A PROCEDURE TO VERIFY THE ACCURACY OF DELIVERY OF PRESCRIBED RADIATION DOSES IN RADIOTHERAPY

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Abstract

In New Zealand there are currently no regular external audits to verify the full treatment chain in radiotherapy. This thesis reports on a project to devise such an audit procedure suitable to assess the accuracy of the delivery of prescribed radiotherapy doses to patients over the full treatment process. The National Radiation Laboratory (NRL), regulatory authority, will use the method developed to conduct biennial audits of all radiotherapy centres.

A commercial chest phantom with a MOSFET dosimetry system was provided for this project. The MOSFETs were commissioned and their characteristics determined, namely reproducibility, energy dependence and angular dependence. The MOSFETs were also tested in a clinical environment with the phantom. Measurements were carried out to test the MOSFET capabilities in both lung and soft tissue in the phantom. Two plans were devised for the audit process, a straightforward one with two parallel opposed beams and a more complex one involving lung tissue and wedges. These plans were designed to test the entire treatment planning and delivery process.

It was found that each MOSFET detector needed to be individually calibrated. Reproducibility was found to have an average standard deviation of 2% on standard sensitivity and 1.2% on high sensitivity. The angular dependence of the detectors showed that when the MOSFET was rotated by 90 degrees to the beam axis a drop in response of 3% was observed with 6 MV. The energy dependence factor was constant within uncertainty for all MOSFETs.

Overall, the MOSFET and phantom dosimetry system was determined to be suitable for the audit. The measurements with phantom showed that doses in high dose regions could be determined accurately. The greatest variation from the Treatment Planning system dose to the measured dose was 6%. The trial runs of the audit in two New Zealand radiotherapy centres showed that the procedure created is able to find discrepancies within the desired 5%, recommended by the ICRU, in the prescribed dose to the phantom.

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1. Introduction

Cancer in New Zealand

In New Zealand, as of 2003, cancer was the leading cause of death, overtaking heart disease [1]. In the same year it accounted for 28.6% of all deaths [1]. In the older age groups cancer occurs at a much higher rate. In the over 75 age group, the rate of cancer is nearly four times higher than in the 45-64 age group [1]. The rate of new registrations of cancer is also increasing; between the period of 1995 and 2003, an increase of 17.2% new cases were registered as is shown in Figure 1.1. As the average age of the population increases, the rate of cancer is also predicted to increase [1]. It then follows that the treatment of cancer will become increasingly important.

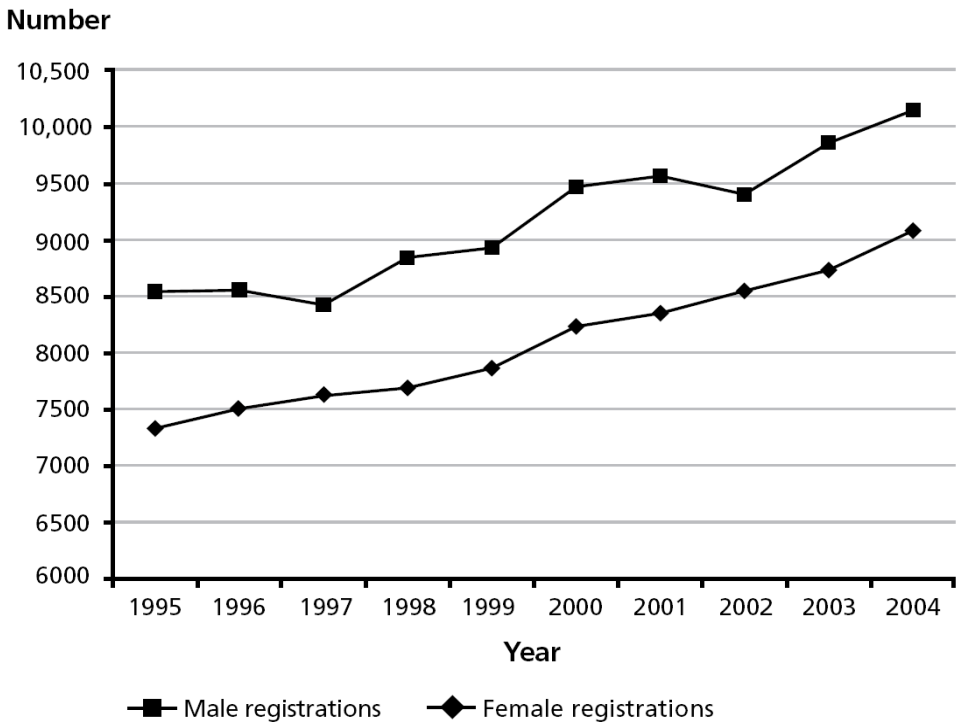

Figure 1.1 The total number of cancer registrations by sex from 1995 to 2004 [1].

A variety of treatment options exist. The common treatments are surgery, chemotherapy and radiotherapy and sometimes a combination of these. Of the

17,000 people newly diagnosed with cancer every year, about 40% (6800) are given radiotherapy treatment [1]. Radiotherapy is used both as a palliative (to relieve symptoms) and curative treatment.

Radiotherapy works by delivering ionising radiation to the tumour volume. The common modalities in New Zealand are photons and electrons [2]. Linear accelerators (linacs) usually deliver the dose and are capable of delivering both electrons and photons. This method of treatment works by damaging the DNA in the cancer cells thus preventing them from reproducing [3]. It is vitally important that the cells in healthy tissue receive minimal damage while the cancer cells receive a sufficient amount to control the cancerous tissue [4]. Small changes in dose can have large effects on patient outcome. Too little means that the cancer is not sufficiently controlled and too much may lead to side affects and permanently damage healthy tissue [5]. For example, an increased dose to the skin can cause deterministic effects such as erythema and necrosis; while too much to the spinal cord can cause irreparable paralysis. Much importance in radiotherapy is therefore placed on accurately delivering the correct dose of ionising radiation to a precise location in the patient [4].

In clinical practice great care is taken to ensure that the administered dose matches the prescribed one. If the discrepancy between these two doses becomes too large a radiation accident has occurred. The International Atomic Energy Agency (IAEA) records these accidents so that other centres around the world may learn from previous mistakes [6, 7]. The IAEA definition of a radiation accident is defined as follows:

- *"Accident refers to any unintended event, including operating errors, equipment failures or other mishaps, the consequences or potential consequences of which are not negligible from the point of view of protection or safety*."

There are currently six radiotherapy treatment centres in New Zealand (all in public hospitals), with a further two private centres planned for 2009. The current six New Zealand centres are in Auckland, Hamilton, Palmerston North, Wellington, Christchurch and Dunedin. The two private centres are planned for Auckland and Christchurch. With nearly 7000 patients to treat every year [1], these facilities are very busy and are running close to capacity.

Planning and delivering a radiotherapy treatment is a complex process and consists of multiple steps that are typically carried out by different hospital staff. Typically the process of radiotherapy is as follows: the patient is imaged using a CT scanner where tattooed set-up marks are applied to the patient to allow accurate positioning to receive the treatment, the data is transferred to the treatment planning system, where the Radiation Therapist (RT) makes a treatment plan based on the contouring of the tumour and the organs at risk by an oncologist, the plan is checked by a medical physicist, which is then sent through a verification system, and then finally the treatment using a linear accelerator (linac), takes place.

In each step of the treatment chain outlined above can introduce potential errors that may affect the success of the treatment. Section 1.1 outlines these potential sources of error.

1.1. Potential sources of error in radiotherapy

There are several potential sources of error in the treatment chain which could affect the radiation dose that the patient receives [8]. The following is a brief overview of the sources of potential errors.

Treatment Planning Systems

The Treatment Planning Systems (TPS) calculates the dose in any place within a patient given a set of specified beam parameters. All of these TPSs account for differences in tissue density and volume along with machine specific output data to calculate the required dose. It is critically important that the CT data of patient density is correct so that the treatment planning system has the correct information to calculate the dose properly and also that the CT data are accurately converted to electron density data in the TPS. In New Zealand the TPSs that are used are Pinnacle (Koninklijke Philips Electronics, Bothell, USA), Eclipse (Varian Medical Systems, Palo Alto, USA) and XIO (CMS, St Louis, USA). The algorithms used for dose calculations differ between TPSs, further detail is given in one of the sections below.

CT number and electron density conversions

CT scans represent the different tissues within a patient by means of Hounsfield numbers (HU[1]) [4]. For the TPS to correct for tissue inhomogeneities the HU need to be converted into electron densities. This is accomplished by means of a CT to electron density conversion table. Since the treatment planning algorithms that calculate the dose depend on the electron densities, it is important that these electron densities are accurate as inaccurate electron densities can lead to an inaccuracy of

[1] X-ray attenuation unit used in CT Scan interpretation which characterises the relative density of the patient. The radio-density of distilled water at standard pressure and water is defined as zero HU, while air is defined as -1000 HU.

the delivered dose to the patient [9]. A CT number to electron density conversion table has to be determined by the medical physicists as each CT scanner can give slightly differing conversions. The information from the scan is then entered into the TPS. If this is carried out incorrectly, it could affect patient dose.

Treatment planning algorithms

In treatment planning, compromises may be made between speed and accuracy [10]. The most accurate algorithms such as Monte Carlo [11, 12] take longer for the system to calculate. In clinical practice, however, Radiation Therapists (RTs) have many patients to design plans for so the use of faster and less accurate algorithms are necessary. Some algorithms, for example, may underestimate the dose distributed by scatter outside of the main beams. Algorithms tend to be the least accurate where there are boundaries of sharp density changes [13] such as lung tissue interfaces in the lung. When a new TPS is commissioned and installed, the linac geometry and output is modelled by medical physicists and from then on only consistency checks are performed, so if the system was commissioned incorrectly at the start, the error might not be realised in later checks. In addition to errors associated with incorrectly commissioning a commercial TPS, a major potential source of error arises when the TPS software is upgraded. This can give rise to changes to the TPS algorithms or stored machine beam data. Some algorithms include; the anisotropic analytical algorithm, the collapsed cone, the multi-grid superposition, fast Fourier transform convolution, and Monte Carlo simulations [14]. Different centres around New Zealand use different TPSs and, therefore, different algorithms.

Machine output

Commissioning a new linac provides the baseline values for the operation of the linac and the data for the TPS. Daily measurements as well as routine quality assurance

(QA) checks by the medical physicists determined by means of measurement verify that the machine output factors are within certain margins of the original machine calibration. Therefore if the original calibrations were incorrect, their daily measurements will also be flawed. The linac output measurements are audited every two years by the National Radiation Laboratory (NRL) and each centre's ionisation chambers are calibrated every other year (also usually by the NRL). Therefore major discrepancies between prescribed and delivered dose are unlikely to occur from this source, though they are not unheard of. In Canada in 1985-1987, six known accidental exposures occurred by the same type of linac [5]. Under specific circumstances when the linac went from x-ray mode to electron mode, the linac operated without the x-ray target and beam flattener which then lead to massive overdoses, causing the deaths of three patients. This continued for this period of time as there was no efficient mechanism to follow up reports of suspected accidents [5]. This was machine failure of a type that would not be picked up in daily measurements due to the specific circumstances in which the overdoses occurred.

Human error

Human error is one of the main sources of radiation accidents in radiotherapy [15]. This section overlaps with the other section in the potential sources of error section as they all have the capacity for human error to be the main factor. For example in Glasgow in 2006 [16] a patient received a 60% overdose due to a radiotherapy accident after a member of staff incorrectly transcribed a number. In a radiotherapy centre in the UK from 1986-1987 the Co 60 calibrations were based on incorrect tables and there was no independent check of the dose rate, with the result being that 207 patients were given a 25% overdose [6]. In the years 1982-1991, again in the UK, a treatment planning system was commissioned inappropriately which caused nearly 1000 patients to be under dosed by 5-30% [6]. In a more serious incident in

Spain, which resulted in at least 18 deaths from radiation, there was an error in the maintenance of a linac and procedures regarding the linac being taken to and from maintenance were not followed. The physicists at this centre were not informed of any maintenance taking place. The procedures for regular machined output verifications were not performed and the display signals were ignored [5, 6]. This is a particularly good example of the importance of communication in radiotherapy [5]. In Costa Rica in 1996, the results from an external audit were ignored and over 100 patients were given an overdose of around 60%. At least 17 people died in this incident due to radiation over exposure [6, 17]. These examples show that it is not only very important to regularly perform QA, but it is also essential to cross-check the work independently to avoid the same human errors being repeated.

Patient set up

The position of the patient in the CT scanner must exactly be reproduced under the linac. This is normally done with the aid of any of the following; room lasers, cone beam CT scanners, image guided radiotherapy, and portal imaging. All of these systems need to be calibrated, and if this is performed incorrectly errors in the location of the delivered dose may occur. Discrepancies between the linac positioning lasers and those of the CT or TPS do occur [18, 19]. For example, if the lasers are misaligned the beams can miss significant proportions of a tumour and damage enough healthy tissue to adversely affect patient outcome. The RTs rely on the lasers and the tattoos to position the patient. The lasers are checked as part of the quality assurance (QA) process (usually daily), but if they get bumped or moved slightly, the positioning of the patient could be out. The laser position is usually checked daily at most centres as part of the QA process.

Other potential sources of error

Other potential sources of error include software and hardware faults. A software error can occur when patient/plan data is transferred from the CT to the TPS or the TPS to the linac, the information transferred could potentially be corrupted. An example of a hardware fault is when a 'sticking' Multi Leaf Collimator (MLC) changes the beam affecting patient treatment.

All the potential sources of error in the various categories mentioned above can cause both dosimetric and/or geometric errors which inevitably affect patient outcome.

1.2. Motivation

Presently in New Zealand there is no external QA process carried out by the NRL other than solely checking the machine outputs. The Trans Tasman Radiation Oncology Group (TROG) group has carried out level three audits (see section 1.2.1) in New Zealand, though these have not been regular [20]. There is a need for regular dose verification in external beam radiation therapy in New Zealand. The ICRU 62 [21] report details a procedure that addresses this need. This procedure has been implemented in other western countries including Australia [20], the United Kingdom [22] and Ireland [23]. The (New Zealand) NRL, as part of its compliance monitoring activities, intends to develop such a national dose survey. This procedure, when implemented in New Zealand intends to verify the accuracy of the planning and delivery of prescribed radiotherapy doses to patients. This survey is to be developed in association with the University of Canterbury and clinical staff in New Zealand's six radiotherapy centres. The work done in this Masters thesis project provides the basis for this audit and is intended to have the capacity to be extended to the auditing of more complex techniques, such as Intensity Modulated Radiation Therapy (IMRT) or

stereotactic radiotherapy, when these high precision treatment approaches become more widely available in New Zealand.

This project creates an approach for an audit that is relevant to radiation therapy in this country. The input of the NRL and the expertise of many clinical staff will be taken into account. The approach is based on an anthropomorphic chest phantom, provided by the NRL, to which a prescribed dose is delivered. Doses received by the chest phantom were measured using a Metal Oxide Semiconductor Field Effect Transistor (MOSFET) dosimetry system [24] which was provided by the NRL. This project includes the commissioning and calibration of the MOSFET dosimetry system, determination of representative tumour volumes, treatment technique and also specifies the process which the phantom will go through at each of the centres in order to test the full treatment planning and delivery chain.

In order to audit the radiotherapy centres around New Zealand it was decided, in conjunction with the NRL, to do an International Commission on Radiation Units (ICRU) level three [21] audit with the anthropomorphic chest phantom. The chest phantom for the audit was made by Computerised Imaging Reference Systems (CIRS, Norfolk, USA). The phantom has the proportions based on the chest of a 70 kg male. Within the phantom there are three polymers approximating three different tissue types. These polymers types are spatially arranged in the phantom to model a person's spinal cord, lung and soft-tissue. The densities and electron densities of the different polymers are similar to that of a real person so that the behaviour of radiation in the phantom will be similar. Figure 1.2 shows the phantom used and Figure 1.3 demonstrates the arrangement of the different polymers within the phantom.

Figure 1.2 CIRS anthropomorphic chest phantom.

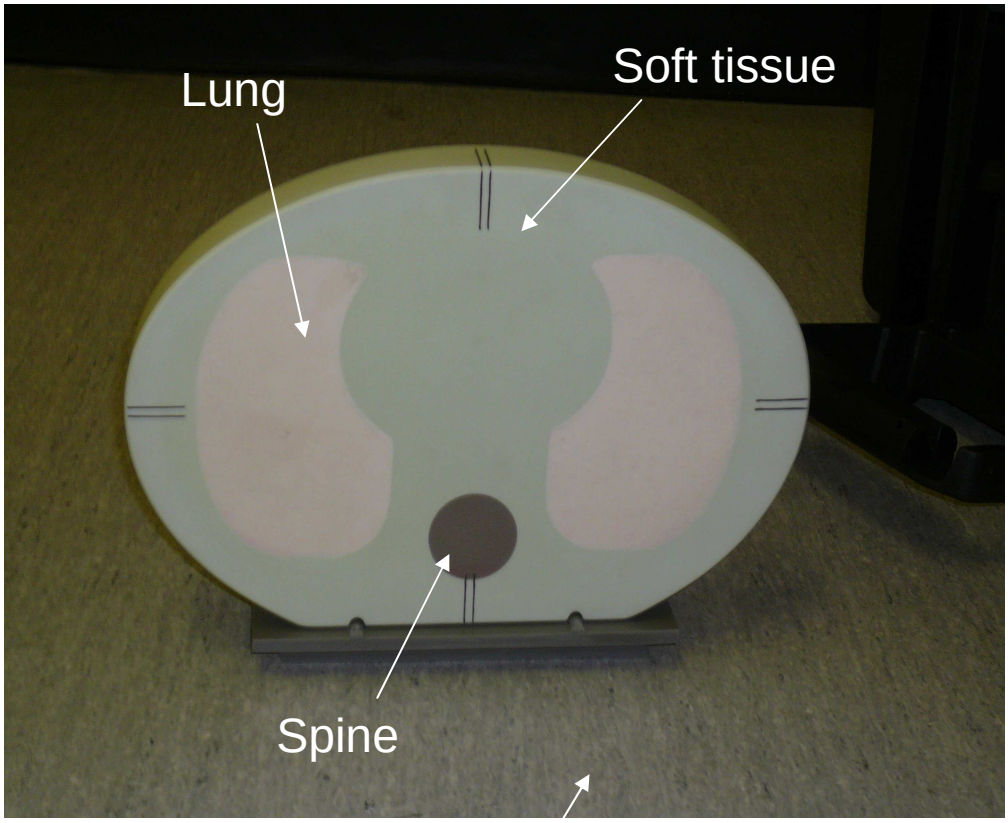

Figure 1.3 The arrangement of the different polymers in the phantom.

1.2.1 The levels of audits

ICRU 62 [21] defines three levels of auditing the dose at radiotherapy centres. Level 3 was decided to be the most relevant for this project. Details of each level are given in Table 1.

Table 1 The ICRU audit levels

Level 1: Basic Techniques *"The minimum requirements for reporting can be followed in all centres including those with restricted therapy equipment, dosimetric computer and staff facilities. This level may be sufficient in a centre when only simple treatments are performed, such as some palliative treatments. For this level it is assumed that the dose at the ICRU Reference point and an estimation of the maximum and the minimum doses to the PTV* (Planned Treatment Volume) *can be determined."*
Level 2: Advanced Techniques *"The standards of Dose planning at this level allow the exchange between different centres of more complete and relevant information. At this level it is assumed that the GTV* (Gross Treatment Volume)*, CTV* (Clinical Target Volume) *and PTV [21] can be defined in one or more planes (sections) using reliable patient data acquisition tools, and modern imaging techniques under reliable conditions (a series of CT and or MRI scans). It is also assumed that complete dose distributions are computed in the central plane and in other planes (sections) using only central axis dose data, and with inhomogeneity corrections, when appropriate."*
Level 3: development of Techniques *Level 3 includes the development of new techniques for which reporting criteria are not yet established (e.g., BNCT[2], intensity modulation, etc.). Some procedures, which are now at level 3, can become level 2 with the development of techniques, equipment and standards.*

[2] BNCT – Boron Neutron Capture Therapy

1.3. The aim and scope of the project

The aim of this project is to investigate and develop procedures to create an ICRU level 3 audit using the CIRS phantom to measure the accuracy of the entire treatment chain from imaging to the delivery of the doses with the linac. This investigation led to the creation of the audit protocol, which contains clear instructions of what the hospitals are required to do to take part in the audit and also contains the information that the NRL needs to collect. In order to create this level 3 audit we were forced to ask the following questions: What exactly do we want to test? How do we test these parameters? Are our resources and equipment suitable? Furthermore, the audit should also be practical in that it could not be allowed to create a heavy burden on the workload of the already busy hospitals, while still getting relevant and meaningful dose information. The protocol also needs to be straightforward in order that it is quick and easy for the NRL to adapt. Part of the project has been the development of spreadsheets so that only the raw information has to be entered and the doses along with the uncertainties are automatically calculated. It is hoped that this will ease the work of conducting the audit and following the protocol.

The NRL has purchased, along with the CIRS chest phantom, a Thompson and Nielsen MOSFET detector system for the purpose of conducting this audit. It is envisaged that in the future this audit could be expanded to encompass more complicated radiotherapy treatments, such as IMRT, stereotactic treatment, etc.

1.4. Outline of the thesis

The commissioning and calibration of the equipment constituted a large part of this project. It was necessary to determine how accurate and reproducible the MOSFETs were. Furthermore, the limitations of the phantom had to be determined. Chapter 2 deals with the MOSFET detectors, their capabilities and uncertainties, while Chapter 3

details the development of the appropriate treatment plans and procedure for the chest phantom that will be carried out for the audit. Chapter 4 details the measurements with the phantom. In both Chapter 3 and 4 the *preliminary phantom measurements* are the measurements of the phantom with the linac to determine the treatments plans and the *practice audit runs* are the runs of the audit that follow the audit protocol in appendix A. Finally, Chapter 5 discusses and summarises the outcomes of this project and discusses future research directions.

2. MOSFET detectors

2.1. Introduction

MOSFETs are solid state transistors whose technical characteristics are altered by exposure to ionising radiation. This property enables them to be used as radiation detectors. There are advantages and disadvantages to using MOSFETs over other dosimeter technologies for this purpose. Some advantages include their small size and their ability to be read immediately after irradiation with a push of a button. They are also relatively inexpensive. They are not more accurate or more reproducible than the more commonly used Thermo-Luminescent Dosimeters (TLDs) [25-27], however, and are also dependent on energy and angle of the radiation beam relative to the MOSFET, which is a disadvantage [28].

This chapter on MOSFET dosimetry includes information on the MOSFET detector system that was used for this project, MOSFET theory, and the assessment of performance characteristics of the MOSFETs. This is followed by the uncertainty analysis of the MOSFETs. In the final part of this chapter, all the components of the uncertainty from the different parameters are combined to find the overall uncertainty in the MOSFETs for the phantom results.

The characteristics and the uncertainty of the MOSFETs had to be assessed to determine the suitability of this equipment for the audit. Uncertainties for the confidence level 68% (1σ) of are given for each section of the chapter, with the combined uncertainty from all the parameters contributing to the uncertainty of the MOSFETs stated for the 95% confidence level (2σ) in the uncertainty analysis section. As each parameter is explained and characterised, the contributing uncertainty from that parameter in both confidence levels is stated. The determination of the experimental uncertainty of the MOSFETs was a vital part of this project [29]. For the

audit to have value, it was necessary to be able to quantitatively state the accuracy of the MOSFETs with respect to a given dose [29].

2.2. MOSFET theory

MOSFET is an acronym of Metal Oxide Semiconductor Field Effect Transistor. These transistors have three metal contacts (called the source, drain and the gate) attached to the layers of semiconductor which have different dopent concentrations. This arrangement is detailed in schematic diagram of a MOSFET in Figure 2.1.

When the voltage of the gate reaches what is called the *threshold* voltage, current can pass through the P type semiconductor from the source to the drain. Irradiating the silicone oxide insulating layer (called gate oxide in Figure 2.1) increases the threshold voltage needed for current to pass from the source to the drain. The threshold voltage which is proportional to radiation dose can then be measured by the MOSFET reader.

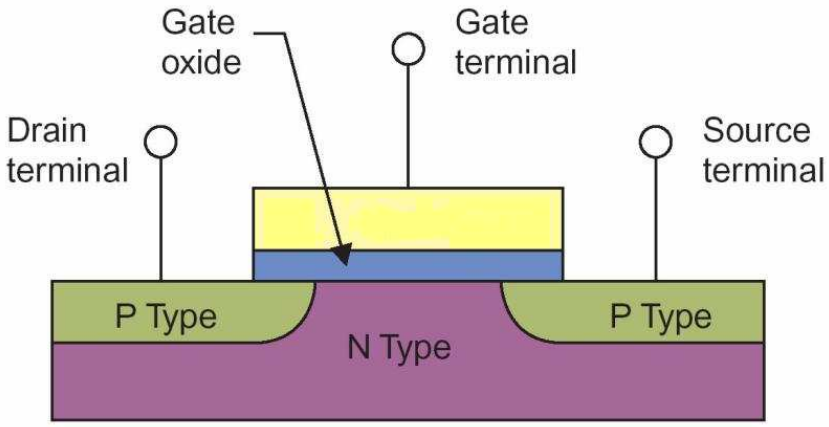

Figure 2.1 A schematic diagram of a MOSFET. For the MOSFET to activate, the voltage on the gate terminal needs to be sufficient for a channel to form through the N type conductor. This threshold voltage is dependent on the properties of the gate oxide layer.

The threshold voltage depends on the properties of the gate oxide layer. When the MOSFET is irradiated holes are trapped in the silicone oxide layer at the edge of the N type semiconductor. In some sense holes are places where electrons would otherwise be. These trapped holes raise the insulating strength of the silicone oxide

layer and consequently the threshold voltage is also raised. During irradiation, a bias voltage is applied between the gate and the drain across the MOSFET. This affects the proportion of holes that are trapped in the silicone oxide layer. For example, a MOSFET with a different bias voltage will also have a different threshold voltage than a MOSFET irradiated with the same radiation and a different bias.

2.3. The MOSFET dosimetry system

The MOSFET system that was used in this project was the Thomson & Nielsen MOSFET 20 Dosimetry System Model TN-RD-50 (Best Medical, Nepean, Canada). This MOSFET 20 system is easily transportable and contains a reader (Figure 2.2), four bias supplies (Figure 2.2) and 20 MOSFET dosimeters (Figure 2.3).

The MOSFETs can be read immediately after irradiation. During the irradiation of the MOSFET (which permanently affects their silicone oxide layer), they are plugged into the bias supply box. The bias supply (Figure 2.2) supplies the bias voltage during irradiation. After irradiation the bias supply box is connected to the reader which gives a reading of the radiation that the MOSFET was exposed to in mV. The reader gives two readings, the total accumulated voltage and the reading for the last irradiation. After each reading the MOSFETs can be zeroed on the reader (though the accumulated mV reading cannot be).

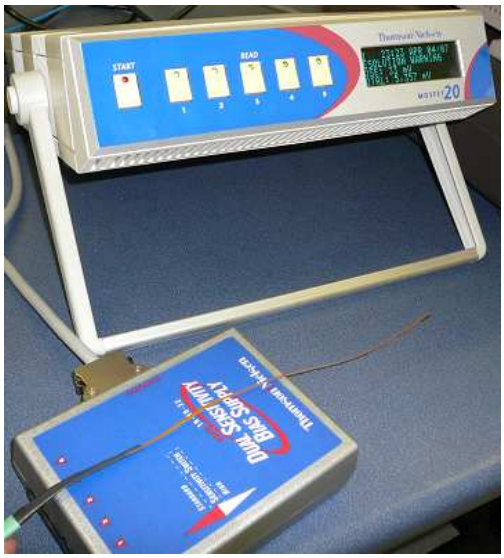

Figure 2.2 MOSFET 20 reader in the background and Bias supply in the foreground.

Figure 2.3 MOSFET dosimeter, the transistor is under the black epoxy bubble at the end of the dosimeter**.**

The dosimetry system has two settings, high sensitivity and standard sensitivity. In these different modes the sensitivity of the MOSFETs vary as follows: In standard sensitivity an absorbed dose of approximately 1 cGy gives a rise of 1 mV of threshold voltage in the MOSFET detector, while in high sensitivity an absorbed dose of approximately 1 cGy raises the threshold voltage of the MOSFET by 3 mV. The voltage across the MOSFETs is different for the two settings during irradiation and causes the differing responses. While investigating the capabilities of the MOSFETs both standard and high sensitivity were used. For the part of this project working with the phantom, the standard sensitivity mode was used as the MOSFETs are disposable and can only be used for a limited life. The higher sensitivity mode would have expended the MOSFETs faster. The MOSFETs can be used to an accumulated voltage of around 20 000 mV [30].

2.4. MOSFET performance assessment

In order for the audit to determine the dose at a point in the phantom, the characteristics of the MOSFETs had to be assessed and calibration factors

determined. Much of the literature about these detectors details the way the MOSFETs capabilities have been assessed previously [26, 28, 31, 32]. The characteristics of the MOSFETs that were tested for this project were: reproducibility, calibration factors (response per Gy), how and if the calibration factors changed over effective dose, energy dependence, fading and, angular dependence. The dose rate with linearity was also considered. These characteristics were tested prior to using the MOSFET system in a clinical environment. They (and other parameters) were tested more thoroughly as more information on specific circumstances of the MOSFET behaviour was required as the project progressed. An example of this increase of testing requirements is the angular dependence measurements. Initially it was assumed that the angular dependence would be constant for all energies, however, after completing the first test audit run, it was realised that the angular dependence must be tested at different energies because the angular dependence is a function of the energy of the radiation.

2.4.1 ^{60}Co performance assessment

The MOSFET system's ^{60}Co performance assessment was performed with the NRL's ^{60}Co exposure facility which was a Theratron 80 teletherapy unit which was manufactured by the Atomic Energy of Canada Limited (AECL). It contains a 40 TBq ^{60}Co source that produces an air kerma rate of approximately 0.16 mGy/min at a source distance of 1 m. To characterise the MOSFETs these detectors were placed in a water phantom at a depth of 5cm in a 10cm by 10cm field at a distance of 1 meter from the ^{60}Co source at the position of the MOSFET. The absorbed dose to water was determined using the IAEA TRS-398 protocol [33] for ^{60}Co measurements. The set up is shown in Figure 2.4. A similar set up was used for the higher energy measurement with the linac. The beam was first measured by using an ionisation chamber with an absorbed dose to water calibration traceable back to a primary

standards laboratory. This gave accurate beam output data of the ^{60}Co source, which enabled the determination of the calibration factors of the individual MOSFETs. The MOSFETs were put into the same cavity as the ionisation chamber in a purpose built perspex sleeve (Figure 2.5). The sleeve was made so the MOSFET would fit snugly in the cavity for the ionisation chamber, as the Ionisation chamber is significantly larger than the MOSFET.

Figure 2.4 The water phantom in which the MOSFETs were calibrated according to TRS-398 protocol.

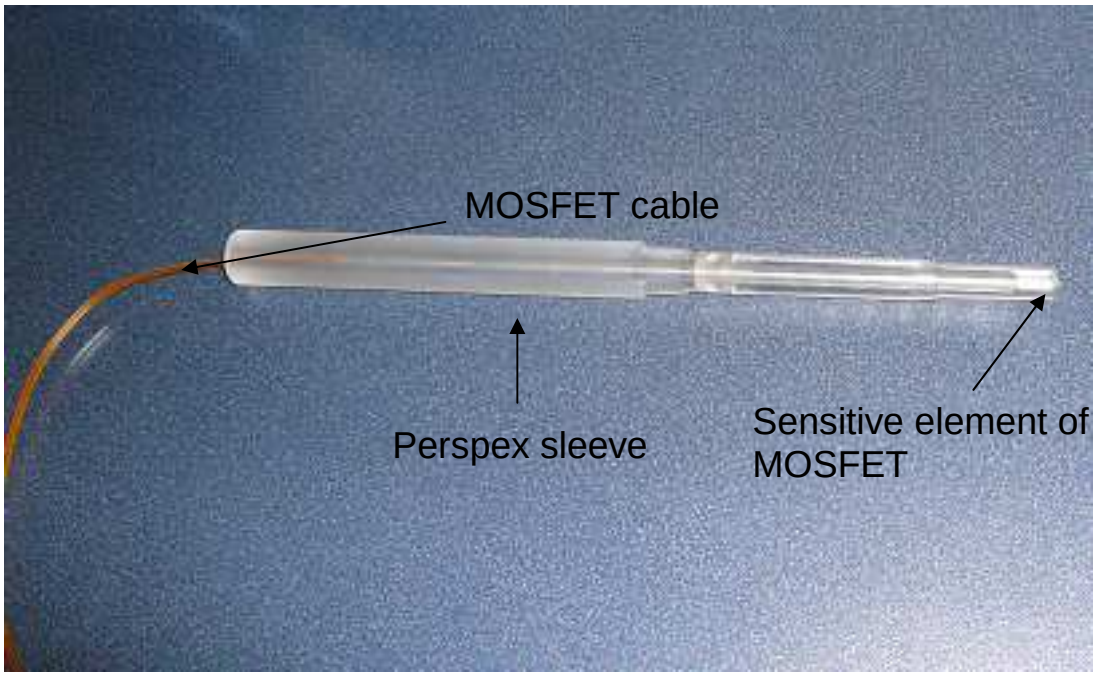

Figure 2.5 Perspex sleeve for MOSFETs which was used in all water phantom measurements with the MOSFETs.

2.4.2 Reproducibility

For a repeated dose given to a MOSFET, the reading obtained for each can vary by several percent. This variation in the reading, or the reproducibility, was investigated. This was an important part of the assessment of the MOSFET detectors as this parameter was likely to be a large component of the combined uncertainty as reported by other researchers [34]. The manufacturer states [30] that these MOSFETs have reproducibility that is shown in Table 2.

Table 2 Reproduction of manufacturers' information [30] on the reproducibility of the MOSFETs. These are to a confidence level of 68% using ^{60}Co.

Dose	Bias supply sensitivity	
	High	Standard
200 cGy	<0.8%	<2%
100 cGy	<1.2%	<3%
20 cGy	<3 %	<8%

To find the reproducibility of the MOSFET, individual MOSFETS were irradiated 10 times with the following doses: 0.5 Gy on high sensitivity and 1.0 Gy on standard sensitivity. These doses were chosen for economic reasons as well as equipment time

restraints. The results for standard and high sensitivity are shown in Table 3 and Table 4 respectively. Note that an individual MOSFET was tested either on high or standard sensitivity, not both. These tables show that some of the MOSFETs were more reproducible than others. The MOSFETs that were determined to be noisy (MOSFETs with a high relative standard deviation) were not used for the practice audit runs. Whether or not a particular MOSFET was too noisy was decided on an individual basis.

Standard sensitivity

Table 3 The reproducibility of the MOSFETs on standard sensitivity irradiated ten times with 1 Gy in ^{60}Co

MOSFET No	Average (mV)	Standard deviation (mV)	Relative std Deviation (%)
2930	115.0	2.1	1.8
2927	115.3	2.9	2.5
2949	113.7	1.4	1.2
2861	113.6	2.5	2.2
2919	116.1	2.3	2.0

The reproducibility in standard sensitivity was found to be similar to the manufacturers' claims [30], with the reproducibility being <3% for a confidence level of 68% on standard sensitivity. To improve the accuracy of the measurements of the doses to the phantom, it was decided that each treatment plan would be repeated 3 times. The contributing uncertainty from the reproducibility was then taken to be the largest standard error[3] from Table 3 the reproducibility measurements on standard sensitivity. The contributing uncertainty that was considered most representative of the MOSFETs that were tested was taken to be MOSFET 2919 with a standard deviation of 2.0%. The standard error of this is 1.15% assuming that three

[3] Please note that the standard error is not the same as the standard deviation, the standard error is the standard deviation divided by square root of the number of measurements.

measurements will be taken in the audit. This is the value that will contribute to the combined uncertainty of the MOSFETs.

High sensitivity

Table 4 The reproducibility of the MOSFETs on standard sensitivity irradiated ten times with 0.5 Gy in ^{60}Co

MOSFET No	Average (mV)	Standard deviation (mV)	Relative std deviation (%)
2933	158.4	2.9	1.8
2932	158.1	1.9	1.2
2931	156.9	2.0	1.3
2912	160.6	1.3	0.8
2935	157.0	1.9	1.2

The MOSFETs on high sensitivity also showed similar results to the manufacturer's claims (Table 2) [30]. The 68% confidence level for the standard error of the MOSFETs on this sensitivity was found to be <0.6%. This value was obtained using the largest relative standard error in the high sensitivity reproducibility measurements.

Measurements in this project on high sensitivity were done for completeness. Since in the audit, or the preliminary phantom measurements, no MOSFETs were used on high sensitivity. Uncertainties from this source (MOSFETs irradiated in the high sensitivity mode), do not contribute to the overall uncertainty of the MOSFETs used in the phantom in this project.

2.4.3 Calibration Factor

Calibration factors were obtained for a total of thirteen MOSFETs. The calibration factor was the dose given divided by the response of the MOSFET. As the calibration factors varied significantly between the MOSFETs it was necessary to obtain a

separate calibration factor for each MOSFET. To obtain these calibration factors, the MOSFETs were irradiated with 1.5 Gy in ^{60}Co three times each. This dose was decided to be a good compromise between expending the MOSFETs too quickly and obtaining accurate calibration factors. Each MOSFET was irradiated three times to obtain the calibration factor, with the calibration factor calculated using the average of the three measurements.

Table 5 Each MOSFET with their own specific calibration factor along with the standard error

MOSFET No	Calibration Factor (cGy/mV)	Relative standard error (%)
2930	0.891	0.8
2861	0.878	0.9
2949	0.896	0.6
2912	0.932	0.9
2935	0.942	0.4
2929	0.882	0.9
2915	0.892	1.1
2032	0.928	0.8
2931	0.945	0.4
2933	0.929	0.8
2947	0.887	1.1
2950	0.907	0.9
2919	0.874	1.6

MOSFET number 2919 was identified as being particularly noisy and so was not used in any of the testing or audits in the phantom. The large standard error of this particular MOSFET was also not considered for the overall uncertainty analysis that is presented at the end of this chapter.

The contributing uncertainty from the MOSFET calibration was taken from the greatest value of the standard errors in Table 5 (except from MOSFET 2914). This makes the contributing uncertainty from the calibration factor to be 1.1% with a confidence level of 68% and 2.2% with a 95% confidence level.

Accumulated dose dependence of the calibration factors

It was necessary to determine whether the calibration factors remained constant as the voltage across a MOSFET increased. To test this MOSFETs that had already been calibrated and in some cases had already had significant dose on them, were re-calibrated after they had acquired more voltage across them. The stability of the MOSFETs' calibration factors' were examined. The uncertainties from both the first and second calibration factors were combined to give the combined uncertainty column in Table 5. The MOSFETs are listed in descending order of accumulated effective dose between the two calibrations. Not all the MOSFETs were recalibrated as it was undesirable to expend too much of their effective dose capacity.

As can be seen from, Table 6 the calibration factor changed, sometimes this change was greater than the combined uncertainty of the calibration factors. To determine if there was any pattern to the change in calibration factors with dose (as there did not seem to be), more measurements of the calibration factor were taken.

Table 6 The change in calibration factor as the MOSFETs acquired more voltage across them.

MOSFET #	1st calibration factor (cGy/mV)	Accumulated voltage (Δ mV)	2nd calibration factor (cGy/mV)	Combined uncertainty (%)	Relative change in calibration factor (%)
2930	0.891	4116	0.926	1.4	3.9
2861	0.878	4900	0.885	1.0	0.8
2912	0.932	3849	0.892	1.1	-4.3
2929	0.882	2461	0.911	1.2	3.2
2914	0.874	1326	0.884	1.9	1.1
2931	0.945	1282	0.923	1.0	-2.3
2933	0.929	814	0.917	1.1	-1.3

Reproducibility over a large dose

To determine if the calibration factor changed as a function of the total voltage across the MOSFET, or the calibration factor changes were solely a function of reproducibility, a MOSFET was irradiated with 1.5 Gy thirty six times with the total dose over 120 Gy. This gave information on the change in response over a large range of accumulated voltage. The MOSFETs' accumulated voltage was equivalent to 60 Gy of dose on standard sensitivity.

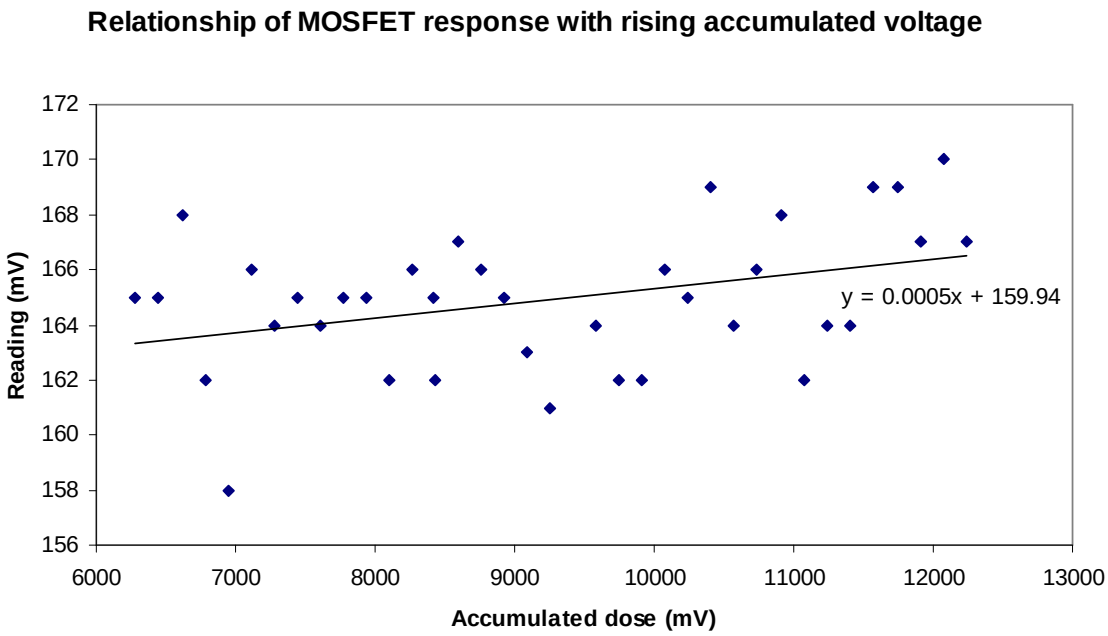

Figure 2.6 This graph suggests that there is an upwards trend in response as the voltage across the MOSFET increases.

There is a slight upwards trend shown in Figure 2.6. The graph shown in Figure 2.7 is perhaps more relevant since it shows the running average of three measurements, which is the number of measurements which will be taken at the audits. It shows that there is a trend upwards that is significant. Consequently it was decided that a MOSFET must be re-calibrated after around 3000 mV of accumulated voltage, which corresponds to an irradiated (effective) dose. The interval of 3000 mV of accumulated voltage was selected the change in response was approximately 2%.

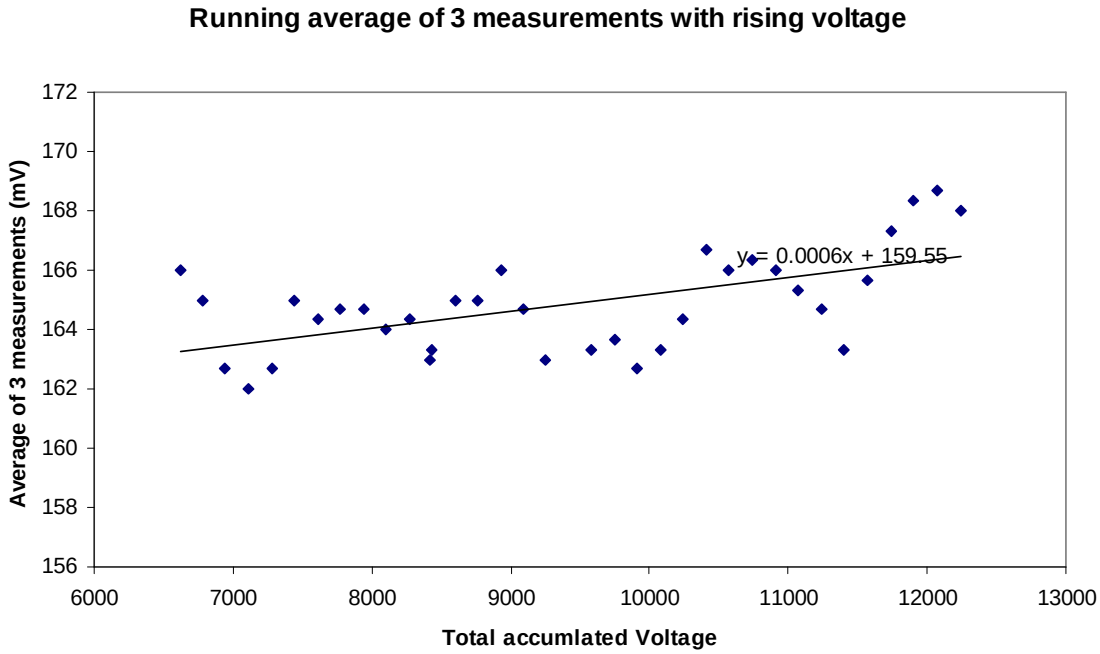

Figure 2.7 This graph shows that the average of three sequential measurements of a MOSFET

To obtain as accurate as possible results for the audit, it was decided that henceforth the MOSFETs should be recalibrated immediately before an audit.

2.4.4 Angular Dependence

MOSFETs are known to have angular dependence, with different researchers having reported a wide range of variations in response to angle [28, 35]. Therefore, it was very important to determine the angular dependence of the MOSFETs that were used for the audit. To perform this, four MOSFETs were irradiated with the ^{60}Co source, in the water phantom set up in section 2.4.1, at angles ranging between 0 and 180 degrees in 45 degree increments. The MOSFETs were rotated counter clockwise (facing the plug end of the MOSFET) around the axis running length wise as shown in Figure 2.8.

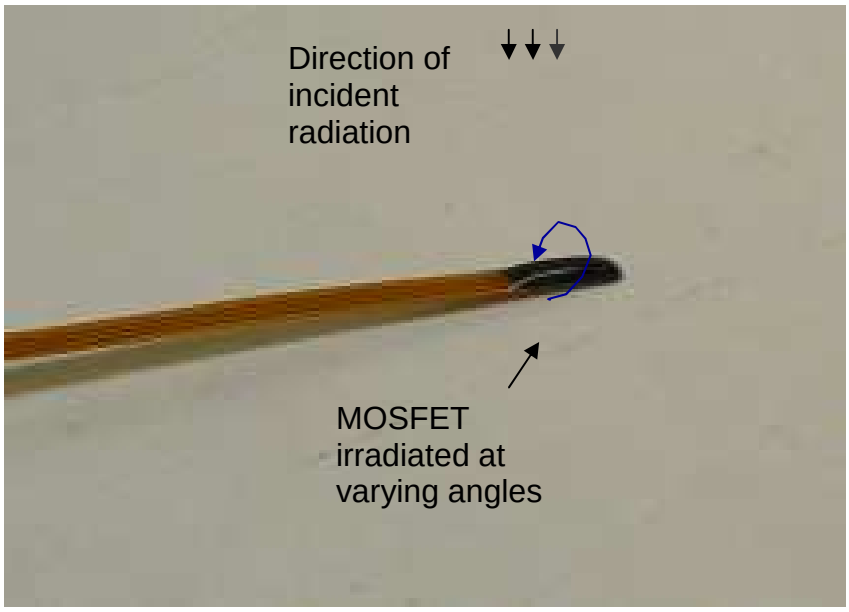

Figure 2.8 Diagram clarifying the angular dependence measurements. The MOSFET was rotated in the direction of the blue arrow and readings of response were taken at various angles.

All irradiations were performed with the bias supply on standard sensitivity. Each MOSFET was irradiated three times with 1.5 Gy at each angle to obtain more accurate results. Figure 2.9 shows the change in response with angle for these trials. The MOSFET response was the least when the MOSFETs epoxy bubble was 90 degrees to the incident radiation beam.

The angular dependence, shown in Figure 2.9, demonstrates that there is a significant contribution to the combined uncertainty from angular dependence of the MOSFETs with the ^{60}Co. The MOSFET manual [30] states that the angular dependence values for irradiations 1-20 MeV is 2% for 360 degrees of rotation. Figure 2.9 shows that the angular dependence for ^{60}Co (with gamma emissions at 1.33 and 1.17 MeV [36]) is more than this. However, after conducting angular dependence measurements with 6 and 18 MV photons (Figure 2.10 and Figure 2.11 respectively), it was found that the results were closer to that of the manufacturers' claims [30] than the ^{60}Co, though the angular dependence remained larger than the manufacturer's claims.

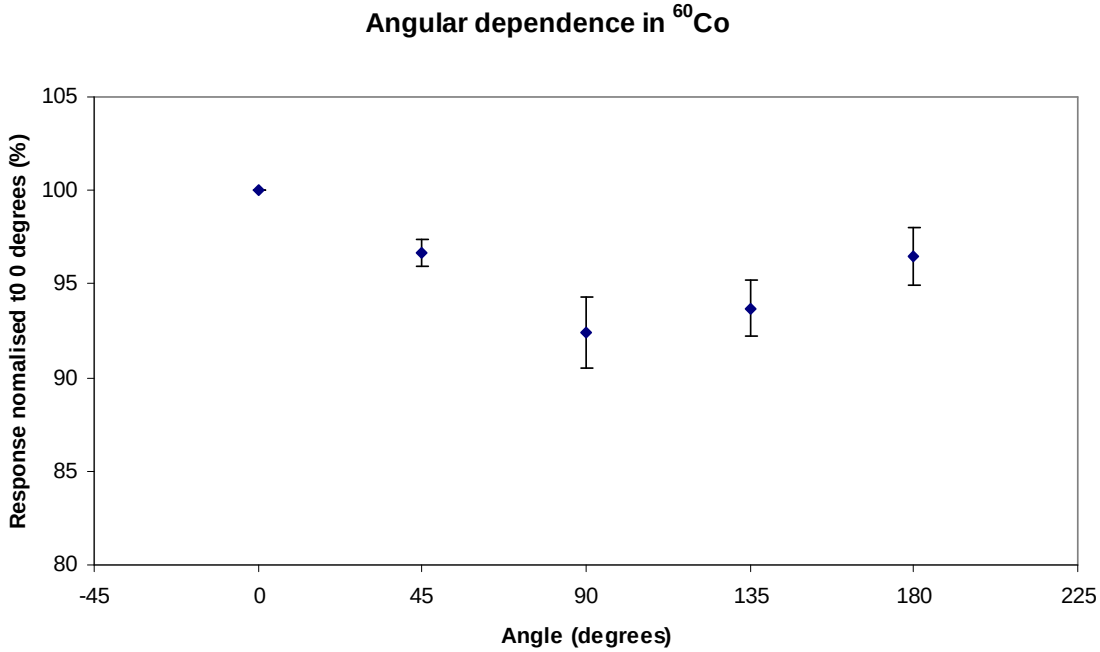

Figure 2.9 This graph shows that the angular dependence of the MOSFETs in ^{60}Co is approximately sinusoidal. There was a variation of the amplitude from 6 to 10 percent for the different MOSFETs

To obtain information on the angular dependence of the MOSFETs that were to be used in the practice audit runs, the MOSFETs used were, for economical reasons only irradiated at 0 and 90 degrees (at both 18 and 6 MV for completeness). This was because it was undesirable to expend the capacity of the MOSFETs more than was necessary.

Figure 2.10 shows the percentage drop in response at 90 degrees with 6 MV photons. It was determined that the drop in response at 90 degrees to the beam for the MOSFETs used in the final stages of the practice audit run ranged from 2.3% to 3.9% with an average of 3.3%. Currently, the treatment plans for the audit solely use 6 MV photons so the correction factor for the angular dependence was determined from these measurements.

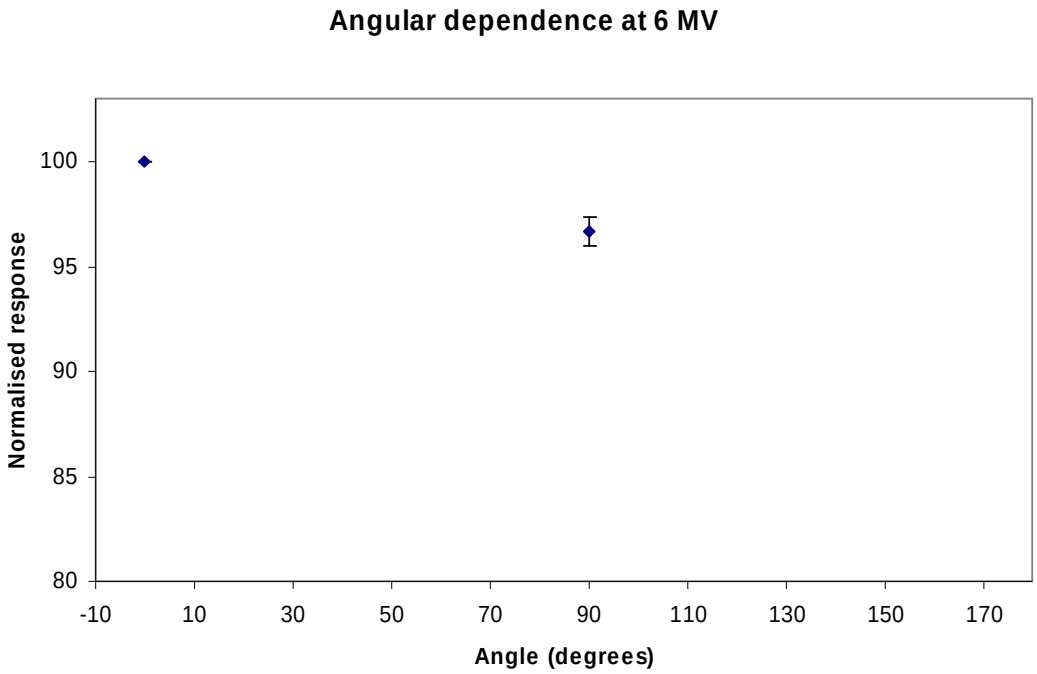

Figure 2.10 Angular dependence of the four MOSFETs at 6 MV. The angular response was tested only at angles of 0 and 90 degrees. The error bars represent one standard deviation

As in the audit the MOSFETs will be used to measure 6 MV, the uncertainty in the angular dependence with this energy counts towards the combined uncertainty in the MOSFETs used in the audit. The contributing uncertainty of the MOSFETs from the angular dependence is 0.33% with a confidence level of 68%

The percentage drop for 18 MV (Figure 2.11) showed similar results with an average drop of 2.8%. This is close to the manufacturers' claims of 2%, but if in the future the audit uses 18 MV in the treatment plans, correction for the angular dependence would be necessary.

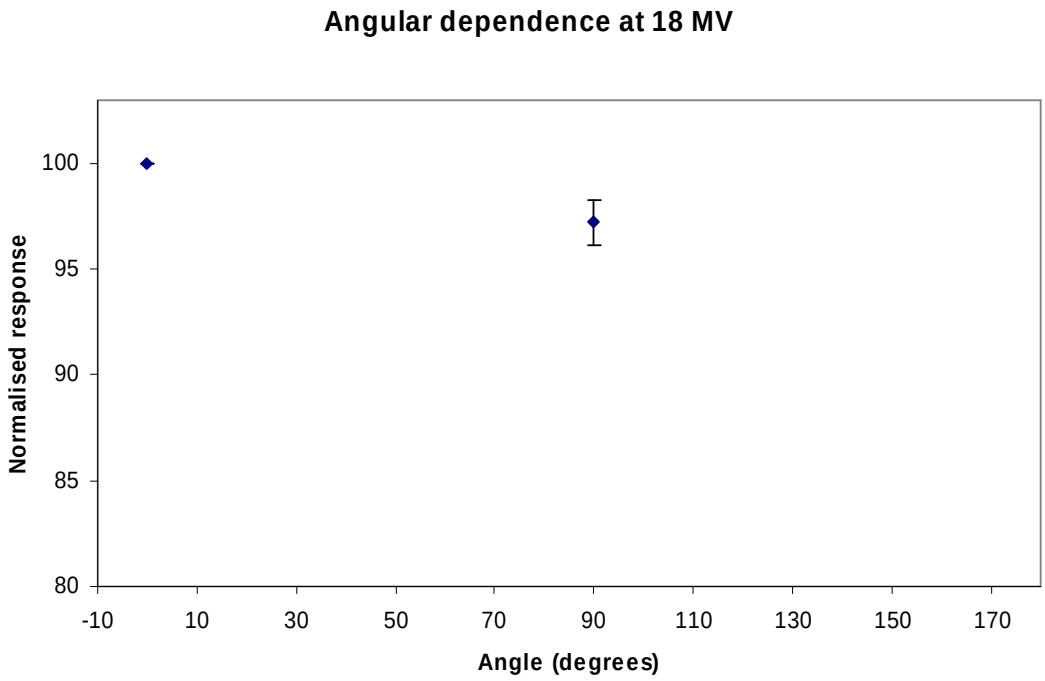

Figure 2.11 Angular dependence of the four MOSFETs tested at 18 MV. These MOSFETs used for the final stages of the preliminary audit run. The angular response was tested only at 0 and 90 degrees. The error bars represent one standard deviation.

The angular dependence was significantly different between ^{60}Co and 6 MV and 18 MV suggesting that the angular dependence varies with energy. This has been reported by other researchers [37], who noted that the angular dependence dropped up to 28% for photons in the keV range. The magnitude of the differences in response indicates that the angular dependence should always be corrected for. That is, at all angles except zero, a factor correcting for angular dependence is required.

Deriving the relationship of the angular dependence

To derive the angular relationship of the angular dependence, it was initially assumed that the angular dependence would be symmetrical around 180 degrees. It was later realised that this symmetry could be further investigated, this was not done in this project due to both time constraints (each measurement took ten minutes with the ^{60}Co source) and also to avoid expending the MOSFETs effective dose capacity. As can be

seen from Figure 2.12, the curve of best fit approximates a sine curve (the curve below is in reality a polynomial).

The stretched out sine curve derived was; $1 + 0.06\sin(0.9 \times 3.1/180)$. It was assumed that the shape of the curve for the angular dependence for ^{60}Co and the 6 MV photons would be the same with the exception of different amplitude.

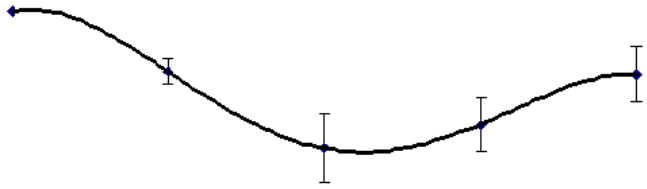

Figure 2.12 The curve above approximates a sine curve, a relationship of the drop on response with angle was derived.

2.4.5 Linearity

Linearity is the response of the MOSFET as a function of the dose irradiated. This was an important characteristic to test as if this function was not linear, the curve would have to be calculated and corrections for this curve applied so accurate doses measured with the MOSFETs could be obtained. The linearity was measured on both high and standard sensitivity.

Standard sensitivity

The linearity of the MOSFET on standard sensitivity was tested in a similar way. Five measurements were taken at five varying doses to obtain the linearity as shown in Figure 2.13. Each error bar in this figure represents one standard error.

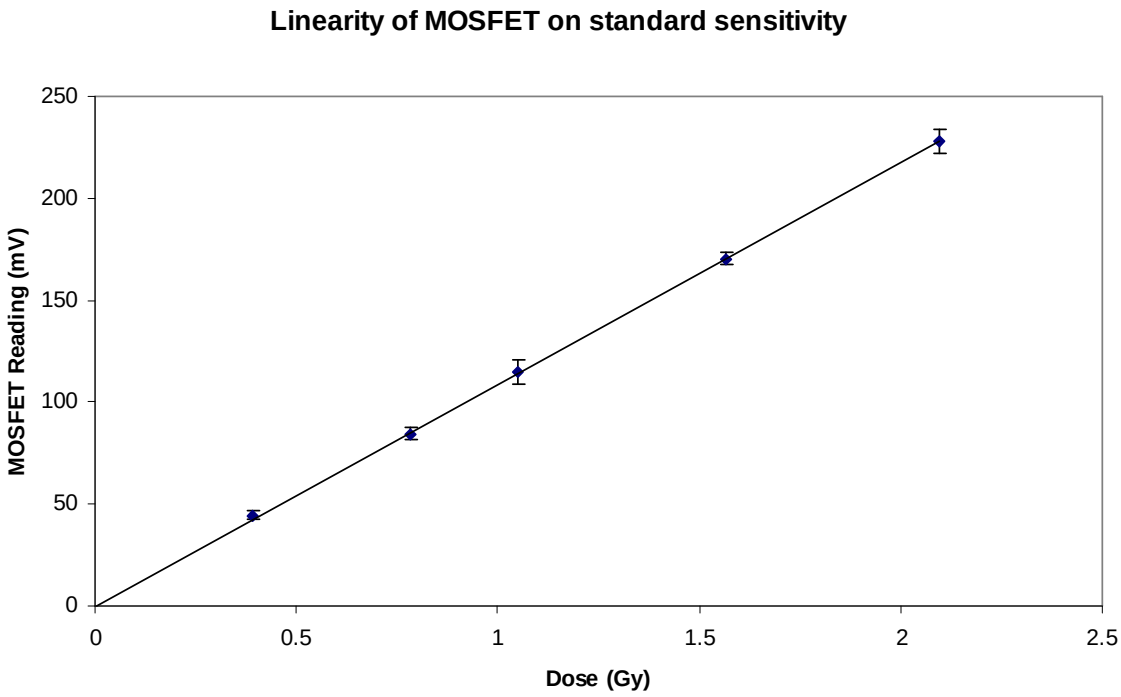

Figure 2.13 Linear relationship of MOSFET on standard sensitivity

It was found on standard sensitivity that there was a strong linear relationship with dose. As with the high sensitivity measurements, the line of best fit passes well within all the error bars for the standard sensitivity measurements. The strong linearity shown by the MOSFETs at both sensitivities shows that, for the doses investigated, the MOSFET radiation detectors will be suitable for the audit without any corrections factors for linearity needed.

The linearity with dose rate was also considered. This was assumed to be taken into account with the comparison (and the consequent correction factor applied) between the ^{60}Co source (low dose rate) and the linac (high dose rate) measurements in section 2.4.7.

High sensitivity

To find the linearity of the MOSFETs on high sensitivity, five measurements were taken at six varying doses using the ^{60}Co source to accurately determine the linearity.

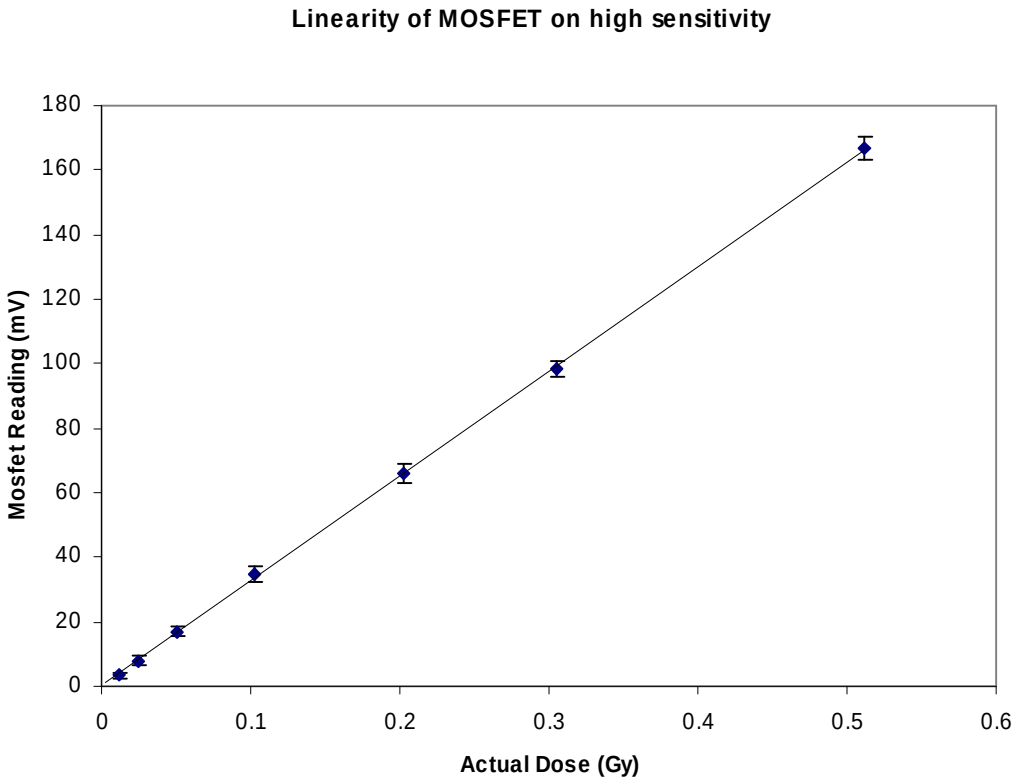

Figure 2.14 The linearity of the MOSFET on high sensitivity

These tests showed that the MOSFET had excellent linearity of response at high sensitivity. Figure 2.14 shows this linear relationship, with the line of best fit passing well within all the error bars, where each error bar represents one standard error.

2.4.6 Fading

MOSFETs can fade if they are not read immediately after irradiation. This means that the response in mV is lower if MOSFETs were read some time after irradiation instead of immediately. According to the MOSFET manual [30] this is less than 3% of 2 Gy when read within 15 minutes of exposure. This was measured with the MOSFETs that

were used for this project. The MOSFETs were irradiated with 1.5 Gy with the ^{60}Co source and then read after five, ten, and fifteen minutes; this was repeated three times. No fading was observed in these measurements.

2.4.7 Energy Dependence

To assess the energy dependence in the response of the MOSFETs, three MOSFETs were irradiated with 1.5 Gy with the ^{60}Co source and also 18 MV and 6 MV photons from the linac which was a Varian 21iX S/N 1009 model. Figure 2.15 shows the relative response of the MOSFETs which demonstrates that 18 MV gave the lowest response and Co 60 the highest.

The contributing standard error to the combined uncertainty of the MOSFETs from the energy dependence is 1.1% with a confidence level of 68%. This is from the standard error of the distribution of the energy dependences from the individual MOSFETs tested. This uncertainty is only relevant to the preliminary phantom measurements, not the practice audit runs as the MOSFETs used in the practice audit runs all had calibrations factors obtained for them in 6 MV photons. This was found to be consistent with other research [38].

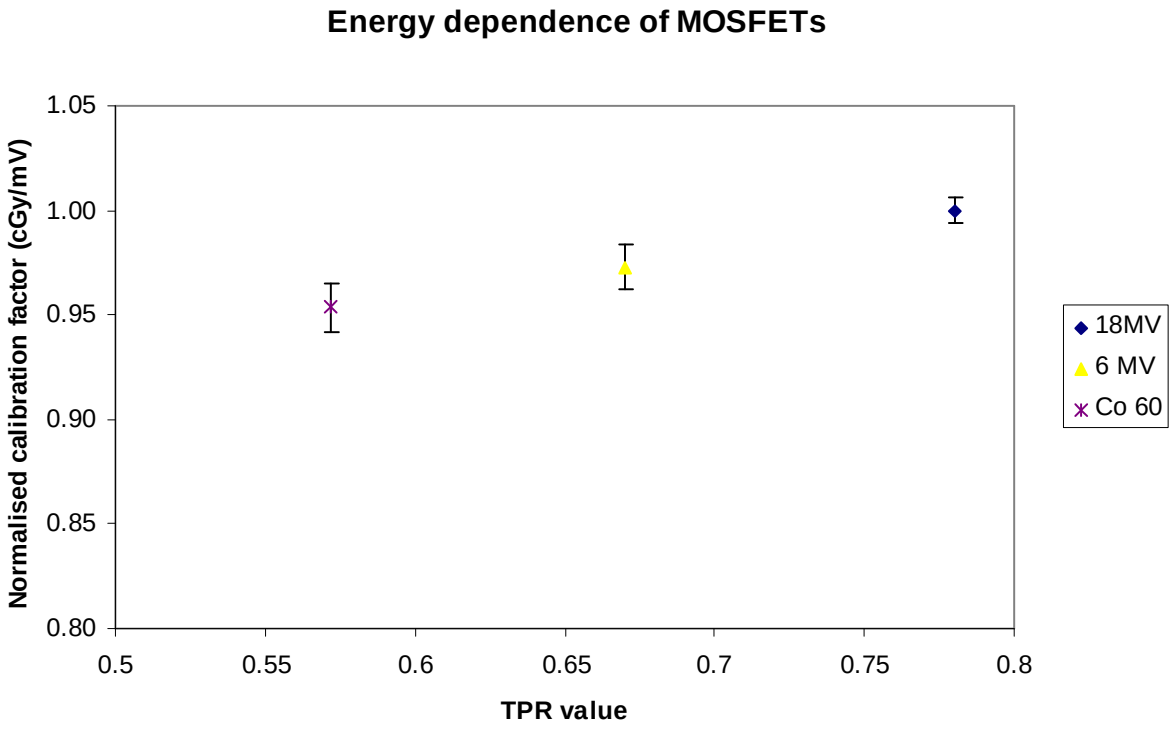

Figure 2.15 Graph showing the relationship between the MOSFET calibration factor with changing energy and TPR[4] [39] value.

2.5. The worksheet

An excel worksheet was created to apply all the corrections from the parameters characterised in this chapter, which calculates the dose to the MOSFET. Only the beam angles, the MOSFET locations in the phantom, the MOSFET number and response, are required to be entered into the worksheet. The worksheets can be found in Appendix B. The worksheet was created for ease of analysis, and also so the results could be entered into the worksheet during irradiations of the phantom with the doses given immediately. It will also lighten the burden for the NRL to analyse the results in future audits.

[4] TPR stands for Tissue Phantom Ratio and is defied as the ratio of the dose at a given point in phantom to the dose at the same point at a fixed reference depth usually 5 cm.

2.6. Uncertainty analysis

The uncertainly analysis was a vital part of this project. In the audit, it needs to be stated within what accuracy the MOSFETs can give information on the dose. To find the total uncertainty of the MOSFETs all the correction factors and their associated uncertainties needed to be taken into account. The uncertainty in the dose determination when using the MOSFETs was found using the principles according to the guidance in International Organisation for Standardisation's '*Guide to the Expression of Uncertainty in Measurement'* (ISO guide) [29].

Not all the MOSFETs were calibrated in the same manner for all the runs in the phantom. For example, the MOSFETS used in the first trials with the phantom under the linac, and the MOSFETs used in the practice audit runs were calibrated slightly differently, thus the uncertainty analysis is not the same for the two sets as one of these sets does not include the uncertainty in the energy dependence.

For the first runs with the phantom under the beam, the MOSFETs were calibrated using the ^{60}Co source and a correction factor for the energy was applied. For the practice audit runs, information was obtained about the characteristics of those particular MOSFETs for the photon energies used in the audit.

2.6.1 General uncertainties

According to the ISO guide [29] there are two types of uncertainties, they are:

A Type A method of evaluation of uncertainty is by the statistical analysis of a series of observations.

B Type B method of uncertainty is by means other than the statistical analysis of a series of observations

With one exception, all the uncertainties of the MOSFETs were type A uncertainties as they were determined by statistical analysis. The exception to this is the uncertainty in the angles that were not in 45 degree increments between 0 and 180 degrees, as the angular dependence between these increments were interpolated between the data points. Since that in both the preliminary phantom measurements and the practice audit runs the MOSFETs were at angles to the beam that were not in 45 degree increments from zero, there is an uncertainty associated with these angles that are not from statistical analysis. Thus an estimated uncertainty of 0.5% was added to cover this type B uncertainty.

In this section all the factors of the uncertainty that were taken into account will be indentified. The errors and where they come from are outlined below.

Table 7 The contributing uncertainties to the combined uncertainty of the MOSFETs

Type A uncertainties	Contributing uncertainty	Type B uncertainties	Contributing uncertainty
Reproducibility	0.8%	Interpolation of angles	0.5%
Calibration factor	1.1%		
Angular Dependence 6 MV	0.3%		
Energy Dependence	1.1%		

All the type A uncertainties were found by finding the standard error of the group of measurements that tested those particular characteristics. The equation to find the delivered dose from the MOSFET response using TRS-398 [33] is:

Equation 1

$$D_{w..Q} = M_Q \times N_{D.W.Q_o} \times k_a \times k_e \times k_r \times k_{ia}$$

Where:

Symbol	Definition
$D_{w.Q}$	the absorbed dose to water in a water phantom at reference depth with a beam quality Q
M_Q	The MOSFET reading
$N_{D.W.Q}$	The calibration factor of the MOSFET
k_a	correction factor of the angular dependence of the MOSFET
k_e	energy correction factor
k_r	Correction factor for the reproducibility, equal to one, but does have a contributing uncertainty
k_{ia}	Correction factor for the interpolations between the data points in the angular dependence, equal to one, but does have a contributing uncertainty

Since all the correction factors are multiplicative, the combined uncertainty, according the *Guide to the Expression of Uncertainty in Measurement* [29], is the positive square root of the combined uncertainties squared. The coverage factor used was two; this was chosen to give a confidence level of approximately 95% in the MOSFET results, though the sample size in the audit may be sufficiently small to warrant a coverage factor in excess of 2.

2.6.2 Combined uncertainty in preliminary phantom measurements

For the uncertainty of the doses given by the MOSFETs used in the first trial runs with the phantom, the energy dependence, angular dependence, reproducibility, and uncertainty in the calibration factor were taken into account:

Table 8 The combined uncertainty for the MOSFETs in the preliminary phantom measurements.

Type A uncertainties	Contributing uncertainty	Type B uncertainties	Contributing uncertainty
Reproducibility	1.15%	Interpolation of angles	0.5%
Calibration factor	1.1%		
Angular Dependence 6 MV	0.33%		
Energy Dependence	1.1%		
Combined uncertainty	2.0%		
Expanded uncertainty	4.0%		

2.6.3 Combined uncertainty in the audit practice run

The MOSFETs used the audit run all had calibration factors obtained with 6 MV photons, hence the energy dependence uncertainty is not factored in. Consequently in this case, the contributing uncertainties are the uncertainties from the angular dependence, reproducibility, and the calibration factor.

Table 9 The combined uncertainty for the MOSFETs in the practice audit run.

Type A uncertainties	Contributing uncertainty	Type B uncertainties	Contributing uncertainty
Reproducibility	1.15%	Interpolation of angles	0.5%
Calibration factor	1.1%		
Angular Dependence 6 MV	0.33%		
Combined uncertainty	1.7%		
Expanded uncertainty	3.4%		

These uncertainties are within the target of ± 5% that the *ICRU Report 62* [21] recommends for the reporting of doses in radiotherapy audits. This means that the MOSFET dosimetry system can be employed for the audit, and any doses outside this recommended range will be detected.

Including both the reproducibility uncertainty and the calibration uncertainty could be considered to be taking the same uncertainty into account twice because both are direct functions of the standard deviation in the MOSFETs measuring a dose under the same conditions. Both of these uncertainties were included to eliminate the possibility of underestimating the combined uncertainty of the MOSFETs.

2.7. Summary and discussion

The MOSFETs were characterised with both ^{60}Co source and 6 MV and 18 MV photons with the linac. To do this the MOSFETs were placed in a water phantom in a depth of 5 cm with the ^{60}Co source, and a depth of 10 cm for the measurements with the linac. The machine and beam outputs were first measured with the NRL's ionisation chamber whose calibration factor can be traced back to a primary standards laboratory.

The MOSFET characteristics that were tested were the reproducibility of response, stability of calibration factors, angular dependence, linearity, and energy dependence. Most of these parameters were tested with the ^{60}Co source, and some parameters should be tested more thoroughly in future work. For example, the linearity at low doses of the MOSFETs should be further characterised and more information about how the angular dependence varies with dose rate and photon energy is also needed.

It was found that the reproducibility of the MOSFETs was within about 2% on standard sensitivity, and it was decided that any phantom measurements should be performed at least three times to obtain accurate results.

The calibration of the factors did not stay stable with large increases of effective dose on the MOSFETs. It is recommended that new calibration factors should be obtained after 3000 mV of effective dose. This is because after this effective dose, the change in response was approximately 2%.

The angular dependence was found to be significant, the drop in response with the MOSFETs at 90 degrees to the beam was found to be approximately 6% with the ^{60}Co source and 3.3% and 2.8% with 6 MV and 18 MV photons respectively.

The linearity of MOSFET detectors was found to be highly linear with dose for both high sensitivity and standard sensitivity measurements. The energy dependence was also characterised. It is recommended that the MOSFETs used for future audit runs should be first calibrated with 6 MV photons and the drop in response with the MOSFET at 90 degrees for each individual MOSFETs should also be determined so more accurate results can be obtained for the future audit runs. No fading was observed in the MOSFETs.

The uncertainty of the MOSFETs was calculated according to the ISO guide, and included a coverage factor of 2 to give a confidence level of approximately 95%. Each MOSFET could potentially have its own uncertainty calculated individually, but in the uncertainty section of this chapter, general uncertainties demonstrative of all the MOSFETs‘ uncertainties were calculated. The main sources of uncertainties were the contributing uncertainties from the reproducibility and the calibration factor. The

calculated combined uncertainty with a coverage factor of two was calculated to be 4.0% and 3.4% for the preliminary phantom measurements and the practice audit runs respectively.

3. Treatment planning and audit protocol

3.1. Rationale

This chapter details the process of determining the treatment plan(s) to be used for the audit. The processes that the phantom will go through for collecting pertinent information at each centre are also addressed. As well as the planning, this chapter also discusses the formation of the audit process, and the creation of the audit protocol (Appendix A). The protocol contains detailed information and instructions to inform the centres of the individual steps of the audit process. The protocol also informs the centres of what they are required to do to participate in the audit. A number of diagrams have been repeated in this chapter and the subsequent one for ease of reference.

Determining a suitable procedure for the audit was the most crucial aspect of this project. Consequently a considerable proportion of the work for this project was spent on this task. The following variables had to be decided upon; the kind of audit employed (see section 1.2.1), the parameters to be measured, the doses (magnitudes and modalities etc) to be delivered, and the manner in which data was to be collected and analysed. The audit had to be practical and an acceptable burden on the busy centres, but it also had to collect enough information to be useful. It was decided that it would be best to keep the audit simple, at least for the first run since it could always be expanded to encompass more complicated parameters in the future.

Developing the procedure for the audit was not a trivial task. It was decided relatively early in the project that the centres would carry out specified treatment plans, since this allows a dose intercomparison between the centres. The alternative was that the

centres created their own plan(s), given a specified diagnosis and dose to a volume to the 'patient'.

Whether the CT data of the phantom would be given to the centres or the centres would collect it themselves required much thought. This is because the possible discrepancies in the CT to density conversion tables in the respective TPSs between the centres could introduce more errors in the final dose. Advantages and disadvantages of including this possible source of error were considered. The advantage was that if the CT data was supplied, it could have been easier on the centres as a CT appointment would not have needed to be made. One disadvantage was that if this contributed an error which affected patient doses, then it would not be found by the audit. Another disadvantage was possible import problems of the CT data into the various TPSs.

Plans also had to be designed to establish under which conditions doses could be measured with the MOSFETs. For example, could doses be measured accurately in the lung in a lung insert, with a different density to that of tissue [40]? A treatment plan had to be designed and executed in order to determine this. This is detailed in section 3.2.1 with the results shown in section 4.1.2.

3.2. Determining the treatment plan

There were two options for the method considered for the audit. One option was to provide the treatment plan and the dose, while the other was to provide the dose only and leave the specific treatment plan parameters up to the centres. It was decided relatively early on that the treatment plans that the centres would use to treat the phantom would be specified. This decision was made so a direct comparison of the doses could be made between the centres, whereas if the plans were designed by

each centre, they would all be different, consequently making a direct intercomparison impossible. While deciding on plans various ideas were discussed, all of which had some advantages and disadvantages. Some involved the inclusion of more complicated beam parameters. Using more complex beam parameters would have tested the capabilities of the TPSs to a greater extent, but would have significantly increased the work load of the centres, as well as introducing more sources of possible errors in the overall doses administered to the phantom.

The first step was to determine a representative tumour volume in the CIRS phantom. The size and location of the tumour volume was restricted by the inserts of the phantom. The treatment volumes had to be the last three centimetres of the cylindrical insert since this is where the MOSFET detectors could be placed within the phantom (see Figures 1.2 and 1.3). Since the tumour volume was specified to be the last three centimetres of the insert, the treatment volume extended to 1.5 cm either side of where the MOSFETs could be placed (Figure 3.1).

Before any on the treatment plans could be finalised, it was necessary to determine whether the dose to the lung could be measured accurately given that the density of the lung material, both in real people and the phantom, is much smaller (0.25 to 0.37 g/cm^{-3} [41]) than that of water. This is because the charged particle equilibrium conditions also differ in this lower density material. This is further outlined in section 3.2.1. The treatment plans were all created using the treatment planning system at Christchurch hospital, which is the CMS XIO Treatment planning system (CMS, St Louis, USA).

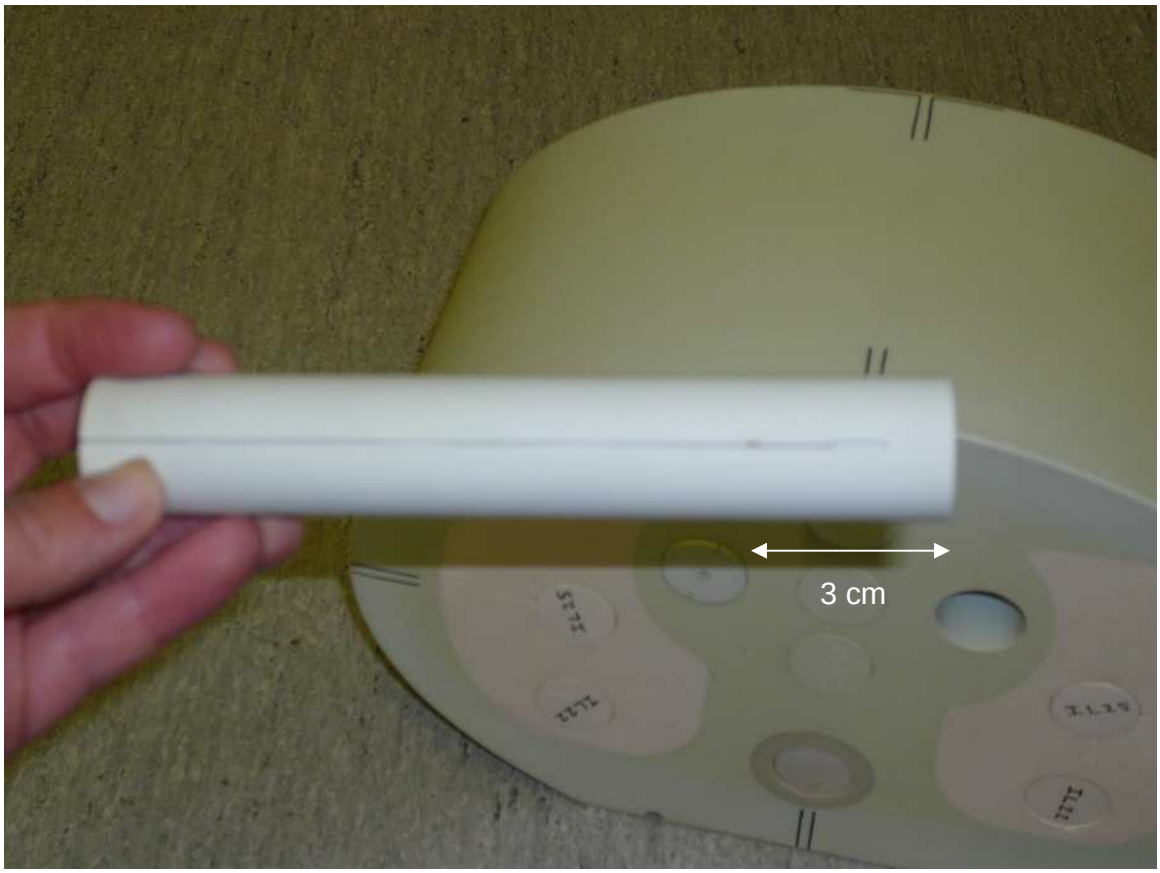

Figure 3.1 The end 3 cm of the insert as shown was chosen to be the tumour volume. Note that only half of the chest phantom is shown here.

3.2.1 Preliminary lung plan

For standard dosimetry measurements in radiotherapy, a certain set-up with defined conditions is used. It was not known how the differing charged particle equilibrium conditions would differ in the lung tissue since the charged particle equilibrium could occur at different depths due to the different density of the lung tissue. Thus, a plan was designed to treat a volume in the lung. Since the preliminary lung treatment plan was only intended to determine whether the MOSFETs could accurately measure doses in this lower density medium, the plan was straightforward. It consisted of two parallel opposed beams using 6 MV photons without wedges or collimators with field sizes of 4cm by 4cm, giving a total dose of 2 Gy to the lower insert in the lung (see

Figure 3.2). Two plans were investigated, one with the lung insert and one with the tissue insert as the tumour volume.

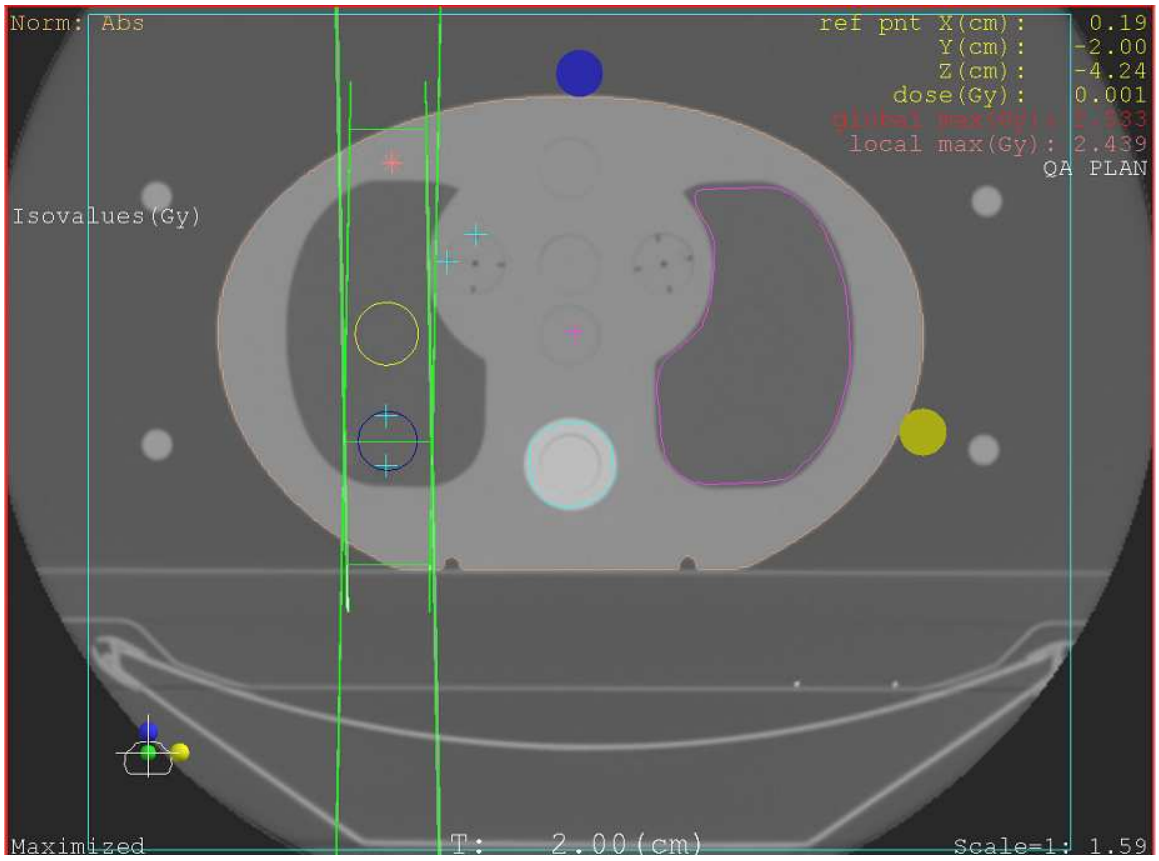

Figure 3.2 The preliminary lung plan. The blue crosses mark the interest points where the doses were measured and compared with the TPS values.

Since there was no cavity in the centre of the lung insert for the MOSFETs to be placed, the dose had to be measured at the interest points at the edge of the tumour volume (see Figure 3.2). The plan was designed such that a uniform dose could be delivered to the tumour volume.

Measurements with the lung insert in place showed that the MOSFETs measured the dose accurately. The measured doses obtained in the treatment volumes were within 2% of the Christchurch's XIO TPS, with the TPS giving 1.96 Gy and 2.07 Gy to near the top and bottom of the tumour volume respectively. Since the results were within 2%, they were well within the uncertainty of the MOSFETs, which is 3%. This plan was

carried out three times to acquire accurate dose information from the MOSFETs. Based on these results, it was decided that a lung plan could be considered for the audit.

TPSs can often have difficulty calculating dose where there are sharp density changes [42]. It was unknown whether charged particle equilibrium existed, consequently a very similar plan to that of the preliminary lung plan was carried out with the only alteration being the use of the tissue insert in the lung instead of the lung insert as the treatment volume (see Figure 3.3). With the use of the tissue insert instead of the lung insert, the doses were 2% and 4% higher for the interest point doses at the top and bottom of the tumour volume respectively.

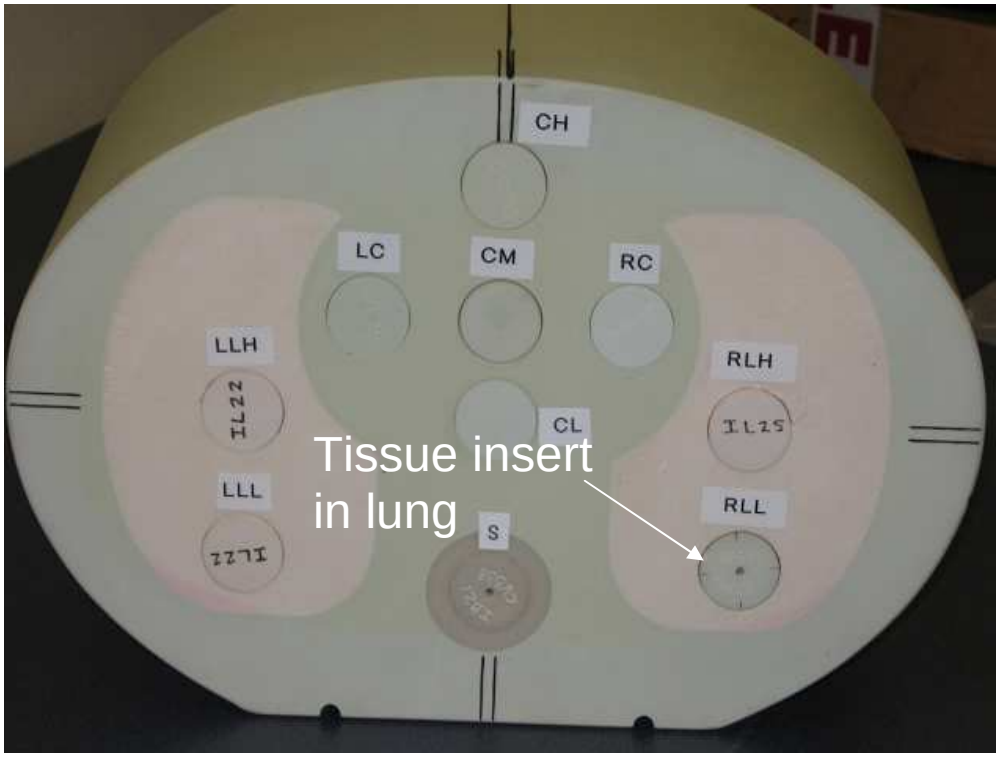

Figure 3.3 The Tissue insert was placed in the lung to compare the values with the lung insert

The use of the tissue insert in the lung could potentially mimic more accurately a typical tumour in the lung than the lung insert would, as tumours quite often have a

density close to that of tissue [43]. Though sometimes the tumours can be strewn out (disseminated) with no significantly sizable mass with a density close to that of tissue [44].

Both of these preliminary lung plans gave excellent results (the results from the parallel opposed lung plan are detailed in section 4.1.2.), demonstrating that the MOSFET detectors can be used in the lung with both the lung and the tissue insert, which means that the use of a lung plan for the audit using the MOSFET detectors was a possible option.

3.2.2 The mediastinum plan

The mediastinum plan was originally part of the preliminary phantom measurements to help decide what treatment plan(s) would be used for the audit. It was named the mediastinum plan as it delivered a dose to the centre of the chest phantom, which in a real person would be called the mediastinum. It was decided after the preliminary phantom measurements that this plan would be included in the audit. It was also elected that one of the plans should be kept relatively simple, with no wedges or complicated beam parameters, and also avoiding any sharp density changes around the treatment volume. The mediastinum plan had all these desired qualities.

The mediastinum plan was designed to deliver a dose of 2 Gy to the centre of the central insert in the phantom using parallel opposed beams going posteriorly and anteriorly with a 5cm by 5cm field with 6 MV photons (Figure 3.4). The beam at gantry angle of 0 degrees (going anteriorly to posteriorly) had a dose weighting of 0.45, whilst the beam at 180 degrees (posteriorly to anteriorly) had a dose weighting of 0.55 relative to the dose prescription point. The mediastinum plan with the isodose lines is shown in Figure 3.5.

Figure 3.4 The mediastinum plan, using 6 MV photons and a 5 cm by 5 cm to the central phantom insert delivering a dose of 2 Gy.

As mentioned previously, a simple treatment plan has advantages. This is because if there are any discrepancies between the TPS dose and the measured dose, then there would be fewer possible sources of error from which such discrepancies might occur. Consequently, any potential sources of error would be easier to isolate.

The main complicating factor in this plan was the different density of the spinal cord through which one of the two beams had to pass. The results from these plans and the practice audit runs are in Chapter 4. Note that the mediastinum plan was originally performed at the same time as the preliminary lung plan, and the mediastinum plan was not altered for the audit. However, the MOSFET detector positions, and thus the acquired interest point doses, were different in the first run of this plan than they were in the final audit protocol. This is because in the preliminary

phantom measurements the purpose was to ensure that the equipment was capable of performing in these conditions, and the point doses were not chosen for the maximum efficacy of gathering information for the audit, but rather to ensure that the equipment was working as it should.

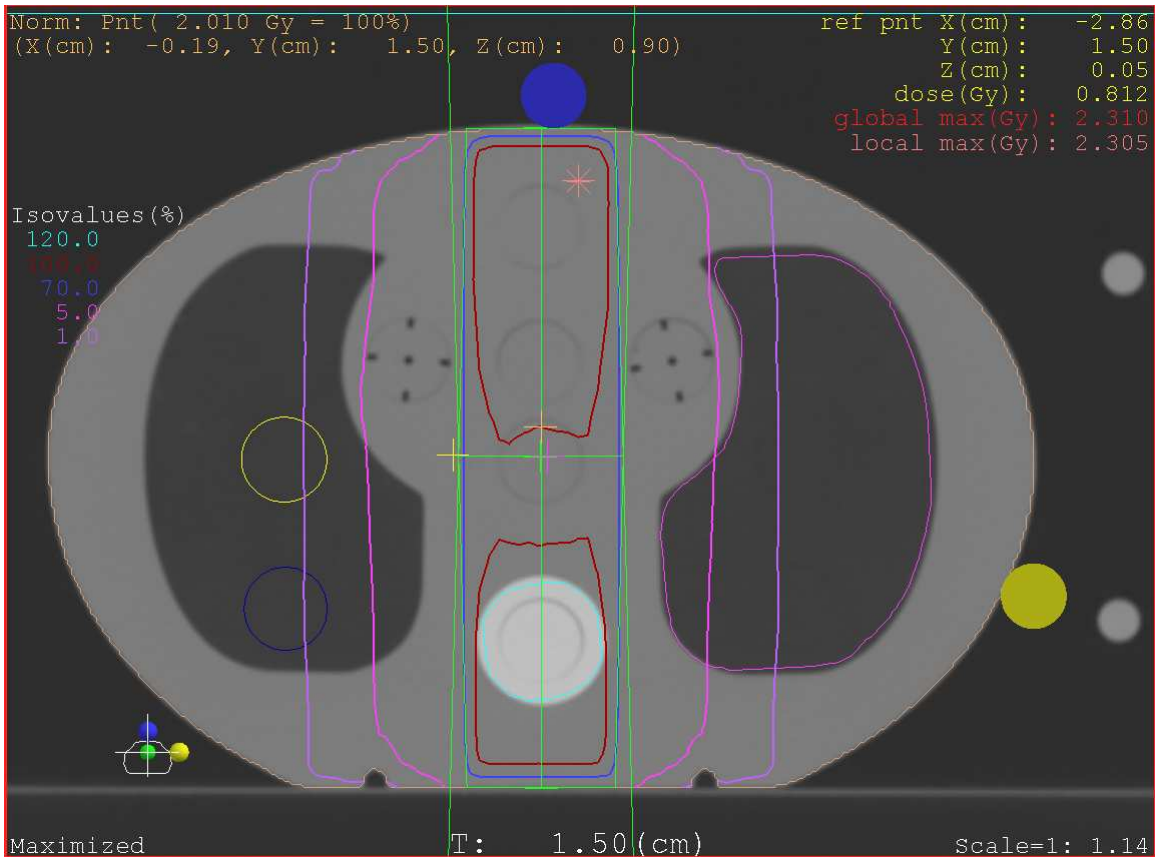

Figure 3.5 Mediastinum plan with isodose lines

For this initial run of the mediastinum plan all the measured doses were in agreement with the TPSs calculated doses of which all were within the uncertainty of the MOSFETs (this is detailed in section 4.1.1). The Dose Volume Histogram (DVH) of the mediastinum plan is included for with the results in Chapter 4.

The DVHs of the various treatment plans are included as they enable the comparison of different treatment plans' dose distributions. The dose distribution is important for patient outcomes as the percentage of a tumour to receive a particular dose impacts tumour control. It is also important to calculate the percentage of a particular organ

receiving a certain dose, as this is a factor in organ toxicity. The dose grid size for all the presented DVHs is 0.3 cm in three dimensions.

3.3. Clinical lung plan

The drawback of having a simple plan was that it did not test any of the more complex dose calculations of the TPSs, for example, dealing with the lung to tissue density changes, or the use of wedges. Thus it was decided that a more complex plan, closer to a real clinical plan, would also be included in the audit.

To construct a good lung plan there are many factors to consider, because although it is important to get the full prescribed dose to the tumour, it is also important to do as little damage as possible to the surrounding tissue. In this case the organs at risk include the lung where the treatment volume lies, the contra-lateral lung, and the spinal cord. Increased dose to the surrounding lung can have adverse side effects such as scarring. While, an increased dose to the spinal cord can be very serious, in extreme cases paralysis may occur [45]. With these factors in mind, MOSFET detectors for this plan were placed in the contra-lateral lung, the spinal cord, and the mediastinum. The DVH of this plan is shown in Figure 4.6 and an image of the plan including isodose lines is shown in Figure 3.7.

This plan was designed to test the TPS's ability to calculate the absorbed dose in the lung along with the use of wedges. As shown in Figure 3.6, one of the beams exits through the spinal cord. It was decided that MOSFETs would be placed in the spinal cord to compare the measured doses with those of the TPS's. The dose to the spinal cord is important to consider while creating a radiotherapy treatment plan as spinal cord toxicity can occur [45, 46]. This was decided to be a relevant part of the audit and the interest point doses in the spinal cord were collected for both treatment plans used

in the audit. The exact location of the interest points for the clinical are demonstrated in Figure 3.6

It was chosen to have a lung plan that would resemble those typically used in clinical cases. This treatment plan consisted of two beams, which both used 60 degree wedges at gantry angles of 200 and 280 degrees and, like the mediastinum plan, uses 6 MV photons. These beams had a relative weighting of 0.48 and 0.55 to the dose prescription point respectively. After consulting a clinical oncologist, Dr. Nik Nedev from Palmerston North Hospital, the proposed plan was slightly modified by increasing the field sizes from 4 cm by 4 cm to 5 cm by 5 cm which was considered to be more realistic.

Figure 3.6 Clinical lung plan, using 5 by 5 cm fields with 60 degree wedges and 6 MV photons

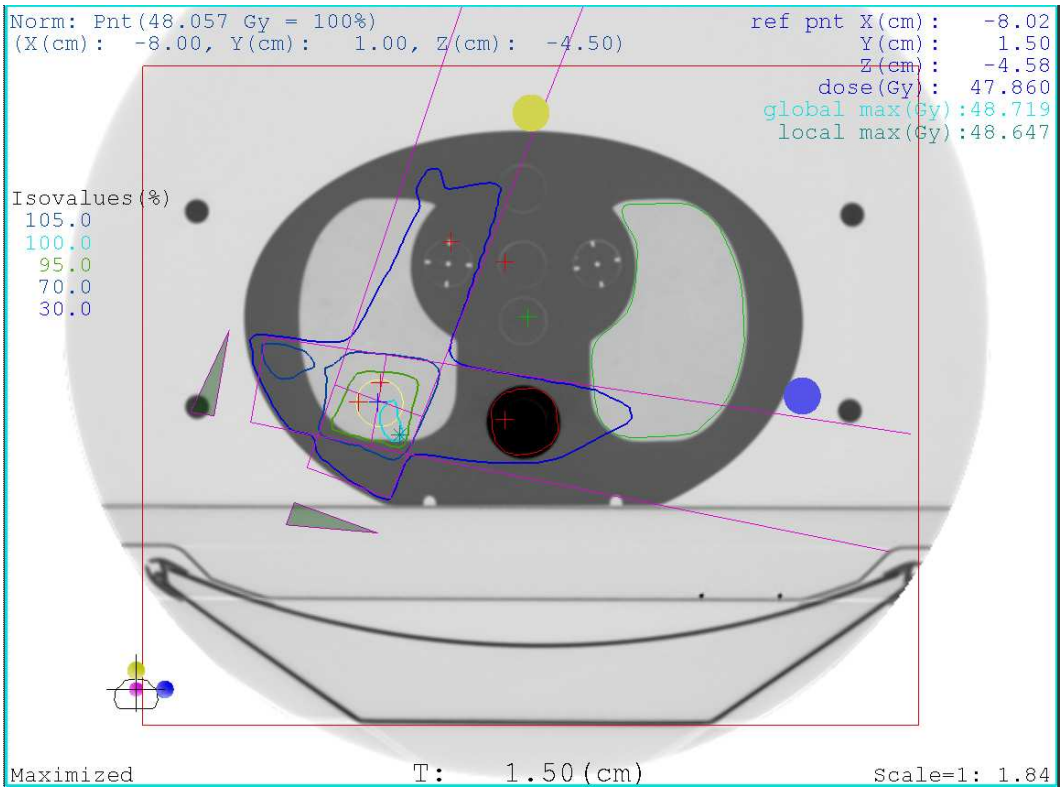

Figure 3.7 The clinical lung plan with isodose lines

3.4. Point doses and contouring

The two plans that would be used for the audit were the parallel opposed mediastinum plan (also referred to as the simple plan) and the clinical lung plan. The decision of where interest point doses were going to be measured within the phantom (where the MOSFETs were going to measure the dose) remained to be made. After some cursory investigation it was decided that there would be two MOSFETs measuring the dose in the tumour volume for each of the plans. This was because it was desired to establish an accurate measured dose in the tumour volumes [47] since the dose to the tumour volume was considered to be the dose of the most importance.

Both plans also included measured doses in the spinal cord, as well as out of the primary beams. These MOSFETs were there to make sure that the calculated dose to the spinal cord was correct. MOSFET detectors were also placed within the

primary beams in places other than the spinal cord and the tumour volumes. This was done to check that the exit and entry doses were calculated accurately by the TPS. The MOSFETs placed outside of the primary beams were there to measure the dose from the scatter of the beams. This was done since some of the treatment planning algorithms underestimate scatter [10]. In total five MOSFET detectors were placed in the phantom for each plan (see Figure 3.8 Figure 3.9). Six interest point doses were collected for each plan from the TPS in the centres. This sixth interest point dose was the dose at the normalisation point which is in the centre of the tumour volume/insert[5]. A TPS interest point dose for the normalisation point could not be verified with a MOSFET detector as the detectors cannot be placed at this point in the phantom. However it was collected from the centres so a comparison of the dose to the prescription point calculated by the centres respective TPSs could be made. It should be noted that the dose at the interest points in the tumour was very close to the prescribed dose in the normalisation point because the dose distribution in the tumour volume was fairly uniform.

To be able to compare the doses of the TPS to the measured doses, the centres were asked to obtain interest point doses from their TPS for where the MOSFETs were to be placed. This was carried out while the centres were creating the treatment plans. The accuracy of the radiotherapy treatment system of each particular centre was determined by comparing the interest points from the TPS to the doses measured by the MOSFETs after irradiation of the phantom.

[5] The tumour volume was defined to be part of the insert.

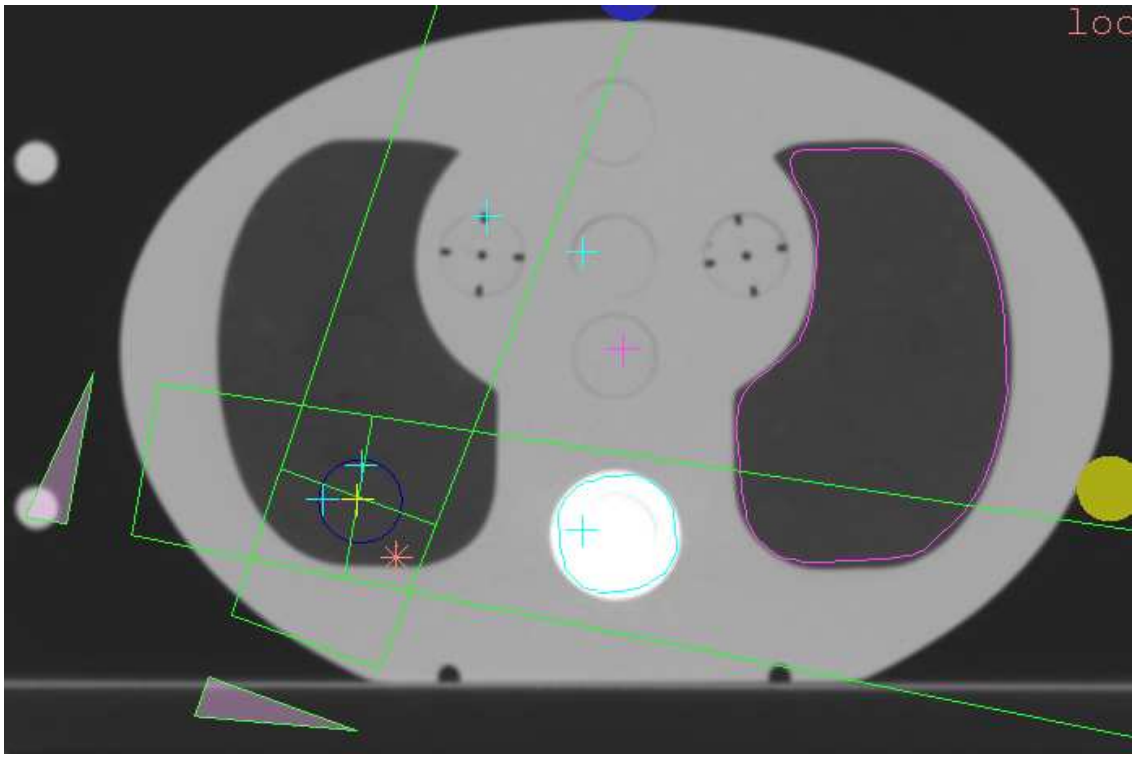

Figure 3.8 The point doses in the clinical lung plan marked by the blue crosses.

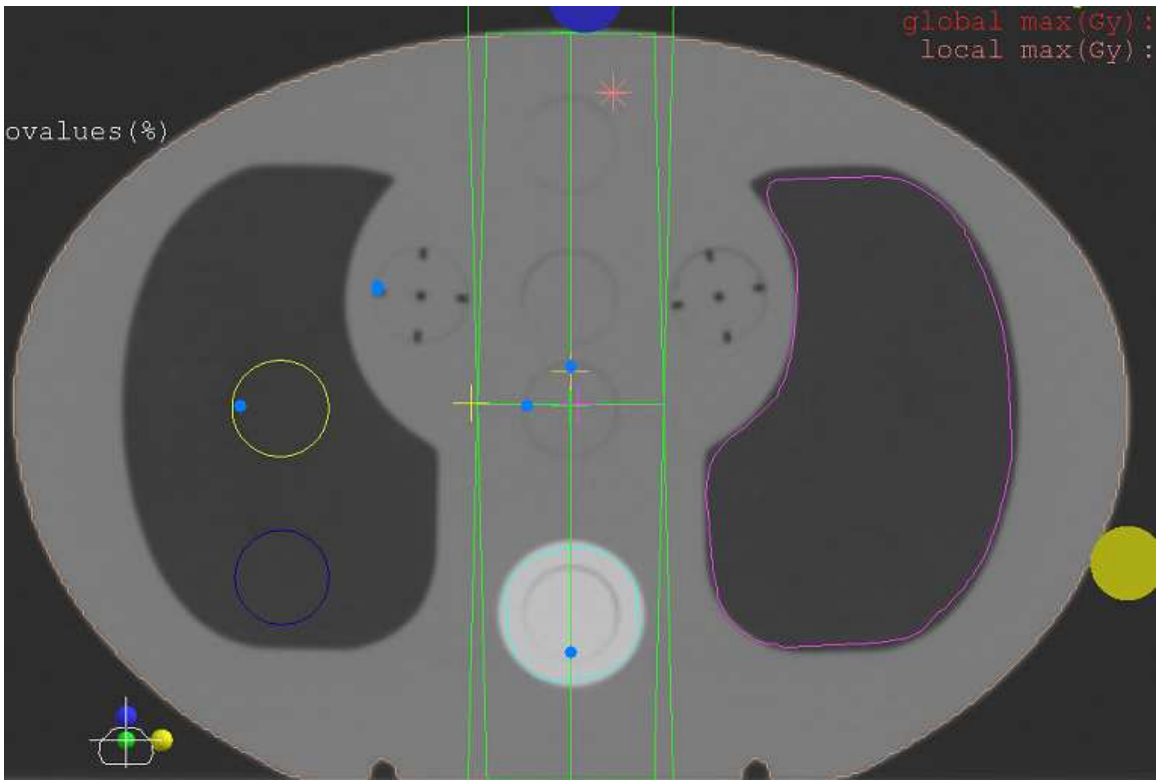

Figure 3.9 The point doses in the mediastinum plan represented by the blue dots

Each centre placed the interest points in the TPS using the co-ordinates (with the co-ordinate system specified in the audit protocol) that was provided in the audit protocol (Appendix A). The interest points were then manually adjusted by precisely placing them visually, which was easily done since the MOSFET slots in the inserts were visible on the CT images in the TPSs. These small adjustments from the co-ordinate points were necessary because some of the interest point doses were not exactly in the right locations (the right locations being in the exact centre of where the MOSFET detectors could be placed). This was possibly due to slightly different positions of the phantom during the CT scans, as it is possible that the phantom was slightly skewed. The calculated dose to the interest points was then compared to the measured results by the MOSFETs.

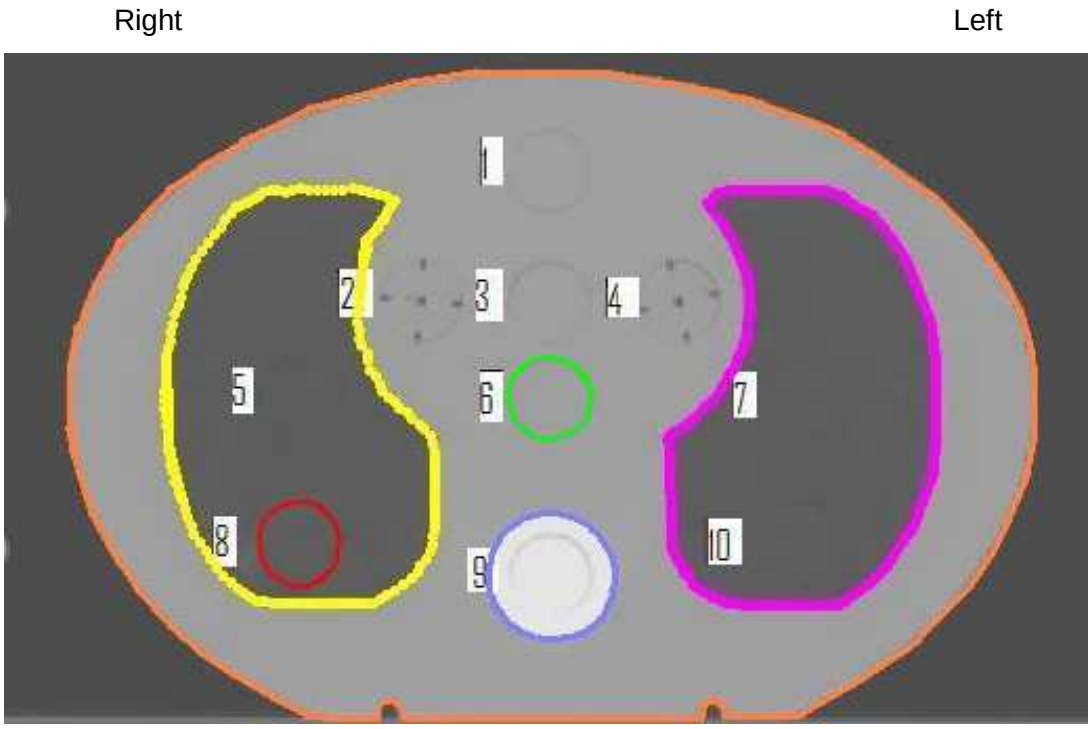

Figure 3.10 This figure shows the contouring that the centres were asked to carry out.

The centres were also asked to contour certain structures in the phantom. These structures were; the left lung, the right lung, the skin of the phantom, the treatment volumes, and the spinal cord. The contoured structures and the labelling of the inserts are shown in Figure 3.10. This information was gathered so that it was

possible to compare the DVHs. This information is important because the integral dose to the organs can have implications on patient outcome; organ toxicity can occur if too much dose is given to too greater proportion of an organ [48].

3.5. Creation of the protocol

The protocol was the list of instructions and information that the centres were to follow in order to participate in the audit. These instructions had to be clear and easy to follow so as not to be a burden on the already busy centres.

The protocol provides step by step instructions for obtaining the CT images of the phantom, creating the plans in the TPS (including interest point doses and contouring), and also the treatment of the phantom. The audit protocol was read and edited by medical physicists and was also discussed with an RT in order to ensure the usefulness of the protocol. The audit protocol was easy to follow for the participating centres and was detailed enough to gather meaningful information. The detailed protocol is given in Appendix A. The following section briefly addresses the individual steps.

3.5.1 The CT parameters

The CT parameters to be used were specified in the protocol. The protocol states that the phantom should be scanned head first in the supine position with the use of standard reconstruction filters.

The central (zero) slice was defined to be at the half way point of the phantom (Figure 3.11). This also is where the inserts end in the phantom as shown in Figure 3.11. The MOSFETs can only be placed in the + 1.5 cm slice in the phantom, because this

is the slice where the inserts have the slots for MOSFETs to be inserted. This is also the slice where all the interest points were placed, including the normalisation points and beam isocentres. The portion of the phantom imaged was 8 cm superiorly and inferiorly of the CT origin.

When the phantom was imaged, it was asked that the slice thickness would be 0.25 cm. If this was not possible the slice thickness was to be a devisor of 1.5 cm so that in the TPS there would be a slice at + 1.5 cm where the beams as well as the interest point doses could be placed. This slice also contained the centre of the tumour volumes.

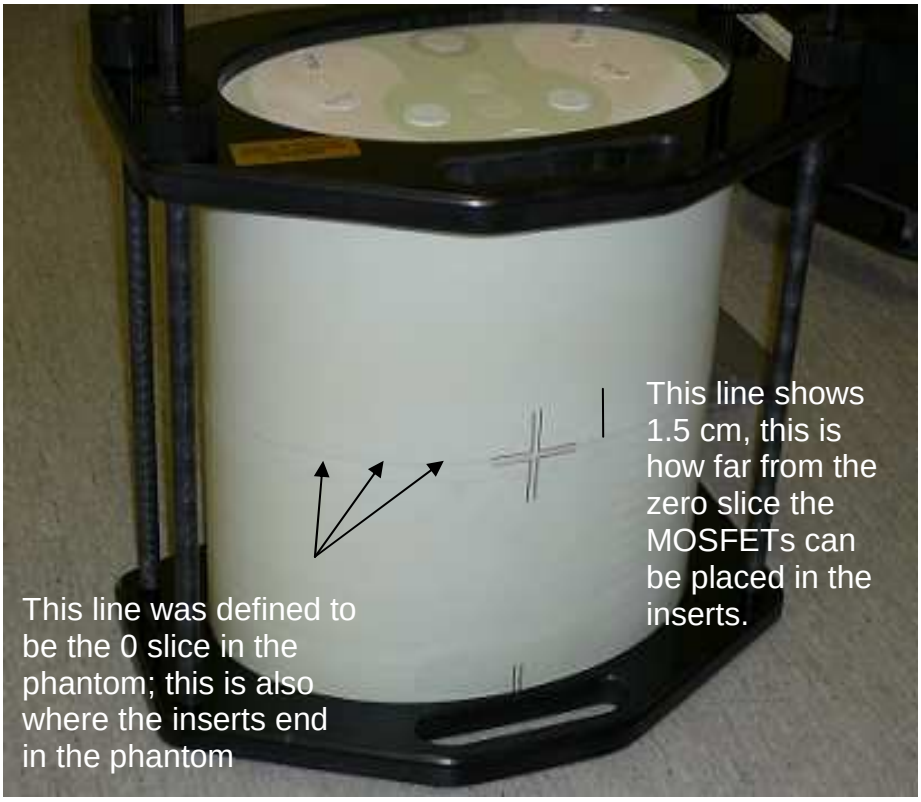

Figure 3.11 The zero slice in the CIRS chest phantom and the location of the MOSFETs

3.5.2 Treatment planning

It was requested that an RT carry out the treatment planning because they generally do planning on a day to day basis and it was thus reasoned that they were more likely

to do the planning more time efficiently, with less errors in the treatment planning parameters, and provide more accurate contouring. However, for the preliminary audit run this was not always possible, and in these cases a medical physicist carried out the planning.

For each of the plans in the audit, the reference point, normalisation point, and the isocentre, were all defined to be the same point in the phantom (each plan had a different normalisation point though, as the treatment volumes are in different parts of the phantom for the two plans). The planned dose to these points for both the plans was 2 Gy. This dose was chosen for three reasons; it was a realistic dose for a fractionated treatment, it was small enough to avoid expending the MOSFETs too quickly, and it was high enough that relatively accurate measurements could be taken with the MOSFETs.

The beam parameters were all specified in the audit, namely, no multileaf collimators, shielding, compensator or bolus were to be used. This was done so that a more effective comparison between the centres could be carried out, as the use of these techniques would add more variables to the audit, which would mean that there would also be more sources of potential error.

After the placement of the interest points in the plans, the dose (in Gy) to these points is written into the tables that are at the end of the protocol (Appendix A). The time that was taken for the treatment planning is also recorded on the same page as the tables with the interest point doses. This data was collected so the extra work load that the centres undergo to partake in the audit could be ascertained.

3.5.3 Transferring of patient data

Once the treatment planning was complete it was requested that the treatment plan data, with the contouring and the interest point doses, were exported in DICOM or RTOG format on to a CD so that the DVHs could be compared retrospectively between the centres.

The audit protocol requests that all the data be transferred between devices in the usual manner. The protocol asks that the CT information is sent to the TPS in the manner in which data is usually transferred while, more importantly, it also requests that the treatment plans are sent to the linac in the usual manner. It was decided that it was not desirable that the linacs be set up completely manually. This was so any potential sources in of error during the transit of information from the TPS to the linac could be discovered in this audit. It also eliminates potential errors from the centre personnel typing in the incorrect monitor units or gantry angle that could occur with a medical physicist or RT entering the beam parameters manually.

3.6. Discussion and summary

The audit protocol was designed to encompass the ability to test various parameters of the radiotherapy treatment process. The tumour volumes had to be decided on, the treatment plans had to be created, and decisions had to be reached concerning the location of the interest point doses and contouring. Perhaps most importantly the treatment protocol had to written in a way that all the centres would find the participation in the audit an undemanding yet clearly useful experience.

The tumour volumes were decided to be the last 3 cm of a phantom insert, as this is where the MOSFETs could be placed without alterations to the phantom. All the

treatment plans were designed to deliver a dose of 2 Gy to the centre of the tumour. This point in the tumour was also the; prescription point, normalisation point and beam isocentre, as these were all defined to be at the same point.

A treatment plan delivering to the mediastinum was created that consisted of two parallel opposed beams, and as it was simple, it would only test simple beam parameters. The possibility of including a lung plan in the audit procedure was then considered so more complex beam parameters would be tested. Any treatment plan in the lung also tests the TPSs ability to accurately calculate the dose with the inclusion of sharp patient density changes.

To determine whether it was possible to include a treatment plan to the lung it had to be discovered whether the altered conditions of charged particle equilibrium would affect the accuracy of the readings of the MOSFET detectors. This was done by means of two parallel opposed beams to a tumour volume in the lung. Once it had been confirmed that the MOSFET detectors could indeed accurately measure the dose in the lung, it was decided to investigate the use of a more complex plan which would test more complicated beam parameters.

The clinical lung plan consisted of two beams at 80 degrees to one another, with each beam using 60 degree wedges. To ensure the clinical lung plan was indeed clinical, it was sent to a clinical oncologist who recommended that the field sizes be enlarged from 4 cm to 5 cm.

The interest point doses in the audit practice run were chosen for the maximum efficacy of collecting relevant information about the delivered doses. Contouring was

also included of the tumour volumes, the right lung, the spinal cord as well as the phantom, so DVHs could potentially be compared between the centres.

Great care was taken to ensure that the audit protocol was easy to follow. To do this a variety of hospital staff was consulted, and changes were made to the protocol as they were recommended.

4. Results and discussion

The first half of this chapter gives the results and discusses the three main preliminary phantom measurements, which included the parallel opposed lung plan, carried out with both the tissue and lung insert, the mediastinum plan, which also consisted of parallel opposed beams, and the more complex lung plan. Draft plans were initially used to help select which plans were going to be used for the main audit. The clinical lung plan was also trialled at Christchurch hospital.

The second half of the chapter gives the results of the audit practice runs at two New Zealand hospitals following the audit protocol developed in this project. The only involvement that NRL and myself had during the audit practice runs was assembling the phantom and the detectors, while the hospital staff executed all the other steps. At the conclusion of this section there is a general discussion about these first practice audit runs.

4.1. Preliminary phantom measurements

In all of these preliminary phantom measurements the phantom was set up manually under the linac in the treatment 2 bunker in the Christchurch hospital's oncology department. This linac is a Varian 21iX S/N 1009 model capable of delivering 6 and 18 MV photon beam energies, with a Varis treatment console. All the measurements were corrected for angular dependence and energy dependence according to Equation 1 in section 2.6.1.

4.1.1 The mediastinum plan

This plan was a parallel opposed beam plan with the tumour volume in the centre of the phantom. The two beams are placed posteriorly and anteriorly. This plan was made and carried out to test the MOSFETs in the phantom with no complications such as wedges or density changes to affect the dose deposited. Two MOSFETs were placed in the tumour volume, one in the spinal cord tissue and two outside the beams. The interest point locations that were measured in this plan differ from those measured in the practice audit run (section 3.2.2). The interest point locations used in the first run of the phantom with the mediastinum plan are shown in Figure 4.1 and the DVHs of the plan are shown in Figure 4.2. The numbered point doses correlate with the location column in Table 10. This table also includes the percentage difference between the TPS calculated dose and the measured.

Figure 4.1 the mediastinum plan with the locations of the interest points that were measured in the first phantom run with the mediastinum plan.

Table 10 the results from the parallel opposed mediastinum plan.

MOSFET No.	location	Average of readings (mV)	Relative std dev (%)	measured dose (Gy	TPS dose (Gy)	% diff between TPS and measured
2914	1	214	1.7%	1.96	1.999	1.8%
2912	2	202.7	0.3%	2.02	2.0002	-1.1%
2931	3	202.7	0.5%	2.12	2.096	-1.0%
2933	4	11.7	13.1%	0.11	0.124	15.1%
2933	5	4	25.0%	0.037	0.010	267%

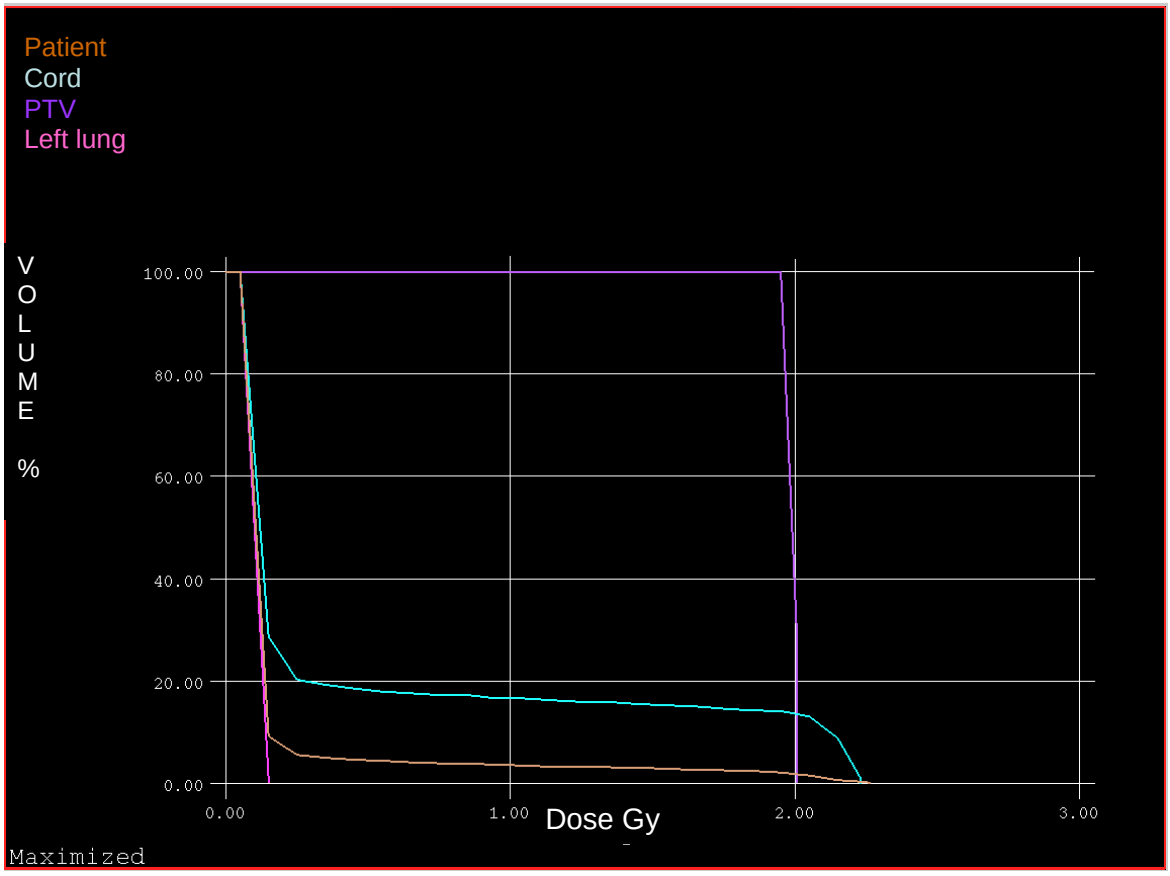

Figure 4.2 DVH of mediastinum plan.

In all of the measured points in the mediastinum plan of around 2 Gy (interest points 1-3), which include the two points in the tumour volume and in the spinal cord, there was a good correlation between the TPS calculated dose and the measured ones

where the greatest difference in absolute dose was 0.04 Gy. In places outside of the beams (points 4 and 5) the TPS doses underestimated the dose (though only by 0.01 Gy at interest point 4). In terms of absolute dose however it is small, with a dose difference of 0.03 and 0.02 Gy for points 4 and 5 respectively (see Table 10). Since the MOSFETs showed a good linear response even at low doses (though the MOSFETs are comparatively noisy), it seems that the TPS is slightly underestimating these doses. However, as the linearity measurements were performed with ^{60}Co, it is possible that the linearity of the MOSFETs is poorer with the comparatively high dose rate of the linac. Another possible cause for the large relative discrepancy between the TPS and the measurements is that some TPS algorithms do not calculate the dose beyond a certain distance from the edges of the field. TPS algorithms can have limited accuracy near changes in tissue density, such as between lung and tissue, which could also explain the discrepancy between these doses.

4.1.2 Parallel opposed lung plan

This plan was to ascertain whether it was possible for the MOSFETs to read properly in the phantom's lung tissue. It was uncertain if there would be charged particle equilibrium in the tumour volume, with the density changes from the tissue to the lung tissue, and how this would affect the measurements with the MOSFETs. Two beams placed posteriorly and anteriorly were used without wedges. This plan was also repeated with a tissue insert in the lung and the results were compared.

The numbered points which represent the locations of the interest points at which doses were measured and compared with the TPS are shown in Figure 4.3. These marked interest points correlate with the location column in Table 7. This also

compares the TPS doses with the measured doses. The DVH for this plan with the lung insert in the lung is shown in Figure 4.4.

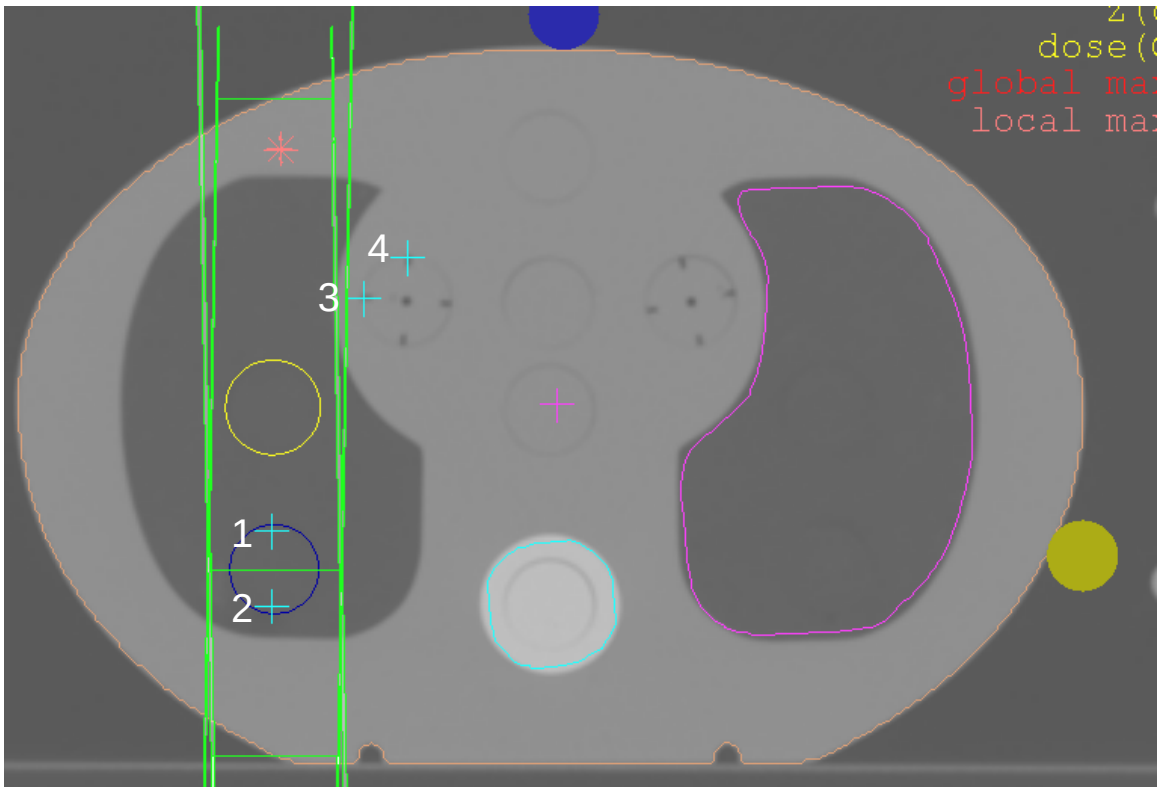

Figure 4.3 The points where the dose was measured in the parallel opposed lung beam are marked by the blue crosses and are also numbered.

Table 11 the results from the parallel opposed lung plan with the lung insert.

MOSFET No.	location	Average of readings (mV)	Relative std dev (%)	measured dose (Gy)	TPS dose (Gy)	% diff between TPS and measured
2914	1	212	1.2%	1.94	1.96	0.8%
2912	2	209	1.3%	2.09	2.07	-0.9%
2931	3	9.7	6.0%	0.090	0.064	-41.4%
2933	4	4.7	12.4%	0.043	0.012	-257.6%

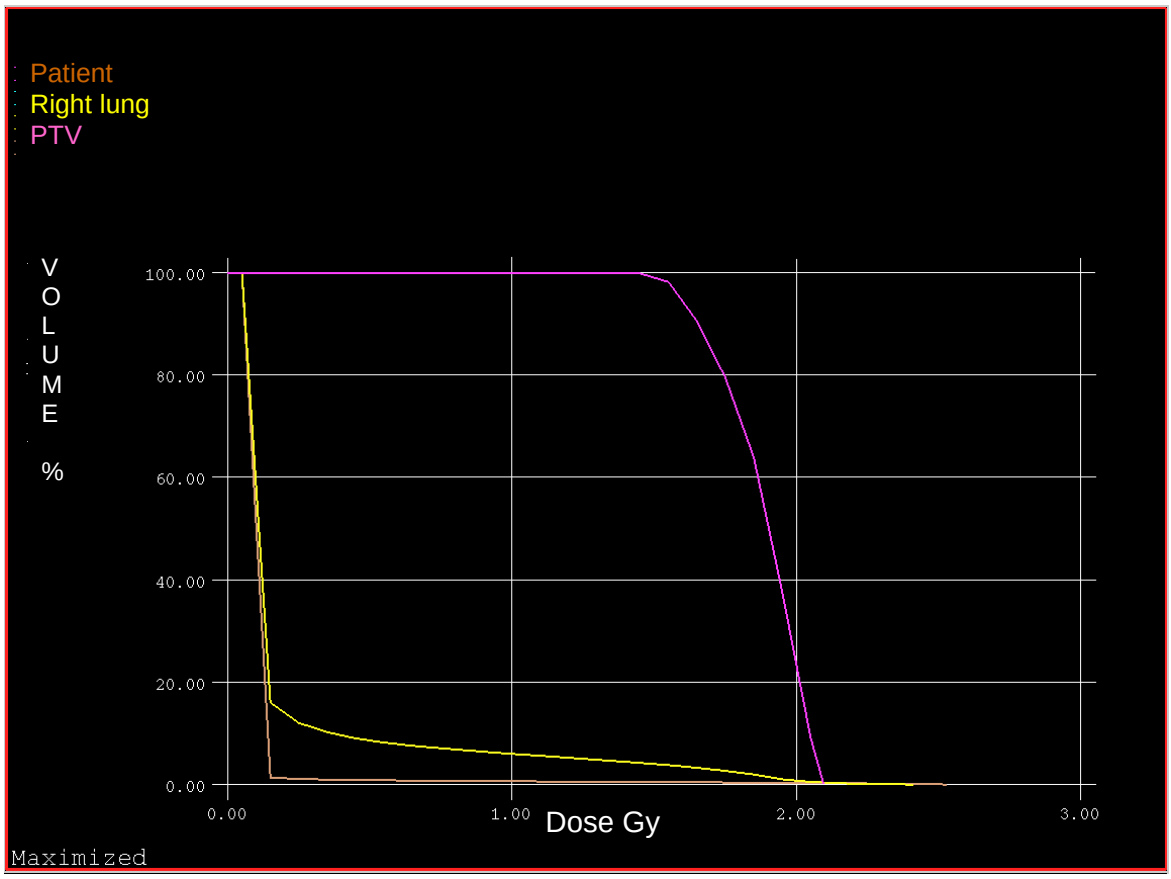

Figure 4.4 The DVH of the parallel opposed lung plan. The treatment volume in this case was PTV 1

The measured doses in the tumour volume were well within the uncertainty of the MOSFETs. The MOSFETs outside of the main beam however showed that the TPS doses outside of the main beam were relatively lower than the measured ones, suggesting that the TPS underestimated the dose from scatter.

While the doses outside the main beams were underestimated by the TPS, in terms of absolute dose, it was very little in both cases; it was less than 0.03 Gy. The MOSFETs are also less proportionally accurate at very low doses, this is what was found in the measurements performed in this project (Chapter 2) and is also reported by other researchers [49, 50].

The excellent results in the high does region obtained in this plan meant that a lung plan could be considered for the audit, as it meant that the MOSFETs could measure

the dose accurately in the lung. Once this was established, which was also outlined in chapter 3, it was possible to consider a more complex lung plan.

4.1.3 The clinical lung plan

This was the plan that was designed to test more of the parameters of the treatment planning system, and was also used for the practice audit run. Like the mediastinum plan, the measured doses were in different locations than those chosen for the audit run. The clinical lung plan was also altered slightly between the preliminary phantom measurements and the practice audit runs. The field sizes of the beams were extended from 4 cm by 4 cm to 5 cm by 5 cm. The results for this plan are shown in Table 2 and Figure 1.1 shows the interest point locations used in these measurements.

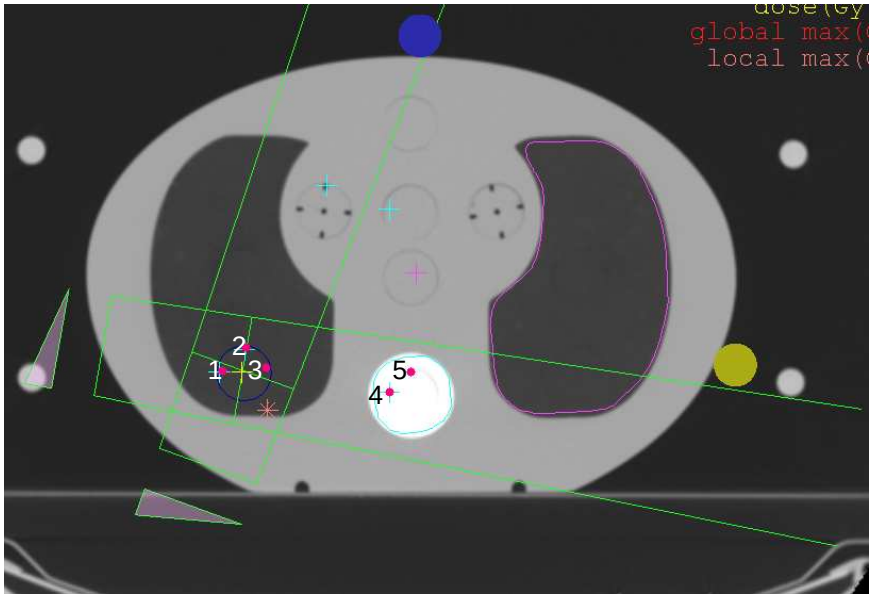

Figure 4.5 the numbered interest points where the doses were measured in the preliminary phantom measurements of the clinical lung plan. The pink dots represent the locations of the measured doses, while the blue crosses represent the point dose measured for the preliminary audit run.

Table 12 The results from the clinical lung plan with the lung insert.

MOSFET No.	location	Average of readings (mV)	Relative std dev (%)	measured dose (Gy)	TPS dose (Gy)	% diff between TPS and measured dose
2914	1	211	2.2%	2.02	1.97	-2.6%
2912	2	208	1.7%	1.99	1.96	-1.3%
2950	3	216	0.1%	2.03	2.01	-1.1%
2931	4	96	1.6%	0.86	0.92	5.9%
2933	5	91	1.7%	0.88	0.87	-0.5%

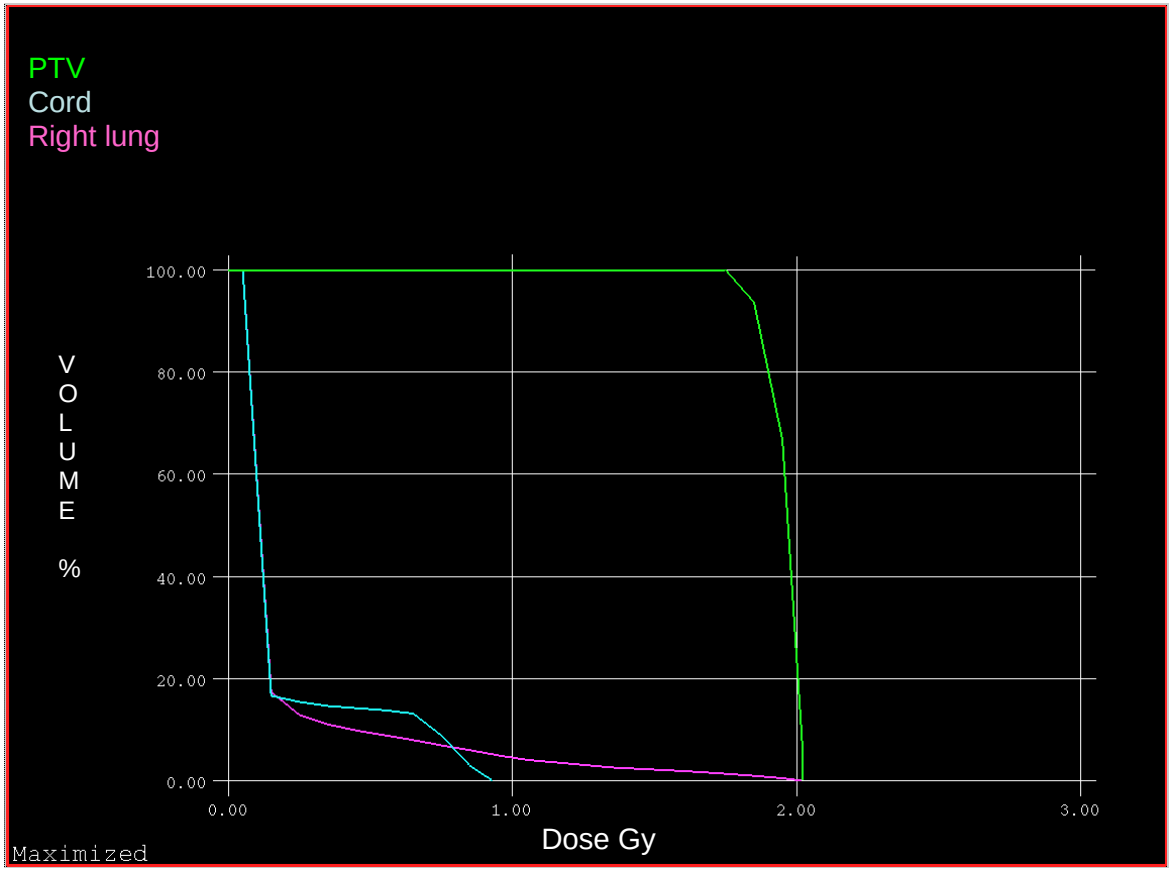

Figure 4.6 DVH of clinical lung plan.

With the exception of MOSFET location number 4, the results in the preliminary phantom measurements of the clinical lung plan also showed that the doses to the interest points were accurate within the uncertainty of the MOSFETs.

At MOSFET location number 4 for this plan, the TPS dose and the measured dose differed by 0.06 Gy. This could be due to several reasons. One possible reason is that the interest point doses in the TPS were not entered at precisely the right location. This would account for the difference as the dose gradient at this area was steep (though no steeper than at point 5), the isodose lines are shown in Figure 4.7. Another possibility is that the phantom was not set up precisely and with the steep dose gradient in this area, a small error in the position of the phantom could account for the discrepancy of the measured dose and the TPS dose.

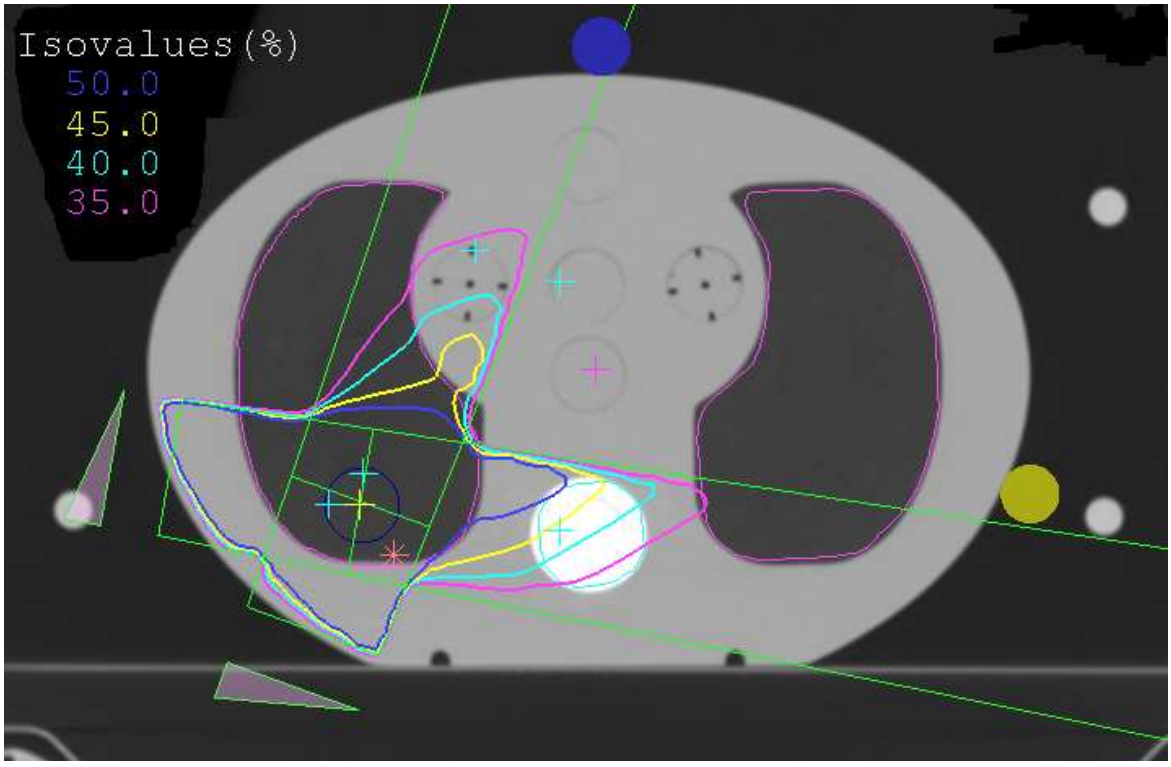

Figure 4.7 The Isodose lines for 50% of the dose to 35% of the dose, showing the dose gradient in the spinal cord.

The results in Table 12 for the preliminary phantom measurements showed that the system that had been created with the MOSFET dosimetry system, the MOSFET commissioning (with the appropriate correction factors derived from this commissioning), and the phantom, could be used with confidence for the audit especially in the high dose regions. The good results in all three of the preliminary phantom measurements showed that any large dose discrepancies could be found in

the audit in the high dose interest point locations. However, care has to be taken to ensure that the geometrical set-up is accurate before any conclusions can be drawn about any possible dose miscalculations of the TPS. The dose discrepancies in low dose interest points also need to be treated with care as more information about the MOSFET response on standard sensitivity at low doses should be collected with high dose rates. The current information and measurements on low doses with the MOSFETs means that these interest point locations can only be discussed in the most general terms [34].

4.2. The audit practice runs

Two centres took part in the practice audit runs that will be referred to as Centre A and Centre B in this thesis. This section will show the comparative interest point doses along with a discussion about the possible improvements of the both the protocol and the procedures of the audit personnel. These practice audit runs were performed in conjunction with the NRL's biennial audit of two of the centres in which they check compliance with New Zealand's radiation regulations and the machine outputs. The protocol was sent to the centres in advance so it could be read and any queries could be answered before the visit. The ease in which the audit protocol (Appendix A) could be followed was an important element of the information gathered in these practice audit runs.

The phantom and the MOSFET dosimetry system were transported without complication to both Centre A and Centre B in specially modified cases which transported the equipment unharmed.

4.2.1 Centre A

Centre A was very busy on the days of the visit as it had been experiencing some problems with the linacs (mostly due to complications from power cuts), but all of this centre's personnel were co-operative and helpful. This centre had no problems following the audit protocol. The scanning of the phantom went smoothly, with the exception of the minor detail of the slice thicknesses allowed by the CT scanner. This centre could not have a slice thickness of 0.25 cm so it was then decided to use a slice thickness of 0.3 cm as this is a devisor 1.5 cm. (This means that there would be a slice at MOSFET locations, see section 3.5.1).

The co-ordinate system that the TPS used at this centre (Pinnacle) labelled the axes differently than the co-ordinate system in the protocol, but this was quickly established and corrected for with minimal loss of time. The interest point positions were corrected manually since some of the interest points did not lie precisely over the places where the detector could be placed. The points were not systematically shifted however (they were not all off in the same direction). This prompted the revision of the interest point co-ordinates in the protocol upon return to Christchurch. It was found that the phantom had been slightly skewed in the CT scanner in Christchurch.

The initial planning at centre A took between 1 and 1½ hours and was carried out by a medical physicist. This was done directly from the instructions of the protocol. A large component of this time consisted of waiting for the TPS to calculate the doses to the interest points. It was realised at this point that the dose grid size was not specified in the protocol, highlighting the need of the audit practice runs to show any inadequacies in the audit protocol. Some further time was spent planning later on in the day, without the presence of any NRL personnel, as the linac that was booked for the audit was not the same linac as the treatment plans were originally made with.

An error was made by Centre A with their planning of the mediastinum plan. The beam at 180 degrees did not have the correction factor to account for the beam passing through the couch applied. This meant that there were not sufficient monitor units to give the prescribed dose to the tumour volume for this beam. This shows that it would be more suitable for someone who plans regularly, normally an RT, to carry out the treatment planning as someone who plans regularly is less likely to make such a mistake that gives defective results for the audit. This mistake also meant that, at first glance, the measured doses to the interest points were far too low, in one case by 7% in the high dose region of the tumour volume. This mistake did not affect the clinical lung plan as this plan did not require any of the beams to pass through the couch as the phantom had been placed with its base right to the edge of the couch.

The exporting of the plan data from the TPS on to CD after the planning was also a slow process, mostly because it took some time to realise that the full plan data could only be transported in RTOG, not DICOM-RT format as the person exporting had limited experience in this procedure.

The entire process of the treatment of the phantom, including setting up the phantom, the MOSFET detectors, and carrying out each of the two plans three times each, took slightly under two hours. The offsets were applied by the medical physicists present. The treatments went without incident, with the small exception of a faulty MOSFET bias supply box (Figure 2.2, section 2.3) which was not recognising the presence of any MOSFET detector plugged into one of its five slots. This bias supply was replaced before any irradiations of the phantom began, and all the MOSFETs were re-calibrated upon return to Christchurch with the particular bias supply box that was used for the irradiations.

The results from the mediastinum plan

Table 13 shows the comparison of the TPS doses and the measured doses for Centre A for the mediastinum plan. These results do not give much constructive value as the correction factor of the couch was not applied in the treatment planning. The point doses are shown in Figure 4.8.

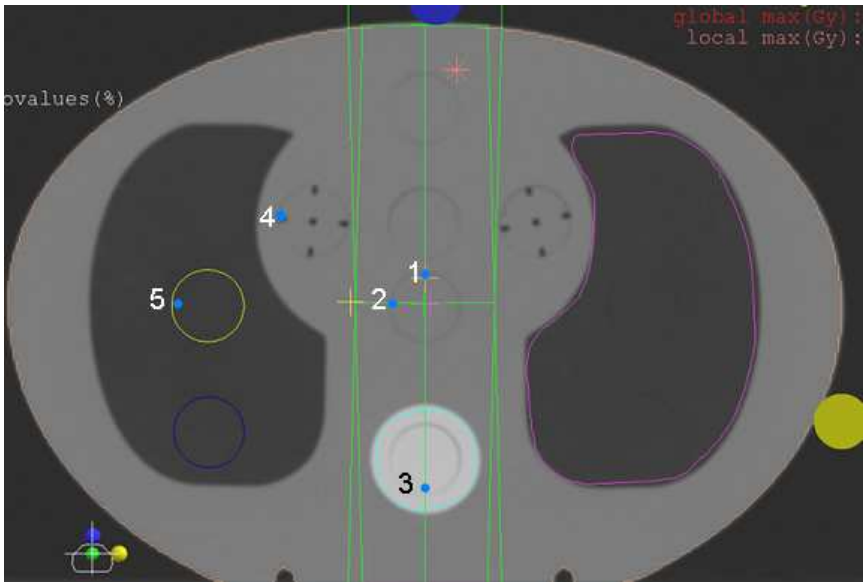

Figure 4.8 The location of the interest points in the mediastinum plan

Table 13 The results from the mediastinum plan at centre A. The percentage differences shown were calculated from the unrounded values.

Mosfet No.	location	Average of readings (mV)	Relative std dev (%)	measured dose (Gy)	TPS dose (Gy)	% diff between TPS and measured dose	% diff between TPS and measured dose with estimated couch correction factor applied
2947	1	204.5	2.5%	1.85	1.99	6.9%	5.8%
2950	2	213.8	1.7%	1.92	1.99	3.3%	2.2%
2948	3	249.8	1.9%	2.21	2.28	3.1%	1.6%
2933	4	5.8	8.7%	0.055	0.09	38%	--
2931	5	1.3	76.6%	0.01	0.03	70%	--

An estimated couch correction factor of 2% was applied for the beam passing through the couch in later analysis. This showed that there was still a slight under dosing, though the MOSFETs at interest point locations two and three were within the

uncertainty of the MOSFETs. The doses outside the main beam remained uncorrected for the couch because, in terms of absolute dose, the discrepancies were very small.

The result from the clinical lung plan

The measured values from the clinical lung plan correlated well with the TPS values at Centre A, as shown in Table 14, with the doses to the tumour volume well within the uncertainty of the MOSFETs. In this plan at Centre A none of the beams passed through the couch to get to the phantom as the phantom was placed with the base at the very edge of the couch. The interest point locations are shown in Figure 4.9.

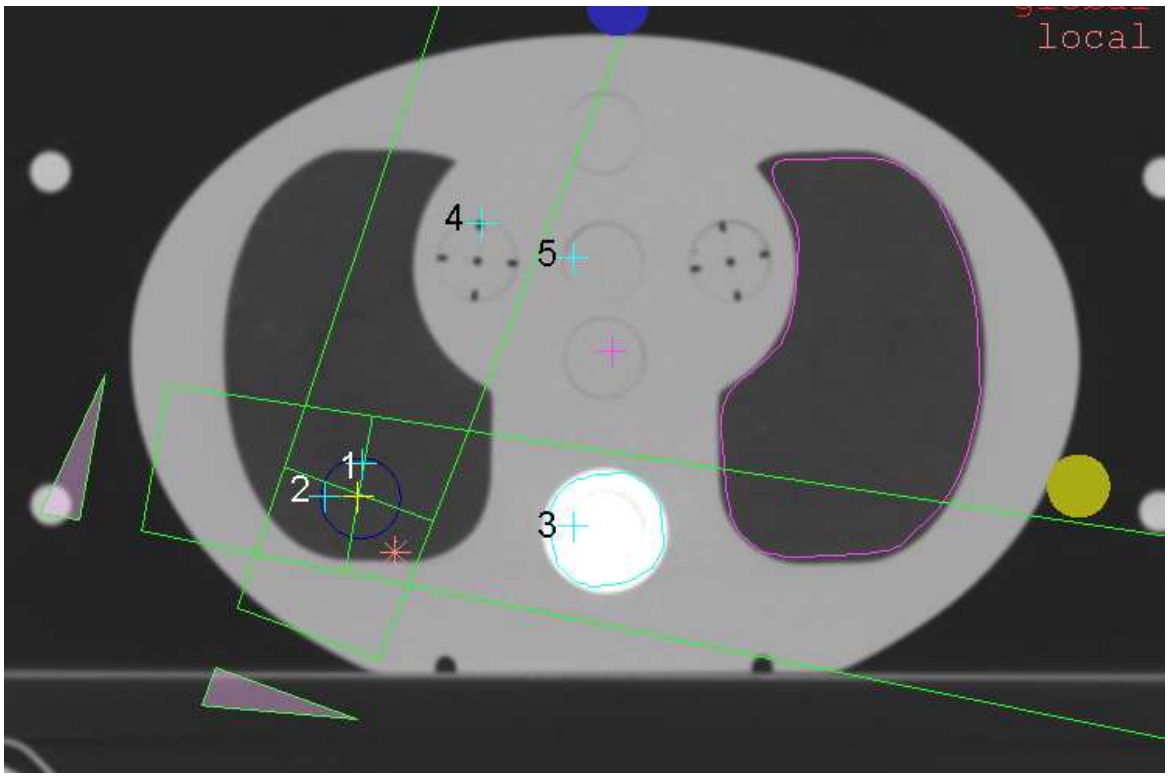

Figure 4.9 The locations of the interest points in the clinical lung plan

Table 14 The results from the clinical lung plan at centre A. The percentage differences shown were calculated from the unrounded values.

Mosfet No.	location	Average of readings (mV)	Relative std dev (%)	measured dose (Gy)	TPS dose (Gy)	% diff between TPS and measured dose
2947	1	222.7	1.4%	2.02	1.97	2.3%
2950	2	219.0	0.9%	1.99	1.98	0.6%
2948	3	97.3	1.6%	0.89	0.86	3.6%
2933	4	74.0	1.4%	0.68	0.72	-5.8%
2931	5	5.3	21.7%	0.048	0.05	-3.4%

4.2.2 The results from centre B

Unlike Centre A, Centre B had a comparatively relaxed atmosphere with a seeming abundance of staff. An RT had been delegated by the head of the medical physics department at this centre to do the treatment planning, but for reasons unknown a medical physicist carried out the planning instead. The physicist did not want to read the protocol, but instead preferred verbal instructions. This was unfortunate because the intention was to see how self-explanatory the audit protocol was. The original planning went quickly and was originally completed in under an hour. Some of the interest points also had to be corrected manually at this centre too.

Upon the first irradiation of the phantom, which was the first irradiation of the mediastinum plan, it was found that there was something not quite right with the delivered doses. The dose delivered was less than ½ of the prescribed dose. After some confusion, it was realised that medical physicist that did the planning did not prescribe the dose to the reference point in the tumour volume. This error did not take long to fix in the treatment planning system, but it did take about half an hour to import the modified plan to the linac.

Once this had been completed, and the first set of good measurements had been taken with the mediastinum plan, the next plan (the clinical lung plan) was immediately loaded onto the linac. As the physicists present had not read the protocol they did not realise that each plan had to be completed three times.

There were more delays trying to get the mediastinum plan re-loaded into the machine. Eventually it was decided to enter the beam parameters manually into the linac. There were also some delays doing this, as the physicist entering the monitor units switched the monitor units for the two beams in one of the irradiations. This was realised upon reading the MOSFETs after this irradiation, so this plan had to be performed yet again.

The clinical lung plan was executed three times without any problems. This centre used physical instead of virtual wedges as they thought that the word 'wedges', without explicitly saying virtual wedges, implied that physical wedges should be used. The use of physical wedges did take longer since at every beam change the wedges had to be moved manually.

At some point in time well into the treatments of the phantom, it was realised by one of the medical physicists that the couch correction factor had not been applied in the treatment planning, which was why the doses recorded were much lower than the TPS doses. Due to the wider couch used at Centre A than Centre B, this affected both the mediastinum plan and the clinical lung plan. This further highlights that ideally an RT would perform the treatment planning for the audit, especially as the same mistake occurred at both centres with the medical physicists performing the planning. Since this error was realised before the final irradiation of the phantom, the beam parameters with and without the couch corrections were printed out so the

corrections could be applied in later analysis to the results that were obtained at this centre. With the setup and delays it took around three hours to treat the phantom.

Centre B, while more relaxed than Centre A, did not seem to take the audit as seriously and thus there were a lot of errors, mostly due to their unwillingness to read the protocol. It also highlights the fact that if the phantom were treated in the same way as a patient is treated, these mistakes would be less likely to occur, and if they did, a more meaningful audit could take place as this would draw attention to any errors in the treatment *system* of that particular centre.

The results from the mediastinum plan

These results were corrected for the couch factor after irradiations, which is why the TPS doses to the tumour volume are less than those in the preliminary phantom measurements and the TPS doses from Centre A. All the measured values matched very well with the corrected TPS values as shown in Table 15. It is worth noting here that the results from the MOSFETs outside the main beam (locations 4 and 5) were as accurate as the MOSFETs could be considering that the MOSFET reader does not show decimal places and the irradiations were repeated three times only.

Table 15 The results from the mediastinum plan at Centre B. The percentage differences shown were calculated from the unrounded values.

Mosfet No.	location	Average of readings (mV)	%std dev	measured dose (Gy)	TPS dose (Gy)	% diff between TPS and measured dose
2947	1	213.7	0.7%	1.97	1.96869	0.2%
2950	2	214.7	1.4%	1.96	1.9795	1.1%
2948	3	256.7	2.3%	2.28	2.29087	0.4%
2933	4	6.3	9.1%	0.06	0.04945	-17%
2931	5	1.7	34%	0.014	0.016	11%

The results from the clinical lung plan

The measured doses and the TPS doses were significantly different in the tumour volume since the difference between the TPS dose and the measured dose was larger than the uncertainty of the MOSFETs as shown in Table 16. The lack of the couch correction factor in the treatment planning was corrected for after irradiation.

Table 16 The results from the clinical lung plan at Centre B. The percentage differences shown were calculated from the unrounded values.

Mosfet No.	location	Average of readings (mV)	%std dev	measured dose (Gy)	TPS dose (Gy)	% diff between TPS and measured dose
2947	1	204.3	3.5%	1.87	2.00	6.3%
2950	2	204.3	1.7%	1.87	1.93	3.1%
2948	3	91.0	1.1%	0.85	0.83	-2.8%
2933	4	71.7	3.5%	0.66	0.69	3.9%
2931	5	5.0	0.0%	0.045	0.05	17.0%

It is not believed that the dose discrepancies measured here are indicative of real patients receiving incorrect doses at this centre, but rather a symptom of the personnel in this centre being unwilling to read the audit protocol. So many mistakes and miscommunications occurred that would not be surprising if some error had occurred amongst this plethora of errors that affected the delivered dose.

4.3. Discussion

The results from all the phantom measurements were promising, showing that a meaningful audit could take place with the equipment that was used in this project, and the audit would pick up any large dose discrepancies in the future. It also showed that the MOSFET dosimetry system's performance assessment was characterised to a degree that allowed any large discrepancies in high dose regions to be apparent. Discrepancies in low does regions can only be stated in general terms [51]. More

information on the linearity at low doses with high dose rates is needed. The use of two bias supply boxes for the audit could also be considered. This would mean that it would be possible to measure the doses with MOSFETs on both the standard and high sensitivity bias supply settings. Other research has found that peripheral radiotherapy doses measured with MOSFETs on high sensitivity measured these doses with a relative standard deviation of only 1% [52].

The practice audit runs gave valuable information on what could be improved in the audit protocol. A main point is that someone who performs patient planning on a regular basis, usually an RT, should perform the planning, which was asked in the protocol, as a lot of the mistakes made in the audit practice runs were due to inexperience with practical treatment planning. This could also reduce the burden on the centres participating in the audit as it would require fewer man-hours to perform the planning. This could also be improved if one person was assigned to perform all the steps in the protocol, as this would limit the errors due to miscommunications between personnel. Though, if this was done any errors due to inter-personnel communication through the various stages of the treatment process would not be found and since human error constitutes a significant proportion of radiation accidents in radiotherapy, this may not a desirable option.

Small adjustments needed to be made to the audit protocol after the audit practice runs, like fixing up the interest point co-ordinates, specifying virtual wedges, including information on dose grid size, and a reminder to include couch correction factors in the treatment planning. The manner in which the uncertainties were calculated could also be expanded to encompass the uncertainty in the MOSFET positions. This would enable the audit to collect more valid information at low dose levels.

It is judged that if the protocol is carried out by one person, or a well co-ordinated team, the total man-hours required from each centre to participate would be around three hours, which would be one hour for the CT scan and treatment planning and two for the treatment of the phantom.

The use of real instead of virtual wedges by Centre B could be a reason for the discrepancies in the doses outside of the main beams. This depends on how the wedges were commissioned in the centre.

5. Discussion and Conclusion

The aim of this project was to create a procedure for the verification of doses in radiotherapy in the New Zealand hospitals that perform such treatments. To do this the equipment that the NRL had purchased, the CIRS phantom and the MOSFET dosimetry system had to be investigated in terms of suitability for such a purpose. The audit protocol also had to be created. This entailed much investigation and thought on what exactly this audit would be testing. It also involved the input from many clinical staff such as RTs, medical physicists, and a clinical oncologist, to ensure that the protocol was testing parameters relevant to radiotherapy while still being easy to follow.

The MOSFET detectors were found to be suitable for measuring the dose in relatively high dose areas, but as the measured dose in low dose areas was comparatively noisy more investigation of the characteristics of the MOSFETs at low doses is needed before any definitive statements about low dose regions can be made.

These detectors also have to be calibrated individually as the calibration factors and angular dependence was determined to vary considerably from one MOSFET to another. The combined uncertainty from the 'noise' of the MOSFETs and the correction factors was found to be 3% with a coverage factor of 2. A more accurate uncertainty could be found if the uncertainty was calculated individually for each MOSFET, including the standard deviation of the three measurements of the MOSFETs in the treatment of the phantom in the audit.

The CIRS phantom provided by the NRL was found to be suitable to conduct this type of audit. Some small alterations had to be made to the phantom throughout the course of the project. The lines for the lasers at the top of the phantom did not match

the ball bearings just under the surface of the exterior of the phantom. The MOSFET cavities in the phantom inserts had to be enlarged as most of the MOSFETs would not fit.

In the preliminary phantom measurements performed at Christchurch Hospital, the measured doses and the TPS doses correlated within the uncertainty of the MOSFETs in the high dose regions. All of these measurements were within 2%, with the exception of one location point in the clinical lung plan that was still within the 3% uncertainty of the MOSFETs. It was also determined that the MOSFETs could measure the dose accurately in the lung despite concerns about how the differing conditions of charged particle equilibrium could affect the measured dose.

The audit practice runs gave valuable information about the audit procedure that had been created. It demonstrated what worked well in the procedure, what did not, and what needed to be clarified in the protocol. Both of the participating centres did not apply the couch correction factors in the treatment planning, so a clear instruction for this was included in the protocol. The TPS doses, for this reason, had to be altered upon return to Christchurch, and in the case of Centre A the couch correction factor was estimated.

Both centres had one plan in which the measured results matched the TPS results within the uncertainty of the MOSFETs, and one plan which did not. The plan which performed well was different for each centre. Centre A failed to apply to the couch correction factor which affected the mediastinum plan, while at Centre B the tumour volume in the clinical lung plan received less dose then the TPS calculated, even with the couch correction factors being applied upon return to Christchurch. For Centre A an estimated couch correction factor was applied, but, except for one interest point

dose, the difference in the measured dose and the TPS dose was still larger than the uncertainty of the MOSFETs at 5.8%. Because the couch factor was only estimated, it cannot be stated definitively if there were any real dose discrepancies at Centre A. For Centre B, the largest discrepancy, which was 6.3%, was for the clinical lung plan. Since so many things went amiss at this centre, especially while the treatment of the phantom was being performed, largely due to miscommunications and choosing not to read the audit protocol, it is unlikely that this discrepancy gives any real information about the treatment abilities of the centre. However, it did give valuable information about how the audit process that had been developed could be improved.

In the future this audit could be expanded to encompass more complicated procedures such as stereotactic or IMRT treatment as they become more common in New Zealand. If the audit was expanded to encompass these treatments, the dose information collected would also need to be expanded to estimate the effects of uncertainty of the detector positions. This could be done by the analysis of two dimensional dose planes at the regions of interest.

This project provides the basis for an external audit of the whole treatment chain for radiotherapy. It also showed that carrying out a meaningful audit is possible with only minimal burden to the busy centres. The lessons learnt in the audit practice runs will enable the smooth operation of this audit in the future. The NRL is currently scheduled to carry out the full official nationwide audit of the verification of the doses in radiotherapy using the procedures created in this project in late 2008 or early 2009.

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Appendix A

This is the audit protocol that was developed during the course of this project. Appendix A1 is also part of the audit protocol.

Audit protocol

Introduction

This project is NRL funded with the purpose of doing two-yearly level three dose audits of NZ Radiation therapy centres. It is being done in conjunction with University of Canterbury.

The project will be carried out using a chest phantom and a MOSFET dosimetry system provided by NRL. Two treatment plans will be carried out, one in the chest and the other more complex one in the lung. The plans have been designed to minimise work for the centres. Below is a brief overview of what is required of the centres:

1. CT the phantom
    - Acquire CT
    - Transfer to TPS
2. TPS
    - Import CT images
    - Place 6 interest points per plan –(*These are outlined later*)
    - Contour 5 structures –(*also outlined below*)
    - Create two treatment plans -(*treatment plan parameters are all specified*)
    - Enter point doses into table
    - Transfer plans to linac
    - Export plans to CD/DVD in DICOM-RT format
3. Linac
    - Irradiate phantom (3 times for each plan)
    - Enter MOSFET readings into table (done by NRL staff)
4. Further evaluation carried out at NRL

Required resources

TPS requirements

The treatment planning system should be capable of performing 3D dose calculation which account for variations in scatter in the presence of heterogeneities.

The treatment planning system at each centre must be capable of exporting data in RTOG or DICOM-RT formats (DICOM-RT if possible). A local physicist should ensure that this procedure can be performed at the centre before the visit.

Radiotherapy Treatment Requirements

Treatment should be delivered on a megavoltage linear accelerator which provides

- Delivery of 6 MV photons
- Beam modifications (virtual wedges - *if virtual wedges are not possible please inform us*)

The phantom will be treated three times for each plan so adequate time for a treatment unit booking should be scheduled for the phantom in the afternoon on the day of the centre visit. This could be arranged out of hours, either a physicist or an RT would need to be there to place and treat the phantom.

The beam parameters are fixed so MLCs should not be used, even though this maybe standard practice. This is so information of the treatment can be compared between the centres. No tweaking of any of the planning parameters is required! If any of the parameters outlined in this protocol are not possible, please inform us.

Detailed description of audit

1. CT the phantom

a. As each centre will CT the phantom, a 30-minute radiotherapy planning CT booking should be scheduled for the phantom at the beginning of the day of the visit.

b. The phantom has inserts in which the MOSFET detectors are placed. There are ten inserts which have been labelled as follows:

Right Left

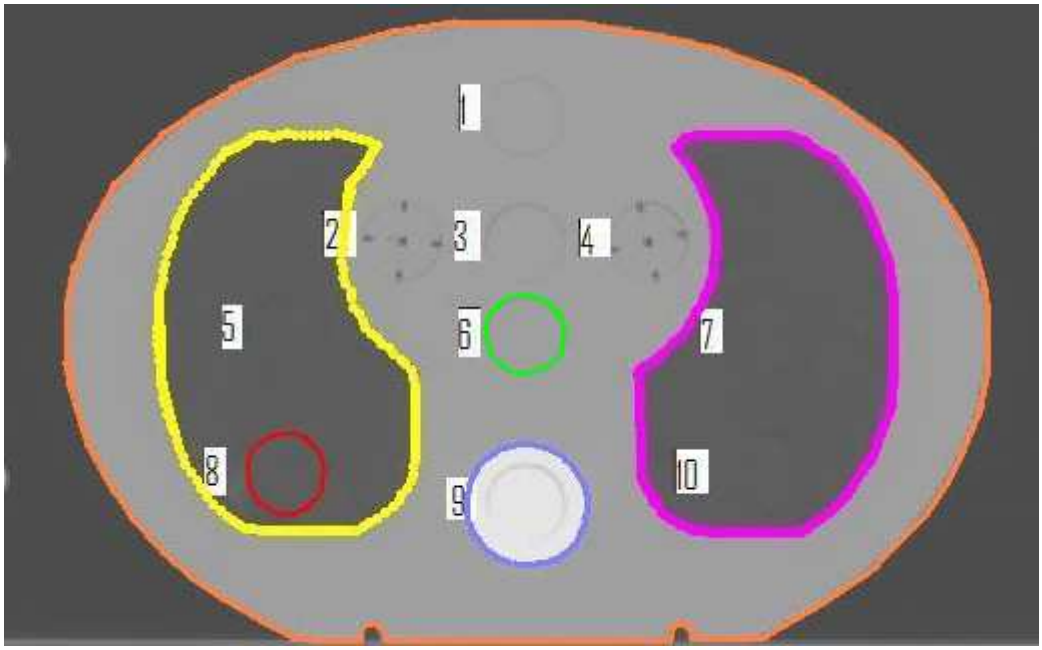

Figure 5.1 This diagram shows illustrates the structures to be contoured. It also shows the insert numbers

c. The inserts can rotate and should be in the correct place as for the treatments. The correct place is with the MOSFET slots to be at 12 o'clock, 3, 6, and 9. (*This will be done by the NRL*)

d. The slices of the phantom should fit together snugly and line up properly so the edges are straight. (*This will be done by NRL*)

e. Align the central markers of the CIRS phantom with the CT room lasers. Do not use the original lines on the top of the phantom!! Use the line which has been scored on the phantom. Note that the orientation of the phantom is in the supine position and note the labels on the phantom.

f. Scan parameters:

   i. Slices thickness: 0.25cm

   ii. Interval: 0.25cm

   iii. Orientation: head first, supine

   iv. Scan length is 8 cm in either direction of the CT origin going superior to inferior.

   v. Reconstruction standard (standard filters)

g. Export CT to TPS

2. TPS

a. Import into TPS. The patient data should be entered into the system as follows

   i. Patient name: CIRSphantom

   ii. Patient id: NRL*ddmmyy*

   iii. StudySet: Chest

b. It does not matter in which order the points and the contouring are done, so where possible follow normal practice.

c. Generate points (Plan 1)

These points will be used to get TPS doses. The CT origin is in the centre of the phantom. (where the placements lasers cross). At this point x, y, and z equal zero on our co-ordinate system. There are ball bearings in the phantom which will help find the origin. In the table below the coordinates of the isocentre is also the normalisation point and the prescription point. After the beams have been placed enter the point doses into the table. The dose grid size should be the usual size which is used. If there is no standard grid size, use 3 mm. Remember to include the *couch correction factors*!

Table of relative co-ordinate positions (Plan 1)

Plan 1 Thoracic plan	Point name	x (cm)	y (cm)	z (cm)	TPS Dose (Gy)
Beam Isocentre and normalisation point	T ISO	0	1.5	0	
interest point 1	T1	0.28	1.5	0.8	
interest point 2	T2	-0.97	1.5	0	
interest point 3	T3	-0.12	1.5	-6.61	
interest point 4	T4	-5.13	1.5	3.11	
interest point 5	T5	-8.98	1.5	0	

d. Generate points (plan 2)

Table of relative co-ordinate positions (plan 2)

Plan 2 Clinical Lung Plan	Point name	x (cm)	y (cm)	z (cm)	TPS Dose (Gy)
Beam Isocentre and normalisation point	L ISO	-7.85	1.5	-4.58	
interest point 1	L1	-7.72	1.5	-3.54	
interest point 2	L2	-8.89	1.5	-4.58	
interest point 3	L3	-1.04	1.5	-5.5	
interest point 4	L4	-3.92	1.5	4.05	
interest point 5	L5	-1.04	1.5	2.95	

e. Enter interest point doses into table (after beams have been placed)

f. Generate contours:

   i. External/patient

   ii. Lung left

   iii. Lung right

   iv. Spinal structure

   v. *Only the tumour volumes outlined below need to be contoured **no** other volumes are required.*

   vi. Tumour volume for **plan 1**:

   - The first 3 cm from insert six – from slice 0 to slice 12 ie, from y=0 to y=3 cm (*figure 2 below*)
   - This is 1.5cm either side of the normalisation point
   - To see the insert we recommend that window=800HU and Level = -100HU

   vii. Tumour volume for **plan 2**:

   - The first 3cm of insert eight – from slice 0 to slice 12
   - This is 1.5cm either side of the normalisation point
   - To see the insert we recommend that window= 200HU and Level = -800HU

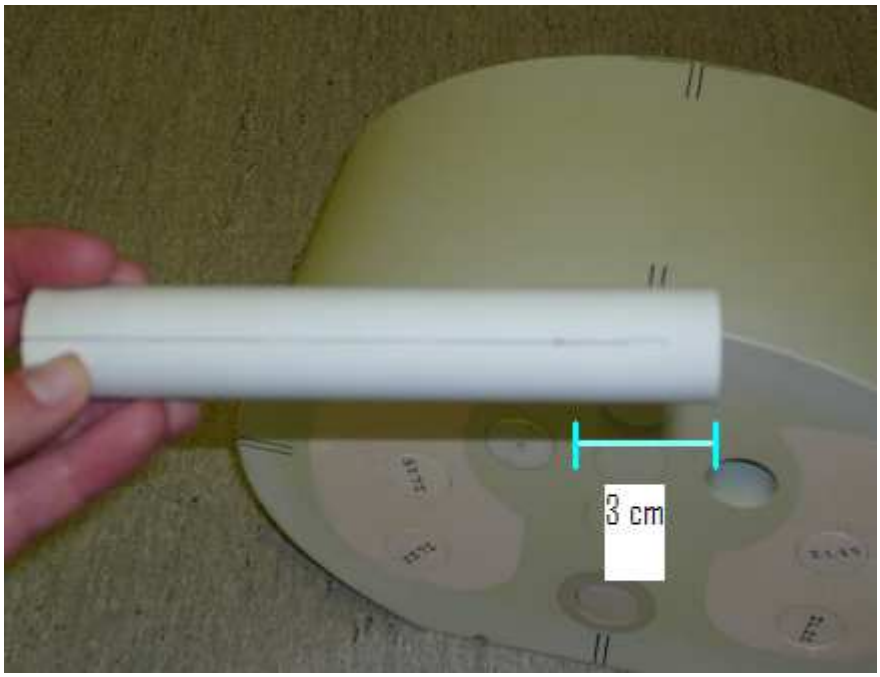

Figure 5.2 This diagram illustrates the tumour volume. It is this first three cm of the insert that should be contoured

Treatment techniques

In both plans the reference point and normalisation point and the isocentre are at the same point and the dose to be delivered to those points is **2 Gy**. The criteria below should not be altered as we **do not** want to test the planning ability of the centres, but rather that the TPS and the detectors are in agreement of the dose. Also make sure that the plans have gone through the appropriate verification process and the plans are ready to use on the linacs when its time to treat the phantom.

Plan one – chest plan

Mandatory Technique Requirements		
Field	**Anterior**	**Posterior**
Density Correction	yes	yes
Isocentric	Yes	Yes
Modality	Photons	Photons
Collimators (cm)	5.0 by 5.0 symmetric	5 by 5 symmetric
Gantry	0^o	180^o
Collimator	0^o	0^o
Couch	0^o	0^o
Shielding	No	No
Compensator	No	No
Bolus	No	No
Recommended Technique requirements		
Energy (MV)	6 MV	6 MV
Wedge	None	None
Relative Weight	45	55
Additional information		
SSD (CM)	90.9	89.0

Plan 2 – Clinical lung plan

Mandatory Technique Requirements		
Field	1	2
Density Correction	yes	yes
Isocentric	yes	yes
Modality	photons	photons
Collimators (cm)	5 by 5 symmetric	5 by 5 symmetric
Gantry*	200	280
Collimator	0	0
Couch	0	0
Shielding	No	No
Compensator	No	No
Bolus	No	No
Recommended Technique requirements		
Energy (MV)	6	6
Wedge	60 ^{o} Thick end lat	60 ^{o} Thick end post
Relative Weight at reference point	48	52
Additional information		
SSD (CM)	95.3	93.4

* In the direction as defined by the IEC

Appendix A1

What the centres need to do. . . . .

1. CT the phantom.
2. The treatment planning (with the CT data), which includes adding beams contours and point doses.
3. Enter Point doses from TPS into the table provided. (*The table is on the next page*)
4. Have the treatment plans ready for treatment i.e., has gone through the verification process and the plans are ready on the machine in time for treatment.
5. Place the phantom for treatment.
6. Export the treatment plans (including structure, dose and images) in DICOM-RT format on to a CD/DVD.

What NRL will do. . . .

1. Put the phantom together ready for treatment with the inserts in the correct locations
2. Arrange inserts with detectors in the correct location for treatment.
3. Take the measurement readings from the irradiations
4. Analyse the results obtained from the information gathered.
5. Inform each centre of their own results

Tables of point doses to fill

Table of relative co-ordinate positions (Plan 1)

Plan 1 Thoracic plan	Point name	x (cm)	y (cm)	z (cm)	TPS Dose (Gy) ***Fill this column***
Beam Isocentre and normalisation point	T ISO	0	1.5	0	
interest point 1	T1	0.28	1.5	0.8	
interest point 2	T2	-0.97	1.5	0	
interest point 3	T3	-0.12	1.5	-6.61	
interest point 4	T4	-5.13	1.5	3.11	
interest point 5	T5	-8.98	1.5	0	

Table of relative co-ordinate positions (plan 2)

Plan 2 Clinical Lung Plan	Point name	x (cm)	y (cm)	z (cm)	TPS Dose (Gy) ***Fill this column***
Beam isocentre and normalisation point	L ISO	-7.85	1.5	-4.58	
interest point 1	L1	-7.72	1.5	-3.54	
interest point 2	L2	-8.89	1.5	-4.58	
interest point 3	L3	-1.04	1.5	-5.5	
interest point 4	L4	-3.92	1.5	4.05	
interest point 5	L5	-1.04	1.5	2.95	

Also please give the following information

How long did the planning take: _____

TPS vendor, software and version used: _____

Calculation planning algorithm used: _____

Appendix B

MOSFET work sheet

This is an example of a filled work sheet from one of the centres, it corrects for energy and angular dependence and calculates the corrected dose for all five points at the end of the sheet. There is two worksheets, one for the chest plan and one for the lung.

chest plan

Beam 1 parameters

Gantry angle	0	degrees
Monitor units	154	MU
Beam energy	6 MV	

Beam 2 parameters

Gantry angle	180	degrees
Monitor units	2	MU
Beam energy	6 MV	

Mosfet data

Mosfet number	Cal factor (Gy/mV)	Insert position	Mosfet position
2947	0.008837578	LC	T
2950	0.00892867	LC	R
2948	0.008838188	S	L
2933	0.009152487	CR	R
2931	0.009019833	RLT	R

Measurement readings:

Mosfet number	reading 1	reading 2	reading 3	reading 4	beam 1 weight	beam 2 weight
2947	203	199	211	205	0.45	0.55
2950	213	212	211	219	0.45	0.55
2948	254	248	244	253	0.1	0.9
2933	6	6	6	5	0.5	0.5
2931	0	1	2	2	0.5	0.5

Mosfet corrections for beam 1

Mosfet number	angle	mod angle	angle correction	energy	weight	Partial dose	
2947	270	**90**	1.05926	0.99	0.45	0.80448914	Gy
2950	0	**0**	1	0.99	0.45	0.84327719	Gy
2948	180	**180**	1.01855	0.99	0.1	0.21757852	Gy
2933	0	**0**	1	0.99	0.5	0.02718289	Gy
2931	0	**0**	1	0.99	0.5	0.00446482	Gy

Mosfet corrections for beam 2

Mosfet number	angle	mod angle	angle correction	energy	weight	Partial dose	
2947	90	**90**	1.05926	0.99	0.55	1.04153365	Gy

2950	180	**180**	1.01855	0.99	0.55	1.04978674	Gy
2948	0	**0**	1	0.99	0.9	1.95820665	Gy
2933	180	**180**	1	0.99	0.5	0.02718289	Gy
2931	180	**180**	1	0.99	0.5	0.00446482	Gy

Mosfet Corrected doses

Mosfet number	Dose	
2947	1.846022784	Gy
2950	1.893063935	Gy
2948	2.175785164	Gy
2933	0.05436577	Gy
2931	0.008929634	Gy

Clinical Lung plan (2 oblique fields with dynamic wedges)

Beam 1 parameters

Gantry angle	200	degrees
Monitor units	154	MU
Beam energy	6 MV	

Beam 2 parameters

Gantry angle	280	degress
Monitor units	2	MU
Beam energy	6 MV	

Mosfet data

Mosfet number	Cal factor (Gy/mV)	Insert position	Mosfet position
2947	0.008837578	RLL	R
2950	0.00892867	RLL	T
2948	0.008838188	S	R
2933	0.009152487	RC	T
2931	0.009019833	S	R

Measurement readings:

Mosfet number	reading 1	reading 2	reading 3	reading 4	beam 1 weight	beam 2 weight
2947	219	224	225		0.48	0.52
2950	217	221	219		0.48	0.52
2948	96	97	99		0	1
2933	73	74	75		1	0
2931	6	6	4		0.5	0.5

Mosfet corrections for beam 1

Mosfet number	angle	mod angle	angle correction	energy	weight	Partial dose

2947	200	**160**	1.035271	0.99	0.48	0.935115	Gy
2950	110	**110**	1.059262	0.99	0.48	0.929196	Gy
2948	200	**160**	1.035271	0.99	0	0	Gy
2933	110	**110**	1.059262	0.99	1	0.670511	Gy
2931	200	**160**	1	0.99	0.5	0.023812	Gy

Mosfet corrections for beam 2

Mosfet number	angle	mod angle	angle correction	energy	weight	Partial dose	
2947	280	**80**	1.057063	0.99	0.52	1.070848	Gy
2950	190	**170**	1.027244	0.99	0.52	1.034053	Gy
2948	280	**80**	1.057063	0.99	1	0.900245	Gy
2933	190	**170**	1.027244	0.99	0	0	Gy
2931	280	**80**	1	0.99	0.5	0.023812	Gy

Mosfet Corrected doses

Mosfet number	Dose		TPS dose	diff	% diff
2947	2.005962531	Gy	1.97	0.035963	1.7928%
2950	1.963249242	Gy	1.98	-0.01675	-0.8532%
2948	0.900245139	Gy	0.86	0.040245	4.4705%
2933	0.670511164	Gy	0.72	-0.04949	-7.3808%
2931	0.047624717	Gy	0.05	-0.00238	-4.9875%