPR}) \). The PEF within the phantom was calculated by back-projecting the PEF at the imaging plane to the phantom plane (the back-projection accounted for the inverse square law and attenuation of the primary photon beam). The order of their algorithm for calculating \textit{in vivo} doses was approximately proportional to

\[
\mathcal{O} \propto N_F N_D N_V
\]  

(2.9)

where \( N_F, N_D, \) and \( N_V \) were defined in equation (2.1).

Spies \textit{et al.} [114] developed a rapid analytical method to calculate the first and second order Compton scatter fluence at the detector. One important simplification was that the second order Compton scatter was assumed to be isotropically distributed around a centre located at the midplane of the phantom. Integrals over the energy spectra were replaced with the average value of the function. As well, it was assumed that the incident primary photon fluence was a parallel beam, and therefore the divergence of the photon beam from the linear accelerator was ignored. Spies \textit{et al.} [114] also hypothesized that for Compton
style detectors [see equation (2.5) and discussion of that equation] the description of scattering by first-order Compton scatter alone may be sufficient in practice.

Spies et al. [114] examined the scatter from small, solid copper cylinders irradiated with radiosurgical\(^9\) fields. They demonstrated absolute differences within 0.02-0.03 between analytical SPRs and SPRs calculated with an in-house developed Monte Carlo simulation code. Direct experimental measurements of the scatter (away from the central beam axis) were also carried out using an in-house developed portal imager consisting of 128 CsI scintillation crystals optically coupled to silicon photodiodes. Although difficulties were present in reconciling the experimental and Monte Carlo results, the Monte Carlo model was concluded to be good for describing the portal scatter to primary dose ratio using small air gaps.

2.3.2 Slab Derived Scatter Kernels

Perhaps the most promising methods for portal scatter dose calculation are those that apply slab derived scatter kernels computed from Monte Carlo simulation [43]. In this approach, cylindrically symmetric scatter kernels are calculated by scoring the scatter dose at the portal imager resulting from a pencil beam traversing a homogeneous water phantom and an air gap. A database of kernels \(k(t)\) is generated for a range of phantom thicknesses \(t\), air gaps, and photon beam energies.

Originally, the method was to be applied for measuring the radiological tissue thickness\(^10\) of breast tissue with portal images by Hansen et al. [43]. Their original research was limited to 6 MV photon beams. The radiological thickness is then used to design cus-

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\(^9\)Radiosurgical fields are usually less than 4 cm in diameter as measured at the isocentre. These fields are used to treat inoperable small malignant and non-malignant lesions of the brain.

\(^10\)The radiological tissue thickness for a single voxel in the phantom is the product of the electron density of the voxel and the photon pathlength through the voxel. The radiological thicknesses for a column of tissue along the photon source to detector ray line is the sum of the radiological thickness for each voxel along the ray.
tomized tissue compensation to improve the dose homogeneity within the breast. Portal images were used to determine the tissue thickness rather than CT densitometry since these patients do not fit into the bore of standard CT scanners when they are in the treatment position (the arm on the same side as the treated breast is raised over the patient's head).

To determine the tissue thickness, an iterative method was developed based on measured portal images with \( I \) and without \( O \) the patient in the photon beam, and the database of scatter kernels \( k(t) \) indexed by tissue thickness, \( t \). The image without the patient in the beam gives the photon fluence distribution incident on the patient, which is nonuniform. Although the method was intended to be iterative, one iteration was found to extract the tissue thickness accurately enough. Consequently, only the non-iterative approach is discussed here.

First, \( I \) is taken as the initial estimate of the dose from primary photons, \( P \). Then the tissue thickness \( t(\vec{r}) \) at a point \( \vec{r} \) is calculated using the relationship

\[
P(\vec{r}) = O(\vec{r}) \exp[-\bar{\mu} t(\vec{r})]
\]

(2.10)

where \( \bar{\mu} \) is the mean attenuation coefficient for the photon beam and \( O \) is the portal image taken without the patient in the beam. For example, in breast tissue the mean attenuation coefficient for adipose tissue could be used. The tissue thickness is then used to estimate the scatter dose at the imager, \( S(\vec{r}) \):

\[
S(\vec{r}) = \int O(\vec{r'}) k_t(\vec{r'}) (\vec{r} - \vec{r'}) d^2 r'.
\]

(2.11)

This estimated scatter is then used to provide a better estimate of the primary dose assuming that the total imager signal is from primary and scatter dose,

\[
I(\vec{r}) = P(\vec{r}) + S(\vec{r}).
\]

(2.12)
Finally, equation (2.10) is solved a second time for $t(r')$ using the estimate for the primary dose $P(r)$ from equation (2.12). This approach was validated for homogeneous water phantoms and several heterogeneous phantoms in [43] with an accuracy of better than 1.5% for the dose from primary radiation and 2.8 mm (one standard deviation) of the true radiological thickness.

The slab derived scatter kernel approach for scatter estimation was extended for air gaps less than 40 cm and photon energies up to 24 MeV by McCurdy and Pistorius [69, 70, 71]. To use the scatter kernels for heterogeneous cases, the CT data was converted into an equivalent homogeneous phantom (EHP). The EHP was calculated by converting each column of CT data (along the source to imager pixel ray line) into an equivalent thickness of water by summing the radiological thickness of each voxel in the column. The maximum deviation between the predicted and Monte Carlo results for beam energies of 6 and 24 MV and air gaps of 10 to 40 cm was 0.5±0.6%, expressed as a percent of the total fluence on the central beam axis. Since the EHP concept was developed by Pasma et al. [97], the EHP concept will be defined in section 2.4.2 when discussing the experimentally derived slab kernels measured by Pasma et al. [97].

One of the main drawbacks of using Monte Carlo codes to develop scatter dose estimation models and to calculate portal scatter kernels is the lack of standard codes that separately score the primary and scatter fluence, or dose at the imager. Consequently, several researchers have developed and validated their own code for this purpose (for example, Jaffray et al. [47], Swindell et al. [118], and McCurdy and Pistorius [70]). All of these codes examine only photon scatter. An advantage of Monte Carlo approaches for developing scatter models is that the scatter dose can be separated according to the scatter mode (single versus multiple scattering, for example). This is important to determine the dominant scatter modes.
2.3.3 Monte Carlo Calculation of the Total Imager Dose

Descalle et al. [26] calculated portal images with the PEREGRINE Monte Carlo code for a phantom used in contrast studies that contained holes of varying diameter and depth. They achieved an accuracy of 1% for the total dose using a lateral grid resolution of 1 mm. This feat is still only possible with computer systems that network a large number of processors to carry out the computations in parallel to reduce the real time for the computation. Good agreement between the measured and calculated images was seen when comparing the contrast and resolution. Measured images were taken with the liquid matrix ionization chamber portal detector.

While Monte Carlo simulation represents the most accurate method for calculating portal scatter, the time for the simulation limits use of this technique to a very small number of research centres. More practical approaches are required for widespread implementation of in vivo dosimetry with portal imagers.

2.4 Empirical Scatter Dose Estimation

The scatter calculation methods presented in this section are all based on (i) a database of measurements of the scatter at the imaging plane, and (ii) an assumed functional form for the scatter. The first technique is the simplest and applies the experimental finding that the scatter dose is uniform across the imager for large air gaps. The second method derives measured scatter kernels as a function of the thickness of homogeneous polystyrene phantoms, analogous to the slab derived scatter kernels calculated with Monte Carlo simulation by Hansen et al. [43]. The third approach assumes a functional form for the change in scatter fluence between the patient and the imager.
2.4.1 Uniform Scatter Dose Approximation

Investigators at the Netherlands Cancer Institute (NCI) developed a method for calculating the midplane dose within the patient using the measured portal image [11, 12, 14, 30, 31]. Currently, their method is applied to in vivo dosimetry for lung radiation therapy at their institute. In this section the aspects of their technique that are relevant here will be briefly reviewed, including terminology, problems, and the consequences of the drawbacks and limits of their approach.

An overview of their method is shown in figure 2.4. To convert the measured portal image to the portal dose image, the pixel values are converted to dose (in step two) with a nonlinear calibration curve relating pixel value to dose. The calibration curve is measured for homogeneous phantoms of varying thickness [31]. In step three, a correction is performed to account for an incorrect commercially applied algorithm to compensate for the differing sensitivity of each detector pixel [31].

The NCI group of Boellaard et al. [12] investigated the portal scatter dose as a function of field area, phantom thickness, air gap, and source-to-phantom surface distance for homogeneous phantoms and a beam energy of 8 MV. The primary dose for each phantom thickness was calculated by extrapolating the total dose versus field area to zero field area, for an air gap of 90 cm. For smaller air gaps, the scatter was computed by subtracting the primary dose (as measured at an air gap of 90 cm and then corrected for beam divergence) from the total dose. At air gaps of less than 50 cm, the scatter dose distribution was approximately Gaussian, with an increasing width as the air gap increased. At the centre of the beam, the scatter dose increased by a factor of 25 when the air gap decreased from 50 cm to 5 cm. At air gaps larger than 50 cm, the portal scatter dose was uniform across the imager. This finding is applied in step four of the NCI in vivo dosimetry method, where the portal scatter dose is approximated by a uniform
1. Measure portal image
2. Convert pixel values to dose: *Portal Dose Image*
3. Compensate for commercial flat field correction
4. Apply look-up table to remove portal scatter dose
5. Apply inverse square law: *Primary Exit dose*
6. Apply convolution model using transmission: *Exit Dose Image*
7. Compare planned and extracted exit dose images
8. Calculate *Midplane Dose Image*
9. Compare planned and extracted midplane dose images

**Figure 2.4.** Overview of the Netherlands Cancer Institute in vivo dosimetry method. *Step six in the chart requires a measurement of dose with and without the patient in the beam to estimate the scatter within the patient.*

distribution.

The portal scatter dose for patients is estimated using a look-up table [14] of the total to primary dose ratio on the central beam axis measured for homogeneous phantoms. This look-up table is measured for a range of field areas, air gaps, and homogeneous phantom thicknesses. To estimate the portal scatter dose for a patient, the field area, air gap, and thickness\(^\text{11}\) of the patient are used to look-up the ratio of the total to primary

\[^\text{11}\text{In this case the thickness } t \text{ was calculated from the beam transmission } T \text{ through the patient: } t = -\ln T/\bar{\mu}. \text{ The transmission was defined as the ratio of the central axis doses with and without the patient in the beam. The data used for calculating the transmission was measured from portal dose images.}\]
dose ratio. The primary dose for the patient portal image is then calculated by dividing the measured total dose for that patient by the ratio of the total to primary dose found from the look-up table. The scatter dose is then equal to the difference of the total and primary dose. Finally, the scatter dose is subtracted from each pixel in the portal dose image. A drawback of the look-up table approach is the time needed for measuring the total to primary dose ratios for a range of field sizes, air gaps, and phantom thicknesses to create the look-up table.

In step five, the inverse square law is applied to calculate the primary exit dose in the patient using the primary portal imager dose (the exit plane is within the patient and is a distance $d_{max}$ from where the beam exits the patient). In step six, the total exit dose is calculated by adding the patient scatter dose at the exit plane to the primary exit dose. The patient scatter dose is estimated using the transmission through the patient, which was defined as the ratio of the primary dose measured with and without the patient in the beam [14]. Measuring a portal image without the patient in the beam for each field requiring in vivo dosimetry is a significant drawback of their method, since it extends the time needed for each treatment, which becomes yet more significant for treatments with multiple fields.

In step seven the in vivo doses at the exit plane are compared over the whole field to the intended doses calculated with their treatment planning system. A better location to compare the intended and in vivo doses is at the tumour plane, which may coincide with the midplane of the patient in some cases. For step eight, estimation of the midplane dose, Boellaard et al. [11] developed a novel method for calculating the midplane dose from the exit dose, patient thickness, and patient transmission. The accuracy of this method was poor (differences of 5 to 10% for a 6x6 cm$^2$ field and 4 to 18 MV beams) for small radiotherapy fields and heterogeneous phantoms irradiated with single fields. While their method for calculating the midplane in vivo dose does not require CT data, it was
recognized in their work that a more accurate estimate of the midplane dose would be achieved with CT data and a treatment planning algorithm that partially accounted for electron transport. Their midplane in vivo dose calculation technique was independent of the CT data to provide an overall check of the treatment planning calculation. The accuracy of their exit in vivo doses using the uniform portal scatter dose approximation and the convolution model for the patient scatter was 2.5% (one standard deviation).

2.4.2 SLAB DERIVED SCATTER KERNELS

Pasma et al. [97, 99] measured slab derived scatter kernels and applied these kernels to predict the portal scatter dose for heterogeneous phantoms and patients. The scatter kernels were extracted from measured ionization chamber data at the imaging plane for homogeneous polystyrene phantoms. In this section, the method for predicting the portal dose is briefly reviewed. The notation used here is a simplification of that used in the original papers in order to highlight the important features of their algorithm.

Pasma et al. [98] calibrated their Philips SRI-100 portal imager so that a portal image could be converted to the dose $D(x,y)$ as measured by an ionization chamber in a buildup cap\footnote{A buildup cap is a sleeve with a water-equivalent thickness equal to the depth of maximum dose in water for the beam. This sleeve is placed over a radiation detector (for example, an ionization chamber).} at the imaging plane. Buildup material is added on top of the imager to filter out electrons generated within the patient. This added buildup material, however, is heavy (for example, the 1 mm sheet of steel used weighed 1.3 kg, while the portal imager originally weighed 15 kg). Over one year, the added weight caused sag of the imager and a change in the position of the central axis by 3%, which had to be corrected for during daily calibration of the imager.

In their method to measure the scatter kernels, first the primary dose $P(x,y)$, and then the scatter dose $S(x,y)$ is calculated. To calculate the primary dose, the total
dose $D(x, y)$ at position $(x, y)$ was measured with an ionization chamber as a function of field area (FA) for each phantom thickness, resulting in a function $D(\text{FA}, x, y)$ for each thickness. This function was then extrapolated to zero field area for each thickness. The extrapolated value of $D(\text{FA}, x, y)$ at $\text{FS}=0$ theoretically contains no scatter dose, and is therefore equal to the primary dose. Although this method has been criticized since the extrapolation is subjective, the function $D(\text{FA}, x, y)$ was shown to be linear at 6 MV for field sizes less than 144 cm$^2$ at the isocentre.

The scatter dose kernels were derived in a more complex manner. Figure 2.5 illustrates the notation used to index the scatter kernel $s(r, t, L)$ as a function of position $r$, tissue column thickness $t$, and air gap $L$. The scatter kernel is assumed to be spatially invariant and rotationally symmetric. The total scatter dose $S$ is assumed to be equal to the sum of the scatter kernels for each tissue column in the irradiated volume of the patient:

$$S(x, y) = \int_{(x', y')_{\text{field}}} s[r(x' - x, y' - y), t(x', y'), L(x', y')] dx' dy'. \quad (2.13)$$

The scatter kernels are derived by solving equation (2.13) for $S(x = 0, y = 0)$ (that is, $S$ at the central axis) using a method similar to that described by Storchi and Woudstra [115, 116].

Since the scatter kernels were derived from measurements for homogeneous water phantoms a method is needed to allow calculation of the scatter dose for heterogeneous cases. Pasma et al. [97] solved this problem by converting the patient computed tomography data (that is, the three-dimensional matrix of densities relative to water) to an equivalent homogeneous polystyrene phantom (EHP). Each column in the EHP is calculated by finding the total radiological thickness of the patient along the source to detector ray, and then dividing this total by the electron density of polystyrene. As well, the distances between the centre of mass and detector plane are equal for the corresponding columns in the patient and the EHP.
Chapter 2. Literature Review

Figure 2.5. Phantom setup for empirical slab derived portal scatter kernels. (a) The total scatter dose in the image under the point \((x, y)\) is equal to the sum of the scatter contributions from each column of tissue [for instance, at \((x', y')\)] in the irradiated part of the field. The field boundaries are indicated by the heavy divergent lines. (b) Scatter kernels \(s[r = (x' - x, y' - y), t, L]\) are derived for a clinically applicable range of tissue column thicknesses \(t(x', y')\) and air gaps \(L(x', y')\) between the exit surface of the phantom and the portal image plane. This figure was adapted from figures in [97].

This approach was verified for the prediction of portal dose images, and the agreement between predicted portal images and ionization chamber measurements was 1% (one standard deviation) for anthropomorphic phantoms [97]. The method has also been used to predict the fluence under dose compensators as a quality assurance check for the inverse dose calculation algorithm and milling machine that design and mill the compensators [99]. The portion of the code that predicts the primary dose component has also been used to verify intensity modulated beams [96]. No patient or phantom was placed between the photon source and portal imager for this verification, so that only the primary dose was required.

Pasma et al. [99] calculated the in vivo dose for prostate cancer patients by scaling the total imager dose by the ratio of the dose at the isocentre in the patient to the dose
at the imager. This group limited their in vivo dosimetry to a single point at 5 cm depth within the patient on the central beam axis. The ratio applied for the clinical calculation was measured on the central axis with a 25 cm homogeneous polystyrene phantom.

A significant drawback of their scatter computation approach is the large workload for measuring the data to derive the scatter kernels. It was estimated that the minimum time required to measure this data was 4 hours per beam energy for a single linear accelerator [97]. Setting up and levelling the watertank\(^{13}\) takes approximately two hours. At present, at the Vancouver Cancer Centre, thirteen individual photon beam energies are used. Therefore, a minimum of 58 hours would be needed to measure the data to apply this method.

2.4.3 EMPirical SCATTER FLUENCE FUNCTION

Bogaerts et al. [15, 16] proposed a method for estimating the scatter dose on the central axis of the portal image for 6 MV beams and air gaps up to 40 cm. Central to their technique is a function that defines the change in scatter fluence between the exit and the imaging planes. This function was assumed to depend on the size of the air gap. This approach is analogous to the inverse square law for primary radiation. Specifically, they assumed that the functional form for the ratio of the scatter fluence between the exit and imaging planes \(F_s\) was given by:

\[
F_s(x, y, g, \alpha) = \frac{xy}{x'y'} = \frac{1}{1 + \frac{z_x^2}{x y}\alpha g + \frac{g y^2}{x y}}
\]

(2.14)

where \(g\) is the size of the air gap, \((x, y)\) the length and width of the field at the exit plane (the exit plane is defined in figure 7.1), \((x', y')\) the length and width of the field

\(^{13}\)A watertank is a 40×40×40 cm\(^3\) acrylic tank filled with water within which an ionization chamber is moved under computer control. Measurements of the dose profiles and depth doses are possible by moving the chamber at constant velocity within the beam and by using a reference ionization chamber to monitor fluctuations in the beam output.
at the imager, and $\alpha$ is a parameter determined from experimental measurements of $F_s$. The value of $\alpha$ ranged from 0.6 to 1.0 and varied for different phantom thicknesses, heterogeneities, and field sizes.

The maximum achievable accuracy for the calculated exit doses was estimated to be 2.4% (one standard deviation), and the maximum deviation for test phantoms was 4.5%. Although the accuracy of their method was less than that for conventional in vivo dosimetry, for example by using diodes, it was concluded that the technique was still useful. Reasons for this finding included satisfactory accuracy, a reduced workload for the radiation therapists administering the treatment, and a smaller increase in the treatment time to perform the in vivo dosimetry.

2.5 SUMMARY

Obstacles still exist that must be overcome before there is widespread availability of scatter dose estimation algorithms for in vivo dosimetry with portal imagers. Our aim is to develop a scatter estimation technique that is an improvement over previous methods.

The uniform portal scatter dose method appears to be a sound approximation both theoretically [117] and experimentally [12], and is efficient to implement. This method is limited to large air gaps (defined as greater than 50 cm) and was applied successfully for patient in vivo dosimetry. The drawback to this approach is the additional time needed for measuring the data for the look-up tables. In the current work the portal scatter dose is approximated by a uniform distribution. The total scatter dose is approximated by the first-order Compton scatter dose, which may be sufficiently accurate for the total scatter dose [114]. The scatter is computed by ray-tracing through the patient, the air gap, and the imager in an approach that is similar to the Delta Volume algorithm [131] or the method described by McCurdy and Pistorius [72].
While some of the previous methods have also focused on ways to calculate the dose from primary radiation, our intention is to use convolution/superposition for this computation. This direction would mean that portal dose images could be calculated with convolution/superposition by (i) modifying the convolution/superposition algorithm for the primary component and (ii) incorporating the analytical method for the scatter estimate.
CHAPTER 3

MATERIALS

The first section in this chapter briefly describes the operation of ionization chambers, which are used later for the experimental validation of the analytical method for calculating the SPR. In the second section, the construction, readout, and calibration of the liquid matrix portal imager are reviewed. This imager was used for the in vivo dose measurements in chapter 7.

3.1 IONIZATION CHAMBERS

Ionization dosimetry is one of the most convenient and most widely used methods for measuring absorbed dose [51]. In most cases, it is also the most accurate. The ionization chamber is the central piece of equipment in this system. A typical thimble-type ionization chamber is shown in figure 3.1.

The goal of ionizing dosimetry is to deduce the absorbed dose or energy absorption per unit mass in the medium surrounding the chamber. Cavity theories (for example, Spencer-Attix [17], Bragg-Gray [66], and Burlin [79]) have been developed to convert the charge measurement to absorbed dose. The particular cavity theory appropriate for a given chamber depends on the dimensions of the cavity and the atomic composition of the wall and the gas.

The Bragg-Gray theory is used for small cavities.\(^1\) Provided that the composition of

\(^1\)For small cavity ionization chambers, the charged particles lose only a small fraction of their energy in crossing the cavity. The ranges of the electrons are assumed to be much larger than the cavity diameter so that most of the charged particles in the cavity originate from the wall or medium.
Chapter 3. Materials

Figure 3.1. Cross-sectional view of a typical thimble-type ionization chamber. A potential difference exists between the inner surface of the wall and the central electrode. As charged particles cross the gas cavity they ionize air particles during transit. The charge is collected using an electrometer connected to the ionization chamber by a shielded cable.

The chamber walls is similar to that of the medium, the absorbed dose in the medium surrounding the ionization chamber is given by the Bragg-Gray equation:

$$\frac{dE}{dm} = W J_g \left( \frac{S_m}{S_g} \right)$$  \hspace{1cm} (3.1)

where

- $dE/dm$ is the energy deposited per unit mass in the surrounding medium
- $W$ is the average energy deposited in the gas per ion pair formed
- $J_g$ is the number of ion pairs formed per unit mass of gas
- $\left( \frac{S_m}{S_g} \right)$ is the mean ratio of the mass collision stopping powers of medium and gas, averaged over the energy distribution of the secondary charged particles crossing the cavity.

The minimum energy required to ionize a gas molecule is considerably smaller than the average energy deposited in the gas per ion pair formed, $W$. This is explained by the secondary particles dissipating much of their energy in non-ionizing collisions and by the scattered electron from an ionizing collision that usually emerges with some surplus
kinetic energy. For electrons in dry air, the value of \( W \) is 33.85±0.15 eV per ion pair. The measured quantity in absorbed dose calculation is usually the charge collected per unit mass, or volume, of the cavity.

For absorbed dose measurements in water, the wall material ideally used is one that has similar properties compared to water or air. The physical properties that are important are the mass energy absorption coefficients, \( \mu_{\text{en}}/\rho \), and the mass collision stopping powers, \( S/\rho \). The mass energy absorption coefficients determine the energy imparted to the medium by the photons, and the mass collision stopping powers determine the energy transfer from the charged particles to the solid or gas.

Ion chambers are operated at high enough potentials so that most of the ions formed in the chamber are collected. The collection efficiency for a particular geometry of ionization chamber and cavity material can be derived by considering the recombination of the ions in the cavity during charge collection.

In pulsed radiation, the pulses are typically short (a few microseconds or less) and the interval between pulses is long compared to the transit time of the ions between the electrodes (for example, in small air ionization chambers the transit time is typically 100 to 300 \( \mu s \)). If these conditions are satisfied, then it may be assumed that the total ionization per pulse occurs instantaneously and that the ions produced by one pulse are collected before the next pulse starts.

### 3.2 The Portal Imager

The electronic portal imaging device used for our work was a liquid matrix ionization chamber system (Varian Associates, Palo Alto, CA) installed on a Varian Clinac 2100C/D linear accelerator. This imager was chosen since we have the most experience with this particular model for portal dosimetry [92, 93, 94] compared to other types of
Figure 3.2. Diagram of the cross-section of the liquid matrix ionization chamber portal imager showing the materials surrounding the liquid layer. (Figure adapted from [31].)

portal imagers at our centre. The technology was developed at the Netherlands Cancer Institute by H. Meertens and M. van Herk [77, 123]. The imager is mounted on the linear accelerator gantry with a motor-driven retractable arm, which allows motion in three-dimensions (see figure 1.5 in section 1.6).

The sensitive area of the detector is 32.5x32.5 cm$^2$. This area is partitioned into a matrix of 256x256 liquid ionization chambers, each with a volume of 1.27x1.27x0.8 mm$^3$. The matrix is formed by crossing two printed circuit boards, each etched with 256 parallel copper strip electrodes, at 90$^\circ$ to one another. The liquid ionization film is sandwiched between the two circuit boards, as shown in figure 3.2.

X-ray photons incident on the imager are converted to electrons, which then ionize the molecules in the liquid. Once an equilibrium is reached between the rates of formation and recombination of these ions, the ionization current is measured from each chamber. The ionization current is amplified and corrections are applied to account for differences between individual electrometers and chambers. The image is then displayed on a terminal outside of the treatment room.
A 1 mm thick plate of plastoferrite$^2$ is used to convert the X-ray photons into electrons. The foam, circuit boards, and plastoferrite in front of the liquid layer are equivalent to $\approx 8 \pm 0.5$ mm of water [13], while the material behind the liquid is equivalent to $\approx 5$ mm of water [31]. By adding material above the top-most circuit board (for example, polystyrene) so that electronic equilibrium is achieved at the liquid layer, the pixel$^3$ signal and signal-to-noise ratio can be maximized.

3.2.1 LIQUID FILM

By using a liquid ionization medium rather than one of gas, the chambers can be very small since the liquid has a higher density and will therefore have a larger signal-to-noise ratio. This imager uses isooctane (2,2,4 trimethylpentane, C$_8$H$_{18}$) for the liquid film. Ionization chambers utilizing organic liquids may be classified as relying on ion transport or electron transport. In this imager, isooctane ions are formed and travel through the liquid to the electrodes. The mobility of these heavy ions is low and hence the transit time for an ion to cross the chamber from one electrode to the other is long: van Herk [123] calculated the transit time to be 0.5 s.

Although the isooctane is pure when the imager is constructed (grade - spectroscopic), impurities probably arise from water diffusing through the circuit boards as well as from interaction between the liquid and the chamber materials. Radiochemical reactions within the liquid will affect the concentration of impurities. Furthermore, a change in the level of purity may affect the signal collected if the impurity is charged, since this will affect the current collected by the electrometer. Since the image gray-scale values remain within 1% over three months [31], the purity of the liquid over time is not a significant problem.

$^2$Plastoferrite is a mixture of plastic and barium ferrite (density, $\rho = 4.75$ g cm$^{-3}$).

$^3$A pixel is a small element of area.
The reason for the choice of isooctane for this imager was not apparent from the literature, however, this liquid has a relatively high free ion yield\(^4\) compared to other hydrocarbons [105].

### 3.2.2 Readout Electronics

Readout of the ionization current for this imager proceeds pixel by pixel, and is shown schematically in figure 3.3. High voltage (typically 250-300 V) is switched from one row electrode to the next on the top printed circuit board. When one high voltage row electrode is switched on, the current is read out from each electrometer attached to the 256 column electrodes on the bottom circuit board.

Activation of the high voltage switches and readout electrometers is synchronized with

\(^4\)The free ion yield, \(G_{fi}\), is the average number of free ion pairs formed per unit of absorbed radiation energy. The value of \(G_{fi}\) is influenced by the recombination of ions formed along the ionization track of the secondary electron, the linear energy transfer of the radiation, and the electric field strength.

---

**Figure 3.3.** Schematic diagram of the readout electronics for the scanning liquid matrix ionization chamber electronic portal imaging device. A high voltage (HV) is switched from one row electrode to the next. The ionization current is read from each of the 256 column electrodes sequentially when one row electrode is polarized. (Figure adapted from [123].)
the internal 60 Hz clock of the linear accelerator used for producing the radiation beam pulses [39]. The time for a single current measurement from all 256 column electrodes is \( \approx 2 \) ms, and current sampling commences \( \approx 8 \) ms after a beam pulse. An additional wait time is included before any measurements are taken to allow the ion concentration within the liquid to reach equilibrium. Equilibrium is reached within \( 1 \) s after turning the beam on [123]. The time to regain equilibrium after readout at the highest dose rates is \( \approx 40 \) ms [123], since only a small fraction of the ions are collected. With four current samples per pixel, the total time for image acquisition is \( (2 \text{ ms sample}^{-1} \text{ row}^{-1} \times 4 \text{ samples} + 8 \text{ ms row}^{-1}) \times 256 \text{ rows} + \text{wait time} \approx 4 \) s. One of the drawbacks for this imager is the long scan time for image readout. The long scan time prevents integration of the signal as a function of time.

3.2.3 Computer

A computer located outside the treatment room and attached to the imaging electronics serves several functions, including automatic position control of the detector and image display. Image analysis algorithms are available on the computer, including image enhancement, edge detection, basic statistics, and image matching.

Raw images are corrected for differences in the pixel sensitivities, electrometer offsets, and leakage currents [18]. The first correction accounts for the electrometer offsets and leakage current and is given the symbol \( E_j \), where the subscript \( j \) represents the \( j^{th} \) electrometer. Measurement of \( E_j \) is carried out quickly without polarizing voltage prior to the acquisition of each image [31]. The magnitude of \( E_j \) is \( \approx 10\% \) [18] of the raw pixel intensity. Second, the ionization chamber offsets are determined from the bias field image or dark current image \( B_{ij} \), which is measured without radiation. This bias correction helps to cope with the artifacts caused by the fast switching of the high-voltage electrodes and its magnitude is \( \approx 1\% \) [18]. Third, the variation in ionization chamber cell sensitivity
is approximately measured from the ratio of the individual pixel response to the average pixel response when the entire matrix is irradiated with a uniform field. The image of the uniform field is termed a flood field, $F_{ij}$. Differences in chamber sensitivity are $\approx 40\%$ in magnitude [18] and arise from differences in electrode shape and electrode surface heterogeneities. The commercially displayed image $W_{ij}$ is calculated from the raw image $I_{ij}$ using the equation

$$W_{ij} = (I_{ij} - E_j - B_{ij})\frac{F}{F_{ij}}$$

(3.2)

where $F$ is the average pixel intensity in the flood field.

### 3.2.4 Calibration for Dosimetry

The differential equation governing the ion-pair concentration $n(t)$ [105] when no polarizing pulse is applied is given by van Herk [123]

$$\frac{dn(t)}{dt} = \dot{N}_{in}(t) - \alpha n^2(t)$$

(3.3)

where $\dot{N}_{in}(t)$ is the ionization rate, $\alpha$ is the ion recombination constant, and $t$ is the time. In both continuous and pulsed radiation beams the imager readout commences after the ion concentration has reached equilibrium. For continuous radiation, as from a $^{60}$Co radiotherapy source, the ionization rate $\dot{N}_{in}(t)$ is constant. At equilibrium, the rate of change of the ion concentration $dn(t)/dt$ is zero, and the equilibrium ion concentration $n_{eq}$ is equal to [123]

$$n_{eq} = \frac{\sqrt{\dot{N}_{in}/\alpha}}.$$

(3.4)

For radiotherapy linear accelerators, having pulsed beams, the approximate solution for the average ion concentration $n_{avg}$ between pulses is given by Boellaard et al. [13]:

$$n_{avg} \approx \left(\frac{\Delta N_{in}}{\alpha \Delta t}\right)^{1/2} - \left(\frac{3}{8}\right) \alpha^{1/2} \Delta t^{1/2} \Delta N_{in}^{3/2}$$

(3.5)
where $\Delta N_{in}$ is the number of free ions produced per pulse and $\Delta t$ is the time between pulses. Boellaard et al. [13] showed that the first term of equation (3.5) was within 1% of the exact solution for $n_{avg}$ for typical clinical dose rates (less than 400 cGy min$^{-1}$).

In this thesis, the portal image is converted to the dose as measured by an ionization chamber within a water-equivalent medium. The experimental relationship between the pixel intensity $W_{ij}$ and the dose rate as measured by an ionization chamber within a water-equivalent medium, $\bar{D}$, has been confirmed by many authors (for example, [31, 132, 134]):

$$W_{ij} = G(\bar{D}_{ij}) = a\sqrt{\bar{D}_{ij}} + b\bar{D}_{ij}$$

(3.6)

where $a$ and $b$ are the calibration constants for dosimetry. This is in agreement with the theoretical predictions, given by equations (3.4) and (3.5). To determine $a$ and $b$, the average pixel value in a small region of interest and the dose rate are measured under the same conditions for a range of dose rates. The dose is measured with an ionization chamber placed at the depth of maximum dose within a block of water-equivalent material (for example, polystyrene). The rectangular block has the same dimensions as the imager. The ionization chamber is placed at the same source to detector distance as the portal imager.

Although water-equivalent material (for example, polystyrene) is added above the ionization chamber, this material is insufficient to stop all the patient-generated electrons. For example, for an 18 MV photon beam, a total thickness of 3.2 cm of water-equivalent plastic would be added above the ionization chamber. Since electrons lose energy at the rate of $\approx 2$ MeV cm$^{-1}$ in water, patient-generated electrons above $\approx 6.4$ MeV will pass through the buildup material and deposit dose in the ionization chamber. This was the reason for tracking the patient-generated electrons in the current study.

When calibrating the pixel signal against the dose measured in a $30 \times 30$ cm$^2$ polystyrene plate, the calibration constants were found to be nearly independent of field size [30].
The region of interest is defined as an area in the image over which the beam may be considered to be uniform. An outer housing for the imager is usually removed when the imager is applied for dose measurements. Sufficient buildup material is added to the imager to maximize the signal.

If the commercially corrected images are used for dosimetry, then errors of several percent result [31], because the flood field produced by the linear accelerator is nonuniform and the commercial imager software assumes that the flood field is uniform. To solve this problem, the dose in the flood field is measured with an ionization chamber and the corrected pixel values \( V_{ij} \) are calculated from

\[
V_{ij} = \frac{W_{ij} G(D_{ij, \text{flood}})}{F_{ij}}
\]

where \( F_{ij} \) is the flood field [see also equations (3.2) and (3.6)].

It is assumed that the calibration constants \( (a, b) \) measured on the central beam axis are a good approximation for all pixels in the image. At non-zero gantry angles (that is, when the beam is pointed in directions other than down), the calibration constants vary within the image, due to changes in thickness of the liquid layer across the detector [132].

The imager warm-up time, or the time to obtain a constant reading, is \( \approx 1 \) hour [31]. The amount of signal change for a chamber varies from pixel to pixel depending on the heat dissipated by nearby electronics. Only those pixels that are located near heat sources exhibit a change in their reading with time. The chamber sensitivity remains constant as long as the acquisition time between images is at least 5 min [31]. The accuracy of calibration is \( \approx 0.6\% \) and the calibration constants \( [a \text{ and } b \text{ in equation (3.6)}] \) remain stable within 1\% for \( \approx 3 \) months [31].
3.3 Summary

The most important part of this chapter was subsection 3.2.4, which presented the method for calibrating the liquid matrix portal imager for use as a dosimeter. When calibrating the imager, the gray-scale image is converted to the dose as measured by an ionization chamber placed at the depth of maximum dose within a water-equivalent rectangular block (for example, within a block of polystyrene). The pixel value to dose calibration curve was given by equation (3.6).

In the following chapter, a new analytical method for calculating the scatter to primary dose ratio is presented. This analytical method is validated against Monte Carlo simulation results and experimental measurements in chapters 5 and 6, respectively.
CHAPTER 4

THEORY: ANALYTIC SPR CALCULATION

This chapter presents the analytical method for calculating the scatter to primary dose ratio (SPR) in radiotherapy portal images. In this method, the imager scatter dose at off-axis points is equal to the scatter calculated at the central beam axis. Hence the portal scatter dose is approximated by a uniform distribution, in accordance with the result found experimentally by Boellaard et al. [12] when measuring the scatter for large air gaps between the phantom and the imager. This approximation greatly reduces the calculation time, since the scatter dose is computed only at one point in the image instead of at each point in the image. The technique accounts for the photon spectrum, the patient tissue density data, and the detector response. Attenuation and divergence of the primary and scattered photons is calculated by ray-tracing from the photon source, through the patient, to the imager. It is assumed that the detector is composed of water-equivalent materials. Since the response of the liquid matrix portal imager used in the current work can be calibrated against the dose as measured by an ionization chamber, the assumption of a water-equivalent detector is appropriate here. Several examples of the analytical calculation for the dose from primary and scatter radiation, as well as the SPR, are presented. This work was published in [90].
4.1 History

McCurdy and Pistorius [72] applied the theory of Compton scattering to predict the first-order portal scatter fluence component. Spies et al. [114] analytically modelled the first and second order Compton scatter from copper cylinders and achieved good agreement between the model and Monte Carlo data. The high-density phantom was chosen to test the model in extreme circumstances. Direct measurements of the scatter at off-axis points were compared to Monte Carlo and analytical results.

Spies et al. [114] hypothesize that, in practice, portal scatter may be well approximated by first-order Compton scatter alone for Compton detectors. Boellaard et al. [12] found that the scatter dose is uniform when the air gap is at least 50 cm: this approximation is also used in the current work and so large air gaps (50 cm or larger) are used. Large air gaps also minimize the SPR from multiply scattered photons (McCurdy and Pistorius [70]).

4.2 Current Development: Analytical Imager Dose Calculation

In this chapter, we developed a new method for calculating the SPR on the central axis. First, the theory for calculating the dose from primary is presented, and then the computation of the dose from scatter is described. The units for each quantity are stated. When an approximation is used, the impact of this simplification on the final SPR is discussed.
4.2.1 IMAGER DOSE FROM PRIMARY RADIATION

In this section the equation for the dose from primary photons at the portal detector, $P^A$, is presented. Throughout this work the superscript $A$ denotes an analytically calculated quantity. The dose from primary radiation for a simplified parallel, monoenergetic photon source is discussed first, and then the more realistic case of a diverging, polyenergetic beam is examined.

The simplified case applies to a broad, parallel-ray, monoenergetic photon source of energy $E$ (units [MeV]) incident on a homogeneous absorber of constant thickness. The direction of the incident photons is parallel to the central axis throughout the beam. The phantom and detector are assumed to be homogeneous and composed of the same material. The dose from primary photons at the detector $P^A$ [MeV g$^{-1}$] can be calculated from the collision KERMA$^1$ $K_c$ [MeV g$^{-1}$] [49],

$$ P^A = \beta K_c = \Phi \exp[-\mu(E)(t + d_{max})] \frac{\mu(E)}{\rho} E_{ab,w}(E) \beta $$  \hspace{1cm} (4.1)

where

$\Phi$ is the photon fluence or photon flux incident on the phantom [photons cm$^{-2}$]

$\mu(E)$ is the linear attenuation coefficient for photons of energy $E$, [cm$^{-1}$]

$t$ is the thickness of the homogeneous absorber [cm]

$\rho$ is physical density of the medium [g cm$^{-3}$]

$d_{max}$ is the depth of maximum dose (see section 1.9.1) at the detector, [cm]

$^1$KERMA is the kinetic energy released per unit mass.
\( \bar{E}_{ab,w}(E) \) is the average energy absorbed per photon (average kinetic energy transferred to electrons that leads to ionization, excluding energy lost to bremsstrahlung) for water [MeV photon\(^{-1}\)]

\( \beta \) accounts for differences between the collision KERMA and the absorbed dose [7, 84] (dimensionless).

The term \( \Phi \mu/\rho \) gives the number of photons that interact per unit mass of material irradiated by a photon fluence \( \Phi \). The average energy absorbed per photon of incident energy \( E \) is given by [49]

\[
\bar{E}_{ab}(E) = \frac{\mu_{ab}(E)}{\mu(E)} E
\]

where \( \mu_{ab} \) is the absorption coefficient [cm\(^{-1}\)] for the detector material.

The detector response function \( R(E) \) [MeV g\(^{-1}\) cm\(^2\) photon\(^{-1}\)] [see equations (2.6) to (2.8) in section 2] was included explicitly in equation (4.1), and was equal to the mean dose absorbed in water for an incident photon of energy \( E \) given by

\[
R(E) = \frac{\mu_w(E)}{\rho_w} \bar{E}_{ab,w}(E) \beta.
\]

This particular response function was chosen since the portal imager is calibrated against the dose measured with an ionization chamber (see subsection 3.2.4).

The transmission of the primary photons in equation (4.1) (specifically, the term \( \exp[-\mu(E)(t+d_{max})] \)) was replaced by a discrete summation along the source to detector ray-line. The exponential transmission of the primary photons \( T_p(\vec{r}_d, E) \) at a particular point on the detector located by vector \( \vec{r}_d \) [illustrated in figure 4.1(a)] is thus given by

\[
T_p(\vec{r}_d, E) = \exp \left[ - \sum_{\vec{r}_n=0}^{\vec{r}_d} \Delta r \mu(\vec{r}_n, E) \right]
\]

where the lower limit of the sum (\( \vec{r}_n = \vec{0} \)) is located at the photon source. Vector \( \vec{r}_n \) is the index of the sum. In the calculation \( \Delta r \) was approximated as \( \Delta z \) (the thickness of...
the voxel\textsuperscript{2} along the Z axis, which was 1 cm).

The ray pathlength for the attenuation of the photons (primary and scattered) was calculated within approximately 1 mm for the homogeneous phantoms, and was exact along the central axis for all phantoms. Voxel coordinates along the ray path were calculated using the angles between the (X,Y) axes and the projection of the ray onto the XY plane. This method of ray tracing accounts for beam divergence. Since the pathlength through the voxel \( \Delta r \) was assumed to be equal to \( \Delta z \) (that is, the voxel size along the Z axis), this ray-trace was approximate for the heterogeneous phantoms away from the central beam axis.

For lung, muscle, and adipose tissue, the exponential transmission was accounted for in the analytical calculation by using an approximation to equation (4.4). In these cases, the mass attenuation coefficient for water \( \mu_w(E)/\rho_w \) [cm\textsuperscript{2} g\textsuperscript{-1}] was substituted for the tissue specific coefficient but it was then multiplied by the actual density of the tissue:

\[
\Delta r \mu(\vec{r}_n, E) = \Delta r \frac{\mu(\vec{r}_n, E)}{\rho(\vec{r}_n)} \rho(\vec{r}_n) \quad \text{(4.5)}
\]

\[
\approx \frac{\mu_w(E)}{\rho_w} \Delta r \rho(\vec{r}_n). \quad \text{(4.6)}
\]

This approach is reasonable for photon beam energies up to 24 MV. For bone, the attenuation was calculated without using this approximation (that is, for a voxel composed of bone, the mass attenuation coefficient for bone was used when calculating the attenuation for that voxel). In the experimental validation described in chapter 6, aluminum is substituted for bone and the attenuation coefficient for aluminum is used for those voxels containing this metal.

The mass attenuation and mass absorption coefficients for each material were calculated by linear interpolation using the data tabulated as a function of energy in [49]. These coefficients are plotted for water in figure 4.2.

\textsuperscript{2}A voxel is an element of volume.
The dose from primary radiation for a parallel photon source [equation (4.1)] can be modified for a point photon source by including the inverse square law that describes the decrease of the primary fluence with distance from the source. Since the fluence was defined at the phantom surface ($\Phi_0$) for the Monte Carlo simulation, the primary fluence at the detector $\Phi$ is given by [49]

$$\Phi = \Phi_0 \left( \frac{\text{SSD}}{\text{SDD}} \right)^2$$  \hspace{1cm} (4.7)

where the SSD is the source to phantom surface distance and the SDD is the source to
detector distance. In the analytical calculation, both the source to surface distance and the source to detector distance were measured along the central beam axis, which is an approximation for points away from the central axis. This approximation introduces a negligible error for the cases considered in this thesis.

Since realistic radiotherapy beams from clinical linear accelerators emit photons over a range of energies, the dose from primary photons is summed over all energies present in the primary photon energy spectrum. The energy spectra used in this work were previously discussed in section 1.5.

In the analytical calculation, the photon source was modeled as an isotropic, point source and the photon spectrum was assumed to be invariant across the beam. Jaffray et al. [47], Partridge and Evans [95], and Swindell and Evans [117] used the same approximate photon source model and obtained good agreement between calculated and measured scatter to primary dose ratios or scatter fraction data.

Figure 4.2. Mass attenuation (\(\bullet\), left vertical axis) and mass absorption (\(\circ\), right vertical axis) coefficients for water. Data from Johns and Cunningham [49]. The smooth curve between the data points is for visual guidance only.
By using equations (4.6) and (4.7) to modify equation (4.1), the expression for the dose from primary photons becomes

$$ P^A(\vec{r}_d) = \beta \left( \frac{SSD}{SDD} \right)^2 \sum_{E_i=E_{\text{min}}}^{E_{\text{max}}} \Phi_o(E_i) T_p(\vec{r}_d, E_i) \frac{\mu_w(E_i)}{\rho_w} \bar{E}_{ab,w}(E_i). \quad (4.8) $$

This expression is for (i) an isotropic, polyenergetic, point source, (ii) a heterogeneous phantom or patient, (iii) a detector assumed to be composed of materials with similar attenuation, scattering, and absorption properties as water, and (iv) a photon source with an energy spectrum that is invariant across the beam. The summation limits for the outer sum \((E_{\text{min}}, E_{\text{max}})\) are from the minimum photon energy to the maximum photon energy present in the photon beam incident on the phantom or patient and \(E_i\) is the index of summation.

### 4.2.2 IMAGER DOSE FROM FIRST ORDER COMPTON SCATTER

The total dose from scatter \(S\) at a pixel on the portal detector can be expressed as the sum of the dose from each scatter mode,

$$ S(\vec{r}_d) = S_F(\vec{r}_d) + S_{MS}(\vec{r}_d) + S_{CP}(\vec{r}_d) \quad (4.9) $$

where

- \(S_F\) is the dose from photons that scatter once within the scattering object and then interact with the detector, which is also termed the dose from first-order Compton scatter

- \(S_{MS}\) is the dose from photons that scatter more than once with the scatter object, and includes the dose from bremsstrahlung and annihilation photons that originate within the scatter object
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\( S_{CP} \) is the dose from charged particles (electrons and positrons) that are set in motion within the scattering object, cross the air gap between the object and the detector, and then deposit dose within the detector.

McCurdy and Pistorius [70] showed that the first-order scatter dominates the scatter fluence for a wide range of beam energies, field sizes, and air gaps. Furthermore, Spies et al. [114] hypothesized that the total scatter dose could, in practice, be approximated by the first-order Compton scatter dose. In the current work, the portal scatter dose from multiply scattered particles (photons and patient-generated electrons) was neglected, which can be stated as:

\[
S^A(\vec{r}_d) \approx S^A_F(\vec{r}_d). \tag{4.10}
\]

Further, air gaps are used that are 50 cm or larger, since Boellaard et al. [12] showed that the scatter dose was uniform for large air gaps. The scatter dose at off-axis points is approximated by the first-order Compton scatter dose at the central beam axis.

The total dose from first-order scatter, \( S^A_F(\vec{r}_d) \), was calculated from the probability that a photon originating from the photon source scatters once within the phantom and then deposits dose within the detector. The total scatter dose from first-order scatter for each pixel on the detector was evaluated by summing the contribution from each scattering voxel within the irradiated volume of the phantom. In this section, the primary photon source is approximated by a model of the energy spectrum that is invariant across the beam. The equation for the scatter dose is for a heterogeneous scattering object (patient or phantom). As well, the detector materials are assumed to have similar radiological properties compared to water.

The photon path for a photon that scatters once and reaches the detector was shown in figure 4.1(b): the primary photon travels from the source to the voxel along \( \vec{r}_v \), scatters at the voxel at \( \vec{r}_v \), and then the scattered photon travels along vector \( \vec{R} \) to the detector pixel.
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located at \( \vec{r}_d \). A monoenergetic source of primary photons (energy \( E \)) is considered first. The detector scatter dose at \( \vec{r}_d \) from first-order Compton scattered photons generated within a scattering voxel located at \( \vec{r}_v \) is expressed as [MeV g\(^{-1}\)]:

\[
S^A_F(\vec{r}_d, \vec{r}_v, E) = \Phi_o(E) F(\vec{r}_v) T_p(\vec{r}_v, E)\varepsilon(\vec{R}, \theta, E)e(\vec{r}_v, \vec{r}_d, E_1) T_s(\vec{r}_v, \vec{r}_d, E_1) \tag{4.11}
\]

where

- \( F(\vec{r}_v) \) is the inverse square law that accounts for the divergence of the photons from the phantom surface to the scattering voxel, \( F(\vec{r}_v) = (SSD/\vec{r}_v)^2 \)
- \( T_p(\vec{r}_v, E) \) is the transmission of the primary photons to the scattering voxel
- \( \varepsilon(\vec{R}, \theta, E) \) is the monoenergetic scatter kernel [MeV g\(^{-1}\) photon\(^{-1}\) cm\(^2\)]
- \( e(\vec{r}_v, \vec{r}_d, E_1) \) is the ratio of the electron density of the voxel to the electron density of water (units of electron density, [electrons cm\(^{-3}\)])
- \( T_s(\vec{r}_v, \vec{r}_d, E_1) \) is the transmission of the scattered photons between the scattering voxel at \( \vec{r}_v \) and the detector point at \( \vec{r}_d \)
- \( E_1 \) is the energy of the scattered photon that originates at the scattering voxel, and is given by \( E_1(E, \theta) = E/\{1 + \alpha[1 - \cos(\theta)]\} \)
- \( \alpha \) is the energy of the incident primary photon divided by the rest mass energy of the electron, \( \alpha = E/(0.511\, \text{MeV}) \)
- \( \theta \) is shown in figure 4.1(b) and is the angle between the incident primary photon direction (given by vector \( \vec{Q} \)) and the direction of the scattered photon (vector \( \vec{Q} \) terminates at the same location as vector \( \vec{r}_d \)).

The photon fluence \( \Phi_o(E) \) was defined at the phantom surface.
In equation (4.11), the transmission of the primary photons between the incident phantom surface and the scattering voxel, \( T_p(\vec{r}_v, E) \), is given by

\[
T_p(\vec{r}_v, E) = \exp \left[ - \sum_{\vec{r}_n = \vec{0}}^{\vec{r}_v} \Delta r \, \mu(\vec{r}_n, E) \right].
\] (4.12)

For voxels containing water-like tissues, the same approximation was made as previously discussed for equation (4.6).

The monoenergetic scatter kernel in equation (4.11), \( \varepsilon(\vec{R}, \theta, E) \), gives the dose per unit incident primary photon fluence at the detector pixel at \( \vec{r}_d \) from primary photons of energy \( E \) that travel from the incident photon source, scatter at the voxel located at \( \vec{r}_v \), and then interact with the detector at \( \vec{r}_d \). The scatter kernel is expressed as

\[
\varepsilon(\vec{R}, \theta, E) = \frac{d\sigma(\alpha, \theta)}{d\Omega} \rho_{e,w} \delta V \frac{dA}{|\vec{R}|^2} \frac{\mu_w(E_1)}{\rho_w} \frac{E_{ab,w}(E_1)}{dA} \beta
\] (4.13)

where

- \( d\sigma(\alpha, \theta)/d\Omega \) is the differential Klein-Nishina cross-section per electron per steradian, which gives the probability that an incident photon of energy \( E \) will scatter at an angle \( \theta \), [cm\(^2\) electron\(^{-1}\) steradian\(^{-1}\)]
- \( \rho_{e,w} \) is the electron density of water, [electrons cm\(^{-3}\)]
- \( \delta V \) is the volume of the scattering voxel at \( \vec{r}_v \) [cm\(^3\)]
- \( \vec{R} \) is the three-dimensional distance between the scattering voxel at \( \vec{r}_v \) and the detector pixel at \( \vec{r}_d \), \( \vec{R} = \vec{r}_d - \vec{r}_v \), and is present to account for the solid angle \( d\Omega = dA/|\vec{R}|^2 \)
- \( dA \) is the pixel area on the detector located by the vector \( \vec{r}_d \), shown in figure 4.1.
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To calculate the scatter from a voxel composed of a specific tissue with electron density $\rho_e(\vec{r}_v)$, the scatter kernel is multiplied by the electron density relative to water for the voxel, $\rho_e(\vec{r}_v)/\rho_e(w)$, in equation (4.11). In the analytical calculation the differential cross-section $d\sigma(\alpha, \theta)/d\Omega$ was calculated directly from the formula stated in [49]. Angle $\theta$ is given by

$$\theta = \arccos \left( \frac{\vec{R} \cdot \vec{Q}}{|\vec{R}||\vec{Q}|} \right). \quad (4.14)$$

In the calculation, the scatter volume $\delta V$ was chosen to be 0.25 cm$^3$ (0.5 cm along the X and Y axes, 1 cm along the Z axis). This choice was based on the result that the absolute error in the scatter to primary dose ratio, due to the size of the voxel, was approximately $\delta$SPR$=0.01\delta V$ (for SPRs equal to $\approx 0.10$).

In equation (4.11) the transmission of the scattered photons between the scattering voxel at $\vec{r}_v$ and the detector pixel at $\vec{r}_d$ was calculated from:

$$T_s(\vec{r}_v, \vec{r}_d, E_1) = \exp \left( - \sum_{\vec{r}_n=\vec{r}_v}^{\vec{r}_d} \Delta r \mu(\vec{r}_n, E_1) \right). \quad (4.15)$$

Again, for voxels with water-like tissues, the approximation presented in equation (4.6) was made, and for voxels containing bone, no approximation was used.

Each voxel in the scattering object is assigned a uniform physical and electron density, and hence the description of the phantom or patient is discrete. Since each voxel within the irradiated part of the phantom or patient contributes scatter dose to each pixel on the portal detector, the total scatter dose (for a monoenergetic incident beam) at the detector pixel $\vec{r}_d$ is the sum of the scatter contribution from each scattering voxel,

$$S^A_F(\vec{r}_d, E) = \sum_{x=z_s}^{z_m} \sum_{y=y_0}^{y_f} \sum_{x=x_l}^{x_r} S^A_F[\vec{r}_d, \vec{r}_v = (x, y, z), E] \quad (4.16)$$

where the $(x, y)$ limits of summation are shown in figure 4.1(c). The limits of summation in the Z direction are from the entrance phantom surface $z_s$ to the exit phantom.
surface $z_m$. Since radiotherapy beams are polyenergetic, the total scatter dose at $\vec{r}_d$ was calculated by summing over the entire incident primary photon energy spectrum

$$S^A_P(\vec{r}_d) = \sum_{z=z_s}^{z_m} \sum_{y=y_b}^{y_f} \sum_{x=x_1}^{x_f} \sum_{E_i=E_{min}}^{E_{max}} S^A_P[\vec{r}_d, \vec{r}_v = (x, y, z), E_i].$$

(4.17)

The energy bin width for the sum over energy varied from 0.20 to 0.25 MeV (the bin widths chosen were the same as the widths for the published photon energy spectra, and hence differed for each spectrum).

4.2.3 Scatter to Primary Dose Ratio

The scatter to primary dose ratio (SPR) was computed by taking the ratio of equations (4.17) and (4.8), which is expressed as

$$\text{SPR}(\vec{r}_d) = \frac{\sum_{z=z_s}^{z_m} \sum_{y=y_b}^{y_f} \sum_{x=x_1}^{x_f} \sum_{E_i=E_{min}}^{E_{max}} S^A_P[\vec{r}_d, \vec{r}_v = (x, y, z), E_i]}{(\text{SSD/SDD})^2 \sum_{E_i=E_{min}}^{E_{max}} \Phi_o(E_i) T_P(\vec{r}_d, E_i) \frac{\mu_w(E_i)}{\rho_w} E_{ab,w}(E_i)}.$$ 

In this expression, the $\beta$ term in the numerator and denominator are not shown since they are assumed to cancel. This approximation, although not explicitly stated, was also used in the work of Jaffray et al. [47], Partridge and Evans [95], and Swindell and Evans [117].

The contribution to the SPR from photons that pass through the secondary photon collimator jaws was found to be negligible. Jaw transmission is $\approx 0.5$ to 1% of the primary photon beam intensity. This effect was also ignored in [47], [95], and [117].

4.2.4 Examples

The analytical forms for the dose from primary and scatter radiation [see equations (4.8) and (4.17) respectively] were evaluated for homogeneous water phantoms and
Table 4.1. Physical and electron density relative to water ($\rho/\rho_w$, $\rho_e/\rho_{e,w}$ respectively) for the simulated lung and bone tissues used in the analytical and Monte Carlo calculation. The effective atomic number, $\bar{Z}$, is also listed. For lung, the value of $\bar{Z}$ is approximated by the value for water.

<table>
<thead>
<tr>
<th>Heterogeneity</th>
<th>$\rho/\rho_w$</th>
<th>$\rho_e/\rho_{e,w}$</th>
<th>$\bar{Z}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lung</td>
<td>0.250</td>
<td>0.248</td>
<td>7.51</td>
</tr>
<tr>
<td>Bone</td>
<td>1.850</td>
<td>1.737</td>
<td>12.31</td>
</tr>
</tbody>
</table>

water phantoms containing lung or bone slabs. The terms to account for beam divergence (SSD/SDD) and (SSD/$\tau_w$) were not included since these examples used a parallel source. These cases were chosen to show briefly how the primary photon energy and the presence of heterogeneities within the phantom affect the portal scatter to primary dose ratio. The lung and bone slab phantoms were chosen since they are low and high density heterogeneities, respectively. The physical and electron densities for the lung and bone are listed in table 4.1 and were taken from [21]. In all cases, the photon sources were broad, parallel, monoenergetic beams and the air gap between the phantom and the imager was equal to 50 cm. This air gap size was chosen since the imager dose from scatter has been shown to be uniform both theoretically [117] and experimentally [12] (for point sources) when the air gap is larger than or equal to 50 cm. In all cases the analytical results were compared to data from Monte Carlo simulation (code SDOSXYZ, which is described later in chapter 5) to evaluate the validity of the analytical approach.

Two photon beam energies were selected for the examples to illustrate the behaviour of the SPRs for low and high energies of the radiotherapy beam. The mean energy for a polyenergetic beam is approximately one third of the accelerating potential. For example, a 6 MV beam has a mean energy of 1.9 MeV, and as a rough guide, the 6 MV beam behaves like a 2 MeV monoenergetic photon source. For the SPR examples here, monoenergetic energies equal to 2 and 8 MeV were chosen since 6 MV and 24 MV are
common low and high-energy beams, respectively.

Relatively large field sizes were used for these examples (16×16 cm² and 20×20 cm²) since the scatter was significant, whereas for smaller fields (for example, 5×5 cm²) the SPR was ≈0.01.

In the analytical method, the dose $D$ from primary and scatter radiation was calculated from the collision KERMA $K_c$, multiplied by the quantity $\beta$, which accounts for the fact that the dose is affected by secondary electrons generated upstream of the dose deposition point [63]. $\beta$ depends on the photon beam energy and beam type (for example, parallel versus point source), the field size, the depth within the phantom, and the phantom material. $\beta$ can be estimated from first principles [63] and through Monte Carlo simulation [41]. The latter approach was chosen for our work since the former approach is restricted to available published data. $\beta$ was calculated on the central axis at the depth of maximum dose from the ratio

$$\beta = \frac{D}{K_c}$$

where the total dose per incident photon fluence $D/\Phi$ was computed with the Monte Carlo simulation code DOSXYZ [103]. The collision KERMA per incident photon fluence $K_c/\Phi$ was found at the depth of maximum dose $d_{max}$ using

$$K_c(d_{max})/\Phi = \exp[-\mu(E)d_{max}] \frac{\mu(E)}{\rho} \bar{E}_{ab,w}(E).$$

The value for $\beta$ calculated at the depth of maximum dose within the imager was assumed to be applicable for the dose from primary as well as scatter. For example, for an 8 MeV photon beam collimated to a 5×5 cm² field, $\beta=1.058\pm0.008$: figure 4.3 shows the absorbed dose calculated with Monte Carlo simulation, the collision KERMA, and the product of $\beta$ and the collision KERMA. The photon cross section data used was the same for the collision KERMA calculation and the Monte Carlo simulation.
In figure 4.4 the dose profiles across the imager for the scatter and primary radiation are illustrated for the homogeneous 20 cm thick water phantom. For both the 2 and 8 MeV photon beams, the scatter dose is uniform and increases with increasing field size since a larger volume is exposed to the primary photon fluence. The dose from both primary and scatter radiation are higher for the 8 MeV beam than for the 2 MeV beam. Table 4.2 tabulates the scatter to primary dose ratios, which shows that the SPRs are slightly lower for the higher beam energy. The probability of Compton scattering at non-zero scattering angles decreases with increasing photon beam energy (see figure 1.9), which is probably why the SPRs are lower for 8 MeV compared to 2 MeV.

The heterogeneous phantoms contained slabs of bone or lung within a 20 cm thick water phantom. The lung slab was 8 cm thick, while the bone slab was 3 cm thick. Figure 4.5 and table 4.2 show the results for the heterogeneous cases. The graphs show that the lung and bone heterogeneities have a strong affect on the dose from primary photons but very little effect on the dose from scattered particles. In the presence of a

**Figure 4.3.** Graph illustrating the relationship between KERMA (---) and dose (histogram) per unit incident photon fluence. The KERMA was calculated for a monoenergetic 8 MeV photon beam incident on water. The dose was computed using Monte Carlo simulation (DOSXYZ) with a monoenergetic, parallel photon beam of energy 8 MeV. Field size 5x5 cm². \( \beta = 1.058 \pm 0.008 \). KERMA multiplied by \( \beta \) (---).
Table 4.2. Scatter to primary dose ratios calculated on the central axis from the Monte Carlo simulation results for a parallel beam incident on a 20 cm thick phantom. The imager was represented as a rectangular block of water and the doses were scored at the depth of maximum dose for the respective beam energy.

<table>
<thead>
<tr>
<th>Case</th>
<th>Energy (MeV)</th>
<th>Field size (cm²)</th>
<th>SPR&lt;sub&gt;MC&lt;/sub&gt;</th>
</tr>
</thead>
<tbody>
<tr>
<td>Homogeneous</td>
<td>2</td>
<td>16×16</td>
<td>0.035</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>20×20</td>
<td>0.050</td>
</tr>
<tr>
<td></td>
<td>8</td>
<td>16×16</td>
<td>0.025</td>
</tr>
<tr>
<td></td>
<td>8</td>
<td>20×20</td>
<td>0.034</td>
</tr>
<tr>
<td>Lung slab</td>
<td>2</td>
<td>20×20</td>
<td>0.035</td>
</tr>
<tr>
<td>Bone slab</td>
<td>2</td>
<td>20×20</td>
<td>0.053</td>
</tr>
<tr>
<td>Lung slab</td>
<td>8</td>
<td>20×20</td>
<td>0.032</td>
</tr>
<tr>
<td>Bone slab</td>
<td>8</td>
<td>20×20</td>
<td>0.045</td>
</tr>
</tbody>
</table>

In the presence of a low density inhomogeneity (lung), the SPR decreases compared to a homogeneous water phantom of the same physical thickness mainly because the dose from primary photons increases due to the decreased attenuation of the primary photon beam. Conversely, in the presence of a high density inhomogeneity (bone), the SPR increases compared to a homogeneous water phantom of the same physical thickness mainly because the dose from primary radiation decreases due to the increased attenuation of the primary beam.
Figure 4.4. Dose profiles at the detector for homogeneous phantoms. The dose profiles from primary photons as calculated from Monte Carlo simulation (—) and using the analytical equations (○) are shown as well as the dose profiles from scatter, Monte Carlo (——), analytical (□). In all cases, the phantom was 20 cm thick (water) and a parallel photon source was used in the simulation and calculation. The photon energies and field sizes for each part in the figure are: (a) 16×16 cm² field, 2 MeV photons; (b) 20×20 cm², 2 MeV; (c) 16×16 cm², 8 MeV; and (d) 20×20 cm², 8 MeV.
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Figure 4.5. Dose profiles at the detector for heterogeneous phantoms. The dose profiles from primary photons as calculated from Monte Carlo simulation (—) and using the analytical calculation (○) are shown as well as the dose profiles from scatter, Monte Carlo (- - -), analytical (□). In all cases, the field size was 20×20 cm², the total physical thickness of the phantom was 20 cm, and a parallel photon source was used in the simulation and model calculation. The photon energy and inhomogeneity for each part of the figure are: (a) 2 MeV photons, lung slab phantom; (b) 2 MeV, bone slab; (c) 8 MeV, lung slab; and (d) 8 MeV, bone slab.
In all of the examples the Monte Carlo results for the scatter to primary dose ratio at the central axis was accurately predicted by the analytical approach. The mean absolute difference between the analytical and Monte Carlo SPRs on the central beam axis was 0.003, and the maximum absolute difference was 0.006.

4.3 SUMMARY

A method was described in this chapter to calculate the imager dose from primary photons as well as from first-order Compton scattered photons. For a simple parallel photon source the analytical method accurately predicted the SPR in comparison to Monte Carlo results. In equation (4.18), multiply scattered particles were neglected and the portal scatter dose was approximated by a uniform distribution. The accuracy of equation (4.18) is unknown. Therefore, it was necessary to validate the equation with both Monte Carlo simulation and with experimental measurements to ensure that the cumulative affect of each approximation is acceptable.

A computer program was written to calculate the dose from primary radiation in equation (4.8), the dose from scatter in equation (4.17), and the scatter to primary dose ratio. In this program the scattering object was described as a three-dimensional matrix with a voxel size of 0.25 cm$^3$, and the detector dose was computed on a rectangular grid with variable pixel spacing. The dose from primary photons, equation (4.8), was evaluated at each pixel on the detector for heterogeneous phantoms and at the central beam axis for homogeneous cases. The scatter dose in equation (4.17) was evaluated at the central beam axis for all cases. When comparing the doses calculated by this analytical method to the Monte Carlo simulation results, the doses from the analytical calculation were converted to dose per photon in units of [Gy photon$^{-1}$] by applying the unit conversion [MeV g$^{-1}$]=[1.602×10$^{-10}$ Gy] to equations (4.8) and (4.17).
In this chapter, the accuracy of the analytical scatter to primary dose ratio (SPR) calculation presented in chapter 4 is examined. Monte Carlo (MC) simulation was chosen for the validation of the analytical method since with its use the detector dose can be separated into different components according to particle type and interaction history, which cannot be determined experimentally. The first section of this chapter briefly reviews current MC simulation times for calculating the dose within body phantoms. Second, the MC code SDOSXYZ that was written and verified for this work is described.

While previous reports have shown that the first-order scatter dominates the photon scatter fluence [47, 70], an estimate of the portal scatter to primary dose ratio from first-order Compton scatter and from multiply scattered particles is lacking. The third section of this chapter presents the simulation to score the SPRs from each scatter mode (first-order Compton, multiple photon scatter, and patient-generated electrons) for homogeneous and anthropomorphic phantoms.

In the fourth section, both the MC simulation results and the validation of the analytical method are discussed. The results for the homogeneous cases cover a wide range of beam energies, phantom thicknesses, field areas, and source to detector distances. A summary of the MC validation of the analytical SPR method is given in the fifth and final section. This work was published [90].
5.1 MONTE CARLO SIMULATION TIME FOR DOSE CALCULATION

The Electron Gamma Shower (EGS4) MC package for simulating photon and electron/positron transport [10] (Rogers and Bielajew, 1989) was chosen for the study since EGS4 is an extensively verified set of codes for simulating radiotherapy beams. The Monte Carlo code written for this work, SDOSXYZ, is a variant of the EGS4 code DOSXYZ. The first S in SDOSXYZ stands for scatter. SDOSXYZ (see section 5.2) uses the same description of the phantoms and photon source as DOSXYZ. Typically, our simulation consists of a phantom representing the patient, a large air gap between the phantom and the detector, and a homogeneous water slab for the portal detector (see figure 5.1). The phantom and detector are defined by a set of voxels, or cubes, of variable length in the X, Y, and Z planes.

When the computation time (or CPU time) required to achieve results of acceptable accuracy with MC simulation is long, as was the case here, the CPU time and methods to minimize this time are important. Therefore previously reported CPU times and techniques to reduce that time are included here. Total CPU time depends on the size of the dose scoring voxels, overall volume of the simulation phantom, number of photon histories, energy of the incident photon beam, and accuracy desired in the final result. The Stanford MC group [25, 65] reported simulation times for photon beam dose calculation within a body phantom using a network of 22 Pentium Pro 200 MHz personal computers (PCs). A rough estimate of the CPU time for photon beams [65] was stated as \( \approx 30 \) minutes using the 22 PC network. This estimate was for dose scoring voxels from 2-5 mm on a side, beam energies from 4 to 15 MV, and uncertainties of 1% (one standard deviation). In the present work, five 333 MHz processors were used to reduce the overall computation time.

The aim of the PEREGRINE project [24, 126] is to provide rapid, accurate MC cal-
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Figure 5.1. The geometry for the Monte Carlo simulation. All field sizes were specified at the isocentre, which was at 100 cm from the photon source. The dose scoring bin was located on the central beam axis at the depth of maximum dose in the detector.

culation. This goal is achieved using multiple computer processors and techniques to increase the efficiency at the expense of reduced accuracy in certain areas. For example, the electrons generated from photons interacting with the secondary collimator jaws are ignored in the simulation. This approximation leads to a loss of fine detail in the tail of the beam profiles under the collimator jaws but the absolute dose estimates from the simulation with this approximation are in good agreement with experimental measurements up to beam energies of 18 MV. In the current MC validation the transmission and scatter from the collimator jaws were ignored to decrease the CPU time.

5.2 The Monte Carlo Code: SDOSXYZ

SDOSXYZ scores the total dose per unit incident fluence in a Cartesian geometry for a specified particle source. Details of the code are given in appendix A. SDOSXYZ scores the primary dose $P$ and the scatter dose $S$ at the portal detector. Separation of the
primary and scatter dose was performed by examining the first interaction of each photon. If a photon interacted first in the detector, then the dose from the Compton scattered electron (or electron/positron from pair production) was scored as dose from primary. If a photon interacted first in the patient phantom, then the entire dose in the detector from the resulting particle shower was scored as dose from scatter. SDOSXYZ also separates $S$ according to particle type and history: charged particles (CP), singly scattered photons (F), and multiply scattered photons (MS). The multiple scatter component includes photons from bremsstrahlung, annihilation, and multiple Compton scattering. SDOSXYZ scores the dose per primary photon [Gy photon$^{-1}$] and results are presented as SPRs (for example, SPR$_{CP}^{MC}$).

SDOSXYZ, like DOSXYZ, divides the total number of photon histories for the simulation into ten equal batches. In this work the superscript $MC$ indicates an MC calculated quantity. After all the photon histories are simulated, the average $\overline{\text{SPR}}^{MC}$ and standard deviation $\text{SPR}_{\sigma}^{MC}$ of the total $\text{SPR}^{MC}$ for the ten batches are computed. As well, the average and standard deviation are also computed for each component of the SPR (that is, for $\text{SPR}_F^{MC}$, $\text{SPR}_{CP}^{MC}$, and $\text{SPR}_{MS}^{MC}$). The standard deviation of the SPR, $\text{SPR}_{\sigma}^{MC}$, was calculated from

$$\text{SPR}_{\sigma}^{MC} = \sqrt{\frac{\sum_{i=1}^{N} (\text{SPR}_i^{MC} - \overline{\text{SPR}}^{MC})^2}{N(N - 1)}}$$  \hspace{1cm} (5.1)$$

where $N$ is the number of batches ($N=10$).

The results from the code SDOSXYZ were verified by comparing simulation results to data from DOSXYZ as well as data reported by Ahnesjo [1]. Each comparison is documented here.

To verify that SDOSXYZ recorded the same total dose as compared to DOSXYZ, the total dose along the central beam axis for a stack of eleven tissue slabs was calculated by both codes and compared. This geometry is 25.6 cm thick and consists of slabs of
cortical bone, adipose tissue, lung, and muscle. The simulation was run for a parallel beam collimated to a 10×10 cm² field using beam energies of 4 MV and 24 MV [80]. This particular slab geometry and photon source were chosen since the results could be compared to results in [1], as discussed below. The voxels were 0.2 cm in the depth direction and 1×1 cm² in area, which is only a small difference from that used in [1] (cylindrical geometry with voxels of radius 0.2 cm). The larger area voxels used here did not affect the results and significantly reduced the simulation time. The total dose along the central beam axis, as computed from SDOSXYZ, is shown in figure 5.2. The root mean square deviation between the total dose from SDOSXYZ and DOSXYZ was found to equal 0.0001%, which was accurate enough for the current investigation.

To validate the separation of the dose from the different scattering modes, the results from SDOSXYZ, just described for the eleven slab thorax phantom, were compared to published results [1]. In this comparison, the dose from primary, first scatter, and multiple scatter were compared along the central beam axis and are shown in figure 5.2. Good quantitative agreement was found between our results and [1] for both 4 and 24 MV. Differences may have arisen between this work and Ahnesjo’s [1] because of the use of an earlier version of the simulation code [10, 83] in [1].

The results from SDOSXYZ were also verified for calculation of the portal scatter dose by calculating the scatter fraction of dose (SF)¹ and comparing these values to data reported by Jaffray et al. [47]. This comparison is shown in figure 5.3. The data of [47] was chosen since the scatter fraction data covered a wide range of beam energies (6 and 24 MV), while other sources of scatter data for portal dosimetry were limited to lower beam energies (see for example [95] and [117]).

The scatter fraction data calculated here used the same photon energy spectra as

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¹The scatter fraction of dose was defined as the ratio of the dose from scatter to the total dose, both measured on the central beam axis at the portal imaging plane.
Jaffray et al. [47], and the simulation setup was a point photon source incident on a 17 cm thick polymethacrylate slab. The field size was $30 \times 30 \text{ cm}^2$ and the photon spectra were from [80]. In the current work, the doses were computed by SDOSXYZ, while in [47], photon fluences were converted to scatter fractions assuming a Compton style detector [see equation (2.5) in section 2.3.1]. In [47] the detector dose from patient-

\[ \begin{align*}
\text{(a)} & \quad \text{Dose along the central beam axis as computed by SDOSXYZ (—) for a stack of slab phantoms irradiated with a 10x10 \text{ cm}^2 \text{ parallel } 4 \text{ MV photon beam. The total dose is separated into the primary, total scatter, and first scatter dose components. Good agreement is obtained with previous results reported by Ahnesjo [1] (○). (b) Same as in (a) except for a 24 \text{ MV photon beam. The doses from primary and first scatter were omitted in (b) for clarity.}
\end{align*} \]
generated electrons was neglected in the SFs. Since the magnitude of the SF from charged particles was $\approx 0.03$ as calculated by SDOSXYZ for the 24 MV beam, this component of the SF was not included in the SFs from SDOSXYZ presented in figure 5.3 to allow for a meaningful comparison with the published data.

![Figure 5.3](image.png)

**Figure 5.3.** Verification of scatter fraction data computed by SDOSXYZ. The scatter fraction is plotted versus air gap for (a) 6 MV and (b) 24 MV photon beams incident on a 17 cm thick polymethacrylate slab. The field area was $30 \times 30$ cm$^2$. The scatter fraction data calculated using SDOSXYZ (•) are compared to measured (□) and calculated (△) results reported in Jaffray et al. [47]. Jaffray et al. [47] did not report measured scatter fraction data for the 24 MV beam.

Figure 5.3 shows that our SF results agree well with the measured and calculated data from [47] for both beam energies. In summary, our results from the MC code SDOSXYZ are in good agreement with previously published data both within heterogeneous slab phantoms and at the portal imaging plane.
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5.3 Simulation Parameters

A point source model was chosen as the photon source for the portal scatter dose simulation. The photon energy spectra used here for beam energies equal to 4, 6, 10, and 24 MV were taken from [80] while the spectrum for the 18 MV beam was from [127]. This is consistent with the choices by previous authors in portal dosimetry, for example [47, 95, 117].

For all cases, the simulation was performed with default values for the Parameter Reduced Electron Step Algorithm (PRESTA) [10] (Bielajew and Rogers, 1989) to reduce the computation time while maintaining accuracy in the model of the physics of the charged particle transport. A photon history was terminated when the energy of the photon fell below PCUT=10 keV, while electrons/positrons were tracked until their total energy fell below ECUT=521 keV.

The following subsection provides a description of the simulation for the homogeneous and anthropomorphic phantoms. Simulation results for these phantoms provided benchmarks with which to debug the code for the analytical SPR method described in Chapter 4 and quantitatively assess the accuracy of the analytical SPR approach. Heterogeneous phantoms were chosen to investigate the influence on the SPRs of tissues and body cavities with physical properties that are significantly different from those of water (for example, bones and air gaps).

5.3.1 Homogeneous Phantoms

Figure 5.1 showed the geometry for the MC simulation of the homogeneous phantoms. We calculated the SPRs (total SPR, as well as the SPR for each of the three scatter modes) for a wide variety of cases for homogeneous phantoms. This data set is unique since patient-generated electrons were included in the SPRs. The SPRs were calculated
for a single voxel on the central axis at the depth of maximum dose for a 10×10 cm² field. A single, square voxel on the central beam axis with a length that was 20% of the field length as projected onto the scoring plane was used to be consistent with data reported by previous authors [117, 118] and to be able to obtain accurate results within a reasonable CPU time. The thickness of the voxel was such that the dose remained within ±1% of the maximum dose throughout this depth. The thickness and depth of the dose scoring voxel was 0.5 cm and 1.5 cm, respectively, for the 6 MV beam, and 0.8 cm and 4.2 cm for the 18 MV beam.

The analytical approach approximates the portal scatter dose by a uniform distribution across the imager, which occurs when the air gap between the patient and the imager is at least 50 cm [12]. Consequently, large source to detector distances (SDDs) were used for the MC studies; the SDDs varied from 150 to 230 cm. Since the tumour volume is usually placed at a distance of 100 cm from the photon source, and the minimum air gap for the analytical method was 50 cm, the minimum SDD for simulation of the homogeneous phantom was selected as 150 cm. A maximum SDD of 230 cm was chosen since this is the source to floor distance.

SPRs were calculated for 6 and 18 MV photon spectra, for field areas from 3×3 cm² to 20×20 cm² at the isocentre, and for homogeneous water slab thicknesses from 10 to 30 cm thick.

5.3.2 Anthropomorphic Phantoms

Three heterogeneous anthropomorphic phantoms were simulated as part of the MC validation (figure 5.4). The treatment sites chosen (neck, thorax, and pelvis) cover a representative range of tissue thicknesses and beam energies encountered clinically. The beam energy, field area, and irradiation setup for each phantom were based on standard patient treatments. The atomic composition of each tissue was taken from the Interna-
Figure 5.4. The anthropomorphic phantoms: (a) neck, (b) thorax, and (c) pelvis. Distances are in centimeters. Each phantom is shown in an axial view. Field areas (measured at the isocentre) and beam orientations were: neck, 8x8 cm² lateral; thorax, 12x12 cm² anterior; and pelvis, 10x10 cm² lateral. For each case, the beam position is illustrated by the divergent solid lines and the central beam axis is indicated by the dashed lines.

The voxel size at the detector was set to 1 cm along the X axis and 2 cm along the Y axis. The anthropomorphic phantoms were symmetric about the Y axis to allow the volume of the voxels to be doubled, which reduced the CPU time² by a factor of $\sqrt{2}$.

²The uncertainty in the dose computed within a voxel is proportional to the square root of the number of photon histories for the simulation. Therefore, by doubling the volume of the voxel, the number of histories can be reduced by a factor of $\sqrt{2}$. 
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A neck phantom was selected since the imager scatter dose from this phantom would be minimal due to the relatively small fields used for neck treatments and the short beam path through the tissue. Lower beam energies (4 and 6 MV) are typically applied for head and neck treatment sites since the depths of the tumours are small. Therefore, the neck phantom was simulated with 4 and 6 MV beam energies (that is, two separate results were obtained, one with the 4 MV beam and another with the 6 MV beam). The field area for the neck case was 8×8 cm² and this was a lateral field passing through the spine and trachea as shown in figure 5.4(a).

A thorax phantom irradiated with an anterior beam through one of the lungs was selected since this treatment configuration involves a large, low-density organ (lung). In [89], the lung density relative to water for cancer patients undergoing radiotherapy was reported to vary from 0.15 to 0.45, with a mean of 0.28±0.03. Lower lung densities are found in emphysema patients, while patients with pneumonia have higher lung densities. Three results were obtained for the thorax phantom to cover the range of lung densities found clinically; the lung densities simulated were 0.1, 0.24, and 0.5 g cm⁻³. A beam energy of 6 MV and a field area of 12×12 cm² were used for these cases.

The third phantom chosen simulated lateral irradiation of the pelvis with the beam passing through the femoral heads. This site was chosen since, for the higher photon energies used for these cases, attenuation of the primary photon beam is greatest due to the large photon path through the tissue and high density bone. As a result of the attenuation of the primary beam, the SPRs from multiple photon scatter and charged particles are large and therefore, this phantom tests the validity of the analytical approach for an extreme case. Tumours within the pelvis are treated with 10, 18 and 24 MV beams. The pelvis phantom was simulated for these three beam energies (10, 18, and 24 MV) with a 10×10 cm² lateral field.
5.4 RESULTS

In this section, the MC SPR results are described in three parts. First, the central axis SPRs calculated using MC simulation for the homogeneous phantoms are presented. The first subsection also examines the SPR for each scatter mode as a function of off-axis distance at the detector for the anthropomorphic phantoms. Second, the accuracy of the analytical method as compared to the MC data is tabulated. In the third subsection, the MC simulation uncertainties and computer simulation times are reported and briefly discussed.

5.4.1 SCATTER TO PRIMARY DOSE RATIOS

In figure 5.5, the sum of the SPR from multiply scattered photons and the SPR from patient-generated electrons, \((\text{SPR}_{\text{MS}}^{MC} + \text{SPR}_{\text{CP}}^{MC})\), is shown for an SDD of 185 cm. Figure 5.6 presents the total SPRs on the central axis as a function of field area, SDD, and beam energy as calculated with SDOSXYZ for the homogeneous water phantoms. This figure shows that the SPR decreases (in almost all cases) with increasing beam energy.

Figure 5.7 reports \(\text{SPR}_{F}^{MC}\), \(\text{SPR}_{\text{MS}}^{MC}\), and \(\text{SPR}_{\text{CP}}^{MC}\) as a function of SDD, beam energy, and field area for a 20 cm thick water phantom. The following observations can be made from this figure: (i) \(\text{SPR}_{F}^{MC}\) decreases most in magnitude with increasing SDD - this decrease is more dramatic for the 6 MV beam; (ii) \(\text{SPR}_{F}^{MC}\) is greater at the lower beam energy; and (iii) \(\text{SPR}_{\text{MS}}^{MC}\) is slightly larger at 6 MV than 18 MV. The trends just described in figures 5.6 and 5.7 are due to the increasing probability of Compton scattering at larger scattering angles as the incident photon energy decreases. The results for \(\text{SPR}_{\text{CP}}^{MC}\) using the 6 and 18 MV beams show that the increase in scoring voxel depth with beam energy was insufficient to compensate for the increased energy of patient-generated electrons.
Figure 5.5. The sum of the scatter to primary dose ratio for multiply scattered photons and patient-generated electrons, \((SPR_{MS}^{MC} + SPR_{CP}^{MC})\), as calculated using Monte Carlo for an SDD of 185 cm. The data is shown for beam energies of (a) 6 MV and (b) 18 MV incident on homogeneous water phantoms.

\(SPR_{CP}^{MC}\) is greater at the higher beam energy in figure 5.7 due to the increase in the energy of the scattered electrons (and hence, path of travel) with increasing photon beam energy.
Figure 5.6. Scatter to primary dose ratios on the central beam axis calculated using Monte Carlo simulation for the homogeneous water phantoms as a function of the area of the square field at the isocentre. SDDs varied from 150 to 230 cm. Beam energies and water phantom thicknesses shown: (a) 6 MV, 10 cm; (b) 6 MV, 20 cm; (c) 6 MV, 30 cm; (d) 18 MV, 10 cm; (e) 18 MV, 20 cm and (f) 18 MV, 30 cm. In parts (d), (e), and (f) the data for SDDs equal to 160, 185, and 230 cm were omitted for clarity.
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Figure 5.7. Contribution of $SPR^MC_F$, $SPR^MC_{MS}$ and $SPR^MC_{CP}$ to the total $SPR^MC$ as a function of SDD and field area as calculated using Monte Carlo simulation for a 20 cm thick water phantom. From the top left: (a) $SPR^MC_F$, 6 MV; (b) $SPR^MC_{MS}$, 6 MV; (c) $SPR^MC_{CP}$, 6 MV; (d) $SPR^MC_F$, 18 MV; (e) $SPR^MC_{MS}$, 18 MV and (f) $SPR^MC_{CP}$, 18 MV.
Figure 5.8 shows the off-axis SPRs for the neck phantom MC results. The SPR profiles were taken along the X axis across the simulated detector at an SDD of 185 cm for all the anthropomorphic phantoms: the coordinate system for the simulation was given in figure 5.4. The simulated detector is always perpendicular to the photon beam as shown in figure 5.1. At each voxel within the detector the scatter and primary doses were scored, and then the SPR calculated. Since the scatter dose is relatively uniform, the structure seen in the SPR is due to changes in the primary dose. For these cases, the SPR from multiply scattered photons and electrons was negligible. Therefore treatment sites involving the neck are potentially good candidates for the analytical SPR method, which accounts for only the first-order scatter dose.

In figure 5.9, the off-axis SPRs for the lateral irradiation of the thorax phantom are plotted. Due to the difference in density between lung and muscle, the edge between the lung and muscle is visible as a jump in the SPR at an off-axis distance of ≈5 cm. Changing the lung density from 0.1 to 0.5 relative to water increased the total SPR by only ≈0.01. The SPR from first-order scatter is within ≈0.005 of the total SPR, so that approximating the total scatter dose by that from first-order Compton scatter is a good approximation for this case.

Figure 5.10 shows the off-axis SPRs for the pelvis phantom. As the beam energy increased from 10 to 24 MV, the total SPR increased for this phantom. The SPR from charged particles was larger than the SPR from multiply scattered photons for all energies. The SPR from charged particles increased with increasing beam energy, however, the SPR from first and multiple photon scatter were indifferent to beam energy. The total SPR is within ≈0.02-0.03 of that from first-order scatter.
Figure 5.8. Scatter to primary dose ratio calculated using Monte Carlo as a function of off-axis distance along the X-axis for the neck phantom at a beam energy of (a) 4 MV and (b) 6 MV. The off-axis distance is measured along the detector, which was at a source to detector distance of 185 cm.
Figure 5.9. Scatter to primary dose ratio as a function of off-axis distance along the X-axis for the thorax phantom at a beam energy of 6 MV for lung densities relative to water of (a) 0.5, (b) 0.25 (normal density), and (c) 0.1. The SPR changes by less than one percent when the relative lung density changes from 0.1 to 0.5. The SDD was equal to 185 cm.
Figure 5.10. Scatter to primary dose ratio as a function of off-axis distance along the X-axis for the pelvis phantom at a beam energy of (a) 10 MV, (b) 18 MV, and (c) 24 MV. The SPR from multiply scattered photons increases negligibly with increasing beam energy compared to the change in the SPR from charged particles. The SDD was equal to 185 cm.
5.4.2 Validation of the Analytical SPR Calculation

Comparison of the SPRs calculated using the analytical method described in chapter 4 to the total SPR from the MC simulation for the neck phantom cases are given in figure 5.11. In the analytical calculation, the scatter dose across the imager was approximated by that on the central beam axis and the primary dose was calculated for each voxel at the detector. As expected from figure 5.8, the analytical method is a good approximation for the total SPR across the X axis for the neck phantom irradiated with a 4 or 6 MV beam. The differences between the analytical and the MC results are mostly due to the main approximation in the analytical approach: multiply scattered particles (photons and electrons) were neglected in the analytical method whereas they are included in the MC data.

Similarly, the results for the thorax cases are shown in figure 5.12 and the analytical SPR is also in good agreement with the MC results. The pelvis cases are illustrated in figure 5.13. For the pelvis cases, the SPR from multiply scattered particles is larger than for the neck and thorax phantoms. In turn, the large SPR from multiple scatter for the pelvis cases is due principally to insufficient buildup material on the imager to stop the patient-generated electrons before they reach the scoring voxel within the detector.

Three quantities were computed when comparing the analytically calculated SPRs to the MC SPRs: (i) the maximum difference between the two sets of SPRs, $\Delta \text{SPR}_{\text{max}}$, (ii) the mean difference,

$$\Delta \text{SPR} = \frac{1}{N} \sum_{i=1}^{N} (\text{SPR}_{\text{A},i} - \text{SPR}_{\text{MC},i}),$$

(5.2)

and (iii) the root mean square of the differences,

$$\Delta \text{SPR}_{\sigma} = \sqrt{\frac{\sum_{i=1}^{N} [(\text{SPR}_{\text{A},i} - \text{SPR}_{\text{MC},i}) - \Delta \text{SPR}]^2}{N-1}}$$

(5.3)
where $N$ was the number of SPRs compared. For the results from the homogeneous phantoms, the SPRs were compared only on the central beam axis. The data for homogeneous cases were analyzed by grouping all the results for one beam energy together (total number of SPRs per beam energy: $N=72$), and then calculating $\Delta\text{SPR}_{\text{max}}$, $\overline{\Delta\text{SPR}}$, and $\Delta\text{SPR}_\sigma$. For the anthropomorphic cases, the MC results were analyzed for each simulation separately and the SPRs were compared for each pixel on the detector; $N$ was equal to the number of pixels on the simulated detector ($N=98$, $200$, and $162$ for the neck, thorax, and pelvis cases respectively).

Good agreement was obtained between $\text{SPR}_F^A$ and $\text{SPR}_F^{MC}$ for the homogeneous water phantoms: at $6 \text{ MV}$, $\overline{\Delta\text{SPR}} \pm \Delta\text{SPR}_\sigma = 0.000 \pm 0.003$ and $\Delta\text{SPR}_{\text{max}} = 0.009$ while at $18 \text{ MV}$, $\overline{\Delta\text{SPR}} \pm \Delta\text{SPR}_\sigma = 0.001 \pm 0.003$ and $\Delta\text{SPR}_{\text{max}} = 0.010$ [see equations (5.2) and (5.3)].

Table 5.1 summarizes the quantitative comparison between $\text{SPR}_F^{MC}$ and $\text{SPR}_F^A$ for the anthropomorphic phantoms. The mean difference between $\text{SPR}_F^{MC}$ and $\text{SPR}_F^A$ is equal to the mean difference over the field between $\text{SPR}_F^{MC}$ and $\text{SPR}_F^{MC}$ (see figures 5.8, 5.9, and 5.10). The mean difference between $\text{SPR}_F^{MC}$ and $\text{SPR}_F^A$ was mainly due to approximating the total scatter dose by the first-order Compton scatter dose only. From table 5.1 the mean difference between the analytical and MC results, $\overline{\Delta\text{SPR}}$, was $0.005$ or less for both the neck and thorax cases. For the pelvis cases, $\overline{\Delta\text{SPR}}$ was $0.03$ or less.

The accuracy of the current analytical method is comparable to that of similar techniques. The SPR model of Swindell and Evans [117] was shown to have a mean experimental difference of $0.005$ or less for an SDD of $200 \text{ cm}$, water phantom thicknesses up to $30 \text{ cm}$, beam areas up to $400 \text{ cm}^2$, and beam energies of $6$ and $10 \text{ MV}$ [95]. The analytical SPR model of Spies et al [114] agreed within $0.02$ to $0.03$ with MC results for off-axis SPRs using air gaps from $6.3$ to $18.3 \text{ cm}$, a $6 \text{ MV}$ radiosurgical field, and copper phantoms up to $3.5 \text{ cm}$ thick.
Figure 5.11. Graph of the scatter to primary dose ratio as a function of off-axis distance along the X-axis calculated from Monte Carlo simulation (—) and analytically (●) for the neck phantom. Results are shown for beam energies of (a) 4 MV and (b) 6 MV.
Figure 5.12. Graph of the scatter to primary dose ratio as a function of off-axis distance along the X-axis calculated from Monte Carlo simulation (—) and analytically (●) for the thorax phantom. Results are shown for lung densities relative to water equal to (a) 0.5, (b) 0.25, and (c) 0.1.
Figure 5.13. Graph of the scatter to primary dose ratio as a function of off-axis distance along the X-axis calculated from Monte Carlo simulation (—) and analytically (●) for the pelvis phantom. Results are shown for beam energies of (a) 10 MV, (b) 18 MV, and (c) 24 MV.
Table 5.1. Agreement between the scatter to primary dose ratios calculated from Monte Carlo simulation ($SPR_{MC}$) and analytically ($SPR_{A}$) for the anthropomorphic phantoms. The second column lists the number of pixels at the detector used in the analysis of the accuracy of the predicted scatter doses, $N$.

<table>
<thead>
<tr>
<th>Case</th>
<th>Number of pixels (N)</th>
<th>$\Delta SPR \pm \Delta SPR_\sigma$</th>
<th>$\Delta SPR_{max}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Neck, 4MV</td>
<td>98</td>
<td>0.000±0.001</td>
<td>0.003</td>
</tr>
<tr>
<td>Neck, 6MV</td>
<td>98</td>
<td>-0.002±0.001</td>
<td>0.005</td>
</tr>
<tr>
<td>Thorax, low density lung</td>
<td>220</td>
<td>-0.004±0.003</td>
<td>0.014</td>
</tr>
<tr>
<td>Thorax, normal density lung</td>
<td>220</td>
<td>-0.005±0.003</td>
<td>0.013</td>
</tr>
<tr>
<td>Thorax, high density lung</td>
<td>220</td>
<td>-0.005±0.003</td>
<td>0.013</td>
</tr>
<tr>
<td>Pelvis, 10 MV</td>
<td>162</td>
<td>-0.014±0.007</td>
<td>0.029</td>
</tr>
<tr>
<td>Pelvis, 18 MV</td>
<td>162</td>
<td>-0.018±0.007</td>
<td>0.038</td>
</tr>
<tr>
<td>Pelvis, 24 MV</td>
<td>162</td>
<td>-0.027±0.008</td>
<td>0.050</td>
</tr>
</tbody>
</table>

5.4.3 MONTE CARLO SIMULATION TIMES AND UNCERTAINTIES

Minimizing the CPU time for the MC simulation was important for the validation described in this chapter. The total simulation time for the homogeneous and anthropomorphic phantom results described in section 5.3 alone required thirteen weeks using a single 333 MHz processor. This estimate neglects the CPU time for debugging and validating the SDOSXYZ results, debugging the simulation files, and the time to obtain simulation results for other parts of this thesis.

Several methods were used to minimize the total CPU time. For the heterogeneous cases, the voxels at the detector were twice as wide along the Y axis (that is, along the axis of symmetry for these phantoms) as that along the X axis. This was possible because of the symmetry of the phantoms. Making use of this symmetry reduced the total number of photon histories by a factor of $\sqrt{2}$. For the homogeneous cases, the SPR was calculated for a single, large voxel on the central axis. The thickness of the dose scoring voxel at the detector along the source to detector ray line was maximized for
Table 5.2. Monte Carlo CPU times and total number of photon histories for the homogeneous water phantoms. This table contains a representative sample of all the cases, since there are too many to list individually. The source to detector distance for these results was 200 cm and the field area was $14 \times 14 \text{ cm}^2$.

<table>
<thead>
<tr>
<th>Beam energy (MV)</th>
<th>Phantom thickness (cm)</th>
<th>Number of histories (millions)</th>
<th>Number of histories/hour (millions/hour)</th>
<th>CPU time (hours)</th>
<th>Absolute uncertainty in the SPR</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>10</td>
<td>25</td>
<td>4.1</td>
<td>6</td>
<td>0.002</td>
</tr>
<tr>
<td>20</td>
<td>40</td>
<td>4.1</td>
<td>6</td>
<td>0.004</td>
<td></td>
</tr>
<tr>
<td>30</td>
<td>45</td>
<td>1.7</td>
<td>17</td>
<td>0.005</td>
<td></td>
</tr>
<tr>
<td>18</td>
<td>10</td>
<td>8</td>
<td>2.4</td>
<td>3</td>
<td>0.003</td>
</tr>
<tr>
<td>20</td>
<td>12</td>
<td>1.3</td>
<td>9</td>
<td>0.003</td>
<td></td>
</tr>
<tr>
<td>30</td>
<td>18</td>
<td>1.0</td>
<td>18</td>
<td>0.004</td>
<td></td>
</tr>
</tbody>
</table>

all cases, which increases the volume of the voxel. Maximizing the dose scoring voxel volume increased the probability of dose deposition within the voxel, and thus, reduced the simulation time.

Each MC simulation was run until the absolute uncertainty in the total SPR was less than or equal to 0.01 (for example, if the SPR was equal to 0.20 and the absolute uncertainty was 0.01, then the SPR was $0.20 \pm 0.01$). Appendix B lists the MC calculated SPRs for the homogeneous water phantoms: these tables include the total SPR, as well as the SPR from each scatter mode (first-order Compton, multiple photon scatter, and patient-generated electrons). Absolute uncertainties are reported for the total SPR and each component of the SPR. The mean MC uncertainties for the SPRs (averaged over the field at the detector) for the anthropomorphic phantoms were 0.001, 0.003, and 0.006 for the neck, thorax, and pelvis cases respectively.

Table 5.2 gives the CPU times and the total number of photon histories for a representative sample of the MC results for the homogeneous phantoms. These times were obtained using a 333 MHz Pentium II processor and the Linux operating environment.
5.5 SUMMARY

In this chapter, the accuracy of the analytical method was examined. The analytical method was found to be a good approximation for the neck and thorax phantoms, since the SPR from multiply scattered particles in these cases was small. For the pelvis case, the mean difference between the analytical SPR was within 0.015-0.030 of the total SPR calculated using MC simulation. To improve the accuracy of the analytical method for such cases, it would be necessary to include the portal scatter dose from patient-generated electrons, multiple Compton scattering, bremsstrahlung, and annihilation.

Since the sensitive area of the imager is fixed and the analytical SPR method requires a 50 cm air gap, application of the analytical method is limited by the maximum area that the imager can measure when using a 50 cm air gap. Reducing the air gap allows larger field areas to be used, however the accuracy of the analytical method will decrease since the SPR from multiply scattered particles increases with decreasing air gap.

The advantage of the analytical method, compared to the semi-empirical method of Boellaard et al. [14], is that the SPRs were calculated for heterogeneous cases without measuring data for a look-up table. If the analytical SPRs are valid for experimental phantoms, then another institution can use the analytical method and reduce the implementation time for calculating the portal scatter dose for heterogeneous cases.

Since the primary energy spectra for the analytical method are different from the experimental photon energy spectra, an experimental validation of the analytical SPR approach is presented in the following chapter. In reality, the mean energy of the photon spectra can decrease by \( \approx 6 \) to 15\% from the central axis to 10 cm off axis [80], as measured at the isocentre. This change in energy is termed off-axis softening. The analytical method ignores off-axis softening, therefore the experimental validation is necessary to determine the accuracy of the method.
An experimental validation is also necessary since the description of the linear accelerator for the Monte Carlo (MC) simulation was also incomplete. For example, the photon source was modeled for the simulation as an isotropic point source. For the clinical linear accelerators, the photon sources are spatially variant and diffuse.

This chapter describes the experiments carried out to measure the SPRs on the central axis for homogeneous and anthropomorphic phantoms. These measurements are quantitatively compared to the predicted SPRs from the analytical method, which was presented in chapter 4.

First, the choice of field sizes and phantom thicknesses for the experimental validation are discussed. Second, the beam characteristics of the clinical linear accelerators are reviewed to understand the limits of the photon source models for the analytical SPR calculation. Third, the phantom designs, tissue substitutes and measurement techniques for the experimental validation are discussed. The fourth section presents the results from the quantitative analysis of the agreement between the experimental and analytical SPRs. A summary is given in the fifth and final section. This work was published [90].

6.1 CLINICALLY RELEVANT CASES

Clinical linear accelerators can produce radiation field sizes up to 40×40 cm² at the isocentre and the physical thickness of the patient can vary by up to 40 cm. While
it may be possible to produce a 40x40 cm$^2$ field and a specific treatment site may be 10 cm thick, one would never encounter such a combination clinically for the treatment of solid, localized tumours. The experimental validation was carried out for a wide range of clinically relevant cases. The largest field areas for the validation corresponded to the largest areas used for each treatment site. The neck region is treated with 4 and 6 MV photon beams and the maximum field area is 14x14 cm$^2$. Tumours in the thorax are typically treated with a 6 MV beam; the maximum field area is usually 20x20 cm$^2$, although the field can be larger. Tumours in the pelvis are treated with anterior, posterior, and lateral fields; the beam energies applied range from 10 to 24 MV. The anterior and posterior fields are typically smaller than 20x20 cm$^2$, while the lateral fields are smaller than 16x16 cm$^2$.

6.2 PHOTON BEAM CHARACTERISTICS

This section briefly reviews several features of clinical linear accelerator photon beams and how these features influence the experimental validation of the analytical SPR calculation method. The topics covered include the photon energy spectra and the collimator scatter factor.

6.2.1 PHOTON ENERGY SPECTRA

The photon energy spectrum is probably the most important feature of the linear accelerator for dose calculation. For the Monte Carlo validation in chapter 5 the photon energy spectra were assumed to be invariant across the beam. Since complete, accurate spectra for real linear accelerators are difficult to obtain or calculate, this section briefly discusses the photon source models that other investigators have applied for portal scatter estimation and the accuracy of each study.
Jaffray et al. [47], Partridge and Evans [95], and Swindell and Evans [117] all measured SPRs or scatter fraction data on the central beam axis at the position of the portal detector, and these studies have several points in common. In these cases, the measured SPR or SF data for 6 or 10 MV beams showed good agreement with MC calculated values computed using the assumptions that (i) the experimental photon source was a point source, and (ii) generic photon energy spectra, such as those reported in Mohan et al. [80], were adequate to describe the photon spectra. Measured SPR or SF data for 6 or 10 MV beams showed good agreement with MC calculated values. Jaffray et al. [47] and Partridge and Evans [95] showed that the experimental and MC data agreed within the uncertainties in the data, and Swindell and Evans [117] report the mean difference between the calculated and measured SPRs to be 0.005. The comparison by Jaffray et al. [47] was carried out for air gaps from 10 to 60 cm and a 17 cm polymethacrylate slab phantom. The source to detector distance for the data reported by Partridge and Evans [95] and Swindell and Evans [117] was 200 cm. Since these previous studies assumed that the photon energy spectra was invariant across the beam and they obtained good agreement between calculated and measured SPRs or SFs on the central axis, this approximation was also made in the current analytical SPR calculation.

In this chapter, measured SPRs are reported for three linear accelerators (all Varian): a Clinac 600C, which produces a 4 MV photon beam; a Clinac 2100C (6 and 10 MV); and a Clinac 21EX (18 MV). The corresponding photon energy spectra for the analytical computation were from Mohan et al. [80] for 4, 6, and 10 MV, and Waggener et al. [127] for 18 MV. The experimental validation was restricted to SPRs measured on the central beam axis because these generic spectra specify the photon spectrum on the central axis only and the mean energy of the primary photons can decrease by \( \approx 6 \) to 15% between the collimator axis of rotation and 10 cm off-axis [80] (this change in beam energy is known as off-axis softening). The energy spectrum was assumed to be invariant across
the beam for the analytical calculation. The 4, 6, and 10 MV spectra were averaged over a beam radius of 3 cm (measured at the isocentre) (Mohan et al. [80]). The 18 MV spectrum was measured on the central axis using a field of diameter 0.2 cm (Waggener et al. [127]). Off-axis softening was not included in the analytical calculation and therefore introduces an error into SPR^A.

To determine the validity of using these generic spectra for our accelerators, percentage depth doses were computed using these spectra (with the EGS4 Monte Carlo simulation code DOSXYZ) and compared to experimentally measured depth doses. To maximize the agreement between the depth doses calculated with DOSXYZ and the measured doses, a small field (3x3 cm^2) was chosen since the spectra are averaged over a circular beam with a radius of 3 cm [80] (for the 18 MV spectrum the beam was 0.2 cm in diameter [127]).

An ionization chamber (IC10, Wellhofer Dosimetrie, Schwarzenbruck, Germany) was used for the measurements, which has a diameter of 0.6 cm. For the simulation, the dose scoring voxels were chosen to be 0.6 cm laterally and 0.5 cm along the depth axis to match the size of the ionization chamber. The simulation phantom was 10x10x40 cm^3 and the source to detector distance was 100 cm to correspond to the measurement setup. The standard deviation of the Monte Carlo calculated doses relative to the total dose was less than 2%. Percent depth doses calculated from MC simulation and measured experimentally are shown in figure 6.1. These curves were normalized to a depth of 10 cm (see section 1.9.1 for choice of normalization depth). Good agreement is seen between the experimental and simulated depth doses for all energies up to a depth of 35 cm, therefore the generic spectra are a good approximation on the central axis for the linear accelerators used here.
Figure 6.1. Comparison of the percentage depth doses measured experimentally (—) and calculated from Monte Carlo simulation (histogram) for photon beam energies of (a) 4 MV, (b) 6 MV, (c) 10 MV, and (d) 18 MV.

6.2.2 Collimator Scatter Factor, $S_c(FA)$

In the Monte Carlo simulation, the beam collimation was assumed to be perfect. With perfect collimation, all of the photons outside of the defined field size would be completely absorbed. In reality, however, photons scatter within the treatment head, which leads to a field size dependence of the dose per monitor unit.$^1$ This effect is included in photon

$^1$A monitor unit is the unit for the radiation dose measured by the ionization chambers in the treatment head of the linear accelerator.
treatment planning systems by measuring a quantity known as the collimator (or head) scatter factor, $S_{c}$ [48, 109, 125].

The collimator scatter factor depends on the photon source, field area (FA) at the isocentre, and the components in the treatment head (for example, the flattening filter, the monitor ionization chamber within the treatment head, and the collimator blocks). Measurement of $S_{c}(FA)$ is carried out using an air ionization chamber in a buildup cap or using a beam-coaxial narrow cylindrical phantom. The collimator scatter factor is defined as

$$S_{c}(FA) = \frac{D_{\text{air}}(FA)}{D_{\text{air}}(FA=10 \times 10 \text{ cm}^2)}$$  \hspace{1cm} (6.1)$$

where $D_{\text{air}}$ is the dose measured by the ionization chamber.

The value of $S_{c}(FA)$ increases nonlinearly from $\approx 0.9$ to $\approx 1.1$ as the field area increases [the range of values for $S_{c}(FA)$ quoted are from the Varian Clinac 2100C linear accelerator at our centre, 6 and 10 MV photon beams]. The analytical method for the SPR calculation neglected collimator transmission since the SPR measurement technique of Swindell and Evans [117] was used, which removes the collimator scatter effect by normalizing each dose to the dose measured in the same configuration but without the phantom in the beam.

### 6.3 Scatter to Primary Dose Ratio Measurements

In this section, the phantom designs and choice of tissue substitutes for the experimental validation are discussed. Following this, the measurement methods for the SPR on the central beam axis for homogeneous and anthropomorphic phantoms are presented.

#### 6.3.1 Phantoms

One advantage of the experimental methods versus Monte Carlo approaches is that the measurement time for each SPR was constant and was much shorter than the time
for calculating an SPR with Monte Carlo simulation. Since (i) a benchmark data set of measured SPRs covering a wide range of radiotherapy energies, phantom thicknesses, and field sizes was unavailable, (ii) this data would be potentially useful in the future for validating absolute scatter dose calculation methods, and (iii) the measurement time for the SPRs was relatively rapid, there was strong motivation to compile a comprehensive SPR database. Therefore, we measured SPRs for homogeneous water-equivalent phantoms using all the megavoltage beam energies at our centre (4, 6, 10, and 18 MV) and for field areas up to 28x28 cm$^2$. The water equivalent plastic blocks used for the measurements are discussed in section 6.3.2. SPRs were measured for water equivalent thicknesses up to 30 cm for 4 and 6 MV, and up to 40 cm for 10 and 18 MV.

Three anthropomorphic phantoms were designed to represent the neck, thorax, and pelvis and are shown in figure 6.2. Measurements were performed for the neck phantom irradiated with a lateral field for beam energies of 4 and 6 MV. For the thorax phantom, an anterior 6 MV beam was used. The pelvis phantom was irradiated with 10 and 18 MV lateral beams. The differences between the phantoms for the Monte Carlo validation in chapter 5 and this chapter included: the position of the photon beams and the heterogeneities, the atomic composition of the tissue substitutes, and the field areas. The position of the beams and heterogeneities was changed for the experimental validation to facilitate measurement of the SPRs; this method will be explained in section 6.3.4.

The dimensions for the phantoms were measured from an anatomical CT atlas [20] and the Rando phantom$^2$ (The Phantom Laboratory, USA). Section 6.3.2 discusses the choices of tissue substitutes for the experimental validation: aluminum was used to replace bone and cork was substituted for lung. The heterogeneities for each phantom were contained within a Lucite box with an open top and the box was filled with water. The density of

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$^2$The Rando phantom is a humanoid phantom constructed from a natural human skeleton cast inside material that has radiological properties similar to soft tissue. Lower-density material fills the rib cage, to simulate human lung at median respiratory state.
Figure 6.2. Size (in cm) and composition of the experimental anthropomorphic phantoms: (a) neck, (b) thorax, and (c) pelvis. The radiation beams are indicated by the divergent lines. In each case the centre of the radiation field in the phantom is shown by the dotted line. The images are axial views. The beam orientation for each case was: lateral for the neck and pelvis; anterior for the thorax. Every phantom consists of an outer Lucite box (sides 3 mm or 9 mm thick) with an open top. Permanent marks were made on the outside of the box for placement of the phantom at the isocentre of the clinical linear accelerator. Heterogeneities were fixed in place with electrical tape. The blocks of cork were wrapped tightly with masking tape and then sealed in plastic bags.

the Lucite walls was included in the analytical calculation.

6.3.2 Tissue Substitutes

Ideally, phantom materials chosen to test the model would be highly tissue equivalent, readily available, and easy to cut or machine. In this section the reasons for using the chosen tissue substitutes are discussed. The properties of interest for the experimental validation are the attenuation and scattering properties, which are described by the attenuation coefficient and electron density of the material, respectively. To compare the tissue equivalence of the substitute with the tissue it replaces, two quantities were examined: the electron density of the tissue substitute relative to water, $\rho_e$, and the mass attenuation coefficient ratio between the substitute and the tissue,

$$\left(\frac{\mu}{\rho}\right)_{\text{substitute}} \frac{\rho}{\text{tissue}} = \frac{(\mu/\rho)_{\text{substitute}}}{(\mu/\rho)_{\text{tissue}}} \quad (6.2)$$
Figure 6.3 shows graphs of the mass attenuation coefficient ratio for the tissue substitutes discussed in this section. Unless stated otherwise, material data presented here was from ICRU 44 [21].

Materials that were compared for muscle tissue substitutes included Solid Water (Gammex RMI, Middleton, WI), polystyrene, and water. Solid Water (or WT1) is an epoxy based material with fillers of polyethylene, phenolic microspheres, and calcium carbonate and was formed into 30×30×5 cm³ or 40×40×5 cm³ blocks. Solid Water was chosen as the tissue substitute for constructing the homogeneous phantoms since this material was available, is easier to handle than water, and the attenuation coefficient is closest to that for muscle (see figure 6.3). Table 6.1 summarizes the elemental composition and physical density of water and Solid Water.
Figure 6.3. The ratio of the mass attenuation coefficient for the tissue substitute to the mass attenuation coefficient for the tissue is plotted versus photon beam energy for alternative tissue substitutes. These ratios are shown for (a) muscle, (b) bone, and (c) lung. As well, in (c), Griffith lung and LN10 (commercial lung substitutes) were included for comparison to cork.
Table 6.1. Physical properties of the phantom materials. The physical densities were measured for these materials. The relative electron density for polyvinylchloride (PVC) was from ICRU 44 [21]. The physical and electron densities for cork were calculated on the basis that cork is a cellulose-based compound with a molecular weight of 162.14 and a chemical composition of $C_6H_{10}O_5$. The number of electrons per gram for cork relative to water was calculated to be 0.956.

<table>
<thead>
<tr>
<th>Material</th>
<th>Percent composition by weight (%)</th>
<th>Physical density $\rho$ (g/cm$^3$)</th>
<th>Electron density $\rho_e$ relative to water</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td></td>
<td>1.00</td>
<td>1.00</td>
</tr>
<tr>
<td>Solid Water</td>
<td>67.2 8.1 19.9 0.1 2.4 2.3</td>
<td>1.02</td>
<td>0.991</td>
</tr>
<tr>
<td>PVC</td>
<td>38.5 4.8</td>
<td>1.32±0.01</td>
<td>1.24±0.02</td>
</tr>
<tr>
<td>Aluminum</td>
<td></td>
<td>2.70</td>
<td>2.35</td>
</tr>
<tr>
<td>Cork, lot 1</td>
<td>44.4 6.2 49.4 — — —</td>
<td>0.518±0.006</td>
<td>0.495±0.006</td>
</tr>
<tr>
<td>Cork, lot 2</td>
<td>44.4 6.2 49.4 — — —</td>
<td>0.16±0.01</td>
<td>0.15±0.01</td>
</tr>
</tbody>
</table>

For the bone substitute, polyvinylchloride$^3$ (PVC) and aluminum$^4$ were compared. The relative electron density of PVC ($\rho_e=1.24$) is between that for cortical bone ($\rho_e=1.78$) and spongy bone ($\rho_e=1.15$), and the relative electron density of aluminum ($\rho_e=2.35$) is higher than that for cortical bone. Aluminum was selected over PVC since bone is a high-density heterogeneity and aluminum would test the algorithm in an extreme case.

Potential lung substitutes included woods (balsa and cork) [46] and expanded polyethylene foams [52]. Foamed materials with fillers, such as Griffiths lung or LN10/75, were also considered. The main challenge in selecting a lung tissue substitute was in obtaining a material that matched the physical density of lung used for treatment planning. Lung density in treatment planning varies from 0.15 to 0.45 g cm$^{-3}$, with a mean of 0.28±0.03 g cm$^{-3}$ [89]. The physical density of Griffiths lung ($\rho=0.26$ g cm$^{-3}$) and LN10/75 ($\rho=0.31$ g cm$^{-3}$) closely match the mean density of lung, however, these mate-

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$^3$Industrial Plastics and Paints, Richmond

$^4$6061 Aluminum, Metal Supermarkets, Richmond
rials were not as readily available as balsa wood or cork. The density of balsa wood can be quite low ($\rho=0.08 \text{ g cm}^{-3}$), which eliminated this choice. Since it was initially desirable to be able to measure the relative electron densities of the phantom materials with computed tomography, and we would have to apply corrections for beam hardening if using non-water equivalent material such as polyethylene [53], the expanded polyethylene foams were eliminated as choices.

Cork was chosen as the best alternative for the lung substitute, and the mass attenuation coefficient between cork and lung is comparable to the foam mixtures as shown in figure 6.3. Two batches of cork were purchased, one with a high density\(^5\) of $\rho \approx 0.5 \text{ g cm}^{-3}$ and one with a lower density\(^6\) of $0.16 \pm 0.01 \text{ g cm}^{-3}$. The lower density batch was used for the experimental validation since the lung is a low density inhomogeneity, and the lower density cork would test the algorithm in an extreme situation. The density of the cork was determined from the mass to volume ratio. The measurement uncertainty for the mass to volume ratio for the current work was $\approx 2\%$, and since the direct measurement of the density was faster than CT scanning the cork, the direct approach was used. Variation in the density of the cork within a batch was the main source of uncertainty in the density for the low-density cork. The uncertainty in the density of cork when determined from CT data, as found by Kohda and Shigematsu [55] for 360 measurements of the same sample of cork, was $\approx 4\%$ (measured density was $0.287 \pm 0.011$).

The mass attenuation coefficients for the chosen tissue substitutes are shown in figure 6.4. The mass attenuation coefficient for cork was calculated from the weighted sum of the mass attenuation coefficients of the elements in cork, where the weights were equal to the percentage of the element by weight (see table 6.1 and reference [49]).

Table 6.2 lists the effective atomic numbers $\bar{Z}$ for muscle and bone as well as for the

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\(^5\)Cork flooring underlay, Banner Carpets, Burnaby.
\(^6\)European Quality Cork Flooring, Port Coquitlam.
Figure 6.4. Graph of the mass attenuation coefficients for Solid Water, aluminum and cork.

Table 6.2. Effective atomic numbers for the tissue substitutes used for this work and for the human tissues they replace. [49].

<table>
<thead>
<tr>
<th>Material</th>
<th>Z</th>
</tr>
</thead>
<tbody>
<tr>
<td>Muscle</td>
<td>7.64</td>
</tr>
<tr>
<td>Solid Water</td>
<td>7.89</td>
</tr>
<tr>
<td>Cellulose</td>
<td>6.97</td>
</tr>
<tr>
<td>Bone</td>
<td>12.31</td>
</tr>
<tr>
<td>Aluminum</td>
<td>13</td>
</tr>
</tbody>
</table>

tissue substitutes.

6.3.3 Detector

The radiation detectors available for measuring the SPRs included three different portal imagers as well as the ionization chambers at our centre. An ionization chamber was chosen for the measurement of the SPRs since the calibration of the portal imagers for
dose measurements is inexact [31], therefore measurement of the SPRs with an ionization chamber was more accurate than with a calibrated portal imager.

A Farmer-type ionization chamber (PTW Freiburg, Germany, model N30001) was chosen for the experimental validation since this is expected to be the most accurate method for measurement of the SPRs. This chamber has a volume of 0.6 cm$^3$. The inner electrode is made of aluminum and the wall is composed of 0.275 mm of polymethacrylate and 0.15 mm of graphite. The electrode is 21.1 mm long and the inner chamber diameter is 6.1 mm. The small variation of the value of the radiation quality factor for this chamber, $k_Q$, versus beam energy means that the value of $k_Q$ in the numerator of the SPR will cancel with $k_Q$ in the denominator [121]. This ionization chamber was used in conjunction with a Victoreen 500 electrometer (Victoreen, Cleveland, Ohio).

6.3.4 Measurement Methods

In this section, the methods for experimentally measuring the SPRs for the homogeneous and heterogeneous phantoms are presented. For homogeneous cases a published approach was used, while for the heterogeneous cases a novel technique for measuring the SPR on the central axis was developed.

Large SDDs minimize the SPRs from multiply scattered photons and charged particles. The largest SDD for our commercial imagers was 185 cm. Investigators at the Royal Marsden Hospital, which is an active centre for the development of portal scatter dose prediction methods [43, 95, 114, 117], use a fixed SDD of 200 cm; SPRs were measured here with an SDD of 200 cm for comparison with the latter studies.

For the homogeneous cases, the SPRs were measured using the method presented by Swindell and Evans [117]. In this technique, the total dose on the central axis at the imaging plane is measured with and without the phantom in the beam. The ratio of the dose with the phantom divided by the dose without the phantom removes the effect
of the change in the machine output with changing field area (see the discussion on the collimator scatter factor in section 6.2.2). This ratio was denoted by $T_N$, where the $T$ stands for total and the $N$ for normalized. The dose from primary, $P_N$, was calculated by extrapolating the total dose $T_N$ versus field area $FA$ to zero field area, where the scatter signal should vanish. $T_N$ was fitted to a quadratic curve and the fitting was performed with the function minimization and error analysis program Minuit (CERN, Geneva, Switzerland). The physical interpretation of $P_N$ is $\exp(-\bar{\mu}t)$, where $\bar{\mu}$ is the mean linear attenuation coefficient for the photon beam along the central axis and $t$ is the thickness of the phantom. For each field area the SPR was then calculated using

$$SPR^{EXP}(FA) = \frac{T_N(FA) - P_N}{P_N}. \quad (6.3)$$

This method for extrapolating the dose from primary from the total dose is illustrated in figure 6.5. The SPR database measured here for the homogeneous cases is included in appendix C for reference.

The anthropomorphic phantoms were deliberately designed so that there were no heterogeneities along the central beam axis. This permitted measurement of the SPRs through a homogeneous section along the central axis (see figure 6.2). This was the main difference between the anthropomorphic phantoms for the experimental validation and those for the Monte Carlo validation discussed in chapter 5.

The SPRs were measured in two steps. First, the Lucite tank only contained water (no heterogeneities) and the normalized dose from primary $P_N$ was derived as previously described for the homogeneous cases. Second, the heterogeneities were placed in the tank and the normalized total dose $T_N$ was measured at the central beam axis for a range of field areas. The SPRs were then calculated according to equation (6.3). Since the SPR measurements were far faster than the Monte Carlo simulation, the field areas could be varied and ranged from $4\times4 \, \text{cm}^2$ up to a maximum field area (determined by
Figure 6.5. Illustration of the extrapolation method for deriving the dose from primary photons. Total measured dose $T_N$ ($\circ$) and the quadratic fit (—). The dose from primary radiation is equal to the total dose for a field area of zero. The error bars were omitted as they were too small to be seen.

the maximum field area used for that treatment site) in 2 cm increments. The maximum field area was equal to $14 \times 14$ cm$^2$, $16 \times 16$ cm$^2$, and $20 \times 20$ cm$^2$ for the neck, pelvis, and thorax cases, respectively. Two measurements for each data point were taken to minimize the effect of random errors.

The random error in the experimental SPRs for all cases is given approximately by

$$\delta_{SPR} = \left[ \left( \frac{\sigma_{T_N} + \sigma_{P_N}}{T_N - P_N} \right)^2 + \left( \frac{\sigma_{P_N}}{P_N} \right)^2 \right]^{1/2} \text{SPR}$$

$$\approx 2\sqrt{2} \frac{\sigma_T}{T} = 0.006$$

(6.4)

where ($\sigma_{T_N}, \sigma_{P_N}$) are the experimental uncertainties for $T_N$ and $P_N$, respectively, while $T$ and $\sigma_T$ are the total dose and standard deviation of the total dose, respectively. The maximum relative standard deviation of repeated measurements of the total dose was $\sigma_T/T=0.002$. The error bars for the experimental SPRs for all cases were equal to $\delta_{SPR} \approx 0.006$. 

6.4 Analysis Method

Differences between the analytical and experimental SPRs arise from both random and systematic errors. The random error in the experimental SPRs is given by equation (6.4). Systematic errors in the analytical SPR arise from neglecting the scatter from multiply scattered particles (multiple Compton scatter, bremsstrahlung, annihilation, and patient-generated electrons) and from assuming the photon energy spectrum is invariant across the beam. The analysis of the agreement between the analytical and experimental SPRs was the same as the analysis for the Monte Carlo validation in chapter 5. For the readers convenience, the equations are repeated here.

Three quantities were computed when comparing the analytically calculated SPRs to the experimental SPRs: (i) the maximum difference between the two sets of SPRs, \( \Delta \text{SPR}_{\text{max}} \), (ii) the mean difference,

\[
\overline{\Delta \text{SPR}} = \frac{1}{N} \sum_{i=1}^{N} (\text{SPR}^A_{F,i} - \text{SPR}^{\text{EXP}}_i),
\]

and (iii) the root mean square of the differences,

\[
\Delta \text{SPR}_\sigma = \sqrt{\frac{\sum_{i=1}^{N} [(\text{SPR}^A_{F,i} - \text{SPR}^{\text{EXP}}_i) - \overline{\Delta \text{SPR}}]^2}{N - 1}}
\]

where \( N \) was the number of SPRs compared and \( \text{EXP} \) stands for experimentally measured.

For the homogeneous phantoms, these three quantities (\( \Delta \text{SPR}_{\text{max}}, \overline{\Delta \text{SPR}}, \) and \( \Delta \text{SPR}_\sigma \)) were computed for each beam energy. \( N \) was different for each beam energy since the number of phantom thicknesses for which the SPRs were measured differed for each energy (see section 6.1). For example, for the SPRs measured using the homogeneous phantoms irradiated with the 18 MV beam, \( N \) was equal to 48 (2 SDDs \( \times \) 6 field areas \( \times \) 4 phantom thicknesses).
For the anthropomorphic phantoms, $\Delta\text{SPR}_{\text{max}}, \overline{\Delta\text{SPR}},$ and $\Delta\text{SPR}_\sigma$ were calculated for each phantom, beam energy, and source to detector distance. In these cases, $N$ was the number of field areas for which the measurements were made. Since the SPRs were measured for (square) field areas in 2 cm increments, $N$ was equal to 6, 7, and 9 for the analysis of the neck, pelvis, and thorax data, respectively. The field areas for the anthropomorphic cases were based on the typical field areas used for these sites as discussed in section 6.1.

6.5 Results and Discussion

In this section, the agreement between the experimental and analytical SPRs is presented and discussed. Calculation of the analytical SPRs requires the patient density data, which can be obtained from a CT scan. Since the density of the phantoms was known in this work, CT scans of the phantoms were unnecessary. The analytical SPRs were computed according to equations (4.8) and (4.17) in chapter 4.

SPRs were measured experimentally on the central axis for homogeneous Solid Water phantoms from 10 to 40 cm thick, beam energies from 4 to 18 MV, and SDDs of 185 and 200 cm. Figure 6.6 illustrates the majority of these SPRs for an SDD of 185 cm. Those SPRs that were omitted from the graphs (for example, SPRs for the 40 cm thick phantom irradiated with the 10 MV beam) exhibited poor agreement with the analytical SPR method. This lack of agreement probably occurred because neglecting multiply scattered photons in the analytical SPR calculation is a poor approximation in these cases.

Good agreement is seen in figure 6.6 between $\text{SPR}^A$ and $\text{SPR}^{\text{EXP}}$ for all four beam energies; this agreement is quantified in table 6.3 [see also equations (6.5) and (6.6)]. For the larger field areas and for the thicker phantoms, the analytical SPRs agree reasonably
well with the experimental data and are within $\approx 0.02$ of the experimental SPRs. The analytical method could be improved by including higher order scattering. The drawback of including higher order scatter, for example second order Compton scatter, however, would be the increased time needed for the ray-tracing calculation.

The SPRs were measured for air gaps equal to 50 cm and larger for the following five anthropomorphic phantom and beam energy configurations: the neck phantom, for

![Graphs showing comparison of analytical and experimental SPRs for different beam energies and phantom configurations.](image)

**Figure 6.6.** Comparison of the analytical (——) and experimental (○) SPRs for the homogeneous Solid Water phantoms. An SDD of 185 cm was used. Beam energies shown: (a) 4 MV, (b) 6 MV, (c) 10 MV, and (d) 18 MV.
Table 6.3. Agreement between the scatter to primary dose ratio measured experimentally \( (SPR_{\text{EXP}}) \) and calculated analytically \( (SPR_{\text{A}}) \) for the homogeneous water phantoms. Maximum field size was 20×20 cm\(^2\) for all cases. SDDs for this data were 185 and 200 cm.

<table>
<thead>
<tr>
<th>Beam energy (MV)</th>
<th>Max. energy thickness (cm)</th>
<th>N</th>
<th>( \Delta SPR \pm \Delta SPR_{\sigma} )</th>
<th>( \Delta SPR_{\text{max}} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>20</td>
<td>24</td>
<td>-0.004±0.006</td>
<td>0.020</td>
</tr>
<tr>
<td>6</td>
<td>30</td>
<td>36</td>
<td>-0.004±0.006</td>
<td>0.019</td>
</tr>
<tr>
<td>10</td>
<td>30</td>
<td>36</td>
<td>0.000±0.003</td>
<td>0.009</td>
</tr>
<tr>
<td>18</td>
<td>40</td>
<td>48</td>
<td>0.000±0.004</td>
<td>0.011</td>
</tr>
</tbody>
</table>

4 and 6 MV beams; the thorax phantom, 6 MV beam; and the pelvis phantom, 10 and 18 MV beams. Figure 6.7 shows the experimental and analytical SPRs for three of the anthropomorphic phantoms using an SDD of 185 cm. The agreement between the analytical and experimental SPRs was calculated for each source to detector distance using equations (6.5) and (6.6).

Figure 6.8 displays the mean difference between the analytical and experimental SPRs, equation (6.5), for the anthropomorphic cases as a function of the source to detector distance. The error bars are equal to the standard deviation of the mean difference, given by equation (6.6).

In figure 6.8 the analytical method is seen to be a good approach for the neck phantom for all SDDs and both 4 and 6 MV beams. These results confirm the MC data that showed that the SPR from multiply scattered particles was small for thin phantoms and small field sizes. For the thorax phantom, the agreement improves with increasing SDD - this also agrees with the MC results that show that the SPR from multiply scattered particles decreases with increasing SDD.

For the pelvis case, the agreement between \( SPR_{\text{EXP}} \) and \( SPR_{\text{A}} \) improved with increas-
Figure 6.7. Graphs of the measured (○) and analytical (•) SPRs for the anthropomorphic phantoms versus beam area at the isocentre. Results are shown for the (a) neck phantom, (b) thorax case, and (c) pelvis phantom. The source to detector distance was 185 cm.

ing beam energy; this trend with energy was opposite to that found for the Monte Carlo results from the pelvis phantom (see figure 6.8 and table 5.1). From the Monte Carlo results in table 5.1 it was found that at 18 MV, \( \text{SPR}^{MC} \) and \( \text{SPR}_p^A \) differed by \( \approx 0.02-0.03 \) for the pelvis case because of the multiply scattered particles. However, the mean difference between \( \text{SPR}^{EXP} \) and \( \text{SPR}_p^A \) was \( \approx 0.01 \) or less for the pelvis case when using the
Figure 6.8. Mean difference between the analytically calculated and experimentally measured SPRs on the central beam axis versus the source to detector distance. Each curve corresponds to a specific site and energy (labeled at right). Error bars represent the standard deviation of the mean difference (field areas ranged from 4×4 cm² to the maximum field size for that site. The maximum field sizes were: 14×14 cm², 20×20 cm², and 16×16 cm² for the neck, thorax, and pelvis cases, respectively). Only one error bar per curve is shown for clarity. The average standard deviation for each site was: neck 4 MV, 0.001; neck 6 MV, 0.003; thorax 0.009; pelvis 10 MV, 0.003; and pelvis 18 MV, 0.003.

18 MV beam (see figure 6.8). SPR_{MC} was also greater than SPR_{EXP} by ≈0.02-0.03 for the 30 cm thick homogeneous phantom irradiated with an 18 MV beam (see figure 6.9). Off-axis softening of the primary photon beam would explain these results. Off-axis softening is due to the shape of the flattening filter in the treatment head of the linear accelerator. The flattening filter, which is composed of tungsten and copper, is thickest at the central beam axis and tapers off with increasing off-axis distance. Low-energy photons are absorbed in the flattening filter. Thus, the primary spectrum has a greater component of low-energy photons below the thinner part of the filter, which explains the softer primary X-ray spectrum off-axis. Since the filter is tapered, the primary photon fluence changes with off-axis distance. In turn, the softening of the primary spectrum affects the mean energy of the scattered photons that reach the central axis at the portal imaging plane.
Figure 6.9. Graph illustrating the effect of off-axis softening on the SPRs for the 18 MV beam. Comparison of the experimental (•) and Monte Carlo (△) scatter to primary dose ratios for a 30 cm thick water phantom, an 18 MV photon beam, and an SDD of 189.2 cm.

The goodness-of-fit of the normalized total doses \( T_N \) to a quadratic was assessed by examining the \( \chi^2 \) values of the fit:

\[
\chi^2 = \sum_{i=1}^{K} \left( \frac{\text{observed } T_N(FA_i) - \text{fitted } T_N(FA_i)}{\sigma_{T_N}(FA_i)} \right)^2
\]  

(6.7)

where \( FA_i \) is the \( i \)th field area and \( K \) is the number of field areas for the fit. In this case, the number of degrees of freedom was \( \nu=3 \) (\( \nu=K-M \) where \( K=6 \) is the number of data points per curve and \( M=3 \) was the number of parameters in the fit). The \( \chi^2 \) value for 3 degrees of freedom and a level of significance of \( \alpha=0.1 \) is \( \chi^2_{0.9}(3)=6.25 \). Since the observed values for \( \chi^2 \) were less than \( \chi^2_{0.9}(3) \), the quadratic model seems quite reasonable for these data.

Figure 6.8 shows that the experimental SPRs on the central axis were modeled by the analytical SPR method to within \( \approx 0.03 \) for phantoms representing the neck, thorax, and pelvis. The accuracy of the current method is comparable with that of similar techniques. The SPR model of Swindell and Evans [117] was shown to have a mean experimental difference of 0.005 or less for an SDD of 200 cm, water phantom thicknesses up to 30 cm, beam areas up to 400 cm\(^2\), and beam energies of 6 and 10 MV (Partridge and Evans [95]). The analytical SPR model of Spies et al. [114] agreed within 0.02 to 0.03 with Monte
Carlo results for off-axis SPRs using air gaps from 6.3 to 18.3 cm, a 6 MV radiosurgical field, and copper phantoms up to 3.5 cm thick.

6.6 SUMMARY

In this chapter, measurements of the scatter to primary dose ratio on the central axis of radiotherapy beams at the position of the portal imager were used to validate the analytical SPR method. Measurements were carried out for a wide range of clinically relevant treatment configurations involving homogeneous Solid Water phantoms as well as phantoms representing the neck, thorax, and pelvis. Tissue substitutes that mimic the attenuation and scattering from bone, muscle, and lung were discussed and selected. Experimental uncertainties were included for the measured SPRs.

It was found that the analytical SPR method predicted the experimental SPRs to within \( \approx 0.03 \). The agreement between the experimentally measured and analytically calculated SPRs was found to be comparable to similar approaches. The mean differences between the analytically calculated SPRs and the Monte Carlo calculated SPRs were also less than \( \approx 0.03 \), even though the simulation made several approximations for the clinical linear accelerator photon source models.

The analytical SPR method can be applied in several ways. The application of the analytical SPR method to calculate the total portal imager dose is presented in the next chapter. The total dose at the portal imager is the sum of the dose from primary and the dose from scatter. Assuming an accurate method is used for calculating the primary component, \( P \), the total \( T(x, y) \) at a point \( (x, y) \) in the image is the sum of \( P(x, y) \) and the scatter dose at the central beam axis, \( S = P(\text{central beam axis}) \times \text{SPR} \).
In vivo dosimetry is the measurement of the dose delivered to the patient during a treatment session. Detectors can be placed on the patient’s skin at the beam entrance and exit surfaces to measure the dose at the central beam axis. As well, measurements can be carried out within body cavities. Doses are then extrapolated in regions where direct measurement is impossible, such as in the tumour or in critical organs. By comparing the measured dose with the calculated or expected dose, errors can be found and corrected.

Errors in dose delivery can occur from systematic and random causes. In vivo dosimetry is most useful for identifying systematic errors [38]. Leunens et al. [58] showed that it was possible to detect large systematic errors with one in vivo measurement, where a large error was defined as greater than 5%.

Noel et al. [86] found that 16.7% of head and neck patients received doses that deviated 5-10% from the planned doses. In the same study, 1.3% of breast cancer patients had dose inaccuracies larger than 10%, and 6.5% had inaccuracies between 5-10%. The dose inaccuracies determined from in vivo dosimetry can result in important changes within radiotherapy departments. In vivo dosimetry has uncovered systematic errors in treatment planning algorithms that were subsequently corrected (for example, [44, 56, 58]). As well, data showing a reduction in errors when two, rather than one, radiotherapist work at a unit has supported the practice of having two radiotherapists cross-check each other’s work to decrease mistakes [60].
Differences between the measured and planned doses can occur due to incorrect measurement of patient data, weight loss or gain during treatment, errors in patient setup, organ motion [2], problems with beam production, and errors in data transfer. Even with record and verify systems,¹ Noel et al. [86] advocate the mandatory use of in vivo dosimetry for dose delivery verification.

Entrance doses are measured using sufficient material over the detector to establish electronic equilibrium at the measurement layer (see figure 7.1 for the definition of the entrance plane). P-type semiconductor diodes are a common choice for in vivo dosimetry since they allow immediate readout, exhibit good reproducibility, and can be calibrated against an ionization chamber to measure dose [64, 85, 100, 101]. Entrance doses can verify the patient setup and accelerator performance, but alone, are insufficient to estimate uncertainty in the dose delivery. Exit doses correspond to a depth equal to $d_{\text{max}}$ upstream of the exit surface of the patient (see figure 7.1), and give additional dose accuracy information related to patient data such as tissue thickness and tissue heterogeneities. The photon beam transmission through the patient, which is defined as the ratio of the exit and entrance doses corrected for the inverse square law, is useful for measuring inaccuracies in the patient contour and heterogeneities used in the dose calculation.

The midline dose can be estimated by assuming a linear [86] or exponential decrease [59] between the exit dose and entrance dose. A common approach for calculating the midline dose from entrance (or entrance and exit) doses is the method of Rizzotti [34, 44, 102] or slight variants of this method [32, 45, 57]. In this technique, depth dose data are used to convert the measured dose(s) to the dose at the patient midline. Figure 7.2 shows an example of an in vivo measurement with a diode. When the central axis passes through

¹Record and verify systems are software programs that check that the parameters entered by the radiotherapists for controlling the linear accelerator are within tolerance of pre-recorded data.
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Isocentre

\[ d_{\text{max}} \]

Entrance plane

Midplane

Exit plane

Imaging plane

Figure 7.1. Location of the entrance, mid, exit, and imaging planes. The midplane and the isocentre coincided for measurements described in this chapter. The entrance plane was at the depth of maximum dose \((d_{\text{max}})\) within the phantom, while the exit plane was at a depth of \((t - d_{\text{max}})\) where \(t\) was the phantom thickness. For this chapter, measurements at the imaging plane were carried out at \(d_{\text{max}}\) within the imager. Doses were measured with both the portal imager and an ionization chamber at the imaging plane. Ionization chamber readings were also recorded at the midplane and exit planes. Doses were normalized to the dose at the isocentre.

heterogeneities, an equivalent pathlength in water is used rather than the physical depth between, for example, the entrance and midline points. In cases where the entrance and exit doses are used, the method in Rizzotti [34, 44, 102] is limited to situations with symmetric tissue heterogeneities with respect to the patient midline.

Since the patient anatomy and dose information are automatically co-registered in a portal image, these imagers are superior to conventional in vivo dosimeters such as diodes [33]. Previous studies that applied the convolution/superposition (CS) algorithm for in vivo dosimetry with portal images are computationally intensive [42, 75]. It would be of interest to pursue faster methods of in vivo dosimetry using the CS algorithm since an advantage of this approach is that the same convolution kernels can be used for computing the dose from primary photons, and so new dose kernels are unnecessary. This is the objective of this chapter. The method presented here uses the CS algorithm [61, 62, 106] for dose calculation within the phantom and for the imager dose from primary photons.
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Figure 7.2. The diagram illustrates measurement of the in vivo dose to the breast with diodes. Diodes are placed at the central axis to measure the entrance and exit dose. The diode is covered by a hemispherical build-up cap with a water-equivalent thickness equal to the depth of maximum dose for this photon energy. (Figure adapted from a similar diagram in [45]).

The imager dose from scatter is approximated by a uniform distribution across the imager and the scatter to primary dose ratio (SPR) on the central axis is estimated analyticalal from Compton kinematics [90]. The phantom in vivo doses were calculated by back-projecting the measured portal dose using pre-computed corrections calculated using the CS algorithm and the analytical SPR method. Calculated portal dose images were compared quantitatively to measured data obtained with (i) a liquid matrix portal imager calibrated to record dose, and (ii) a Farmer-type ionization chamber (IC). Homogeneous and heterogeneous phantoms were investigated. An illustrative example of the data is included as well as a discussion of the approximations for the technique.
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7.1 MATERIALS AND METHODS

7.1.1 AN ILLUSTRATIVE EXAMPLE

An example of the experimental data used for this study is given in figure 7.3. The image in figure 7.3(b) was measured with the liquid matrix portal imager. Conversion of the image pixel values to dose is described in section 7.1.5. The phantom configuration for this example was a neck case irradiated with a lateral field as shown in figure 7.3(a). The total dose profile at the imager along the posterior-anterior direction of the phantom is graphed in figure 7.3(c). The dose from scatter was calculated by multiplying the total dose on the central axis by \( \frac{\text{SPR}}{(\text{SPR}+1)} \), where the SPR is the scatter to primary dose ratio on the central axis, as discussed in section 7.1.2. The air gap between the phantom and the imager was equal to 75 cm. For large air gaps (defined as greater than 50 cm), the dose from scatter is uniform \([12, 117]\). In figure 7.3(c) the doses from scatter and primary are graphed. In this chapter a method is described to calculate the total imager dose (at any point on the imager) normalized to the isocentre dose, which is then compared to measured data.

Throughout this chapter, the isocentre was chosen as the normalization point for the dose profiles. Consequently, the current work examines both the relative amplitude and the intensity of the calculated imager dose, \( D_T(r_d) \). The profiles at the imaging plane could have instead been normalized to the dose on the central axis at the imager. If the normalization point had been located at the imaging plane, as was the case in \([71]\) and \([76]\), then only the relative amplitudes of the calculated profiles would have been investigated.
Figure 7.3. Illustration of the imager dose from primary photons and scatter radiation. (a) The neck phantom irradiation configuration for this example. The central beam axis (---) as well as the boundaries of the lateral field (—) are indicated for the left lateral field. (b) The portal image of the neck phantom for a 6 MV 8×8 cm² field. An air gap of 75 cm was used. (c) Dose profiles for the total dose, as well as the dose from primary and scatter. The direction of the profile is indicated by the horizontal line in part (b).
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7.1.2 Dose Calculation Methods

The code chosen for calculation of the dose within the phantom is a set of state-of-the-art subroutines based on the convolution/superposition algorithm [61, 62] (see also [67, 68, 106]). These routines are suitable for computing the dose within heterogeneous phantoms when the atomic number of the heterogeneity is close to the effective atomic number for water [49]. This particular set of routines was selected for our work over another set\(^2\) since the former includes a more advanced model of the photon energy spectrum that is based on Monte Carlo simulation of the linear accelerator treatment head [62].

Dose computation using convolution/superposition has four basic steps. For simplicity, the dose deposited in one voxel (at the dose deposition site, \(\vec{r}\)) by photons interacting in another voxel (at the interaction site, \(\vec{r}'\)) is considered (figure 2.1 illustrates vectors \(\vec{r}\) and \(\vec{r}'\)). First, the photon fluence spectrum is determined at the interaction site (\(\vec{r}'\)), through knowledge of the fluence at the isocentre without the phantom present, \(\Phi(\vec{r}' = 0, 0, 100)\), and the transmission and divergence of the primary photons between the photon source and \(\vec{r}'\). Second, the total energy released per unit mass [TERMA, \(T(\vec{r}')\)], by photons interacting at \(\vec{r}'\) is determined from the probability of photon interaction (that is, the attenuation coefficient) and the energy of the incident primary photon. Third, the product of the average electron density between the interaction and dose deposition sites, \(\bar{\rho}_e\), and the distance between the interaction and dose deposition voxels, \((\vec{r} - \vec{r}')\) is computed. Finally, the dose deposited at \(\vec{r}\) is determined from a look-up table for the dose to water (or dose kernel, \(A\)) at the same value of \(\bar{\rho}_e|\vec{r} - \vec{r}'|\) and angle \((\vec{r} - \vec{r}')\) as calculated between the voxels at \(\vec{r}\) and \(\vec{r}'\). The total dose at \(\vec{r}\) is calculated by repeating this process for each voxel within the irradiated volume of the phantom.

\(^2\)Available from http://www-madrad.radiology.wisc.edu/penbeam/index.html
Mathematically, the total dose $D(r) \text{[MeV g}^{-1}]$ can be expressed as

$$D_{CS}(r) = \sum_{x'=x_{s}}^{x_{m}} \sum_{y'=y_{b}}^{y_{f}} \sum_{z'=z_{s}}^{z_{m}} T(r') A[\vec{r}_{e} \cdot |\vec{r} - \vec{r}'|, (\vec{r} - \vec{r}')]$$

(7.1)

where the TERMA $T \text{[MeV g}^{-1}]$ is given by

$$T(\vec{r}) = \sum_{E_{i}=E_{\text{min}}}^{E_{\text{max}}} \frac{\mu(E_{i})}{\rho} E_{i} \Phi(\vec{r}, E_{i})$$

(7.2)

and $A$ is the dose deposition kernel, which is discussed below. In equation (7.1) the three-dimensional summation is carried out for all interaction voxels $\vec{r}' = (x', y', z')$ within the photon field. In equation (7.1), the term $\vec{r} - \vec{r}'$ is the angle between $\vec{r}$ and $\vec{r}'$ while the term $|\vec{r} - \vec{r}'|$ is the distance between the two vectors.

Photon attenuation within the phantom causes the kernel to be variant with depth in the phantom. Calculating kernels at different depths within a water phantom, and then interpolating between the kernels can correct this. This technique is termed kernel hardening and TERMA-weighted [106] kernels at depths $d=0$, 20, and 40 cm were calculated for our work:

$$A[(\vec{r} = 0, 0, d)] = \frac{\sum_{E_{i}=E_{\text{min}}}^{E_{\text{max}}} \frac{\mu(E_{i})}{\rho} E_{i} \Phi(\vec{r}, E_{i}) A[(0, 0, 0), E_{i}]}{\sum_{E_{i}=E_{\text{min}}}^{E_{\text{max}}} \frac{\mu(E_{i})}{\rho} E_{i} \Phi(\vec{r}, E_{i})}$$

(7.3)

As well, the axis of the kernel was rotated to align the axis along vector $\vec{r}'$ (that is, along the ray joining the photon source and the interaction voxel) [61]. The kernels were originally calculated with a vertical axis. Since the photon source is a diverging source (rather than a parallel beam), the kernels are rotated so that they are aligned along $\vec{r}'$.

The remaining quantities in equations (7.1) and (7.2) are defined as:

- $z_{s}, z_{m}$ are the top and bottom surface of the phantom, respectively
- $y_{b}, y_{f}$ are the back and front limits of the radiation field, respectively, and depend on depth
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\[ x_l, x_r \] are the left and right limits of the field; also depth dependent

\[ \rho \] is the physical density [g cm\(^{-3}\)]

\[ \bar{\rho}_e \] is the average electron density between the interaction and dose deposition sites, relative to water

\[ E \] is the energy of the primary photon [MeV]

\[ E_{\text{min}}, E_{\text{max}} \] are the lower and upper energy limits for the photon spectrum [MeV]

\[ \mu(E) \] is the linear attenuation coefficient for photons of energy \( E \), [cm\(^{-1}\)]

\[ \Phi \] is the photon fluence, [photons cm\(^{-2}\)].

In the calculation, the photon fluence was normalized to 1 at the isocentre:

\[
\sum_{E_i=E_{\text{min}}}^{E_{\text{max}}} \Phi[(0, 0, 100), E_i] = 1.
\] (7.4)

**Extracted Dose, \( D_{\text{EX}}(\vec{r}_v) \)**

The extracted dose \( D_{\text{EX}} \) at the point \( \vec{r}_v \) (see figure 4.1) within the phantom was given by:

\[
D_{\text{EX}}(\vec{r}_v) = D_{\text{PI}}(\vec{r}_d) \cdot \frac{D_{\text{CS}}(\vec{r}_v)}{D_{\text{T}}(\vec{r}_d)}
\] (7.5)

where \( D_{\text{PI}} \) was the dose measured by the portal imager at point \( \vec{r}_d \), \( D_{\text{CS}} \) the phantom dose computed with CS using equation (7.1), and \( D_{\text{T}} \) the total calculated dose at the portal imager [given by equation (7.6) below]. Vectors \( \vec{r}_d \) and \( \vec{r}_v \) were both co-linear with the same source to detector ray. Extracting the phantom dose in this manner is analogous to the method of Rizzotti [102], where the phantom dose is determined using (i) a measured dose at an external point outside the phantom, and (ii) the ratio of the dose at the internal point to the dose at the external point, which was predetermined.
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**Imager Dose, $D_T(r_d)$**

The total imager dose $D_T$ [MeV g$^{-1}$] at the point $r_d$ on the imager was calculated in two steps. First, the dose from primary $P_C(r_d)$ was computed with the convolution algorithm using Monte Carlo calculated dose deposition kernels [61, 62, 106]. Second, the scatter to primary dose ratio on the central beam axis $SPR[r_d = (0, 0, SDD)]$ was computed with the analytical method described in [90] (the SDD is the source to detector distance). The total dose was then equal to

$$D_T(r_d) = P_C(r_d) + P_C[r_d = (0, 0, SDD)] SPR[r_d = (0, 0, SDD)].$$

(7.6)

The dose from primary was given by [MeV g$^{-1}$]:

$$P_C(r_d) = \sum_{x''=z_{d,s}}^{z_{d,m}} \sum_{y''=y_b}^{y_f} \sum_{x''=x_1}^{x_f} T[r'' = (x', y', z')] A[\rho_e \cdot |r''|, (r_d - r'')]$$

(7.7)

where $z_{d,s}, z_{d,m}$ are the top and bottom surface of the detector, respectively.

For the cases investigated in this chapter, the mean and maximum difference between measured and analytically calculated SPRs were generally less than 0.01 and 0.03, respectively.

**Phantom Transmission, $T_M(r_d)$ and $T_C(r_d)$**

The phantom transmission as measured at the imager, $T_M(r_d)$, was computed from:

$$T_M(r_d) = \frac{D_{PI,phantom}(r_d)/[1 + SPR(0, 0, SDD)]}{D_{PI, no \phantom{phantom}}(r_d)}$$

(7.8)

where $D_{PI,phantom}$ is the imager dose with the phantom and $D_{PI, no \phantom{phantom}}$ the imager dose without the phantom. The term in the denominator, $D_{PI, no \phantom{phantom}}$, was present to remove the effect of the change in the machine output factor with field area. The numerator was divided by the SPR to remove the scatter dose: $T$ is then the transmission
of the primary beam through the phantom. The corresponding equation for the calculated phantom transmission \( T_C \) was

\[
T_C(r_d) = \frac{P_{C,\text{phantom}}(r_d)}{P_{C,\text{no phantom}}(r_d)}
\]  

where \( P_C \) is the dose from primary calculated with equation (7.7).

### 7.1.3 Accuracy of the Photon Source Model

Liu et al. [62] analyzed the photon fluence at the isocentre as coming from two sources: (i) a primary source, for photons created through bremsstrahlung in the target, and (ii) an extra-focal source for photons that interact within the primary collimator and flattening filter. The primary source was modeled by a point distribution, while a Gaussian distribution modeled the extra-focal source. The effect of the electron target, primary collimator, flattening filter, monitor ionization chamber, and collimator jaws were included in the Monte Carlo simulation. The photon sources were radially symmetric, which is appropriate since the target, primary collimator, and flattening filter have cylindrical symmetry.

The target, primary collimator, and flattening filter for our 2100C/D linear accelerator are the same as that used by Liu et al. [62]. The only difference between our linear accelerator and that modeled by Lui is the energy of the electrons that impinge on the target [119]. The percent depth doses and profiles are directly affected by the energy of the electron beam. It was important to determine the accuracy of these photon source models for our linear accelerators. If the doses we calculated using Liu’s photon source models agreed well with our measurements, then the modeling of our linear accelerator treatment head with Monte Carlo simulation to generate the photon source model for the CS codes would be unnecessary.

The accuracy of the doses calculated from Liu’s models was determined from the
mean ratio $\overline{R}$ and standard deviation of the mean ratio $\overline{R}_\sigma$ between the calculated $D_{cal}$ and measured $D_{exp}$ doses:

$$
\overline{R} = \frac{1}{N} \sum_{i=1}^{N} \left( \frac{D_{cal,i}}{D_{exp,i}} \right), \quad (7.10)
$$

$$
\overline{R}_\sigma = \sqrt{\frac{\sum_{i=1}^{N} \left( \frac{D_{cal,i}}{D_{exp,i}} - \overline{R}_i \right)^2}{N-1}}, \quad (7.11)
$$

$\overline{R}$ and $\overline{R}_\sigma$ were calculated separately for each profile and depth dose; $N$ was the number of data points for the depth dose curve or profile respectively.

Our data was measured with an ionization chamber (IC10, Wellhofer, Schwarzenbruck, Germany) in a 40×40×40 cm$^3$ water tank and compared to calculated data in a phantom of the same size. Percent depth doses were compared for field sizes from 3×3 to 20×20 cm$^2$, while profiles were compared for radiation fields in the range 3×3 and 20×20 cm$^2$. Representative results are graphed in figure 7.4. Depth doses were normalized at 10 cm depth [111]. Profiles were normalized on the central beam axis. For depths up to 25 cm, $\overline{R} \pm R_\sigma = 1.01 \pm 0.01$ or better for the depth doses and $\overline{R} \pm R_\sigma = 0.991 \pm 0.004$ at worst for the profiles. The good agreement between the two data sets is fortunate so that modeling of our 2100C/D clinical linear accelerator with Monte Carlo was unnecessary. The small deviation of $\overline{R} = 1.01$ from 1 for the depth doses most likely indicates that the electron energy in our case is slightly lower than the energy Liu et al. [62] used.

### 7.1.4 TEST PHANTOMS

Figure 7.5 shows the phantoms designed to illustrate the methods described to calculate the total imager dose $D_T(\vec{r}_d)$ and to extract the phantom dose $D_{EX}(\vec{r}_v)$ using the measured portal image $D_{PF}$. Blocks of water equivalent plastic (Solid Water, Gammex RMI, Middleton, WI) were stacked to form the homogeneous phantoms. Figure 7.5(b) depicts the cork slab phantom, constructed from water equivalent blocks and a 12 cm
thick slab of low density cork (physical density relative to water equal to 0.16±0.01; electron density relative to water equal to 0.15±0.01). A slab phantom containing a high density heterogeneity (aluminum, physical density relative to water 2.70 and electron density relative to water 2.35) was also tested [see figure 7.5(c)]. All phantoms were imaged with the centre of the phantom at the isocentre.

7.1.5 IMAGER DOSE CALIBRATION

Conversion of the gray-scale pixel values to dose was achieved using the methods described in [13, 30, 31, 39, 123, 132, 134] that were previously discussed in chapter 3. Briefly, a calibration curve to convert the pixel gray-scale values to dose at the central axis was measured for each field size and beam energy. This calibration curve is nonlinear [square root plus linear term, see equation (3.6)] for the liquid matrix portal imager and an example is given in figure 7.6. The dose at the imaging plane was varied by attenuating the photon beam with blocks of Solid Water in 1 cm increments. Average pixel values were calculated over a 5×5 pixel region of interest. The IC (ionization chamber) measurements for the pixel value to dose calibration curves were measured in a Solid Water phantom of the same water equivalent dimensions as the imager. To reduce the statistical noise in the IC data, three measurements were taken for each point. To reduce the statistical noise in the imager profiles, dose profiles were averaged over 5 pixels in the direction perpendicular to the profile. Imager dose profiles were corrected for the flat field calibration applied by the commercial image display software [31, 132, 134] (see figure 7.7). All measurements were carried out with the imager beneath the photon source using a source to detector distance of 185 cm.
Figure 7.4. Comparison of the convolution calculation (●) and measured data (+) for a 10×10 cm² radiation beam. From the top: (a) percentage depth dose, 6 MV beam; (b) depth dose, 10 MV and (c) profile at a depth of 10.5 cm for a 10 MV beam.
Figure 7.5. Schematic diagram of the phantoms used for the experimental measurements. Sizes are in cm. (a) A homogeneous water equivalent phantom: the range of phantom thicknesses $t$ and beam areas $A$ are indicated. (b) A cork slab phantom, and (c) the aluminum slab phantom.

Figure 7.6. An example of a calibration curve for the liquid matrix electronic portal imager. The error bars are too small to be seen for both the pixel value and dose rate. Raw data ($\bullet$), fitted curve ($-$). The equation for the fitted curve was $W(D) = a\sqrt{D} + bD$ where $W$ was the pixel value, $D$ the dose rate measured with the ionization chamber, and $(a,b)$ the calibration constants. For this curve $a=100.9\pm0.3\text{ min}^{1/2}\text{ cGy}^{-1/2}$ and $b=0.61\pm0.02\text{ min cGy}^{-1}$. 
Chapter 7. Application - In vivo Dosimetry

Figure 7.7. Correction for the flat field calibration applied by the commercial image display software. (a) Comparison of the dose profile measured with the imager (—) to ionization chamber measurements (+) for a 10×10 cm², 6 MV beam. The pixel value to dose conversion curve (see for example figure 7.6) was applied to calculate the dose profile. Off-axis doses measured by the imager (—) underestimate the ionization chamber measurements (+). This problem occurs because the commercial software assumes that the flat field is uniform across the imager. (b) Off-axis correction to rectify the problem caused by the commercial software. Data shown for 10 MV (—) and 6 MV (−−−) photon beams. This correction is equal to the ratio of the ionization chamber measurement to the uncorrected imager dose profile for a large field that completely irradiates the field. (c) Corrected imager dose profile (—) compared to the ionization chamber measurements (+). By multiplying the imager dose profile in part (a) by the off-axis correction in part (b), the problem caused by the commercial software is rectified.
7.1.6 Ionization Chamber Measurements

Ionization chamber measurements were performed to determine the accuracy of: (i) the calibration of the imager for dosimetry, (see section 7.1.5), (ii) the total imager dose $D_T(\bar{r}_d)$ calculated using equation (7.6), (iii) the extracted dose $D_{EX}(\bar{r}_v)$ computed with equation (7.5), and (iv) the phantom transmission $[T_C(\bar{r}_d)$, see equation (7.9)]. Measurements with a Farmer-type IC (PTW, New York) were taken at 1 cm increments along dose profiles at the depth of the midplane and exit plane within the phantom and at the imaging plane (see figure 7.1). The midplane of the phantom was located at the isocentre. The exit plane was located at a depth of $(t - d_{max})$ where $t$ is the phantom thickness and $d_{max}$ the depth of maximum dose. To reduce the statistical noise in the IC data, each point was the average of two measurements.

7.1.7 Analysis

Measurements taken with the Farmer IC were compared quantitatively to the following data sets: (i) portal dose profiles measured with the liquid matrix portal imager, $D_{PI}(\bar{r}_d)$; (ii) the total imager dose $D_T(\bar{r}_d)$ calculated using equation (7.6), and (iii) the extracted exit and midplane doses $D_{EX}(\bar{r}_v)$ computed with equation (7.5). Each comparison was carried out for the three test phantoms (see section 7.1.4). Measured imager dose profiles $D_{PI}(\bar{r}_d)$ were normalized to the isocentre dose measured by the ionization chamber. Extracted dose profiles $D_{EX}(\bar{r}_v)$ were also normalized to the isocentre dose measured by the ionization chamber. The phantom doses calculated with CS, $D_{CS}(\bar{r}_v)$, and the computed total imager dose $D_T(\bar{r}_d)$ were normalized to the isocentre dose computed by CS. Doses measured by the ionization chamber were normalized to the IC isocentre dose.

The accuracy was determined by calculating the mean and standard deviation of the
ratio between the comparison data set and the IC data (both data sets were normalized). For example, the calculated imager dose profile \( D_T(\vec{r}_d) \) could be the comparison data set. The ratio of the normalized dose for the comparison data set and the normalized IC data for each point along the profile was equal to:

\[
R = \frac{\text{Dose(comparison data)}}{\text{Dose(IC)}}.
\]  

(7.12)

The mean ratio and standard deviation of the mean ratio were given by:

\[
\bar{R} = \frac{\sum_{i=1}^{N} R_i}{N}
\]  

(7.13)

and

\[
\bar{R}_\sigma = \sqrt{\frac{\sum_{i=1}^{N} (R_i - \bar{R})^2}{N - 1}}
\]

respectively where \( N \) was equal to the number of IC measurements compared. All the data collected for one particular phantom and beam energy was analyzed as a single group. For example, all data measured for the homogeneous water equivalent phantoms irradiated with the 6 MV beam were analyzed together. In this case, \( N \) was equal to the sum of the number of IC data points measured for the 3×3 cm\(^2\), 10×10 cm\(^2\), and 17×17 cm\(^2\) fields. Relative errors, expressed as percents, were equal to:

\[
\text{relative error} = \frac{\bar{R}_\sigma}{\bar{R}} \times 100\%.
\]  

(7.15)

### 7.2 Results

In this chapter, a method was presented to extract the dose within the phantom \( D_{EX}(\vec{r}_v) \) using a portal imager as a dosimeter. An example illustrating the total imager dose as well as the dose from primary and scatter at the imager was included (see figure 7.3). The method to extract the dose \( D_{EX}(\vec{r}_v) \) uses a pair of measured \( D_{PI}(\vec{r}_d) \) and calculated \( D_T(\vec{r}_d) \) portal dose images. To test this idea, a liquid matrix portal imager
was calibrated for dosimetry and dose profiles $D_{PI}(\tilde{r}_d)$ were measured with this imager at a source to detector distance of 185 cm. Data were collected for three types of phantoms (homogeneous Solid Water, cork and aluminum slab phantoms). The agreement was determined between calculated portal doses $D_T(\tilde{r}_d)$ and portal doses measured with (i) the imager $D_{PI}(\tilde{r}_d)$ and (ii) an IC. Data were normalized to the isocentre dose in all cases. Further comparisons were made between the extracted doses $D_{EX}(\tilde{r}_w)$ for the three phantoms and IC measurements.

Typical results for this study are given in figures 7.8, 7.9, 7.10, and 7.11. These figures show the computed dose at the imaging plane $D_T(\tilde{r}_d)$ calculated using equation (7.6), the extracted midplane dose $D_{EX}(\tilde{r}_w)$ computed from equation (7.5), and the computed phantom transmission $T_C(\tilde{r}_d)$ given by equation (7.9). Table 7.1 presents the results from the quantitative analysis to determine the accuracy of the calculated and extracted doses (see subsection 7.1.7).

### Table 7.1. Number of measurements ($N$), mean ratio ($\bar{R}$), and standard deviation of the mean ratio ($\bar{R}_\sigma$) between the Farmer ion chamber measurements and: (i) the total dose calculated at the imaging plane $D_T(\tilde{r}_d)$, (ii) the portal imager measurements $D_{PI}(\tilde{r}_d)$, (iii) and the extracted dose $D_{EX}(\tilde{r}_w)$ at the exit and midplane of the phantom. Measurements were made using water equivalent plastic blocks (Solid Water, Gammex RMI, Middleton, WI) (SW) and two slab phantoms.

<table>
<thead>
<tr>
<th>Beam Energy (MV)</th>
<th>Phantom</th>
<th>N</th>
<th>$D_T$ $\bar{R} \pm \bar{R}_\sigma$</th>
<th>$D_{PI}$ $\bar{R} \pm \bar{R}_\sigma$</th>
<th>$D_{EX}$ exit $\bar{R} \pm \bar{R}_\sigma$</th>
<th>$D_{EX}$ midplane $\bar{R} \pm \bar{R}_\sigma$</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>SW</td>
<td>141</td>
<td>1.02±0.01</td>
<td>1.00±0.01</td>
<td>0.99±0.02</td>
<td>0.98±0.02</td>
</tr>
<tr>
<td>10</td>
<td>SW</td>
<td>141</td>
<td>1.00±0.02</td>
<td>1.00±0.02</td>
<td>1.00±0.03</td>
<td>1.00±0.03</td>
</tr>
<tr>
<td>6</td>
<td>Cork Slab</td>
<td>51</td>
<td>0.99±0.01</td>
<td>1.00±0.01</td>
<td>1.02±0.01</td>
<td>1.01±0.01</td>
</tr>
<tr>
<td>10</td>
<td>Al Slab</td>
<td>51</td>
<td>0.99±0.01</td>
<td>1.01±0.01</td>
<td>1.03±0.01</td>
<td>1.02±0.01</td>
</tr>
</tbody>
</table>
Figure 7.8. Comparison of the extracted and measured doses for a 25 cm thick solid water phantom [see figure 7.5(a)] irradiated with a 6 MV, 10×10 cm² photon beam. Description of symbols for all graphs: ionization chamber measurements (+); extracted dose (□) using portal imager measurements, analytical SPR approximation, and convolution/superposition calculation; portal imager measurements (—); and convolution/superposition computations (●). Doses were normalized to the isocentre dose, and the location of the isocentre is indicated in part (a). From the top left: comparison of normalized doses at the (a) midplane, (b) imaging plane, and (c) exit plane. For (d), the measured and calculated transmissions are graphed at the imaging plane.
Figure 7.9. Comparison of the extracted and measured doses for a 25 cm thick solid water phantom [see figure 7.5(a)] irradiated with a 10 MV, 10×10 cm² photon beam. Description of symbols for all graphs: ionization chamber measurements (+); extracted dose (□) using portal imager measurements, analytical SPR approximation, and convolution/superposition calculation; portal imager measurements (—); and convolution/superposition computations (●). Doses were normalized to the isocentre dose at midplane. From the top left: comparison of normalized doses at the (a) midplane, (b) imaging plane, and (c) exit plane. For (d), the measured and calculated transmissions are graphed at the imaging plane.
Figure 7.10. Comparison of the extracted and measured doses for the cork slab phantom [see figure 7.5(b)] irradiated with a 6 MV, 10×10 cm² photon beam. Description of symbols for all graphs: ionization chamber measurements (+); extracted dose (□) using portal imager measurements, analytical SPR approximation, and convolution/superposition calculation; portal imager measurements (—); and convolution/superposition computations (●). Doses were normalized to the isocentre dose at midplane. From the top left: comparison of normalized doses at the (a) midplane, (b) imaging plane, and (c) exit plane.
Figure 7.11. Comparison of the extracted and measured doses for the aluminum slab phantom [see figure 7.5(c)] irradiated with a 10 MV, 10×10 cm² photon beam. Description of symbols for all graphs: ionization chamber measurements (+); extracted dose (□) using portal imager measurements, analytical SPR approximation, and convolution/superposition calculation; portal imager measurements (—); and convolution/superposition computations (○). Doses were normalized to the isocentre dose. From the top left: comparison of normalized doses at the (a) midplane, (b) imaging plane, and (c) exit plane.
7.3 DISCUSSION

Throughout this chapter, quantitative comparison was made to IC measurements to investigate the accuracy of each part of the study. This analysis was carried out at 1 cm increments along a single beam profile. Doses were normalized first to the isocentre dose to have a common reference point before the accuracy was calculated as described in section 7.1.7.

The total imager dose $D_T(r_d)$ given by equation (7.6) agreed with the IC measurements to within $\approx 3\%$ (see fourth column of Table 7.1). Several reasons may be given for differences between the calculated imager dose $D_T(r_d)$ and the IC measurements. The energy of the electron beam before it hits the target is likely different for our linear accelerator and for that used in the Monte Carlo simulation by Liu et al. [62] to derive the photon source models. Discrepancies also arise because of the limits of the kernel tilting and beam hardening algorithms within the convolution code. To assess the impact of the electron energy differences and the limits of the beam hardening and kernel tilting algorithms, CS calculated $D_{CS}(r_v)$ and measured doses were compared for a water phantom: these doses were found to agree within $\approx 1-2\%$ (see section 7.1.2). Differences between the calculated and measured imager doses [$D_T(r_d)$ and $D_{PI}(r_d)$ respectively] were also caused by each approximation within the analytical SPR method. For the cases studied here, the mean and maximum differences between measured and analytically calculated SPRs on the central axis were better than 0.01 and 0.02, respectively [90]. The systematic errors contributed by the differing electron energies, the limits of the kernel tilting and beam hardening algorithms, and the analytical SPR method fully explain the small disagreement between the calculated imager dose and that measured with the IC.

Total portal dose images measured with the liquid matrix portal imager $D_{PI}(r_d)$ agreed with IC measurements to within $\approx 1-2\%$. This finding was consistent with previous
Chapter 7. Application - In vivo Dosimetry

The method for calculating the extracted dose $D_{EX}(r_v)$ was presented in section 7.1.2. These doses were computed from the measured portal image, the CS algorithm, and the SPR method. Extracted doses $D_{EX}(r_v)$ agreed with those measured using the IC to within $\approx 3\%$. The reasons for systematic errors in the extracted doses include the different electron beam energies for the photon source model and the linear accelerator, each approximation in the SPR and convolution/superposition algorithms, and the imager dose calibration. Since the liquid matrix portal imager only measured dose rate images, rather than integrated dose for the entire time the beam was on (100 monitor units), the current work was limited to investigating the extracted dose rates within the phantom.

The advantage of the current method to calculate portal dose images compared to semi-empirical approaches is that the dose images, normalized to the isocentre, were calculated without a database of measured scatter dose data. This reduces the time to implement the algorithm and may be potentially beneficial to other institutions who wish to calculate portal dose images. The relative error of the current approach for in vivo dosimetry ($\approx 3\%$) is comparable to the relative error for in vivo measurements with thermoluminescent dosimeters ($\approx 3\%$).

The accuracy of the extracted doses, $D_{EX}(r_v)$, as compared to IC measurements was within $\approx 3\%$. This accuracy is comparable to similar techniques. McNutt et al. [73, 74, 75, 76] and Hansen et al. [42] also applied the CS algorithm for in vivo dosimetry. McNutt et al. [73, 75] compared planned doses to in vivo doses and found a mean difference of $\approx 1-2\%$ with a standard deviation of $\approx 1\%$ (dose differences were expressed as a percentage of the isocentre dose). Hansen et al. [42] compared measurements to in vivo doses and found a maximum difference of 3\% (differences were also expressed as a percentage of the isocentre dose).

The advantage of the current method for computing the extracted doses $D_{EX}(r_v)$,
compared to previous approaches for in vivo dosimetry that apply the CS algorithm, is that the current technique is faster. An approximate estimate of the order $O$ of our algorithm, and several previous methods, is presented here to illustrate this point. The number of computations for the current in vivo method was proportional to

$$O \propto N_F N_D$$  \hspace{1cm} (7.16)

where $N_F$ is the number of fields for the treatment and $N_D$ the number of points within the phantom at which the in vivo dose is to be calculated. Methods that convolve the primary photon energy fluence with the dose deposition kernels (for example, Hansen et al. [42] or McNutt et al. [75]) are (very roughly) proportional to

$$O \propto N_F N_D N_V$$  \hspace{1cm} (7.17)

where $N_V$ is the number of voxels within the irradiated part of the phantom. The additional factor $N_V$ in equation (7.17) compared to equation (7.16) occurs because of the convolution. The value of $N_V$ for a typical treatment (20 cm thick patient, 10×10 cm$^2$ field area, 0.5×0.5×0.5 cm$^3$ voxel volume) would be $N_V=16000$.

### 7.4 Summary

This chapter described a method to calculate portal dose images without the use of a database of measured scatter dose data. With the current increases in treatment complexity, such as intensity modulated radiation therapy, the ability to independently verify the dose delivery across the whole radiation field would be advantageous. The current technique for in vivo dosimetry uses generic Monte Carlo calculated photon spectra and dose deposition kernels, as well as an analytical method to estimate the central axis scatter to primary dose ratio. Portal dose images were calculated and normalized to the
phantom isocentre dose computed with the convolution/superposition algorithm. Examples were presented of the use of this method to extract the phantom dose and to calculate the phantom transmission. The method was tested by comparing calculated profiles to data measured with an ionization chamber as well as with a calibrated commercial portal imager. Measurements were carried out for two beam energies (6 and 10 MV), three phantoms (homogeneous water equivalent, cork slab, and aluminum slab), and three field sizes for each phantom. Calculated portal dose profiles agreed with ionization chamber measurements to within $\approx 3\%$. The reasons for differences between the calculated profiles and those measured with the ionization chamber were discussed. Extracted doses at the midplane and exit plane agreed with ionization chamber measurements to within $\approx 3\%$. The current method for extracting the in vivo dose is faster than previous methods that also applied the convolution/superposition algorithm. In summary, the method presented here to calculate portal dose images is an attractive alternative for in vivo dosimetry.
CHAPTER 8

CONCLUSIONS

In this thesis, an analytical method was validated for estimating the portal scatter to primary dose ratio on the central axis for heterogeneous phantoms. The first section of this chapter summarizes the accomplishments reported in this thesis. Future areas that could be explored to continue the development of the work here are discussed next. Finally, a summary is given of the major conclusions drawn from this research.

8.1 SUMMARY OF WORK

Chapter 2 presented a literature review of current methods for calculating the portal imager dose from scatter radiation. This review revealed that the uniform scatter dose approximation was a powerful simplification for dosimetry with portal imagers. Portal imagers have been applied for the design of customized breast compensators at the Royal Marsden Hospital (RMH) [36, 95, 117] and in vivo dosimetry at the Netherlands Cancer Institute [11, 12, 30, 31] and at the RMH [36]. At the RMH, the central axis scatter to primary dose ratio at the imaging plane was calculated using a simple model based on the probability of first and second order Compton scatter [117]. At the Netherlands Cancer Institute, the ratio of the total to primary dose at the imaging plane was estimated using a look-up table of this ratio measured for homogeneous water equivalent phantoms [14]. Both the SPR model and the look-up table approach were limited to large air gaps (defined in [12] as greater than 50 cm). The SPR model was also limited to homogeneous or nearly homogeneous scattering objects [43].
Chapter 2 also discussed a problem with semi-empirical scatter estimation techniques, such as the method based on empirical slab derived scatter kernels [97]. Scatter estimation methods that use measured data have the drawback of requiring, in some cases, substantial time for measurement of the data for the scatter dose calculation algorithm.

Chapter 3 presented a review of the literature for calibration of the liquid matrix portal imager for dosimetry. This calibration converts the pixel gray-scale image to a dose image. The dose in this case is the dose as measured by an ionization chamber within a rectangular block of Solid Water. The Solid Water is placed at the same source to detector distance as the portal imager. Buildup material is added to both the portal imager and the ionization chamber for this calibration procedure.

In chapter 4 an analytical method for estimating the central axis SPR at the imaging plane for heterogeneous scattering objects was presented. The analytical expression for the SPR from first-order Compton scatter, which was given the symbol $\text{SPR}_C$, was discussed and several examples of the use of this expression were given. This technique uses generic data for the photon source models. $\text{SPR}_C$ was limited to large air gaps, approximates the portal scatter dose by a uniform distribution, and neglects multiply scattered particles. One of the advantages of this method is that it could be used elsewhere without measuring a database of scatter doses.

A Monte Carlo validation of the analytical method for $\text{SPR}_C$ was described in chapter 5. For this validation, the EGS4 code DOSXYZ was modified to calculate the contribution to the SPR from each scatter mode. The resulting code, SDOSXYZ, computes the SPR from first-order Compton scatter, multiple photon scatter, and patient-generated electrons, as well as the uncertainties for each scatter mode. Comparison of the doses calculated with SDOSXYZ to published data was presented. A total of 576 SPRs were calculated for homogeneous water phantoms: these SPRs were computed for a point photon source and the SPRs are included in appendix B. The total SPRs are new since
they included the contribution from patient-generated electrons. A statistical analysis of the agreement between $\text{SPR}^A$ and the Monte Carlo calculated SPRs for three anthropomorphic phantoms (representing the neck, thorax, and pelvis) found that the accuracy of the analytical SPR method was comparable to similar analytical approaches that are limited to homogeneous phantoms.

Chapter 6 discussed an experimental validation of $\text{SPR}^A$. SPRs were measured using an ionization chamber for Varian linear accelerators with beam energies from 4 to 18 MV. The experimental uncertainty in the SPRs was estimated and the data for homogeneous water-equivalent phantoms was included in appendix C: this data is new. SPRs were also measured for heterogeneous phantoms representing the neck, thorax, and pelvis. In chapter 6, $\text{SPR}^A$ was computed on the central beam axis using generic isotropic point photon source models for these linear accelerators. A quantitative comparison of $\text{SPR}^A$ and measured SPRs for the three anthropomorphic phantoms showed that the accuracy of the analytical SPR method was comparable to similar techniques.

Finally, chapter 7 illustrated how $\text{SPR}^A$ can be applied for calculation of portal dose images and extraction of the dose within the phantom using a measured portal image. The method described in chapter 7 for calculation of the portal dose image used only published data for the photon source model. Thus, an advantage of this approach is that implementation of the method elsewhere could be possible without using a database of measured scatter dose data. Calculated portal dose images were compared to measurements taken with a calibrated commercial portal imager as well as with an ionization chamber. Extracted doses within the phantom were compared to doses measured using an ionization chamber. Measured and extracted doses agreed to within $\approx 3\%$ (one standard deviation). The method described here to extract the phantom dose is faster than previous methods for in vivo dosimetry that also apply the convolution/superposition algorithm.
8.2 Future Research

The experimental validation in this thesis was limited to Varian linear accelerators. The database of experimental SPRs could be expanded to include measured SPRs for:
(i) source to detector distances less than 185 cm for the homogeneous phantoms, and
(ii) linear accelerators manufactured by Elekta (Crawley, UK) and Siemens (Concord, CA). SPRs for different brands of accelerators could be inter-compared. A statistical analysis of the agreement between the analytical SPR method and the experimental SPRs could also be carried out when the air gap is equal to 50 cm or larger (for SPRs other than those reported in appendix C).

The application of the analytical SPR method was restricted to the Varian liquid matrix portal imager. Other imagers can also be calibrated to convert the pixel values to dose. Portal dose images could be measured using an amorphous silicon flat-panel array [4, 5, 6, 29, 82] and then compared to calculated images.

Clinical application of the portal dose image calculation and dose extraction methods could be achieved by modifying commercial software written to apply the convolution/superposition algorithm for dose calculation within the patient. Specifically, the code could be modified to calculate the dose from primary photons [see equation (7.7)] and the analytical SPR method [see equation (4.18)]. Suitable commercial software includes the codes marketed by MDS Nordion (Kanata, Canada; code marketed under the name Helax) and Philips (Amsterdam, Netherlands). An interface could also be written to allow CT data to be used with the current codes. The methods described in this thesis could then be applied for quality assurance of patient treatments in several ways. A pair of calculated and measured portal dose images can be compared. As well, the extracted and intended patient doses can be analyzed statistically.

Newer treatment modalities, called intensity modulated radiation therapy or IMRT,
Chapter 8. Conclusions

control the field area and dose rate dynamically during the treatment. Rapid and accurate evaluation of the two-dimensional dose data from IMRT fields is a current challenge in radiotherapy. Due to the complexity of the resulting dose delivery, it is of interest to be able to calculate IMRT portal dose images and extract the patient dose using a portal image for quality assurance during IMRT. Application of the analytical SPR method for quality assurance of IMRT would likely be the most significant potential application for future research.

8.3 Summary

The goal of this thesis, which was successfully achieved, was to extract the phantom dose using a measured portal image. New Monte Carlo calculated SPRs and new measured SPRs were reported, as well as the uncertainties on these quantities. An analytical method for calculating the SPR on the central axis for heterogeneous cases was quantitatively validated for a wide range of clinically relevant phantoms. The method for calculating the portal dose images presented here relies only on previously calculated photon source models and dose kernels, and not on measured scatter dose databases as in some semi-empirical portal dose calculation algorithms. In the future, this approach could be applied for in vivo dosimetry for verification of the radiation dose to patients. This chapter presented several avenues for future research in this direction.
APPENDIX A

SDOSXYZ

Following are the important modifications of the subroutines within DOSXYZ. The programming language is MORTRAN. Some parts of the original DOSXYZ code are included so that the code makes sense.

Within the AUSGAB subroutine:

"Turn on flags for keeping track of particle interactions"
   iausfl(8)=1; "Bremsstrahlung"
   iausfl(14)=1; iausfl(15)=1; "Annihilation"
   iausfl(19)=1; "Compton"
   iausfl(21)=1; "Photoelectric"
   iausfl(17)=1; "Pair production"

IF ( ( (ir(np) > 1) & (edep~ 0.0) ) & (iarg < 5) )
   [ "Score dose"
     IF ( LATCH(np)=3 ) "score the primary dose"
        [ edoseis(ir(np)-1,is) = edoseis(ir(np)-1,is) + edep*wt(np); ]
     ELSEIF ( LATCH(np)=5 ) "score the dose from first Compton scatter"
        [ sdoseis(ir(np)-1,is) = sdoseis(ir(np)-1,is) + edep*wt(np); ]
     ELSEIF ( LATCH(np)=4 ) "score the dose from multiple photon scatter"
        [ mdoseis(ir(np)-1,is) = mdoseis(ir(np)-1,is) + edep*wt(np); ]
   ]
ELSEIF ( (iarg=18) & (Z(np) < botob)) & ( (E(np)+E(np-1)=etotin+RM) )
   [ "First order Compton has occurred"
     IF (iq(np) = 0)
        [ ipoint=np; ]
     ELSE
        [ ipoint=np-1; ]
     LATCH(ipoint)=5; "Tag the photon"
     IF (iq(np) = 0) "Tag the electron"
[ LATCH(np-1)=3; ]
ELSE
[ LATCH(np)=3; ]
]
ELSEIF ( ((iarg=18) & (Z(np) < botob)) & ( (E(np)+E(np-1)~etotin+RM) ) )
[ "Higher order Compton has occured, tag the photon and electron"
  IF (iq(np) = 0)
  [ ipoint=np; ]
ELSE
  [ ipoint=np-1; ]
IF ( LATCH(ipoint)=5 )
  [ " Second scatter "
    "photon is part of multiple scatter, electron is first scatter"
    LATCH(ipoint)=4;
    IF (iq(np) = 0)
    [ LATCH(np-1)=5; ]
    ELSE
    [ LATCH(np)=5; ]
  ]
ELSE
[ "Third or higher scatter"
  "Both particles are now part of the multiple scatter dose"
  LATCH(ipoint)=4;
  IF (iq(np) = 0)
  [ LATCH(np-1)=4; ]
  ELSE
  [ LATCH(np)=4; ]
]
ELSEIF ( (iarg=16) & ( Z(np)<botob ) )
[ "Pair production has occured, tag the positron and electron"
  IF (LATCH(np-1)=0) [ LATCH(np)=3; LATCH(np-1)=3;]
  IF (LATCH(np-1)=5) [ LATCH(np)=5;]
  IF (LATCH(np-1)=4) [ LATCH(np)=4;]
]
ELSEIF ( (iarg=20) & ( Z(np)<botob ) & (LATCH(np)=0) )
"Photoelectric event, tag electron" [ LATCH(np)=3; ]
ELSEIF ( (iarg=20) & ( Z(np)<botob ) & (LATCH(np)~0) )
"Photoelectric event from scattered photon, tag electron" [ LATCH(np)=4; ]
ELSEIF ( (iarg=7) & ( Z(np)<botob ) ) "Bremsstrahlung"
[
IF (iq(np) = 0)
[ LATCH(np)=4; ]
ELSE
[ LATCH(np-1)=4; ]
]
ELSEIF (( (iarg=13) | (iarg=14) ) & ( Z(np)<botob ) ) "Annihilation"
[ LATCH(np)=4; LATCH(np-1)=4;
]

In subroutine HOWFAR, discard primary photons at the bottom of the scattering object.

[ IDISC=1; RETURN; ]
# Appendix B

## Monte Carlo Calculated SPRs

Table B.1. Portal scatter to primary dose ratios (SPRs) calculated using SDOSXYZ for a 6 MV photon beam. The source to detector distance (SDD) is the distance to the top of the detector: the dose scoring voxel is at a distance of \((\text{SDD}+\text{d}_{\text{max}})\) from the photon source. \(\text{SPR}^{\text{MC}}\) is the total SPR at the imaging plane. \(\text{SPR}^{\text{MC}}_F\) is the SPR from photons that Compton scatter once within the scattering object. \(\text{SPR}^{\text{MC}}_M\) is the SPR from photons that scatter more than once with the scatter object, and includes the dose from bremsstrahlung and annihilation photons. \(\text{SPR}^{\text{MC}}_{\text{CP}}\) is the SPR from patient-generated electrons.

<table>
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<tr>
<th>Phantom thickness (cm)</th>
<th>Field length (cm)</th>
<th>SDD (cm)</th>
<th>(\text{SPR}^{\text{MC}})</th>
<th>(\text{SPR}^{\text{MC}}_F)</th>
<th>(\text{SPR}^{\text{MC}}_M)</th>
<th>(\text{SPR}^{\text{MC}}_{\text{CP}})</th>
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### Appendix B. SPRs from Monte Carlo Simulation

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<th>Phantom thickness (cm)</th>
<th>Field length (cm)</th>
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<th>SPR(^{MC})</th>
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<th>Phantom thickness (cm)</th>
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Table B.2. Scatter to primary dose ratios calculated using SDOSXYZ for an 18 MV photon beam. The source to detector distance (SDD) is the distance to the top of the detector: the dose scoring voxel is at a distance of \((SDD+d_{\text{max}})\) from the photon source.

<table>
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<th>Field length (cm)</th>
<th>SDD (cm)</th>
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<th>(\text{SPR}^{MC}_F)</th>
<th>(\text{SPR}^{MC}_{MS})</th>
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<td>SPR\textsuperscript{MC}</td>
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Appendix B. SPRs from Monte Carlo Simulation

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# Appendix C

## Measured Scatter to Primary Dose Ratios

*Table C.1.* Scatter to primary dose ratios measured for a 4 MV photon beam. The absolute uncertainty for all cases was 0.006. The source to detector distance is the distance to the mid-point of the ionization chamber within the solid water.

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Table C.2. Scatter to primary dose ratios measured for a 6 MV photon beam. The absolute uncertainty for all cases was 0.006. The source to detector distance is the distance to the mid-point of the ionization chamber within the solid water.

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### Table C.3

Scatter to primary dose ratios measured for a 10 MV photon beam. The absolute uncertainty for all cases was 0.006. The source to detector distance is the distance to the mid-point of the ionization chamber within the solid water.

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Table C.4. Scatter to primary dose ratios measured for an 18 MV photon beam. The absolute uncertainty for all cases was 0.006. The source to detector distance is the distance to the mid-point of the ionization chamber within the solid water.

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SPR 26
SSD 65
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