Investigation of the radiation dose from cone beam CT for image guided radiotherapy: a comparison of methodologies

Jarryd Buckley
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INVESTIGATION OF THE RADIATION DOSE FROM CONE BEAM CT FOR IMAGE GUIDED RADIOTHERAPY: A COMPARISON OF METHODOLOGIES

A Thesis Submitted in Partial Fulfilment of the Requirements for the Award of the Degree of

Master of Science - Research

from

UNIVERSITY OF WOLLONGONG

by

Jarryd Buckley

Bachelor of Medical and Radiation Physics

School of Physics
Faculty of Engineering & Information Sciences

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INVESTIGATION OF THE RADIATION DOSE FROM CONE BEAM CT FOR IMAGE GUIDED RADIOTHERAPY: A COMPARISON OF METHODOLOGIES

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A Thesis for Master of Science - Research

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ABSTRACT

Image guidance is now an integral component of radiation therapy. It allows visualisation and verification of the target volume and surrounding structures during the course of treatment allowing for tighter dose margins and subsequently better treatment outcomes. This thesis will focus on evaluating the radiation dose from one image guidance modality, kV cone-beam computed tomography (CBCT).

The first section of this thesis compares four alternative methodologies for quantifying kV CBCT dose; the Cone-Beam Dose Index (CBDI), the International Atomic Energy Agency Report 5 recommended methodology (IAEA) and the American Association of Physicists in Medicine Task Group 111 approach (TG111) with the current methodology, the Computed Tomography Dose Index (CTDI). The methodologies were evaluated for two kV CBCT imaging systems, the Varian Clinac® iX On-Board Imager (OBI) and Varian Truebeam™ X-Ray Imager (XI). Dose measurements were taken in a CTDI cylindrical Poly-Methyl Methacrylate (PMMA) phantom using a Pencil type ionisation chamber with a sensitive length of 100 mm for the CTDI, CBDI and IAEA methods. The TG111 method used a custom PMMA phantom constructed in-house with the same radial dimensions as the CTDI phantom but increased in length from 16 cm to 46 cm. Measurements were taken using a Farmer ionisation chamber with a 2.4 cm sensitive length.

The methodologies were evaluated for clinically relevant scan modes on each imaging system. Pelvis, Thorax, High Dose, Low Dose and Standard Dose Head protocols were evaluated on the OBI system and Pelvis, Pelvis Obese, Thorax and Head protocols on the XI system. The TG111 method gave the highest dose for all the scan modes across both OBI and XI systems, followed by CBDI, IAEA then the CTDI. The Pelvis scan measured the highest dose across the methodologies for OBI with values of 21.199 ± 0.014 mGy, 19.381 ± 0.029 mGy, 18.343 ± 0.002 mGy and 9.408 ± 0.029 mGy for TG111, CBDI, IAEA and CTDI respectively. The Pelvis Obese mode on the XI system delivered the highest dose of all the scan modes with 48.529 ± 0.012 mGy, 44.390 ± 1.475 mGy, 32.295 ± 0.017 mGy and 20.743 ± 1.475 mGy for TG111, CBDI, IAEA and CTDI methods.

The two imaging systems were compared for Pelvis, Thorax and equivalent Head protocols. The OBI system gave a higher overall dose than XI. The Pelvis scan was higher for the OBI by 2.95 mGy and Thorax OBI higher by 1.42 mGy from the TG111 method. For Head scans, the OBI also gave the higher dose with 4.605 ± 0.003 mGy compared with 3.527 ± 0.002 mGy on XI from the CBDI method.
The effect of beam width was also investigated. Each methodology was evaluated on the OBI systems Pelvis mode for beam widths from 2 cm to 40 cm. The TG111 and CBDI methods both approached a limiting value for increasing beam width, with the TG111 being higher than CBDI by 2.258 mGy at 40 cm beam width. The IAEA method plateaued at 15 cm beam width, measuring 18.545 mGy and dropping to 18.260 mGy at 40 cm beam width. The CTDI reached a maximum of 14.246 mGy at 10 cm beam width then decreased for increasing beam width falling to 5.047 mGy at 40 cm.

The Integral Dose ($E_{tot}$), a term from the TG111 which describes the total energy absorbed in the phantom, was determined for the Pelvis and Thorax modes on the OBI system, and the Pelvis, Pelvis Obese and Thorax modes on the XI system. The measured values were 445.69 $\pm$ 4.46 mJ and 128.89 $\pm$ 1.29 mJ for Pelvis and Thorax OBI, and 462.41 $\pm$ 4.62 mJ, 999.81 $\pm$ 10.00 mJ and 114.49 $\pm$ 1.15 mJ for Pelvis, Pelvis Obese and Thorax XI. The integral dose was also evaluated for beam widths ranging from 2 cm to 40 cm. The integral dose increased linearly with beam width. $E_{tot}$ increased by 822.49 mJ and 237.75 mJ for Pelvis and Thorax OBI modes respectively. For XI $E_{tot}$ increased by 821.09 mJ, 1,775.38 mJ and 203.30 mJ for Pelvis, Pelvis Obese and Thorax modes.

Dose profiles of the OBI and XI Pelvis modes were measured in the TG111 phantom using the Farmer chamber and XR-QA2 Gafchromic$^{TM}$ film. Both the OBI and XI profiles extended beyond the collimated beam width at isocentre, with the Farmer chamber measurements only falling to 62% and 58% of the maximum dose at the collimated beam edge for OBI and XI systems respectively. The dose in the centre of the beam was 3.2 mGy higher for the XI system compared with OBI, however the difference decreased to 1.2 mGy at $\pm$ 12 cm. The film dose was higher than the Farmer chamber measurements by 0.302 cGy for OBI and 0.245 cGy for XI at the centre of the beam. Several factors were considered for the difference between the ionisation chamber and film but no specific factor could be attributed to the variation.

CTDI measurements were taken in the TG111 phantom using normalised film strips with sufficient length to capture the full dose profile and the results compared to the methodologies. The CTDI$_{film}^{w}$ was calculated to be 20.47 $\pm$ 0.71 mGy for OBI and 21.28 $\pm$ 0.44 mGy for XI. These were in good agreement with the TG111 protocol values of 21.20 $\pm$ 0.01 mGy and 22.42 $\pm$ 0.01 mGy respectively. The agreement between CTDI$_{film}^{w}$ and TG111 validated the TG111 method as a accurate representation of CTDI with sufficient detector and phantom length. The result also validated the CBDI as a good approximation to the CTDI using the existing CTDI$_{100}$ phantom and 100mm Pencil ionisation chamber.

This thesis provides a quantitative comparison between four methodologies on two commercial kV CBCT imaging systems for quantifying dose, and the effects of beam width on dose and total deposited energy. The findings may assist clinical physicists in choosing the appropriate scan parameters and frequency of CBCT acquisitions to optimise the benefits of image guidance while limiting additional dose and in selecting an appropriate methodology for quantifying the output from kV CBCT systems.
KEYWORDS: Cone Beam Computed Tomography, Radiotherapy, Image Guided Radiotherapy
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Chapter 1

Introduction

In recent years there has been rapid development of modulated dose delivery methods in radiotherapy such as Intensity Modulated Radiation Therapy (IMRT), Volumetric Modulated Arc Therapy (VMAT) and Tomotherapy. With these new methods the prescribed treatment dose can be delivered to the target with a high degree of conformity while minimising the dose to healthy tissue.

However these techniques cannot be utilised to their full potential without adequate verification of patient set up. The use of modulated delivery gives highly conformal dose with steep dose gradients, so any variation in patient set up or movement can lead to significant variation in dose delivered to both the tumour and surrounding tissue.

To verify patient set up with the required precision, Image Guided Radiation Therapy (IGRT) is being utilised. Traditionally, position verification was performed with the MV treatment source and an electronic portal imaging device (EPID) or film placed opposite the treatment beam and beneath the patient. However at these MV energies the inherent Compton scatter results in poor soft tissue contrast, limiting reference points within the body to higher Z tissue such as bone, or internal fiducial markers. To resolve this many LINACs now have a built in kV imager which can produce images with improved soft tissue contrast to correct for internal organ motion and patient
set up errors. Examples include the On-Board Imager (OBI) and the Truebeam X-Ray imaging system (XI) of Varian Medical Systems (Palo Alto, CA) and the X-ray Volumetric Imager (XVI) of Elekta Synergy (Crawley, United Kingdom).

These devices consist of a kV X-ray source and an amorphous silicon detector mounted to the LINAC gantry on extendable robotic arms 90 degrees to the treatment beam. These devices can acquire 3D cone-beam CT (CBCT) images of the patient typically within 60 seconds. Volumetric images of the patient can be obtained in a single rotation of the gantry and registered with the radiotherapy planning CT to check for positional errors and make corrections as necessary with a high degree of accuracy [1] [2].

Currently, imaging dose is largely omitted from treatment plans since it’s typically two orders of magnitude smaller than the therapeutic doses, typically of 1-2 Gy for a whole treatment schedule or 1-20 cGy for a single scan [3] [4] [5] [6] [7] [8]. However, during an imaging procedure large portions of the body are irradiated, including radiosensitive structures such as lung, breast, thyroid and reproductive organs. Bone structures also receive higher doses at kV energies due to increased photoelectric effect and their high atomic number Z.

Each clinic has its own protocols for frequency of CBCT depending on tumor site and experience in expected daily shift. A typical CBCT might be once per day for the first week then once per week. For Stereotactic Ablative Radiotherapy (SABR) at least two CBCTs are performed in many centres and often three or four per fraction, i.e one CBCT mid fraction. During a treatment of 30-40 fractions the imaging dose has been shown to be significant [9] [1] and could lead to an increased risk of a patient developing a secondary primary malignancy [10]. Therefore a method for quantifying the imaging dose is necessary to evaluate any increased risk to the patient and make an informed decision on the trade off between imaging dose and position verification.
Clinically, the current CTDI paradigm for defining fan-beam CT dose does not give an accurate estimate of the dose delivered to the patient from daily imaging, particularly for the wide-beam scanning such as CBCT where beam widths can be in excess of 40 cm [11]. Three alternative protocols have emerged in recent years which attempt to better quantify the imaging dose for wide beam scanning [12, 11] however it is not clear if these new methods offer a better representation of patient dose.

1.0.1 Research Aims

This thesis aims to address the following aspects of CBCT dosimetry:

- Evaluate and compare the current CTDI protocol with three alternate methodologies: the Cone-Beam Dose Index (CBDI) adapted for wide beam scanning [13], the International Atomic Energy Agency Report No. 5 (IAEA) suggested approach for wide beam dosimetry [11] and the American Association of Physicists in Medicine Task Group 111 methodology (TG111) [12].

- Evaluate and compare methodologies for varying beam width, specifically for clinical settings and compare between the listed protocols, and

- Compare dose between two Varian imaging systems: the OBI® on the C-Series platform and XI® on Truebeam.
2.1 Radiotherapy

Radiotherapy is the use of ionising radiation for the treatment of cancer. It may be used as an adjuvant therapy with surgery and or chemotherapy, as an ablative treatment for locally advanced inoperable cancers or for palliation of advanced incurable disease. Radiotherapy in combination with surgery and or chemotherapy has been shown to improve local tumor control for a number of tumor types [14, 15, 16].

Radiotherapy may be delivered in 3 ways;

- **external beam therapy** through the use of a high energy linear accelerator or orthovoltage and kilovoltage machines for superficial tumors,

- **encapsulated sources** which are placed within the patient either temporarily such as in High Dose Rate brachytherapy or ‘permanently’ for Low Dose Rate brachytherapy, and

- **unencapsulated sources** for targeted radionuclide therapy where the radioactive material is administered to the patient, usually bound to a molecule that seeks out tumor locations.
2.1. Radiotherapy

For the purpose of this thesis, only external beam therapy with X-rays will be discussed.

2.1.1 External Beam Therapy

The most common modality in radiotherapy is external beam therapy. For deep seated tumors this requires mega voltage (MV) photons. These photons are generated and delivered to the target using a linear accelerator or LINAC as shown in 2.1.

![Truebeam® LINAC](image)

Figure 2.1: Truebeam® LINAC is capable of delivering MV X-Ray and electron therapy as well as IGRT via its X-Ray imaging unit. Varian Medical Systems®

Generally speaking, clinical LINACs deliver radiotherapy as follows. Electrons generated from an electron gun are accelerated in a waveguide under microwave power to relativistic velocities. The electrons are then bent through a bending magnet onto a tungsten target producing MV photons through the Bremsstrahlung process. The beam is then shaped and filtered by a set of collimators and a flattening filter to reduce out of field scatter and flatten the beam profile. The field is then shaped by a secondary set of collimators (or jaws) and a set of multi-leaf collimators (MLC) to create fields of varying intensity. A detailed description of external beam radiotherapy can be found in the literature [17].
2.1.2 Interactions of kV Ionising Radiation

2.1.2.1 Interactions of Dose Deposition

With the exception of imaging with the MV treatment beam, imaging of the patient during the course of radiotherapy occurs in the kilovoltage (kV) energy range, typically between 70 kV and 150 kV. At these energies, the dominant interactions within the body are Compton scattering and photo-electric absorption as shown in figure 2.2.

Figure 2.2: Schematic diagram of dominant photon interactions vs incident photon energy. For energy ranges typical of imaging and radiotherapy (above 80-100 keV), the Compton and photoelectric effects are the dominant interactions, and the only interactions for imaging energies as pair production doesn’t occur in the kilovoltage energy range. Also visible is the increased contribution of photo-electric absorption with increasing atomic number of absorber. Source: *Encyclopaedia of Occupational Health and Safety. Fourth Edition*

The dominant interaction of photons within tissue depends on their incident energy and the composition of the material. For radiotherapy treatment, where MV energies are used, the Compton effect dominates for most materials and dose deposition is relatively consistent in soft tissue, muscle and bone [18]. However for diagnostic energies in the kV range (such as for kV CBCT), the dominant interaction varies depending on the energy and mass absorption coefficient of the material. For the lower energies or higher Z materials such as bone, the photoelectric effect dominates due to its \( \approx Z^4 \) dependance and hence absorption is greater. It’s this variation in Z which allows
contrast in diagnostic images. However, regardless of which interaction the photon undergoes, the result is ultimately a fast electron depositing energy in matter \[18\].

**Compton Scattering**

The Compton Effect occurs when an incident photon of energy \(E_p\) undergoes an elastic collision with a free electron in the outer shell of an atom. For radiotherapy applications it is the dominant interaction, with it being the only interaction of significance within tissue in the 200 keV to 2 MeV energy range \[17\]. During the Compton process, some of the photon’s energy is transferred to the electron causing it to be ejected with energy \(E_k\).

Compton scatter reduces the signal to noise ratio (SNR) for imaging, particularly for soft tissue. It presents the main challenge for CBCT reconstruction, especially for MV CBCT where higher doses are required to achieve sufficient image quality \[19\] due to detector quantum efficiency at higher energies producing increased noise. At kV energies, the SNR is improved so kV CBCT requires less dose to achieve images of sufficient image quality for IGRT as shown in figure 2.3.

**Photoelectric Effect**

The photoelectric effect is a phenomena by which a photon is completely absorbed by a bound electron. If the energy of the photon is sufficient, the electron will be ejected from the atom. The energy of the emitted electron depends on the incident photon energy and any energy losses in escaping the atom.

If it is an inner shell electron that is ejected, then an outer shell electron will replace it, emitting a photon with energy characteristic of the energy gap between the electron shells or characteristic X-rays. However, since most tissue contains low-Z elements the energy of such photons is sufficiently small (carbon = 0.3keV) such that they are locally re-absorbed \[17\].
2.1. Radiotherapy

The high Z of bone causes higher photoelectric absorption and high contrast compared with soft tissue in kV CBCT images. The higher absorption also causes higher dose in the bone, reportedly three times the dose to soft tissue [20].

2.1.2.2 Bremmstrahlung Radiation and Characteristic X-Rays

X-Rays are created when the kinetic energy of electrons is converted to electromagnetic radiation when decelerated by a target material. The process is known as Bremmstrahlung radiation and is the source of X-rays for MV external beam radiotherapy and kV imaging.

The energy is lost via coulomb interactions with the target nucleus, which causes the electrons to decelerate and change direction with the loss of kinetic energy being dissipated as heat and the production of an X-ray photon.

In addition to Bremmstrahlung emission, characteristic emissions are also pro-
duced. These emissions arise when a high energy electron strikes a bound electron in the target material. Provided the kinetic energy of the electron exceeds the binding energy of the atomic electron, it can be ejected from the atom. The vacancy created by the ejected electron is replaced by an electron from an outer shell with lower binding energy, releasing a characteristic X-ray with energy equal to the difference between the binding energies of the two shells. Characteristic X-Rays can also be produced from Bremsstrahlung photons striking the filter.

Electron transitions can occur from multiple energy levels, giving rise to different characteristic X-ray energies. The energy spectrum of 90 kVp electrons striking a tungsten target is shown in figure 2.4.

Figure 2.4: Filtered spectrum of Bremsstrahlung and characteristic radiation from a tungsten target for a 90 kV potential. Source: [21].

2.1.3 Dosimeters

The purpose of a dosimeter is to estimate the absorbed dose in the medium to which they are placed. In radiotherapy they are used for determining beam profiles as well as verification of dose deposition with treatment planning.
With the exception of calorimetry, dose cannot be directly measured, but rather some property of ionising radiation is exploited to obtain an output which can be converted to dose.

An ideal dosimeter should:

- Give an accurate and precise (reproducible) measurements over a large measurement range
- Have a low detection limit
- Be linearly proportional to the delivered dose as well as being dose rate independent.
- Be equivalent to the tissue in which the dose is being determined, and;
- Have a good spatial resolution

In reality however, no dosimeter possess all these attributes. A particular dosimeter should be carefully chosen for its appropriate application where its properties are most suited to the task. In this thesis, ionisation chambers and radiochromic film are used to measure kV CBCT dose. Both dosimeters are discussed below.

### 2.1.3.1 Film

Film dosimetry is one of the oldest methods for the detection of X-rays [17]. All films provide a 2D pixel intensity map that with appropriate calibration can be converted into a 2D dose map. In recent times conventional radiographic film has been phased out in favor of digital radiography where real-time dosimetry can be performed and records are not subject to deterioration over time or if exposed to light and/or humidity [17].

Radiochromic film has been replacing radiographic film of late for its self developing and near tissue equivalence, with current film dosimetry in radiotherapy being increasingly dominated by the use of Gafchromic film [22], which also has the benefit of being independent of dose rate [23]. However for energies below 100 keV, the film
2.1. Radiotherapy

displays an energy dependency of $\approx 14\%$ \cite{24, 25} while for energies below 40 keV variations of as much as 170% have been reported \cite{26}. For energies in the range of 80-145 kVp typical of CBCT however, the variation in sensitivity has been shown to be less than 5\%\cite{27, 28}. For a non-homogeneous kV source, the effective energy of the beam is dependent on both the kVp and HVL which must be considered in making an assessment on the response of the film. The use of radiochromic film in the investigation of CT and CBCT has been previously investigated with good agreement between film response and measured dose provided the film has been calibrated for the given beam energy and quality\cite{4, 29, 30}.

2.1.3.2 Ionisation Chamber

The ionisation chamber is the most common dosimeter used in radiotherapy departments. It is considered the gold standard of dosimeters and calibration of any radioactive sources mostly use this dosimeter, which is traceable to a primary standard. Schematically, an ionisation chamber consists of two electrodes in a gas (e.g. air) volume where an electric field is induced. When ionising radiation enters the sensitive volume it ionises the gas producing electrons and positive ions. These ions are attracted to their respective electrodes by the electric field producing a current. The measured current can then be converted to dose. A more rigorous description regarding the operation and physics of the ionisation chamber is given by Metcalfe et. al. \cite{17}.

At diagnostic energy levels (80-140 keV), ionisation chambers have very little energy dependance, and can be directly traceable to secondary (or primary) standards, whereas other detectors, including film and solid state dosimeters are not similarly traceable and need additional calibration steps to establish traceability. Due to their availability in the clinic, ease of use and versatility in their application, clinical protocols developed to measure radiation dose use some version of the ionisation chamber,
such as the 100mm pencil ionisation chamber in the CTDI methodology \cite{31} and the smaller volume thimble chamber in the TG-111 methodology \cite{12} for quantifying CT imaging dose.

CBCT dose has also been investigated with small volume dosimeters such as Thermo-luminescent Dosimeters and Metal Oxide Silicon Field Effect Transistors, with good agreement to ionisation chamber measurements \cite{5, 7, 32, 13, 1, 6, 33, 34, 35}. Their small size make these dosimeters useful for skin dosimetry and placement within anthropomorphic phantoms for organ dose measurements during CBCT scans \cite{13, 1, 9, 36}.

## 2.2 Image Guided Radiation Therapy

With the development of modern treatment modalities such as Intensity Modulated Radiation Therapy (IMRT) and Volumetric Modulated Radiation Therapy (VMAT) which deliver highly conformal dose with steep dose gradients\cite{37, 17, 38, 39, 40}, accurate verification of the position of the target volume and critical structures just prior to treatment is required. Image Guided Radiation Therapy (IGRT) utilises imaging technology in the treatment room to verify positional information prior to or during treatment. IGRT systems allow widespread management of geometric variations in patient set-up and internal organ motion. The clinical introduction of these guidance systems has allowed the assessment and correction of patient positioning uncertainties, revealed internal organ motion and deformation and is paving the way towards advanced and adaptive radiation therapy \cite{41}. By improving the geometrical accuracy of radiation therapy, planning volumes can be reduced leading to improvements in tumor control probability and reduction in toxicity or dose escalation with the same toxicity. Image guidance may include ultrasound and in the future MRI, but the most common modality is kV X-ray.
2.2.1 Image Guidance Modalities

2.2.1.1 Electronic Portal Imaging Device

This imaging modality utilizes the linac MV treatment beam and a flat-panel detector mounted on a retractable arm to acquire 2D images of the patient. The detector consists of an amorphous silicon detector array overlaid with a fluorescent layer. Due to the high energy, these images are of a poorer quality than other imaging modalities from Compton effects. However it’s sufficient for detecting motion of bony structures and internal fiducial markers. The use of the treatment beam for imaging purposes is advantageous as no additional imaging system is required, and imaging dose may be incorporated into the treatment planning. However it has been shown to give the highest dose of any imaging modality. Ding et. al. compared 50% DVL volumes for several organ sites and found MV imaging dose ranged between 3-8 cGy compared with 0.04-1.6 cGy and 0.07-0.12 cGy for kV-CBCT and kV radiograph for the same anatomical structures respectively [42].

2.2.1.2 kV Radiograph

These images are acquired via the linac mounted imaging device consisting of a kV X-ray source and flat panel detector. Two orthogonal images are taken for patient position verification. kV radiographs have been shown to give the lowest dose of the listed modalities [42].

2.2.1.3 Fan-Beam CT

Unlike the other imaging modalities, FBCT utilizes a fan-beam geometry to acquire images analogous to a conventional CT scanner. The X-ray source is coupled to a detector array 180 degrees from the source which both rotate around the patient.

MV FBCT images may also be acquired using the treatment beam of the tomother-
apy treatment machine such as the Hi-Art® system from Accuray and a conventional CT detector (Xe filled ionisation chamber array). Beam energy is reduced from 6 MV to 3.5 MV and pulse repetition frequency reduced to minimise dose [17]. MV FBCT allows for easy comparison to planning CT and produces less dose than multi-slice CT scanning [9]. The dose from MVCT has been reported and is typically 1 cGy per scan. These scans taken on a daily basis during a course of breast radiotherapy have been shown to increase dose to controlateral breast by 28 cGy and 22 cGy to the contralateral lung [43].

2.2.1.4 MV CBCT

MV CBCT images are acquired using the MV treatment beam and existing EPID device in a rotation around the patient. Unlike FBCT the beam width for CBCT is in the order of 20 centimeters (cm) and an image can be acquired in a single rotation of the X-ray tube. It has been reported that MV CBCT images acquired on a linac are of high enough quality for patient alignment [44] however generally with a higher dose than kV CBCT [17, 42, 13, 45, 9].

2.2.1.5 kV CBCT

kV CBCT imaging is typically acquired with a gantry mounted kV imaging device such as the Varian OBI or Elekta XVI. In both cases a kV X-ray source and flat-panel detector are mounted on the gantry 180 degrees to one another and orthogonal to the treatment beam. As with other 3D modalities the system rotates around the patient to acquire CBCT images. It has been shown that kV CBCT results in a lower dose to the patient compared with MV flat-panel imaging with superior image quality [42, 13, 45, 9]. However the image quality for kV CBCT is poorer than conventional fan-beam CT and delivers a higher dose [11, 9], with the exception of multi-slice CT which delivers more dose than kV CBCT [6]. Both MV and kV CBCT are based on
treatment isocentre, so for clinical implementation calibration procedures are required to account for component sags and flexes from the weight of the system on the gantry.

The relative dose delivered from each of the imaging modalities are summarised in table 2.1.

| Modality | Eye Dose | Bladder Dose | Heart Dose |
|----------|----------|--------------|------------|
| MVCT Tomo | - - | - - | 1.1-3.7 |
| MV EPID | 4.3-4.8 | 3.3 | 3.5 |
| MV CBCT | 0.08 (brain) | 0.13 | 6.35 |
| kV CBCT | 0.04-0.06 | 1.0-1.6 | 0.2-0.4 |
| kV Planar | 0.12 | 0.07 | 0.07 |

Table 2.1: Summary of doses in cGy from imaging modalities used in radiotherapy.

### 2.2.2 kV Imaging System

LINAC mounted kV imaging systems such as Varian OBI®, Truebeam XI® and the Elekta XVI® consist of a kV X-ray generator, kV X-ray tube, beam collimator and filters and a kV norm chamber. The system is mounted on a retractable arm which houses the high-voltage and oil-cooling cables for the X-ray tube. 180 degrees to the treatment beam is the kV imager, also on a retractable arm. The flat-panel imager detects low dose X-rays from the kV source and can acquire fluoroscopic and kV digital radiographic images as well as CBCT. The OBI unit mounted on a Varian Clinac IX® accelerator is shown in figure 2.5.

The kV system shares the same isocentre as the MV treatment beam. As such, online corrections can be made to patient set-up with remote couch movement.
2.2. Image Guided Radiation Therapy

Figure 2.5: Varian Clinic IX® with OBI imaging system. The X-ray tube and flat-panel detector are mounted to the gantry 180 degrees to one another on retractable arms which are extended for kV imaging as shown. Courtesy: www.Varian.com

2.2.2.1 kV Delivery System

The kV X-ray generator is located in the stand and delivers the high-voltage power supply to the X-ray tube. The generator also monitors and controls performance of the tube such as temperature and oil pressure as well as tube current, voltage and exposure time.

The kV X-ray tube is mounted on the left extendable source arm and provides the kV X-rays via the production of Bremsstrahlung radiation and characteristic X-rays. A simple diagram of the GS-1542 X-ray tube used by the XI imaging system is shown in figure 2.6.

The tube consists of a cathode that supplies the electrons and an anode target 1-2cm from the cathode that rotates to dissipate heat. The anode and cathode are encased in a tube housing to provide protection from radiation leakage and encapsulate the cooling oil [48].

Electrons from the cathode are accelerated by an electric field between the cathode and anode, which is controlled by the X-ray generator to produce the required photon energy. For example, to generate 120 kVp photons, 120 kV will be applied between the cathode and anode. This will accelerate electrons to 120 keV since kinetic energy
2.2. Image Guided Radiation Therapy

2.2.2.2 Collimation and Filters

The kV delivery system uses a series of collimators and filters to adjust field size and modify the energy spectrum. An overview of the collimation system on the XI system is shown in figure 2.7.

The beam first passes through inherent filtration from the exit window (C) which removes low energy photons from the spectrum before passing a set of primary lead collimators that limit beam scatter by restricting the maximum field size of the beam (J). The norm chamber sits next to the primary collimators and monitors pulse by pulse fluctuations in the secondary radiation of the kV beam during CBCT acquisitions (D).

The image size is regulated by the collimators (K). The collimators consist of two sets of lead blades, one set to regulate field size in the X direction and the other in the
2.2. Image Guided Radiation Therapy

Figure 2.7: kV collimation and filtration on the XI imaging system. Source: [48]

Y direction. The blades move independently of each other allowing for asymmetric field shapes to limit exposure to the areas that require imaging.

An improvement that was made from OBI to the XI system was the addition of a 0.89 mm titanium foil filter to harden the beam by absorbing low-energy photons and subsequently reduce the imaging dose and improve image quality (L) [48].

Bow-tie filters non-uniformly shape the kV beam to improve the quality of CBCT projections (E). They reduce skin dose to the patient, reduce X-ray scatter to improve image quality and enable larger imaging fields than the maximum width of the detector for CBCT acquisitions. The filters are constructed from aluminum and vary in thickness between 2 mm and 28 mm.

For small fields the symmetrical full bowtie filter is used while for large field sizes half bowtie filter produces an asymmetric field in the x-y plane, as shown in figure 2.8.

The bow-tie filter is in-built into the XI system and will insert the appropriate filter for the chosen scan, while the filter is applied to the kV generator manually for the OBI system.
2.2. Image Guided Radiation Therapy

Figure 2.8: (a) Bow-tie filters used for CBCT filtration. The size of the FOV dictates the filter with small FOV head scans using the full-fan (A) while larger FOV scans use the half-fan (B). Source: [48]. (b) Diagrammatic representation of full-fan vs half-fan scanning. Full-fan scanning (a-partial scan) (b-full scan) is used for scanning smaller targets. The collimation is symmetric and the beam is centred on isocentre. Half-fan scanning (c)(d) acquires a larger FOV using an asymmetric beam offset from centre. Source: [30]

2.2.2.3 Flat Panel Detector System

Real time imaging of the patient for image guidance is possible through flat-panel Thin-Film Transistor (TFT) array detectors which convert the X-ray intensities to an electric readout. The PaxScan 4030CB and MV IDU 20 model detectors are used by Varian for kV and MV imaging respectively [48].

The detector consists of an array of individual display elements arranged in a matrix. They are constructed of amorphous Silicon etched to deposit the necessary electrical components and connections. Each individual detector element contains a TFT, charge collection electrode and a storage capacitor as shown in figure 2.9.

The TFT is an electronic switch containing three connections: the source, drain and gate. Gate and drain lines connect the gate and drains of each detector element along a row. X-ray imaging devices with TFT’s can be separated into two categories: those which incorporate a photoconductor to produce electrical charges on the detection of
2.2. Image Guided Radiation Therapy

Figure 2.9: Schematic overview of the flat panel TFT array detector. Each detector element consists of a TFT, a charge collection electrode and a storage capacitor.Courtesy: www.arosystems.com.au

an X-ray, known as direct conversion, and those which use a scintillator material, such as $CsI$ or $Gd_2O_2S$, to produce visible photons which are then collected and converted to charge by the charge collection electrode via the photoelectric process and stored in the storage capacitor. Both processes are illustrated in figure 2.10 [51].

Figure 2.10: Large area active matrix array for X-ray TFT detector with (a) direct conversion using a photo conductor layer with pixel electrodes and (b) indirect conversion using a phosphor layer and photo diodes. Source: [51] figure 7.9 page 173.

During the irradiation the TFT switch is closed, allowing charge to be accumulated
2.3. Radiation Dose from kV CBCT Image Guided Radiotherapy

and stored in each capacitor. Following the exposure the gate line to each detector row is turned on sequentially, allowing the stored charge to flow through the drain line and to the connected charge amplifiers. The charge amplifiers amplify the charge, convert it to a proportional voltage and digitise the voltage level to produce a grey scale value for each detector row. The process is repeated for each row to read out the array [21].

2.3 Radiation Dose from kV CBCT Image Guided Radiotherapy

The imaging dose received as part of radiotherapy procedures has previously been given little consideration in planning as it was considered negligible in comparison to therapeutic dose. However with the development of more advanced imaging procedures during image guided therapy such as CBCT, imaging dose can not be considered negligible and staff should be aware of the magnitude of dose delivered, particularly after claims of 2-3 Gy from CBCT imaging alone during a course of radiotherapy was reported by Spezi et.al. [52, 53].

Many studies have sought to investigate the dose delivered from CBCT imaging with varied results [54]. Typically CBCT dose studies use phantoms such as cylindrical or elliptical designs or Anthropomorphic phantoms [1, 55, 5, 35, 56, 36, 9, 7, 4]. CBCT dose has also been calculated using collected beam data through Monte Carlo methods such as EGSnrc and Geant4 [3, 20, 57, 58, 59, 53, 42], and treatment planning systems (TPS) such as Philips Pinnacle® TPS [60]. Effective doses from CBCT head, thorax and pelvis scans are summarised in table 2.2 and absorbed dose studies in table 2.3. The effective dose is calculated from the summation of doses to irradiated organs, scaled by tissue weighting factors usually taken from International Commission on
Radiological Protection (ICRP) report 60 [61] or more recently report 103, where some weighting factors were revised [62]. A more detailed description of effective dose as defined by the ICRP can be found in section 2.4.3.1. The relationship between effective dose for diagnostic doses due to weighting factors has been previously investigated [63].

| Study             | Head (mSv) | Scan | Thorax (mSv) | Pelvis (mSv) | Phantom                  | Dosimeter          |
|-------------------|------------|------|--------------|--------------|--------------------------|--------------------|
| Sawyer et. al.    | 2.83 ± 0.03|      | 13.4 ± 0.5   | Anthropomorphic | TLD                     |
| Hyer et. al.      | 0.04 *     | 7.15 *| 3.78 *       | Anthropomorphic | Fibre Optic Coupled     |
| Hyer et. al. (OBI)| 0.12 *     | 1.82 *| 4.34 *       | Anthropomorphic | Fibre Optic Coupled     |
| Kan et. al. (std dose) | 10.26 ± 0.46| 23.56 ± 0.35| 22.72 ± 0.29| Anthropomorphic | TLD                     |
| Kan et. al. (low dose) | 2.10 ± 0.08| 5.23 ± 0.122| 4.89 ± 0.163| Anthropomorphic | TLD                     |
| Halg et. al.      |            |      | 4.05 *       | Anthropomorphic | TLD                     |
| Cheng et. al.     | 1.65 ± 0.01|      | 8.21 ± 0.04  | Anthropomorphic | TLD                     |
| Gu et. al.        | 8.54 *     |      | 6.25 *       | VIP-man phantom | Monte Carlo Simulation |

Table 2.2: Effective dose per scan kV CBCT for head and neck, thorax and pelvis protocols. (std dose) and (low dose) refer to different scan settings, with low dose using a lower mAs. The order of magnitude variation between studies is likely due to older scan algorithms having much higher exposures compared with newer versions. Studies which did not provide an uncertainty for their values are indicated by *.

The effective dose values vary within a wide range of 0.04-10.26 mSv for head and neck imaging and 1.82-23.56 mSv for pelvis and thorax imaging. These varied results can be attributed to several factors such as variations in anatomical site, measured dose, relative tissue weighting factor used, a low or high dose imaging protocol and make of the imager which may influence the size of the imaged volume and filtration systems including bow tie filters. Different software versions may have been used, with the newer versions tending to reduce concomitant dose while maintaining similar image quality[54].

Of particular interest is the dose received by organs most sensitive to radiation such as the eye or the spinal cord. Monte Carlo studies have shown the dose to the eye, spinal cord, brain and cervical vertebrae can be as high as 80, 60, 50 and 180 mGy respectively from a head and neck CBCT scan [3]. For bone, doses as high as 250 mGy were predicted due to the increased photoelectric effect. Abdominal scans showed mean doses of 30 and 70 mGy to the prostate and femoral heads respectively [3, 20]. By comparison, Hyer et. al. measured doses in an anthropomorphic phantom
and found only 1.07 mGy for the lens and 0.70 mGy for the brain following a head scan, while following a pelvic scan the prostate received 27.63 mGy suggesting some variation between simulation and measurement [36]. The Hyer study also reports its measurements were taken with a more recent version of the OBI software which reduces tube current and uses a 200 degree gantry rotation for head scanning to reduce dose to the eye [36].

Patient based CBCT dose studies form 20% of the literature [54] and generally employ TLD’s or something similar to measure skin dose, which ranged from from a fraction of a cGy to 7 cGy [5, 13, 35, 56]. Dose to the rectum has also been measured in vivo and indicate 2-3 cGy per CBCT acquisition [45, 65].

Out of field doses from CBCT imaging are of some concern, particularly when combined with head leakage and scatter from the treatment beam. Perks et. al. measured the peripheral kV-CBCT dose and compared with IMRT treatment alone and found the two were of the same order of magnitude [33]. Qui et. al. also investigated the peripheral kV-CBCT dose with head leakage and scatter through Monte Carlo simulation, concluding the peripheral CBCT dose is of the same order of magnitude as LINAC leakage and an order or magnitude smaller than IMRT Arc therapy scatter dose [66].

### 2.3.1 Computed Tomography Dose Index

The current paradigm for characterising radiation dose in CT scanners is based on the computed tomography dose index (CTDI) introduced over 30 years ago [31]. The CTDI quantifies the radiation output of the CT scanner, giving a dose index which relates to the phantom used, mAs and kV for quality assurance purposes. However it does not represent the patient dose [67].

As outlined by Shope et. al., the average dose in the centre of a multi-slice CT
2.3. Radiation Dose from kV CBCT Image Guided Radiotherapy

| Study          | Phantom                  | Dosimeter              | Dose (mGy) |
|----------------|--------------------------|------------------------|------------|
| Hu 2014        | Cylindrical PMMA 45cm    | 0.6cc Thimble chamber, AAPM | 22.70 (Pelvis) |
| Hu 2014        | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 18.06 (Pelvis) |
| Hu 2014        | Cylindrical PMMA 15cm    | 100cm pencil chamber, IAEA Formalism | 15.88 (Pelvis) |
| Hu 2014        | Cylindrical PMMA 45cm    | 0.6cc Thimble chamber, AAPM | 28.11 (High Quality Head) |
| Hu 2014        | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 27.10 (High Quality Head) |
| Hu 2014        | Cylindrical PMMA 15cm    | 100cm pencil chamber, IAEA Formalism | 21.49 (High Quality Head) |
| Hu 2014        | Cylindrical PMMA 45cm    | 0.6cc Thimble chamber, AAPM | 5.49 (Standard Dose Head) |
| Hu 2014        | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 5.53 (Standard Dose Head) |
| Hu 2014        | Cylindrical PMMA 15cm    | 100cm pencil chamber, IAEA Formalism | 4.28 (Standard Dose Head) |
| Ding 2013      | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 4.28 (Standard Dose Head) |
| Kan 2007       | Anthropomorphic phantom  | TLD Skin dose measurement | 67 (Standard Dose Head) |
| Kan 2007       | Anthropomorphic phantom  | TLD Skin dose measurement | 64 (Thorax) |
| Kan 2007       | Anthropomorphic phantom  | TLD Skin dose measurement | 54 (Pelvis) |
| Spezi 2011     | Patient data-skin dose   | Monte Carlo            | 20-60 (Pelvis-femoral heads) |
| Spezi 2011     | Patient data             | Monte Carlo            | 25.50 (Thorax-PTV) |
|                |                          |                        | 300 (Head and Neck-mean dose) |
| Hyer 2010      | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 0.98 (Head XVI) |
| Hyer 2010      | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 16.62 (Thorax XVI) |
| Hyer 2010      | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 24.13 (Pelvis XVI) |
| Hyer 2010      | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 5.17 (Head OBI) |
| Hyer 2010      | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 6.14 (Thorax OBI) |
| Hyer 2010      | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 21.57 (Pelvis OBI) |
| Wen 2007       | IMRT QA phantom, 30cm    | TLD’s                  | 30-60 AP skin dose |
| Wen 2007       | Rando Phantom            | TLD’s                  | 100-110 (femoral heads) |
| Alaei 2010     | Rando Phantom            | TLD’s                  | 21-45 (Soft tissue regions) |
| Alaei 2010     | Rando Phantom            | TLD’s                  | 32-102 (In and near bone areas) |
| Islam 2006     | 30cm diameter cylindrical water phantom | MOSFET detectors | 16 (centre position) |
| Islam 2006     | 30cm diameter cylindrical water phantom | MOSFET detectors | 23 (peripheral average) |
| Amer 2006      | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 1.6 (Head) |
| Amer 2006      | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 6.0 (Thorax) |
| Amer 2006      | Cylindrical PMMA 15cm    | 100cm pencil chamber, CTDI Formalism | 25.0 (Pelvis) |
| Ding 2013      | Patient CT data          | Monte Carlo            | 4.4-6.6 (Head-Eye 50% dose) |
| Ding 2013      | Patient CT data          | Monte Carlo            | 10-16 (Pelvis-Bladder 50% dose) |
| Ding 2013      | Patient CT data          | Monte Carlo            | 2.4-2 (Thorax-Heart 50% dose) |

Table 2.3: Summary of publications evaluating dose from kV CBCT imaging.

scan is mathematically equivalent to the single rotation dose profile per total beam width integrated over \((-L/2, L/2)]\), termed CTDI. Here L is the integration length of the ionisation chamber along the z-axis, as illustrated in figure 2.11. Hence, average dose for a multi-slice scan can be conveniently determined with a single rotation of the X-ray tube. Typically the CTDI is measured with a 100mm length ionisation chamber
symmetrically centered across the fan plane.

Figure 2.11: Diagram of axis for imaging. The axial plane is defined as the plane in the x-y axis. Linac schematic courtesy of http://www.suggest-keywords.com

This $CTDI_{100}$ method is mathematically defined as:

$$CTDI_{100} = \frac{1}{nT} \int_{-50mm}^{50mm} D(z)dz$$  \hspace{1cm} (2.1)

Where $D(z)$ is the dose profile originating from a single axial rotation, the dose being expressed as absorbed dose in air. $n$ is the number of active acquisition channels (detector rows) and $T$ is the nominal slice thickness of each channel.

2.3.1.1 CTDI Phantom

The cylindrical CTDI phantom is constructed of Poly Methyl Methacrylate (PMMA). The body phantom is 32 cm in diameter while the head phantom has a diameter of 16 cm. The phantom is 15 cm in length [68]. Both phantoms are shown in figure 2.12.

Both phantoms have five holes for CTDI measurements, one in the centre and four on the periphery. Each hole not housing the pencil chamber holds a PMMA plug to eliminate air gaps.
2.3. Radiation Dose from kV CBCT Image Guided Radiotherapy

2.3.1.2 \textit{CTDI}_w

The weighted \textit{CTDI}_w describes the average weighted dose in the axial plane. It is calculated using 5 measurements from each position in the phantom, 1 in the centre and 4 on the periphery. Each measurement is weighted according to its position within the phantom as shown:

\[
\textit{CTDI}_w = \frac{1}{3} \textit{CTDI}_{100,\text{centre}} + \frac{2}{3} \textit{CTDI}_{100,\text{periphery}} \quad (2.2)
\]

This concept is only applicable for a series of contiguous axial scans or helical scanning with a pitch of unity. For helical scanning with pitch not equal to unity, the \textit{CTDI}_vol may be calculated.

2.3.1.3 \textit{CTDI}_vol

With the introduction of the helical CT scanner, volume CTDI or \textit{CTDI}_vol was developed and takes into account the effect of couch translation during irradiation and
the associated helical pitch. Its defined as follows:

\[ CTDI_{vol} = \frac{CTDI_w}{\text{helical pitch}} \]  

(2.3)

The \( CTDI_{vol} \) represents the average dose in the central z-axis region of the scanned volume equivalent to the integration length of the CTDI equation, defined by the couch increment per rotation (helical) or couch incrimination in axial scanning.

### 2.3.1.4 Dose Length Integral and Dose Length Product

The dose length integral is, as the name suggests, the integration of the dose profile over the scan length. This may be calculated via summation of point dose measurements stepped through the beam \[69\] or via a long integrating dosimeter such as the 100mm pencil chamber and a single rotation of the X-ray tube.

The dose length product (DLP) is an approximation of the total dose from an entire CT examination and is calculated as follows:

\[ DLP = CTDI_w \times scanlength \]  

(2.4)

Since the DLP gives an estimate of the total absorbed dose, it can potentially be related to the effective dose which has been investigated for diagnostic CT \[70\].

### 2.3.1.5 Integral Dose \( E_{tot} \)

Integral ‘dose’ \( E_{tot} \) is defined by the AAPM Task Group 111 as the total energy absorbed within the phantom measured in Joules\[12\]. It provides a simplified indicator for patient risk and is derived from the volume and density of the phantom multiplied with the planar average equilibrium dose \( D_{eq} \), which is defined in section \[2.3.3.3\].

Some definitions of the planar dose including the TG111 described below, determine the planar dose in the CTDI phantom from uniform weighting of the central and
2.3. Radiation Dose from kV CBCT Image Guided Radiotherapy

peripheral doses based on a $\frac{1}{R^2}$ relationship where $R$ is the phantom radius. However it has been shown that the $CTDI_w$ weighted approach gives a better estimate of planar dose \(^{71}\). The integral is then defined as:

$$E_{tot} = \rho \pi R^2 LD_{eq} \quad (2.5)$$

where $\rho$ is the phantom density, $D_{eq}$ is the radially averaged equilibrium dose and $L$ is the total scan length.

### 2.3.2 Limitations of CTDI

With advances in CT technology since the introduction of CTDI including helical scanning and wide cone-beam scanning, the validity of the CTDI has become increasingly limited \(^{72}\). Additionally, as trends towards wider collimation in the z-axis and longer scan lengths continue, increased amounts of scattered radiation are not detected by the 100mm pencil chamber resulting in underestimation of the dose \(^{72}\). It has been suggested that integration lengths for CBCT should be greater than 300mm for beam widths greater than 20mm to capture the full dose profile\(^{73, 74}\).

When the CTDI is measured, both the primary beam and scattered radiation within the phantom contribute to the measurement. In the IAEA Report No. 5, it’s suggested that provided the actual beam width does not exceed 80mm at isocentre and a focus to isocentre distance of 600mm, the primary beam width will not exceed the 100mm ionisation chamber for $CTDI_{100}$ measurements\(^{11}\). However, the scattered radiation field extends throughout the entire CTDI phantom and into the surrounding air. Hence, the 100mm chamber (100mm scanning length) will not collect all the scattered radiation and will underestimate the delivered dose, by as much as 40 to 50 percent compared to a 450 mm chamber, irrespective of beam width \(^{12, 29}\). Additionally, the phantom used in $CTDI_{100}$ is of insufficient length to produce scatter
equilibrium in the centre of the beam with approximately a 30 percent underestimation relative to a phantom of sufficient length for scatter equilibrium \[74\ \text{[29]}. The ratio of the \( CTDI_{100} \) to \( CTDI_{\infty} \) quantifies the variation of the \( CTDI_{100} \) compared with infinite phantom and detector length and is known as the \( CTDI \) measurement efficiency.

The relevance of CTDI for Cone-Beam CT is particularly questionable. In CBCT, the beam is sufficiently broad to image anatomy of interest in a single rotation without table translation. Thus, the use of CTDI to quantify dose isn’t appropriate as it’s defined as an integration over table translation of length L. CBCT scans are typically broader than the 100mm active length of the chamber itself for pelvis and thorax scanning modes. The beam also diverges to even greater widths due to scatter within the phantom, so the chamber does not capture a large amount of the dose profile. The literature also suggests that the CTDI concept breaks down due to beam non-uniformity and the weightings of central and peripheral dose in \( CTDI_w \) not being an accurate representation of average dose across the volume \[68\].

### 2.3.3 Alternative Protocols for Wide-Beam Dosimetry

Due to the limitations of the CTDI to quantify dose from wide-beam CT scanners, new alternative methods of determining the dose for these wide-beam scanners have been investigated, typically by either increasing the length of the ionisation chamber to capture the full dose profile or utilising a small volume detector to measure cumulative dose in the centre of the phantom \[75\ \text{[74] \[76] \[77]. This has resulted in three new protocols being developed which give an alternative method of quantifying the imaging dose for wide beam dosimetry.
2.3. Radiation Dose from kV CBCT Image Guided Radiotherapy

2.3.3.1 Cone Beam Dose Index

The Cone Beam Dose Index (CBDI) was introduced by Amer et. al. to give a better estimate of the dose from wide-beam scanning using the existing 100 mm chamber and CTDI phantoms [13].

The CBDI is representative of an average dose across the central 10 cm portion of the central axis. Mathematically, it is defined as follows:

\[
CBDI = \frac{1}{L} \int_{-50\text{mm}}^{50\text{mm}} D(z) \, dz
\]

where \(D(z)\) was previously defined in (2.1) and \(L\) in this case representing the chamber length of 100mm. The \(CBDI_w\) is defined in exactly the same way as \(CTDI_w\) (2.2). The \(CBDI_w\) can also be normalised to 100mAs by multiplying by 100 and dividing by the exposure used to measure the CBDI. Some published dose values for head, thorax and pelvis scanning using the CBDI normalised to 100mAs are listed in table 2.4.

| Study            | Head Scan (mGy/100mAs) | Thorax (mGy/100mAs) | Pelvis (mGy/100mAs) |
|------------------|-------------------------|---------------------|---------------------|
| Amer et. al. [13]| 4.3                     | 3.8                 | 5.5                 |
| Cheng et. al. [64]| 5.1                     | –                   | 4.3                 |
| Hyer et. al. (XVI)[36]| 2.73                  | 1.62               | 1.47               |
| Hyer et. al. (OBI)[36]| 3.57                   | 2.34               | 3.17               |

Table 2.4: \(CBDI_w\) studies for head, thorax and pelvis scans in mGy normalised per 100mAs. Normalised values provide a means of comparison between different imaging systems with varying mAs.

The CBDI approach has been shown to give good agreement with CTDI measurements taken with longer ionisation chambers that can measure the full profile from wider beam scanning [78], making it a useful method for quantifying the dose output from CBCT scanning using existing ionisation chambers and phantoms.
2.3.3.2 IAEA Report No. 5

Due to the limitations of the CTDI for wide beam scanning, the International Atomic Energy Agency (IAEA) recommended a protocol developed by the International Electrotechnical Commission (IEC)\(^\text{[79]}\). The protocol is based on the relatively constant efficiency of the \(CTDI_{100}\) to \(CTDI_{\infty}\) for beam widths < 60\(mm\), as shown in figure 2.13\(^\text{[72]}\).

\[
CTDI_{IAEA} = CTDI_{ref} \times \left(\frac{CTDI_{\text{in-air}}^{\text{protocol}}}{CTDI_{\text{in-air}}^{\text{ref}}}\right) 
\]

(2.7)

Figure 2.13: \(CTDI_{100}\) measurement efficiency for a 32 cm PMMA phantom as a function of scanning length for a 120kVp beam. Source \(^\text{[72]}\)

The protocol suggests a two tiered approach. For beam widths < 60\(mm\), the standard \(CTDI_{100}\) should be applied. For longer scanning lengths, the \(CTDI_{100}\) is determined for a reference beam width, recommended to be 20\(mm\). This is then scaled by the free in-air ratio of CTDI for the protocol and reference beam widths, shown below (2.7).
2.3. Radiation Dose from kV CBCT Image Guided Radiotherapy

The protocol defines minimum scanning length of 100 mm for beam widths < 60mm and 20mm either side of beam width for beams > 60mm to ensure all the primary beam is included in the $CTDI_{\text{free-in-air}}$ measurement.

Since this protocol does not include scatter contributions it still underestimates the dose in the central scan region compared to an infinite scanning and phantom length $CTDI_\infty$. Abuhaimed et. al. [30] showed the underestimation to be $\approx 18\%$ in the head phantom and $\approx 24\%$ in the body phantom for a clinical beam width of 198 mm. However, the efficiency remains constant across all beam widths beyond 40 mm, verified by Monte Carlo simulations and measurements for wide beam CT and CBCT as shown in figure 2.14 [30, 79]. By comparison, for the same conditions described above, Abuhaimed showed the underestimation of the $CTDI_{100}$ to be $\approx 55\%$ and $\approx 56\%$ respectively.

2.3.3.3 AAPM Task Group 111

The TG111 report from the American Association of Physicists in Medicine (AAPM) introduces a new method of determining CT dose following several studies investigating an alternative dosimetry approach to wide beam scanning [75, 76, 77]. Its advantage lies in its applicability to all CT scanner types [12]. Unlike the CTDI paradigm, the TG111 method stipulates the use of a small volume dosimeter for a point dose measurement at the centre of the scan length. The dose at the centre of the scan length is formed from contributions by both the primary beam and scatter tails lying beyond the primary beam. These measurements are taken for increasing scanning length to determine the equilibrium dose $D_{eq}$, where dose to the central region is maximum. Further increases in scanning length will not increase dose at the centre of the phantom as the scatter tails are sufficiently remote to make a negligible contribution. This scan length for the equilibrium dose is the equilibrium scan length $L_{eq}$. From this other parameters may be determined such as equilibrium dose constant and equilibrium
2.3. Radiation Dose from kV CBCT Image Guided Radiotherapy

Figure 2.14: Efficiency of the IAEA method compared with $CTDI_\infty$ for head(a-b) and body(c-d) phantoms for 200 and 360 degree rotations. Source: [30]

dose pitch product, which are independent of scanning parameters like scan length, pitch and collimator width [12].

The dose at $z=0$ is characterised by both the equilibrium dose $D_{eq}$ and an approach to equilibrium function $h(L)$ which describes the increasing dose at $z=0$ with increasing scan length ($L$) as shown in figure 2.15 and equation (2.9).

$$D_L(z = 0) = h(L) \times D_{eq} \quad (2.8)$$
where;

$$h(L) ≈ 1 - α exp\left(\frac{-4L}{L_{eq}}\right)$$  \hspace{1cm} (2.9)

Figure 2.15: Approach to equilibrium at the central and peripheral locations within a body PMMA phantom measured with a farmer chamber. Also shown are the approach to equilibrium functions with their respective coefficients. Source [77]

For CBCT, the protocol is simplified somewhat given there is no table motion or pitch. The scan is acquired in a single rotation and scan length is determined by collimation in the z-axis. Preliminary studies following the TG111 method have been published showing an increased dose relative to the CTDI protocol, by as much as 40% compared with the vendor stated dose [80] [8]
2.4 Low Dose Radiation and Secondary Cancer Risk

Due to advances in cancer treatment and earlier detection, cancer survival rates have dramatically improved with 5 year survival rates in the US increasing from 50% to 66% among adults and from 61% to 79% among children from 1975-79 to 1996-2002[81]. Although CBCT has not been in use clinically for sufficient time to evaluate increased cancer risk due to imaging from patient data, many studies have investigated increased secondary cancer risk from radiation exposure data and radiotherapy treatments, which have out of field dose comparable to CBCT [82, 83, 84, 85]. These doses have been linked to a statistically significant increase in secondary cancer risk [86]. The application of risk models to additional kV CBCT dose to assess increased risk from imaging has also been investigated [87].

Our understanding of radiation induced cancers is limited to studies of atomic bomb survivors, nuclear accidents, radiotherapy patients and occupationally exposed radiation workers. Conclusions drawn from these studies are presumptuous however. Underlying factors such as lifestyle or genetic pre-disposition may outweigh any increase in risk of developing a secondary malignancy due to low doses of radiation, particularly for patients already being treated for cancer. This is especially relevant to low dose studies, such as imaging [88]. Quinn et. al. [89] reported an uncertainty of 40% when estimating the Excess Absolute Risk (EAR) and Excess Relative Risk (ERR) using the Preston method for kV CBCT imaging during breast radiotherapy.

2.4.1 Studies of Atomic Bomb Survivors

Life span studies of survivors from the atomic bombings of Hiroshima and Nagasaki provide the best qualitative risk estimates for secondary malignancies following low-LET radiation exposure to humans. Unlike studies of cancer patients, it includes survivors of varying age, gender and received dose. All subjects are from a similar
demographic and considered healthy at the time of the bombings. The studies included enough members to be of statistical relevance, particularly for studying the effects of low doses [90]. They also include enough people from the area of the bombings, but not present at the time of the bombings or are believed to have received a negligible dose to make an assessment of the risk of cancer incidence independent of demographic influence. Studies based on follow up survival data from 1958 onwards form the basis for dose-risk models and radiation protection guidelines [91, 92].

However the accuracy of the studies are limited by the models used to estimate received dose, as well as tracking of survivors over time. As studies continue to be published with longer follow up and improved dose estimates risk models will however continue to improve [93].

From the survival data, Preston et. al. [93] determined \textit{Excess Relative Risk} $ERR$ and \textit{Excess Absolute Risk} $EAR$. The results took into consideration dose ($d$), age at exposure ($e$), gender ($s$), age from birth ($b$), city ($c$), location at time of A-Bomb ($l$) and attained age ($a$). $\lambda_0$ is the background cancer incidence rate (zero dose assumed).

\begin{equation}
\lambda_0(c, s, a, b, l)[1 + ERR(d, e, s, a)]
\end{equation}

\begin{equation}
\lambda_0(c, s, a, b, l) + EAR(d, e, s, a)
\end{equation}

Dose-response relationships considered were linear, linear quadratic, quadratic, linear threshold and a nonparametric model. The data indicated that a linear no-threshold model best describes the dose response curve for low dose [93, 94].
2.4.2 Secondary Cancer Risk from Radiotherapy Procedures

Unlike a cohort study, such as that of atomic bomb survival data where people of varying lifestyle, age, genetic factors etc are exposed to varying doses [88, 93], studies of radiotherapy patients are of similar dose values and the risk of a secondary cancer developing is usually higher. The relative risk of developing a secondary cancer following radiotherapy is inversely related to age of treatment and appears to increase with time following radiotherapy [95, 96, 97]. Several studies have investigated the onset of secondary cancers following radiotherapy for several cancer sites, with cancers of the cervix and the prostate having the largest amount of follow up data [96, 97, 95, 98, 99, 100].

A study of 104,760 survivors of cervical cancer found the majority of secondary malignancies were observed within the pelvic region where the highest dose occurs, including the rectum, vagina, vulva, ovary and bladder [97]. Increases in lymphatic leukemia and cancers of the bone and kidney were also observed. A smaller amount of cases were seen for moderate and lightly irradiated sites. An increased risk of lung cancer was also observed, however this is thought to be more closely related to the effects of cigarette smoking and the presence of human papillomavirus (HPV) [95, 96, 97].

Some debate exists on the risk, if any, of a secondary malignancy occurring as a result of radiotherapy for prostate cancer [100, 101]. The associated risk of a secondary malignancy following prostate radiotherapy is small, with Kendal et. al. suggesting that there is no detectable risk associated with prostate radiotherapy [102]. Rather, an increase in observed renal cancers (the most common secondary cancer site following prostate therapy) being associated with other factors such as smoking. However when considering all cancers following prostate radiotherapy compared with surgery alone, Brenner et. al. [98] reported a small but statistically significant increase in secondary
cancers, which Brenner attributes to radiation effects [101]. A 6% increase in risk of solid tumor relative to surgery was observed, while increased relative risks reached 15% and 34% following 5 and 10 years follow up respectively.

2.4.3 Risk Models for Low Dose Radiation

As mentioned above, treatment with radiotherapy has been shown to increase the risk of a secondary cancer to a statistically significant extent. Hence, it’s advantageous to limit the dose received to normal tissue including OAR as much as possible during treatment. This extends to the use of image guidance during treatment. Currently, there are no epidemiological studies relating imaging and increased secondary cancer risks, partly due to the lack of sufficient patients and follow up times for study but also due to the low dose from the imaging relative to the treatment. Indeed, any increased risk due to imaging may be negated by other factors as outlined above and a large sample size is required for adequate statistics [103] [88].

In light of this, the only method available to quantitatively assess increased risk from imaging is through the use of radiation-induced cancer risk estimation models [104] [62]. These models have been developed based on data obtained from radiation workers, medical exposure data and atomic bomb survival data [93].

The difficulty with model based estimates is the uncertainty in the dose response curve for low dose. Because of the low doses involved, to determine accurately the shape of the response curve would require an impractical number of patients for an epidemiological study to detect an increased risks since the excess risk is less for smaller doses. Alternatively, low dose response may be predicted based on the more easily obtained responses for higher doses. However such extrapolations are highly presumptuous [90]. The limit of dose uncertainty has however reduced over time. Land et. al. defined low dose as below 2,500 mSv in 1980 [90] while more recently
Brenner et. al. defines uncertainty in dose risk estimates to be below 50 mSv [94]. From atomic bomb survival data, a linear relation of dose to excess risk for the range of 5cGy-2.5Gy is observed. However, the use of a linear response for dose values below this is presumptuous due to underlying biological mechanisms that could lead to an under or overestimation of risk compared to the linear no threshold approximation [94]. Potential responses are demonstrated in figure 2.16.

![Figure 2.16: Schematic diagram of different possible radiation risks extrapolated from measured radiation risk to low doses. Curve a-linear extrapolation, curve b-downwardly curving, curve c-upwardly curving, curve d-threshold and curve e-hormetic. Data points indicate the lower limit of doses where significant risks have been demonstrated in human populations (10-50 mSv for acute exposure). [94]](image)

The bystander effect is a phenomenon where signals are sent out by radiation damaged cells to adjacent cells, causing oncogenic damage to these cells. This effect predominates at low doses and saturates at higher doses [105]. The cells may have an adaptive response, where by DNA repair mechanisms are up-regulated from a small radiation dose, reducing the radio sensitivity to larger subsequent doses [106, 107]. The bystander effect is being studied in pre-clinical trials, with responses measured for exposures as low as a single proton or He ion [108]. Its impact on current radiation therapy is yet to be investigated. These effects cause an increased risk, more representative of the downward curve b.
2.4. Low Dose Radiation and Secondary Cancer Risk

Curve c is representative of a quadratic relationship which increases with dose and has been shown to provide a good description of acute dose-effect relations for radiation-induced leukemia \cite{84}. Curve d demonstrates a threshold response, below which the risk is zero. This response is supported by the lack of radiation-induced sarcomas observed at low doses \cite{109}. Curve e represents a hormetic response, which reduced the background incidence of some deleterious endpoint.

2.4.3.1 International Commission on Radiological Protection Report 103

The International Commission on Radiological Protection (ICRP) provides a set of guidelines and recommendations for radiation protection. The report lists dose limits for radiation workers and the general public based on risk of excess fatal cancer following exposure to low-level ionizing radiation \cite{62}. The risk is determined by multiplying the effective dose following exposure by a nominal probability coefficient. The probability coefficients are primarily derived from Atomic-bomb survivor data by applying age and gender averaging for an occupational worker.

The effective dose, given in Sieverts, is defined as follows:

$$ E = \sum_T w_T \sum_R w_R D_{T,R} $$  \tag{2.12} 

where $w_T$ is the tissue weighting factor for tissue T and the sum of all tissue weighting factors $= 1$. The tissue weighting factor depends on the relative radio sensitivity of each organ. A summary of tissue weighting factors from Report 103 is given in table 2.5.

$$ \sum_R w_R D_{T,R} $$ is the equivalent dose in Sieverts ($J.Kg^{-1}$), denoted $H_T$. The equivalent dose takes the dose received by the organ and scales it by the radiation weighting
2.4. Low Dose Radiation and Secondary Cancer Risk

| Tissue                                                                 | $w_T$ | $\sum w_T$ |
|------------------------------------------------------------------------|-------|------------|
| Bone-marrow (red), Colon, Lung, Stomach, Breast, Remainder tissues*   | 0.12  | 0.72       |
| Gonads                                                                 | 0.08  | 0.08       |
| Bladder, Oesophagus, Liver, Thyroid                                    | 0.04  | 0.16       |
| Bone surface, Brain, Salivary glands, Skin                            | 0.01  | 0.04       |
| Total                                                                  | 1.00  |            |

* Remainder tissues: Adrenals, Extrathoracic (ET) region, Gall bladder, Heart, Kidneys, Lymphatic nodes, Muscle, Oral mucosa, Pancreas, Prostate, Small intestine, Spleen, Thymus, Uterus/cervix

Table 2.5: ICRP 103 Tissue Weighting Factors

factor $w_R$ which depends on the relative quality of the radiation which delivered the dose. For photons, electrons and muons, the *Linear Energy Transfer* is less than $10 \text{ keV} \mu\text{m}^{-1}$ and has a $w_R$ value of 1. This value is higher than 1 for higher LET particles such as protons, ions and neutrons.

The Report 103 models were developed for radiation protection purposes in the low-dose range (less than 2.5 Sv). During radiotherapy, organs close to the target volume may receive moderate to high doses, which is not included in the risk estimates. Additionally, cancer patients receiving radiotherapy are generally older so the age averaged coefficient may not be appropriate.

2.4.3.2 Biological effects of ionising radiation report VII phase 2

The BEIR Report VII published in 2006, applies a linear-no-threshold model to assess the ERR and EAR following exposure to low-dose, low LET radiations [104]. Unlike the ICRP models, age, sex and irradiated tissue effect the risk calculations, rather than taking an average of epidemiological data. The EAR and ERR are combined to estimate the likelihood of a radiation induced cancer over the lifetime of the individual,
2.4. Low Dose Radiation and Secondary Cancer Risk

so called the Lifetime Attributable Risk (LAR) and is defined as:

\[ LAR(D, e, a) = \sum_{a}^{100} M(D, e, a) \frac{S(a)}{S(e)} \]  

(2.13)

Where \( e \) and \( a \) are age at exposure and attained age, respectively. \( M(D, e, a) \) is the logarithmic weighted average of ERR and EAR, and \( \frac{S(a)}{S(e)} \) is the probability of surviving from ages \( e \) to \( a \).

To use risk analysis a clinical site is usually chosen for parameter selection. Anthropomorphic dosimetry phantom or Monte Carlo simulations provide more detailed dose for a given organ site. Because measurements in this thesis are restricted to central and peripheral doses in a cylindrical phantom and dose profile measurements not in organ specific sites, we have not calculated risk from dose measurement in this thesis.
Chapter 3

Evaluating CTDI, CBDI, IAEA and TG111 Methodologies for Clinical CBCT Settings

Three alternative methodologies to the CTDI, the CBDI, IAEA and TG111 were evaluated on two commercial CBCT imaging systems, the Varian OBI\textsuperscript{®} on the C-Series platform and XI\textsuperscript{®} on Truebeam. Each methodology was compared for the main clinical settings on each system for head and neck, thorax and pelvis imaging.

3.1 Measuring the CTDI and CBDI

3.1.1 Phantom Design and Materials

A standard 32cm diameter body phantom and 16cm diameter head phantom were used for CTDI and CBDI measurements, as shown in figure 3.1. The phantom is constructed from Poly methyl Methacrylate (PMMA) with a density of 1.18 $\frac{g}{cm^3}$ and an electron density of $3.65 \times 10^{23} \frac{electrons}{cm^3}$.
Both phantoms have insert spaces for a 100mm pencil ionisation chamber at the centre and at the four peripheral locations shown in figure 3.1. The phantom was placed on the couch and held in place with tape and low density foam blocks during measurements. The phantom was aligned to the centre markings on the phantom using the lasers in the room.

![PMMA 32cm CTDI body phantom with the pencil chamber in the central position. For head protocols the 16cm inner phantom was used.](image)

Figure 3.1: PMMA 32cm CTDI body phantom with the pencil chamber in the central position. For head protocols the 16cm inner phantom was used.

### 3.1.1.1 UNFORS RaySafe\textsuperscript{TM} Xi Measurement Device

CTDI and CBDI measurements were taken using the UNFORS Xi detector system from RaySafe\textsuperscript{TM}. The system includes a base unit which connects to several detectors including a CT detector and HVL measurement tool for kV energies shown in figure 3.2. The base unit supports both RS-232/USB and blue tooth for data communication.

The pencil ionisation chamber for CT measurement has a sensitive length of 100mm. The ionisation chamber and electronics are combined, with temperature and pressure
3.1. Measuring the CTDI and CBDI

Figure 3.2: The RaySafe Xi system used for CTDI and CBDI measurements. The CT pencil chamber detector has a sensitive length of 100mm. The HVL tool measures beam quality for kV energies. Both connect to the base unit which gives a readout with temperature and pressure corrections applied. Source: www.raysafe.com

measured by the device and the dose adjusted automatically. The temperature is measured inside the chamber sensitive length ensuring precise temperature measurements in a CTDI phantom or free in air. The detector has a range of 10 $\mu$Gy to 9999 Gy with an uncertainty of $\pm$5%. Its energy dependance is $< 5\%$ with an axial and radial uniformity of $\pm 2\%$ and $\pm 3\%$ respectively [110].

Measured data can be viewed on the base units back lit alphanumerical display and is stored in XML or CSV format. With the Xi view program installed, data can be transferred to a computer directly into a Microsoft Excel$^{\text{TM}}$ document.

3.1.2 OBI and XI systems

Both Varian$^\text{®}$ kilovoltage CBCT imaging systems (OBI and XI) are mounted on the LINAC gantry orthogonal to the treatment beam. Both systems consist of a kV X-ray
3.1. Measuring the CTDI and CBDI

tube and a flat panel detector 180 degrees to the tube. Each system can acquire 2D kV and 3D CBCT images by acquiring projections as the source and detector are rotated either 200 or 360 degrees. The parameters for each clinical mode for the 2 scanners are given in tables [3.1] and [3.2].

|                          | Standard-Dose Head | Low-Dose Head | High-Quality Head | Pelvis | Low-Dose Thorax |
|--------------------------|--------------------|---------------|-------------------|--------|-----------------|
| X-ray Voltage (kVp)      | 100                | 100           | 100               | 125    | 110             |
| X-ray Current (mAs)      | 20                 | 10            | 80                | 80     | 20              |
| X-ray Milliseconds (ms)  | 20                 | 20            | 25                | 13     | 20              |
| Gantry Rotation (deg)    | 200                | 200           | 200               | 360    | 360             |
| Number of Projections    | 360                | 360           | 360               | 665    | 665             |
| Exposure (mAs)           | 145                | 72            | 720               | 680    | 262             |
| $CTD_{w}$ (mGy/100 mAs)  | 2.7                | 2.7           | 2.7               | 2.6    | 1.8             |
| Dose $CTD_{w}$ (mGy)     | 3.9                | 2.0           | 19.4              | 17.7   | 4.7             |
| Fan Type                 | Full Fan           | Full Fan      | Full Fan          | Half Fan | Half Fan       |
| Bow-tie Filter           | Full               | Full          | Full              | Half    | Half            |

Table 3.1: Modes and parameters for the OBI CBCT imaging system. [49]

|                          | Head | Pelvis | Thorax |
|--------------------------|------|--------|--------|
| X-ray Voltage (kVp)      | 100  | 125    | 125    |
| X-ray Current (mAs)      | 20   | 80     | 20     |
| X-ray Milliseconds (ms)  | 20   | 20     | 20     |
| Gantry Rotation (deg)    | 200  | 360    | 360    |
| Number of Projections    | 660  | 660    | 660    |
| Exposure (mAs)           | 147  | 1056   | 264    |
| $CTD_{w}$ (mGy/100 mAs)  | 1.94 | 1.32   | 1.32   |
| Dose $CTD_{w}$ (mGy)     | 2.85 | 13.94  | 3.48   |

Table 3.2: Modes and parameters for the Truebeam XI CBCT imaging system. [48]

The beam width is modified with independently adjustable X and Y lead blade
collimators. The field size at isocentre can be varied from 2.0 mm x 2.0 mm to 50.0 cm x 50.0 cm for both the XI and OBI systems. The X1,Y1 collimators have a range of -25.0 cm to +3.5 cm and the X2, Y2 from -3.5 cm to +25 cm. In the XI system, a titanium beam hardening foil filter further hardens the X-ray spectrum to reduce low energy photons.

The axial plane is further attenuated via a bow-tie filter. It shapes and filters the beam to reduce patient skin dose, X-ray scatter and compensates for the curvature of the patient body. The filters are constructed from aluminum and vary in thickness from 2mm to 28mm. For head imaging a full bow-tie filter is used to provide symmetric field scans while for body scans such as pelvis and thorax the half-fan is used with an asymmetric field. The bow-tie filter is in-built into the XI system and will insert the appropriate filter for the chosen scan, while the filter is applied to the kV unit manually for the OBI system so care must be taken to ensure the appropriate filter is in place when scanning.

### 3.1.3 Determination of CTDI and CBDI

Dose was measured by the pencil chamber for a CBCT scan in each position within the phantom. This dose value represents the cumulative dose across the 100mm length and multiplying by the chamber length yields the dose length integral (DLI).

\[
DLI[\text{mGy.mm}] = D_{\text{measured}} \times L_c = \int_{-50\text{mm}}^{+50\text{mm}} D(z)dz
\]

(3.1)

where \(D_{\text{measured}}\) is the integral dose collected in scanning length \(L_c = 100\text{mm}\).

This value was determined for each position within the phantom. To obtain the CTDI, the DLI was divided by the collimation in the z-axis:

\[
CTDI[\text{mGy}] = \frac{1}{\text{coll}} \times DLI
\]

(3.2)
where \( \text{coll} \) is the collimation width at isocentre and \( D(z) \) the dose profile.

To obtain the CBDI, the DLI was divided by the 100mm chamber length \( L_c \) following the CBDI protocol:

\[
CBDI [\text{mGy}] = \frac{1}{100 \text{mm}} \times DLI
\]  

(3.3)

Hence, the CBDI values for the protocol scans are given simply by the dose measured in the chamber since scan length = chamber length.

\[
CBDI [\text{mGy}] = D_{\text{measured}}
\]  

(3.4)

The weighted \( CTDI_w \) was determined by applying the weighted CTDI formalism defined in the literature for CTDI and CBDI measurements:

\[
CTDI_w [\text{mGy}] = \frac{1}{3} CTDI_c + \frac{2}{3} CTDI_p
\]  

(3.5)

These values were then normalised per 100mAs using equation (3.6):

\[
nCTDI_w [\text{mGy.100mAs}^{-1}] = \frac{CTDI_w \times 100}{E}
\]  

(3.6)

Where \( E \) is the scan mAs. The mAs for each of the protocols was taken from the Varian® imaging manuals.

### 3.1.3.1 Fan-Beam CT Measurements

Initial CTDI measurements were taken on a Siemens SOMATOM CT scanner for several scan settings to compare measured CTDI values with scanner predicted values for narrow beam axial scanning. It was noted that the scanner value was quoted as a \( CTDI_{\text{vol}} \), however since the scans were acquired with a single rotation in axial mode there was no pitch to divide by, hence \( CTDI_w \) was taken as \( CTDI_{\text{vol}} \). The phantom
was fixed to the couch with tape and held in position with low density foam blocks as shown in figure 3.1. The scan parameters are given in table 3.3.

| Protocol                  | kV  | mAs | slice width (mm) | width |
|---------------------------|-----|-----|------------------|-------|
| Prostate Seeds 1          | 120 | 180 | 5                |       |
| Prostate Seeds 2          | 120 | 180 | 10               |       |
| Extreme Clinical Limits   | 140 | 300 | 5                |       |
| Head and Neck             | 120 | 220 | 5                |       |

Table 3.3: Scan parameters for CTDI calculations on SOMATOM CT

An average of five measurements was taken for the five positions within the CTDI phantom for $CTDI_w$ calculations. The $CTDI_w$ and $^nCTDI_w$ were then calculated for four clinically applicable settings and are presented in table 3.4.

| Mode                              | $CTDI_w$ (mGy) | $^nCTDI_w$ (mGy.100mAs$^{-1}$) |
|-----------------------------------|----------------|--------------------------------|
| Prostate Seeds 5mm slice          | 12.11 ± 0.03   | 6.73 ± 0.02                    |
| Prostate Seeds 10mm slice         | 12.75 ± 0.04   | 7.08 ± 0.02                    |
| Head and Neck-body phantom        | 14.98 ± 0.03   | 6.81 ± 0.01                    |
| Head and Neck                     | 29.84 ± 0.03   | 13.56 ± 0.01                   |
| Extreme Clinical Limits           | 29.88 ± 0.05   | 9.96 ± 0.02                    |

Table 3.4: Calculated $CTDI_w$ and $^nCTDI_w$ in mGy for clinical protocols on a SOMATOM CT scanner. The normalized values are per 100mAs. Quoted uncertainties represent 1 standard deviation.

The Extreme Clinical Limits and Head and Neck (measured in the 16 cm diameter phantom) protocols gave the highest values of 29.88 ± 0.05 mGy and 29.84 ± 0.03 mGy respectively, more than double the other measured protocols which varied between 12.11 ± 0.03 mGy and 14.98 ± 0.03 mGy. The higher dose from the Extreme Clinical Limits and Head and Neck (measured in head phantom) protocols are attributed to a higher mAs and less attenuation from to the smaller phantom in the case of the Head and Neck protocol scan.

The $^nCTDI_w$ values varied by 6.84 mGy. The Head and Neck protocol in the 16 cm phantom gave the highest value of 13.56 ± 0.01 mGy.
3.1. Measuring the CTDI and CBDI

A comparison between the measured $CTDI_w$ with the $CTDI_{vol}$ values from by the CT scanner for each mode are presented in figure 3.3.

![Figure 3.3: Measured CTDIw and CTDIvol values taken from the scanner for Prostate Seeds with 5mm and 10mm slice width, Head and Neck and Extreme Clinical Limits modes. The measurements for the Head and Neck protocol were taken in the 32cm diameter CTDI phantom. The error bars represent 2 standard deviations or 95% CI.](image)

The quoted scanner values were higher than the measured CTDI values across all clinical settings by an average of 2.65 mGy. The highest variation was 5.01 mGy for the Extreme Clinical Limits scan due to its higher mAs.

3.1.3.2 CBCT Measurements

The $CTDI$, $CTDI_w$ and $CTDI_{vol}$ were determined on two Varian® imaging systems, the On-Board Imager® and the Truebeam X-Ray Imaging System (XI)®. The CTDI body phantom was used for pelvis and thorax scans, and the CTDI-head phantom for all head protocols. These values were compared to determine the difference in dose
3.1. Measuring the CTDI and CBDI

between the two imaging systems. The kV, mAs and collimation widths are given in tables 3.5 and 3.6.

When the imaging system begins, there is a small delay between the beam starting and the gantry rotating, so the side of the phantom where the tube begins its rotation receives a higher dose. To reduce the effect of gantry start/stop position on the measurement, two scans were acquired for each chamber position in the phantom, one with the gantry rotated clockwise and the other counter-clockwise. An average of these values was then taken to determine \( CTDI \), \( CTDI_w \) and \( nCTDI_w \).

| Protocol          | kV   | mAs | collimation width (mm) | collimation width (mm) |
|-------------------|------|-----|------------------------|------------------------|
| Pelvis            | 125  | 680 | 206                    | 303                    |
| Thorax            | 110  | 262 | 206                    | 303                    |
| Standard Head     | 100  | 145 | 184                    | 272                    |
| High Dose Head    | 100  | 720 | 184                    | 272                    |
| Low Dose Head     | 100  | 72  | 184                    | 272                    |

Table 3.5: Scan parameters for CBCT protocols on the OBI system

| Protocol           | kV   | mAs | collimation width (mm) | collimation width (mm) |
|--------------------|------|-----|------------------------|------------------------|
| Pelvis             | 125  | 1056| 214                    | 281                    |
| Pelvis Obese       | 140  | -   | 214                    | 281                    |
| Thorax             | 125  | 264 | 214                    | 281                    |
| Head               | 100  | 147 | 214                    | 280                    |

Table 3.6: Scan parameters for CBCT protocols on the XI system. The mAs for the Pelvis Obese scan couldn’t be determined as the parameters were not available from the Truebeam® imaging manual.

The calculated \( CTDI_w \) and normalised \( nCTDI_w \) for OBI and XI are presented in table 3.7.

The XI Pelvis Obese protocol measured the highest with \( 20.743 \pm 1.475 \) mGy. The Pelvis protocols gave similar values on both systems, with the XI system measuring
3.1. Measuring the CTDI and CBDI

| Protocol           | $CTDI_w$ (mGy)    | $^{n}CTDI_w$ (mGy.100mAs$^{-1}$) |
|--------------------|------------------|----------------------------------|
| Pelvis OBI         | 9.408 ± 0.029    | 1.384 ± 0.004                    |
| Pelvis XI          | 9.646 ± 0.134    | 0.913 ± 0.056                    |
| Pelvis Obese XI    | 20.743 ± 1.475   | - -                              |
| Thorax OBI         | 2.536 ± 0.004    | 0.968 ± 0.002                    |
| Thorax XI          | 2.302 ± 0.014    | 0.879 ± 0.006                    |
| Std Head OBI       | 2.236 ± 0.003    | 1.542 ± 0.002                    |
| Low Head OBI       | 1.184 ± 0.005    | 1.644 ± 0.007                    |
| High Head OBI      | 11.349 ± 0.046   | 1.576 ± 0.006                    |
| Head XI            | 1.648 ± 0.002    | 1.121 ± 0.005                    |

Table 3.7: Calculated $CTDI_w$ and $^{n}CTDI_w$ values in mGy for clinical CBCT protocols on the OBI and XI systems. Uncertainties represent 1 standard deviation.

slightly higher with 9.646 ± 0.134 mGy compared with 9.408 ± 0.029 mGy on the OBI. For the Thorax protocols the OBI system gave a slightly higher value of 2.536 ± 0.004 mGy compared with the XI value of 2.302 ± 0.014 mGy. The OBI system has three Head Protocols available with varying mAs, given in table 3.5. The differences in mAs were reflected in the measured dose values which varied between 11.349 ± 0.046 mGy for High Dose Head down to 1.184 ± 0.005 mGy for the Low Dose Head protocol. Comparing the mid range Standard Dose Head OBI protocol with the Head protocol of the XI system, the OBI protocol recorded the higher value of 2.236 ± 0.003 mGy compared with 1.648 ± 0.002 mGy for the XI system.

The normalized $^{n}CTDI_w$ values were more consistent across all protocols, varying between 0.879 ± 0.006 $mGy.100mAs^{-1}$ for XI Thorax to 1.644 ± 0.007$mGy.100mAs^{-1}$ for Low Dose Head OBI. Generally the OBI protocols gave higher $^{n}CTDI_w$ values with an average of 1.298 $mGy.100mAs^{-1}$ across the Pelvis, Thorax and Standard Head protocols compared with 0.971 $mGy.100mAs^{-1}$ for the corresponding XI protocols. The mAs for the Pelvis Obese protocol was not given by the Varian Imaging manual so its normalized value was not determined. The higher normalised values for the OBI system reflect the higher contribution of low energy photons in the spectrum to delivered dose, given the measurement conditions were identical and the kVp between
the two imaging systems are equivalent for pelvis and head protocols, and 15 kV different for thorax.

The $CBDI_w$ and $^nCBDI_w$ were calculated using equations 3.5 and 3.6 and are presented in table 3.8.

| Protocol       | $CBDI_w$ (mGy) | $^nCBDI_w$ (mGy.100mA$^{-1}$) |
|----------------|---------------|-------------------------------|
| Pelvis OBI     | 19.381 ± 0.029| 2.850 ± 0.004                 |
| Pelvis XI      | 20.642 ± 0.134| 1.955 ± 0.056                 |
| Pelvis Obese XI| 44.390 ± 1.475| - -                           |
| Thorax OBI     | 5.225 ± 0.004 | 1.994 ± 0.002                 |
| Thorax XI      | 4.927 ± 0.014 | 1.866 ± 0.006                 |
| Std Head OBI   | 4.605 ± 0.003 | 3.176 ± 0.002                 |
| Low Head OBI   | 2.439 ± 0.005 | 3.388 ± 0.007                 |
| High Head OBI  | 23.380 ± 0.005| 3.247 ± 0.006                 |
| Head XI        | 3.527 ± 0.002 | 2.399 ± 0.005                 |

Table 3.8: Calculated $CBDI_w$ and $^nCBDI_w$ values in mGy for clinical CBCT protocols on the OBI and XI systems.

The trends described above for $CTDI_w$ and $^nCTDI_w$ also follow for CBDI. However the CBDI values are more than twice the CTDI, with an increase of 106 % for OBI and 114 % for XI protocols.

The significant difference between CTDI and CBDI is attributed to the breakdown of the CTDI principle of measuring the integral dose profile, since for CBCT the primary beam width exceeds the integration length of the ionization chamber. By dividing the DLI by the chamber sensitive length rather than beam width as shown in equation 3.3 the values are up scaled and hence give a more realistic dose estimate. For clinical practicality the CBDI was performed with only the CTDI phantom and no additional scatter material, as described in the original procedure [13]. The lack of scatter will produce a small decrease in the measured dose, and should be taken into consideration when interpreting these results.
3.2 IAEA Report 5 Protocol

The biggest limitation of the CTDI method is its underestimation of patient dose due to scatter dose beyond the 100 mm detector sensitive volume not being included in the measurement. It has been shown that this underestimation is relatively constant for beam widths up to 40 mm at isocentre, but decreases for beam widths greater than this as shown in figure 2.13 [72]. The IAEA recommends a new protocol which takes advantage of the constant efficiency of CTDI for beam widths < 40 mm and translates it to wide beam scanning, as described in section 2.3.3.2.

The protocol stipulates CTDI be determined for a reference beam width not exceeding 40 mm. This value is then scaled by the ratio of the free in-air CTDI values for the same reference beam, and the width of the beam for which the CTDI is to be determined:

\[
CTDI_{IAEA} = CTDI_{ref} \times \left( \frac{CTDI_{in-air}^{protocol}}{CTDI_{in-air}^{ref}} \right)
\]  

(3.7)

Reference scans were taken for the same kV, mAs and axial collimation as the protocol scans, but the z-axis collimation was set to the 20 mm, recommended by the IAEA [11] for the reference width, as it’s well within the linear efficiency range for CTDI.

The existing 100 mm ionisation chamber was used for the free in-air measurements. For the protocol widths, the IAEA specifies a minimum scan length of the beam width + 40 mm, or 20 mm either side of the beam. For half-fan protocols the beam width was 206 mm for OBI and 214 mm for the XI system, with a minimum scan requirement of 246 mm and 254 mm respectively. Given the chamber length of 100 mm, the chamber was stepped through the beam in 100 mm increments for a total scan length of 300 mm, as shown below:

The dose for each position, multiplied by the chamber length \( L_c \) is then summed.
Figure 3.4: Diagram demonstrating the measurement of CTDI free in-air stepping the 100 mm chamber in increments equal to the chamber length. Source: [11]

to give the total DLI, which then divided by the slice width yields $CTDI_{in-air}$

$$CTDI_{free\text{--}air} = \frac{L_c}{n \times T} \sum_{i=1}^{i=3} D_i$$  \hspace{1cm} (3.8)

The free in-air measurement assumes no scatter contribution to the dose measurements. Hence, the ionisation chamber was held away from the couch by a retort stand and support rod. The distance from the couch was made to be greater than half the beam width, as specified by the IAEA. The set up is shown in figure 3.5.

Free in-air measurements were taken for reference and protocol width beams for the major clinical protocols on the OBI and IX imaging systems. The $CTDI_{IAEA}$ was then determined from equation (3.7) and the reference width CTDI values. The weighted $CTDI_{IAEA}$ was calculated from equation (3.5). The peripheral values were scaled by the same in-air CTDI since there is no phantom attenuation.
3.2. IAEA Report 5 Protocol

Figure 3.5: 100 mm chamber extended from couch by support structure for free in-air measurements. The distance of the chamber from the couch is greater than half the beam width

3.2.1 IAEA Measurements

The $CTDI_{in-air}$ values and their ratios for the OBI and XI systems are presented in table 3.9 and the weighted $CTDI_{IAEA}$ and normalised $CBDI_{IAEA}$ for the OBI and XI imaging protocols in table 3.10. The Thorax mode on the OBI system could not be evaluated for the IAEA protocol due to signal from the reference width scan in the PMMA phantom being too small to be detected by the UNFORS detector.

The average ratio for the OBI system was 16% higher than the XI system. The difference between OBI and XI is due to the XI system having additional beam filtration from the titanium filter which hardens the spectra, removing the lower energy photons. Thus the photon fluence is larger in the OBI beam and more energy is deposited within the pencil chamber. The difference is further emphasized by the long
3.2. IAEA Report 5 Protocol

| Protocol       | $CTDI_{in-air}^{ref}$ | $CTDI_{in-air}^{protocol}$ | Ratio  |
|----------------|-----------------------|-----------------------------|--------|
| Pelvis OBI     | 57.875                | 74.041                      | 1.279  |
| Pelvis XI      | 67.100                | 69.127                      | 1.030  |
| Pelvis Obese XI| 139.150               | 143.511                     | 1.031  |
| Thorax OBI     | 17.005                | 21.058                      | 1.238  |
| Thorax XI      | 16.415                | 16.935                      | 1.032  |
| Std Head OBI   | 7.890                 | 9.489                       | 1.203  |
| Low Head OBI   | 4.081                 | 5.024                       | 1.231  |
| High Head OBI  | 38.730                | 47.332                      | 1.222  |
| Head XI        | 5.105                 | 5.291                       | 1.036  |

Table 3.9: Calculated $CTDI_{in-air}^{ref}$ and $CTDI_{in-air}^{protocol}$ values in mGy for clinical CBCT protocols and the ratio of the two for the OBI and XI systems.

Overall the Pelvis Obese mode delivered the highest value of $32.295 \pm 0.017$ mGy due to its high mAs. The Low Dose Head mode recorded the lowest value of $1.726 \pm 0.007$ mGy, with 72 mAs per scan. The High Dose Head mode delivered a much higher dose of $15.316 \pm 0.002$ mGy compared with the other Head modes, but delivered the lowest dose per mAs of all the head protocols measured.

The $IAEA_w$ values were $3.572$ mGy higher for OBI pelvis compared with XI pelvis. OBI standard head mode and XI head mode were within $0.01$ mGy. The higher OBI values for pelvis modes were attributed to lower $CTDI_{in-air}^{ref}$ values and higher

| Protocol       | $IAEA_w$     | "$IAEA_w$"    |
|----------------|--------------|---------------|
| Pelvis OBI     | 18.343 ± 0.002 | 2.698 ± 0.004 |
| Pelvis XI      | 14.771 ± 0.001 | 1.399 ± 0.000 |
| Pelvis Obese XI| 32.295 ± 0.017 | - -           |
| Thorax XI      | 4.441 ± 0.001  | 1.682 ± 0.000 |
| Std Head OBI   | 3.747 ± 0.018  | 2.584 ± 0.002 |
| Low Head OBI   | 1.726 ± 0.007  | 2.397 ± 0.007 |
| High Head OBI  | 15.316 ± 0.002 | 2.127 ± 0.006 |
| Head XI        | 3.741 ± 0.033  | 2.545 ± 0.111 |

Table 3.10: Calculated $IAEA_w$ and "$IAEA_w$" values in mGy and mGy.100mAs$^{-1}$ for clinical CBCT protocols on the OBI and XI systems. Uncertainties represent 1 standard deviation.
3.3 AAPM TG111 Methodology

The TG111 report describes a new method for quantifying dose for any form of computed tomography imaging. Unlike the methods described above, the dose about the center of the scan length $z = 0$ is determined by a point dose measurement.

For CBCT, where there is no table translation and the image is acquired in a single rotation of the x-ray tube, the methodology is simplified to measuring the central dose $d(z=0)$.

3.3.1 Phantom Design and Materials

3.3.1.1 TG111 Phantom design and Construction

The TG111 protocol allows flexibility in phantom material, radial dimensions and even shape of the phantom. It also does not specify to which medium the dose values are reported. The only requirement is that the phantom be of sufficient length to achieve scatter equilibrium and be of uniform composition.

To fulfill the scatter requirements of the TG111 protocol, a new phantom was manufactured at the University of Wollongong and is shown in figure 3.6.

The cylindrical PMMA phantom was constructed to match the 32cm diameter of
3.3. AAPM TG111 Methodology

(a)

(b)

Figure 3.6: TG111 phantom constructed in-house at the University of Wollongong. (a) The length was made to be 45cm to reach scatter equilibrium in the central position. (b) The cylindrical shape and diameter of 32cm is the same as the CTDI phantom. There are 5 plugs within the phantom for $CTDI_w$ measurements.

The phantom was constructed from 10 cylindrical slices 4.7cm in thickness. Each slice was milled and the holes drilled independently, then glued together using custom made metal plugs to align all the slices as shown in figure 3.7.

3.3.1.2 Chamber Plug Design

The plug housing the ionisation chamber was constructed to conform to the chamber geometry to eliminate air gaps around the sensitive volume and minimise air gaps around the cable. The plug contains three sections. The rear section has a bore hole through which the cable feeds out of the phantom, the middle section which houses the chamber and the end section which is solid. The other 4 holes contain solid rods that can be taken out and rotated with the chamber plug.
3.3. AAPM TG111 Methodology

Figure 3.7: (a) Each slice was individually cut and milled to house plus for ionisation chamber measurements. (b) One of the 4 PMMA plugs that fill the unused chamber holes. (c) Single phantom slice side on. The width of each slice was 4.7 cm.

Figure 3.8: Plug housing 0.6cc farmer ionisation chamber in the TG111 phantom. The plug contains 3 sections, one to run the cable out of the phantom, a middle section milled to conform to the chamber and a solid front section to fill the plug housing.

3.3.1.3 Charge Collection

The chamber used for the TG111 measurements was a 0.6cc NE 2571 Farmer ionisation chamber. The sensitive air volume has a length of 24mm and radius 3.2mm. It is enclosed by a graphite thimble of thickness $0.065g.cm^{-2}$. The chamber operates at a bias voltage of 300V between the stem and chamber wall.
The chamber was connected to a PTW UNIDOSE Webline Type T10023 electrometer. The ionisation chamber and electrometer combination are calibrated in the kV energy range against the Australian primary standard at the Australian Radiation Protection and Nuclear Safety Agency (ARPANSA).

Following the TG111 protocol for stationary table translation, a single CBCT was acquired for each clinical protocol on the OBI and XI imaging systems. Only half-fan protocols were tested as no head phantom was constructed. Two measurements performed at different rotation direction were taken for each position within the phantom and averaged to eliminate gantry start-stop effects.

The weighted value was determined in the same way as for the CBDI and IAEA protocols using equation 3.5.

### 3.3.1.4 Conversion of Charge Readings to Dose

The reading from the ionisation chamber is given as a charge value in nano-coulombs. The conversion from charge to dose was performed in accordance with the AAPM TG-61 protocol [111] and can be determined for absorbed dose to a given material as follows:

\[
D_w \approx qN_K \left( \frac{\bar{\rho}}{\rho} \right)_{\text{material}}^{\text{air}}
\]

(3.9)

where \(q\) is the total charge collected by the ionisation chamber (C) (with the appropriate temperature and pressure corrections) and \(N_K\) is the air KERMA calibration factor (mGy.nC\(^{-1}\)) for the ionisation chamber. \(N_K\) is dependent on beam quality measured by HVL. \(N_K\) was interpolated from the ARPANSA calibration certificate for the appropriate HVL of the imaging beams, however \(N_K\) did not vary significantly within this range of HVL’s. The \(\left( \frac{\bar{\rho}}{\rho} \right)_{\text{material}}\) term is the ratio of the spectrally averaged mass-energy absorption coefficients of the phantom material to air. Unless otherwise
stated, all dose measurements are quoted for air as the reference material. Hence \( \frac{\rho_{\text{material}}}{\rho_{\text{air}}} = 1 \). This approach is consistent with IAEA dose formalisms in diagnostic radiology [112].

The temperature and pressure corrections were applied following the TG61 protocol [111]:

\[
\text{Correction factor} = \frac{P_{\text{ref}} (T[\degree C] + 273.2)}{P (T_{\text{ref}}[\degree C] + 273.2)}
\] (3.10)

The HVL of the imaging beams was measured using the UNFORS HVL measurement tool shown in figure 3.2. The Unfors greatly simplified HVL determination compared with a direct measurement, which requires several measurements with increasing amounts of lead attenuator [111]. The accuracy of the device was verified by measuring HVL for several beam qualities on an orthovoltage unit with known HVL values previously determined following the TG61 protocol [111].

Figure 3.9: (a) Determination of the HVL on the orthovoltage unit using the UNFORS HVL tool. A 5cm diameter cone is attached at 30cm SSD. (b) Back scatter correction measurements were taken with the 10 cm diameter cone on a low-scatter stand.

The set up is shown in figure 3.9 (a). The HVL was determined for 3 filters to
cover a broad range of HVL’s corresponding to the expected HVL’s for CBCT beams. 3 measurements were taken for each filter and an average value determined.

The HVL tool had a trigger delay set to 2s to avoid measuring in the ‘ramp up’ phase of beam delivery as shown in figure 3.10. 25MU was delivered in the measurement.

Figure 3.10: kV tube output as a function of time for the ortho-voltage measurement. By introducing a 2s trigger delay a uniform kV was obtained for the HVL measurement.

Measurements were also taken with the HVL tool elevated with a low-scatter stand to evaluate back-scatter contributions as shown in figure 3.9 (b). Variation in HVL due to residual scatter was in the order of 0.1mmAl so was assumed small enough not to be included in error calculations.

The measured HVL’s from the UNFORS and the reference HVL values are shown in table 3.11.

| Filter | kV  | mA | 5cm collimator (mmAl) | expected (mmAl) |
|--------|-----|----|-----------------------|----------------|
| 1      | 100 | 28 | 3.58                  | 3.67           |
| 2      | 125 | 23 | 6.15                  | 6.30           |
| 3      | 150 | 14 | 9.67                  | 9.67           |

Table 3.11: Calibration HVL values for settings tested on the Ortho-Voltage unit. The calibrated values were determined from machine QA following the TG61 protocol.
After verifying the UNFORS HVL tool was in agreement with the known HVL of the ortho-voltage beam, measurements were acquired for the OBI and XI systems. The tool was placed at isocentre on the couch with the cross-hairs aligned with the sensitive region of the detector as shown in 3.11.

Figure 3.11: Set-up for the measurement of HVL on the CBCT imaging systems. The UNFORS HVL tool was placed on the treatment couch at isocentre and scanned under CBCT clinical mode parameters in kV fluoroscopy mode. The detector was oriented orthogonally to the cathode-anode direction.

Two measurements were taken for each set of clinical CBCT modes and the values averaged. The scan was acquired in kV fluoroscopy mode.

### 3.3.2 Determining air KERMA Conversion Values

The ARPANSA air KERMA calibration factors were entered into MATLAB® and a curve fit to the data using the MATLAB curve fitting program as shown in figure 3.12.
3.3. AAPM TG111 Methodology

Figure 3.12: Curve fit of air KERMA per PC from ARPANSA calibration report.

Using the fit function air KERMA conversion factors were determined from the measured HVL values and are listed in table 3.12.

| Protocol       | HVL (mmAl) | Correction Factor $N_K$ (mGy.nC$^{-1}$) |
|----------------|------------|-----------------------------------------|
| OBI Thorax     | 5.20       | 41.629                                  |
| OBI Pelvis     | 5.84       | 41.618                                  |
| XI Pelvis      | 8.43       | 41.591                                  |
| XI Pelvis Obese| 8.94       | 41.565                                  |
| XI Thorax      | 8.23       | 41.578                                  |

Table 3.12: HVL and corresponding air KERMA corrections for the farmer chamber. The corrections were determined from the curve fit of the ARPANSA data.

3.3.3 TG111 Measurements

The weighted TG111 values are shown in table 3.13 for OBI and XI systems.

The highest $\text{TG111}_w$ measured was 48.529 mGy for XI Pelvis Obese mode and the lowest value 5.540 mGy for Thorax XI mode. The XI measured 5% higher than OBI for Pelvis mode while for Thorax the OBI system was higher by 20%. The normalised $n\text{TG111}_w$ was higher for Pelvis and Thorax modes on the OBI system by 32% and 21% respectively. The reasons behind the variations for Thorax modes and normalised values are due to the larger component of low energy photons in the OBI system. For
3.4. Comparison of Methodologies

A comparison of CTDI, CBDI, IAEA and TG111 for Pelvis and Thorax modes on OBI and XI systems are illustrated in figure 3.13. A summary of the four protocols for each imaging mode are shown in table 3.14.

Table 3.13: Calculated TG111\(_w\) and "TG111\(_w\) values in mGy and mGy.100mAs\(^{-1}\) for clinical CBCT protocols on the OBI and XI systems. Uncertainties represent 1 standard deviation.

| Protocol         | CTDI\(_w\)      | CBDI\(_w\)      | IAEA\(_w\)     | TG111\(_w\)    |
|------------------|-----------------|-----------------|----------------|----------------|
| Pelvis OBI       | 9.408 ± 0.029   | 19.381 ± 0.029  | 18.343 ± 0.002 | 21.199 ± 0.014 |
| Pelvis XI        | 9.646 ± 0.134   | 20.642 ± 0.134  | 14.771 ± 0.001 | 22.420 ± 0.001 |
| Pelvis Obese XI  | 20.743 ± 1.475  | 44.390 ± 1.475  | 32.295 ± 0.017 | 48.529 ± 0.012 |
| Thorax OBI       | 2.536 ± 0.004   | 5.225 ± 0.004   | -  -           | 6.963 ± 0.001  |
| Thorax XI        | 2.302 ± 0.014   | 4.927 ± 0.014   | 4.441 ± 0.001  | 5.540 ± 0.000  |
| Std Head OBI     | 2.236 ± 0.003   | 4.605 ± 0.003   | 3.747 ± 0.018  | -  -           |
| Low Head OBI     | 1.184 ± 0.005   | 2.439 ± 0.005   | 1.726 ± 0.007  | -  -           |
| High Head OBI    | 11.349 ± 0.005  | 23.380 ± 0.005  | 15.316 ± 0.002 | -  -           |
| Head XI          | 1.648 ± 0.002   | 3.527 ± 0.002   | 3.746 ± 0.033  | -  -           |

Table 3.14: Calculated CTDI, CBDI, IAEA\(_w\) and TG111\(_w\) values in mGy for clinical CBCT protocols on the OBI system. Uncertainties represent 1 standard deviation.

The TG111 methodology measured the highest dose for all half-fan CBCT modes
3.4. Comparison of Methodologies

Figure 3.13: A comparison of CTDI, CBDI, IAEA and TG111 weighted values on the four full-fan CBCT modes studied. Error bars represent 1 standard deviation.

due to full scatter being achieved in the centre of the TG111 phantom. The CBDI, which up scales CTDI to better approximate CBCT dose gave the second highest dose for half-fan modes. For full-fan scans where TG111 was not evaluated, the CBDI gave the highest dose across all scan modes except for the XI head protocol. The slightly lower CBDI values compared with TG111 is due to CTDI phantom not having sufficient length to achieve scatter equilibrium. As previously described in 3.1.3.2, this would not have been the case if additional scatter material was included.

The IAEA method gave the third highest dose across the half-fan modes and the second highest reading for full-fan modes behind the CBDI. The lower doses from the IAEA method are attributed to the lack of scatter dose contributing to the in-phantom measurements compared with the TG111 and CBDI methods. The larger difference between IAEA with the CBDI and TG111 methods on the XI system was due to
variations in the free-air beams. The lower energy photons on the OBI system leads to a lower reference width dose in air compared with XI, but a larger dose for protocol beam width. Thus, the ratio of in-air CTDI values is greater for the OBI system which gives a higher overall value for IAEA. The CTDI method gave the lowest dose for all CBCT scans due to the CTDI phantom lacking the sufficient length to achieve full scatter and the 100mm sensitive length of the ionisation chamber being inadequate to capture the full beam width leading to an underestimation of the integral dose.

The TG111 values closely match those of Hu et al., who quotes a weighted pelvis TG111 value of 22.70 mGy compared with an average of 21.80 mGy measured across the two CBCT imaging systems in this study. Conversely, $CTDI_{100}$ values from this study were lower, with a pelvis value of 18.06 mGy compared with an average value of 9.53 mGy in this study. For head protocols, Hu measured 5.53 mGy compared with an average of 1.94 mGy. IAEA values were also slightly lower for our study. Hu measured IAEA values for pelvis and head of 4.89 mGy and 15.86 mGy, compared with average values of 16.60 mGy and 3.75 mGy, respectively. Interestingly, the mAs and kV settings were similar between the two studies, particularly for the OBI system, with the collimation width being wider for the protocols in this study compared with Hu et al. These results may suggest a variation in beam quality between the two studies, particularly with the XI imaging system, resulting in a greater scatter contribution to measurements in Hu’s work, particularly when considering protocols with larger chamber lengths (CTDI, IAEA).

The TG111 methodology delivers the highest CBCT dose due to the full scatter of the 45 cm TG111 phantom. The CBDI methodology provides a good estimate to the TG111 methodology using the existing CTDI phantom and pencil chamber and may offer a more clinically applicable determination for CBCT dose.

It’s worth noting that these values are also higher than the quoted $CTDI_w$ values
3.4. Comparison of Methodologies

given by the OBI and XI systems as shown in figures 3.1 and 3.2 respectively. These quoted values are evaluated for a reference beam width, similar to the process described in the IAEA approach. However, the reference beam width is determined by measuring the FWHM of the reference beam in-air rather than using the specified collimator width. For this reference width, the full dose profile can be captured with the Pencil chamber and hence gives an accurate quantitative comparison of dose for different scanner settings. However it has been demonstrated that these values do not reflect the dose from the corresponding clinical CBCT protocols.
Chapter 4

Beam Profile Measurements and the Dosimetric Impact of Increased Beam Width

4.1 Variation of Central Dose with Increasing Scan Length

To investigate the limitations of the current methodology for wide beam scanning, the four protocols were tested for increasing CBCT beam width in the z-axis. The procedure was carried out as previously described using the Pelvis protocol on the OBI system. The beam widths tested were 2, 5, 10, 15, 20.6, 25 and 40 cm. The beam width for the clinical pelvis protocol was 20.6cm, so this was chosen instead of 20 cm. The distance between successive measurements was increased for the wider beam widths as variation in dose became much smaller due to equilibrium dose being approached.

Measurements were taken in the central and peripheral positions within the re-
spective phantoms and used to calculate the weighted values for each scan width for
the CTDI, CBDI, IAEA and TG111 protocols.

4.1.1 Beam Width Measurements

The weighted protocol measurements are presented in table 4.1 and illustrated in figure

| Beam Width (cm) | CTDI (mGy) | CBDI (mGy) | IAEA (mGy) | TG111 (mGy) |
|----------------|------------|------------|------------|-------------|
| 2              | 14.338     | 2.545      | 14.338     | 5.838       |
| 5              | 15.142     | 7.100      | 15.142     | 11.790      |
| 10             | 14.246     | 14.246     | 16.262     | 15.904      |
| 15             | 12.144     | 18.217     | 18.545     | 18.387      |
| 20.6           | 9.430      | 19.427     | 18.343     | 20.064      |
| 25             | 7.933      | 19.833     | 18.538     | 21.122      |
| 40             | 5.047      | 20.187     | 18.260     | 22.445      |

Table 4.1: Comparison of weighted measurements for CTDI, CBDI, IAEA and TG111
protocols in mGy for the OBI Pelvis scan mode. A CTDI phantom was used for
CTDI, CBDI and IAEA measurements, while the TG111 phantom was used for TG111
measurements.

For beam widths less than 10 cm the CTDI recorded the highest dose, measuring
14.338 mGy and 15.142 mGy at 2 cm and 5 cm respectively. The IAEA and CTDI
values are equivalent for beam widths less than 60 mm. The CBDI and TG111 pro-
tocols recorded lower doses, with TG111 being slightly higher than CBDI. The CBDI
is lower as the dose is divided by the 10 cm chamber length, rather than collimation
width. For collimation widths below 10 cm the chamber is measuring the average of
the profile, which includes in-beam and out of beam and hence, it is not measuring
dose in the centre of the collimated beam width. This averaging is less pronounced in
TG111 as the Farmer chamber has a shorter sensitive length of 2.4 cm. However aver-
aging will still occur and a smaller chamber would be required to accurately determine
peak dose for a 2 cm beam width.
4.1. Variation of Central Dose with Increasing Scan Length

Figure 4.1: Weighted dose measurements with increasing z-axis beam width for CTDI, CBDI, IAEA and TG111 protocols. The OBI Pelvis scan mode was used for all measurements.

The difference between CTDI and IAEA with CBCT and TG111 for narrow beam widths can be attributed to the reported overestimation of the average dose in the phantom by the CTDI protocol for narrow scan lengths [31], which counters the reported underestimation of the CTDI, even for narrow beam widths [72]. At 10 cm beam width, the CTDI and CBDI protocols both measured values of 14.246 mGy which was the maximum value for CTDI across all beam widths. For beam widths of between 10 cm and 15 cm, the CBDI will overestimate the dose due to scatter contributions at the far ends of the 100 cm chamber, which do not reach the centre of the volume (and hence not detected by the Farmer chamber in the TG111 measurement). This is however offset by the lack of additional scatter in the CBDI measurements compared with TG111. If the CBDI was performed with additional scatter material this would not be the case, and the CBDI would measure higher than the TG111 for these beam widths.
As the beam width increased further the CBDI and TG111 continued to increase before plateauing. The maximum values for TG111 and CBDI occurred at 40 cm measuring 22.445 mGy and 20.187 mGy respectively. The CBDI plateaued from 20.6 cm beam width with measurements increasing by only 0.760 mGy beyond 20.6 cm. The TG111 plateau occurred closer to 40 cm beam width. The higher dose measured by TG111 and longer beam width required to reach equilibrium compared with CBDI is attributed to the increased scatter from the longer TG111 phantom and is consistent with other results in the literature [80]. As previously stated in 3.1.3.2, the CBDI method used here does not include additional scatter material.

The IAEA reached a plateau at 15 cm measuring 18.545 mGy. Beyond 15 cm beam width the IAEA varied by only 0.285 mGy. At 40 cm beam width, a slight drop in the IAEA was observed. This is likely due to a limit being reached where increase in beam fluence with increased collimator width is less than the increase in collimator width itself, the divisor in the IAEA calculation. The IAEA protocol does not take into account increased phantom scatter for longer scan lengths, however underestimates $CTDI_\infty$ by the same fraction for all beam widths [11, 30]. Hence, the IAEA remains steady when beam width is increased.

Beyond 10 cm beam width the CTDI gradually decreased to 3.520 mGy at 40 cm. The CTDI is an integral dose metric which approximates the dose in the center of a scan by integrating the full dose profile and dividing by collimation width as described in section 2.3.1. As the beam is attenuated by the phantom, the dose profile in the central position is spread due to phantom scatter. As the beam width is increased the underestimation becomes more pronounced as larger portions of the beam fall beyond the 100 mm sensitive length of the Pencil ionisation chamber. Additionally the collimation width, which is the divisor in CTDI, continues to increase further compounding the underestimation of CTDI. Hence, the CTDI decreases with increasing...
beam width rather than an increase as was observed for the other protocols.

4.2 Integral Dose

The integral dose was calculated from equation 4.1 and represents the total energy absorbed in the phantom:

\[ E_{\text{tot}} = \rho \pi R^2 L D_{eq} \]  

(4.1)

The TG111 calculated values at 40 cm beam width were used for equilibrium dose \( D_{eq} \) for their respective scan modes, where \( D_{eq} \) is the radially averaged dose in the axial plane using the same weighting methodology described by equation (2.2). The collimation width for the respective scan modes were used for the scan length \( L \) and the \( \rho \) value used was the PMMA density of 1.19 \( g.cm^{-3} \) from the NIST database [114]. R represents the radius of the phantom.

4.2.1 Integral Dose Measurements

The Equilibrium dose \( D_{eq} \) and integral dose \( E_{\text{tot}} \) for each scan mode are presented in table 4.2.

| Protocol         | \( D_{eq} \) (mGy) | Integral Dose (mJ) |
|------------------|---------------------|--------------------|
| Thorax OBI       | 6.54 ± 0.07         | 128.89 ± 1.29      |
| Thorax XI        | 5.59 ± 0.06         | 114.49 ± 1.15      |
| Pelvis OBI       | 22.61 ± 0.23        | 445.69 ± 4.46      |
| Pelvis XI        | 22.58 ± 0.23        | 462.41 ± 4.62      |
| Pelvis Obese XI  | 48.82 ± 0.49        | 999.81 ± 10.00     |

Table 4.2: \( D_{eq} \) (mGy) and \( E_{\text{tot}} \) (mJ) for the Pelvis and Thorax modes on the OBI and XI imaging systems. A 1 percent error was applied to for \( D_{eq} \) measurements.

The Pelvis Obese XI mode delivered the highest \( E_{\text{tot}} \) of 999.81 ± 10.00 mJ due
to its higher mAs and kV producing an increased amount of scatter photons to the chamber in addition to an increased number of primary photons. Both Pelvis modes gave similar values, with XI delivering a slightly higher 462.41 ± 4.62 mJ compared with 445.69 ± 4.46 mJ from OBI. The higher XI value is due to the 211 mm collimation width of the XI being slightly longer than the 206 mm collimation width of the OBI scan. The XI system delivers 1056 mAs compared with 680 mAs for the OBI system however the beam is hardened in the XI system by an additional titanium filter. The filter removes low energy photons from the spectrum and increases the effective beam energy. The overall effect is a similar dose deposition between the two systems for the Pelvis scan with $D_{eq}$ varying by 0.03 mGy.

The Thorax measurement was higher for OBI with 128.89 ± 1.29 mJ compared with 114.49 ± 1.15 mJ from the XI. Both Thorax modes have a similar mAs, however due to the additional filtration on the XI system the OBI beam contains more photons. These low energy photons increase the scatter dose, particularly for wide beam widths where $D_{eq}$ is measured.

### 4.2.2 Integral Dose for Increasing Beam Width

The Integral Dose for varying beam width for each mode is presented in table 4.3 and illustrated in figures 4.2 and 4.3.

| Scan (cm) | Length | Pelvis OBI | Pelvis XI | Pelvis Obese XI | Thorax OBI | Thorax XI |
|-----------|--------|------------|-----------|-----------------|------------|-----------|
| 2         | 42.96  | 43.22      | 93.44     | 12.51           | 10.70      |
| 10        | 216.35 | 216.08     | 467.21    | 62.57           | 53.50      |
| 15        | 324.53 | 324.12     | 700.81    | 93.85           | 80.25      |
| 20.6      | 445.69 | 445.12     | 700.81    | 128.89          | 110.21     |
| 25        | 540.88 | 540.19     | 1168.01   | 156.42          | 133.75     |
| 40        | 865.41 | 864.31     | 1868.82   | 250.27          | 214.00     |

Table 4.3: $E_{tot}$ in mJ for increasing beam width for OBI Pelvis, OBI Thorax, XI Pelvis, XI Pelvis Obese and XI Thorax.
The expected linear relationship was observed for Integral Dose with increasing scan width. The higher mAs scans corresponded to a more dramatic increase in the integral dose with the Pelvis Obese XI mode increasing from 93.44 mJ to 1868.82 mJ from 2 cm to 40 cm. Both Pelvis OBI and Pelvis XI protocols had similar increases.
of 42.96 mJ to 865.41 mJ and 43.22 mJ to 864.31 mJ respectively. The increase was higher for Thorax OBI compared with Thorax XI protocol, increasing from 12.51 mJ to 250.27 mJ compared with 10.70 mJ to 214.00 mJ due to the higher scatter dose from low energy photons in the OBI scan.

The increase was greater than for dose in the central slice of the phantom as shown in figure 4.1, particularly for wide beam widths when the Equilibrium Dose is approached. This result identifies the necessity to limit the beam width for CBCT imaging to what is necessary, particularly for higher mAs scans such as the Pelvis Obese protocol.

4.3 CBCT Beam Profiles

Beam profile measurements were taken for the half-fan pelvis mode in the TG111 phantom for XI and OBI imaging systems. The Farmer chamber was placed in the central position within the phantom and stepped through the beam by moving the plug housing the chamber within the phantom by 2.5 cm intervals, the sensitive length of the chamber. Temperature and pressure corrections were applied to the measurements and converted to dose to air using $N_K$ values from the chamber calibration certificate.

Profile measurements were also taken using Gafchromic® XR-QA2 film which is designed for diagnostic energies of 20kVp-200kVp with a dose range of 0.1 cGy-20 cGy. The film consists of a 97 micron polyester layer, 20 micron adhesive layer, 25 micron active layer and a 97 micron white polyester backing layer. The film is self developing and can be handled in room light.

4.3.1 Ionisation Chamber Measurements

The chamber measurements of the OBI and XI Pelvis dose profiles are illustrated in figure 4.4.
4.3. CBCT Beam Profiles

Figure 4.4: Farmer chamber measurements converted to mGy of the beam profiles along the z-axis for OBI and XI Pelvis modes in the centre of the TG111 phantom. The average collimated field width is shown with blue vertical lines (206mm for OBI and 214mm for XI). The asymmetry in the XI profile is due to the heel effect. Both profiles extend beyond the collimation width due to scatter within the phantom. The OBI and XI measuring 62% and 58% of the central value at 10 cm beam width respectively. The central dose is 3.2 mGy higher for the XI compared with OBI due to the harder spectrum produced by the titanium filter in the XI system. However further from the central region the dose becomes similar for the two systems, with the average dose at the 12 cm position varying by an average of 1.2 mGy between OBI and XI. There is an asymmetry in the XI profile due to the X-ray target running along the z-axis and producing a heel effect. The target runs along the axial plane in the OBI system. Some volume averaging over the ionisation chamber sensitive length may also occur in the penumbral region where the dose gradient is steepest.
4.3.2 Film Measurements

4.3.2.1 Film Calibration

Due to the known energy dependance of film at low photon energies\cite{28, 24, 25} the film was calibrated for both the OBI and XI imaging systems since the XI spectrum is hardened with additional filtration compared with OBI.

The film was scanned prior to, and 24 hours post irradiation on an Epson Expression 10000 XL flat bed scanner at 72 dpi resolution in 48 bit colour RGB mode. The scanner was operated in reflection mode with all colour corrections switched off. Prior to scanning the film, six runs of the scanner were taken to warm up the lamp. Three ‘dummy’ images were taken of each batch of calibration film and 3 ‘keep’ images. The red channel of the scanner was used for image analysis as this encompasses the wavelength component associated with the most change in film colour \cite{115}.

Three 2 cm by 3 cm film pieces were placed along the z-axis for each exposure, with one at isocentre and 1 piece adjacent on either side. The film was calibrated against air KERMA measured with a 0.6 cc Farmer Ionisation chamber with calibration traceable to a primary standard. The film was exposed to doses from 0cGy-4cGy. Exposures were taken in steps of 0.5cGy from 0cGy-3cGy and then a single 1cGy step from 3cGy to 4cGy. To avoid any angular dependence in film response due to beam rotation, the calibration exposures were taken in kV fluoroscopy mode with the kVp, mAs, collimator and bow-tie filter parameters as per a pelvis kV CBCT acquisition.

Because the length of exposure in fluoroscopy mode is controlled by the user, in order to determine the dose delivered to the film, a reference detector was used during the Farmer chamber and film exposures. The detector was placed in the beam a fixed distance away from isocentre. An average ratio between the two detectors was then used to determine isocentre dose when the Farmer chamber was replaced with the calibration film as shown in figure 4.7.
The post exposure films were analysed using ImageJ. Regions of interest (ROI) were selected for each film piece and an average pixel intensity measured. The ROI was chosen to maximise the ROI area while avoiding areas close to the film edges. A macro was used to replicate the ROI’s across the three scans.

The mean pixel intensity for the three film pieces at each exposure were compared on both the OBI and XI systems. In the XI system, the X-Ray tube runs along the z-axis along with the three film pieces and it has been previously shown that the heel effect can influence dose in air for kV CBCT scans [116, 117]. The average pixel intensities for the 3 pieces across the measured doses are shown in figures 4.5 for the OBI and 4.6 for the XI systems.

![Figure 4.5: Comparison of mean pixel intensity across the selected ROI on the calibration film pieces along the z-axis for measured doses on the OBI system.](image)

No significant variation in the dose across the three film pieces was observed for any of the doses measured for both the OBI and XI systems. It was concluded the distance across the three film pieces was too small to be influenced by the heel effect and would not influence the calibration.

The net reflectance (net$\Delta R$) for each exposure was then calculated using the average pixel value from the 3 pieces of film for each exposure using the method described
Figure 4.6: Comparison of mean pixel intensity across the selected ROI on the calibration film pieces along the z-axis for measured doses on the XI system.

by Tomic et. al. [118]. The first step is to calculate the change in reflectance:

$$\Delta R_i = R_{i\text{before}} - R_{i\text{after}} = \frac{1}{2^{16}} [PV_{i\text{before}} - PV_{i\text{after}}]$$  \hspace{1cm} (4.2)

where PV is the average pixel value from the selected ROI of film piece i analysed using the red channel and reflectance is calculated as $\frac{PV}{2^{16}}$. The error on reflectance change for film piece i is given by:

$$\sigma_{\Delta R_i} = \frac{1}{2^{16}} \sqrt{(\sigma(PV_{i\text{before}}))^2 + (\sigma(PV_{i\text{after}}))^2}$$  \hspace{1cm} (4.3)

The average change in reflectance ($\Delta R$) for each exposure was then determined using a weighting factor:

$$\Delta \bar{R} = \sum_{i=1}^{3} (\omega^i \cdot \Delta R^i)$$  \hspace{1cm} (4.4)
where the weighting factors were calculated as:

$$\omega^i = \frac{1}{\sum_{i=1}^{3} \left(\frac{1}{\sigma_{\Delta R^i}}\right)^2}$$  \hspace{1cm} (4.5)

The corresponding standard deviation on $\Delta R$ is determined as:

$$\sigma_{\Delta R} = \left[ \sum_{i=1}^{3} \left(\frac{1}{\sigma_{\Delta R^i}}\right)^2 \right]^{-\frac{1}{2}}$$  \hspace{1cm} (4.6)

The process was repeated for three control pieces which received no dose and the result subtracted to obtain the final reflectance change $\text{net} \Delta R$:

$$\text{net} \Delta R = \Delta R - \Delta R_{\text{control}}$$  \hspace{1cm} (4.7)

with the corresponding standard deviation given by:

$$\sigma_{\text{net} \Delta R} = \sqrt{(\sigma_{\Delta R})^2 + (\sigma_{\Delta R_{\text{control}}})^2}$$  \hspace{1cm} (4.8)

The $\text{net} \Delta R$ values with the corresponding doses were imported into MATLAB and a curve fitted using the curve fitting application for both the OBI and XI systems. The fit equation used by Giaddui et. al. [25] of the form $y = \frac{ax}{b-x}$ was applied, where $a$ and $b$ are coefficients and $x$ and $y$ represent $\text{net} \Delta R$ and dose to film respectively. This fit function has the benefit of linearity, is monotonically increasing and returns a zero value for zero dose.

Previous studies with XR-QA film have suggested the response of the film in-air compared with PMMA scatter conditions is within ±2%[29]. To verify the response of the film did not vary between free air and in the PMMA phantom, two calibrations for each condition were produced on the OBI and XI systems. For in air calibration the film pieces were placed on low density foam blocks as shown in figure 4.7(a).
4.3. CBCT Beam Profiles

primary chamber was held at isocentre using a retort stand with the reference chamber
fixed to the couch as shown in figure 4.7(b).

Figure 4.7: (a) In-air calibration of XR-QA2 film. Three pieces of film were used for
each exposure. The pieces were placed along the z-axis, with one at isocentre and one
directly above and below. (b) Calibration of Farmer chamber at isocentre with R/F
detector fixed to couch away from isocentre. By calibrating the isocentre dose with
the R/F detector, dose to film during exposures was able to be determined.

For in phantom calibration, the film pieces were held in place with tape within a
PMMA rod cut along its length as shown in figure 4.8(a).

The reference chamber was held in the beam away from isocentre using a retort
stand and a plastic tool which came with the Unfors package as shown in figure 4.8(b).
The film was calibrated at the centre of the PMMA phantom.

A comparison of isocentre dose to \( \Delta R \) for the free-air and in phantom calibration
methods are presented in figure 4.9 for OBI and figure 4.10 for XI.

No difference in film response was observed for the OBI system between the two
calibration methods. However for the XI system, the film response was greater for
calibration in the phantom compared with free-air. The additional titanium filter in
the XI system removes low energy photons from the incident beam spectrum, however
4.3. CBCT Beam Profiles

Figure 4.8: (a) In phantom calibration of XR-QA2 film. Three pieces of film were taped in place within a PMMA plug that was sliced along its length. The orientation of the film pieces relative to isocentre was the same as for the in-air method. (b) Calibration of Farmer chamber at isocentre with R/F detector held in the beam away from isocentre.

Figure 4.9: Comparison of the calibration data for in-air and in-phantom methods for the OBI system. No difference in film response was observed between the two methods.
4.3. CBCT Beam Profiles

Figure 4.10: Comparison of the calibration data for in-air and in-phantom methods for the XI system. The response of the film was greater when calibrating in phantom compared with calibration in air.

In the centre of the PMMA phantom lower energy scattered photons provide a significant contribution to the measured dose \[117\]. The response of XR-QA2 film has been previously shown to be around 10% higher at low energies \[25\] and hence, for the XI system the response changes in air compared with in phantom. Following this result, and the invariance of the OBI system between calibration methods, the in-phantom calibration method was chosen.

The fitted calibration curves from the in-phantom calibration data are shown in figure 4.11 for the OBI system and 4.12 for the XI system.

4.3.2.2 Profile Measurements

The film was cut into 30 cm by 3 cm strips for the profile measurements. The film was scanned prior to, and post exposure following the same procedure as described in section 4.3.2.1. Small arrows were marked in the film corners and numbered to ensure the film position on the scanner bed was consistent as shown in figure 4.13. For each
4.3. CBCT Beam Profiles

Figure 4.11: Calibration curve for the OBI imaging system. Error bars represent the standard deviation in the net reflectance $\sigma_{net\Delta R}$

Figure 4.12: Calibration curve for the XI imaging system. Error bars represent the standard deviation in the net reflectance $\sigma_{net\Delta R}$
exposure the film strip was placed into the centre of the TG111 phantom using the custom halved plug. Two strips were exposed on both the OBI and XI systems in pelvis mode. Two CBCT scans were acquired for each strip to deliver a higher dose to the film and remove variations due to the start/stop position of the X-Ray tube. After film response was converted to dose from the calibration curve, the value was halved to obtain dose from a single CBCT scan.

Figure 4.13: Film strips for beam profile measurements. The strips were cut to 3 cm by 30 cm and placed in the centre of the TG111 phantom in a custom plug to reduce air gaps.

The pre- and post-exposed film scans were imported into ImageJ. Line profiles were taken for each strip and averaged for the two OBI and two XI strips. The net reflectance was calculated from the pre and post scan data and then imported to MATLAB and converted to dose from the OBI and XI calibration curves. The dose profiles are shown in figure 4.14 for the OBI system and figure 4.15 for XI. The chamber profile measurements described in section 4.3.1 are also overlayed for comparison.
4.3. CBCT Beam Profiles

Figure 4.14: Dose profile (in cGy) along the z-axis for the OBI Pelvis scan in the centre of the TG111 phantom measured using the Farmer chamber and 3cm by 30cm film strip.

The film measured a higher dose for both the OBI and XI systems, particularly in the central region where the dose was highest. The variation for OBI at the centre
position was 1.827 cGy for film compared with 1.525 cGy (17%) for the ionisation chamber. The variation was less for XI with the film measuring 2.089 cGy compared to 1.844 cGy (12%) with the ionisation chamber.

It was hypothesised the high sensitivity of XR-QA2 film to beam energy could be causing the variation between the chamber and film. Tomic et. al. [24] showed for beam qualities ranging from 4.03-8.25 mm Al, commonly used for CT scanning, air KERMA varied by 14% for a dose of 8 cGy. Lillo et. al. [26] reported an increased XR-QA2 film by as much as 170% in the energy range of 18-39 keV, while in the energy range 38-46 keV the film response decreased. It’s worth noting that no other literature observed variations in response above 81%. Given the scatter dose contribution to the film likely contains energies such as these, particularly in the centre of the PMMA phantom, high variability of film response for CBCT scans is possible. Lillo also reports significant variation in XR-QA2 film response between studies published in the literature for similar beam HVL values as shown in figure 4.16.

![Figure 4.16: XR-QA2 dose response curves from Giaddui [25, 4], Tomic [24] and Lillo [26]. Each curve was obtained from a polychromatic source with 120kVp. Source [26]](image)

Variations up to 55% in film response were observed between studies by Giaddui
and Lillo [26] for similar beam HVLs. Lillo suggests such large variations between batches could be attributed to differences in chemical composition of sensitive and adhesive layers regarding the lithium concentration. Further to this, Giaddui observed two different film response curves from the same film batch of film [25, 4].

A change in response of the film due to the effects of the beam rotating compared with a non rotating beam for calibration was also considered. When the beam is rotated the angle of incidence of the beam on the film changes and also which side of the film receives the primary beam (yellow face of film facing the beam to white backing layer of film facing the beam). The angular dependence of XR-QA2 film and effect of film facing up or down on its response has been previously investigated. Giaddui et. al. [25] reported a variation of less than 5% for angles of 0-150 degrees excluding 90 degrees. They also reported an identical response between film facing up and film facing down. Rampado et. al. [119] reported similar findings for XR-QA film with response varying by less that 1% for angles excluding 90 degrees. At 90 degrees, when the beam axis is parallel to the film surface, Giaddui reported a decrease in response of 76% at 5.4 cGy and 81% at 2.87 cGy. Rampado reported a 49% decrease in response at 90 degrees. The change in response at 90 degrees was attributed to several factors including attenuation of the beam when traversing the film and a smaller fraction of the beam hitting the film surface. Following from these results the effect of beam rotation would cause a decrease in film response, rather than an increase which is observed.

Following maintenance of the X-Ray tube it was noted that the output of the X-Ray tube on the OBI system measured by the Farmer chamber had increased by 8% compared with previous measurements taken 12 months earlier. An increase in dose should not affect the response of the film, however any change in the energy spectrum, particularly for low energies, could have a significant impact on film response
To investigate any change in film response of the film strips due to variation in tube output, the film profile measurements were re-taken following the methodology previously described in 4.3.2.2. Though no change in tube output was observed for the XI system, strip measurements were also repeated to observe whether there was consistency in film results. The re-measured strips in the centre of the TG111 phantom are shown in figures 4.17 and 4.18 for OBI and XI systems respectively.

A 9% and 20% decrease in film response was observed for the OBI and XI systems respectively. Good agreement was achieved between the re-measured Farmer chamber values the and XR-QA2 film across the CBCT profile for the OBI system. The film dose was slightly higher than the chamber in the centre of the scan due to film uncertainty and volume averaging in the chamber, however the difference was less than 4%. For the XI system, the film under responded compared with the Farmer chamber. In the
4.3. CBCT Beam Profiles

Figure 4.18: Re-measured film strip in the centre of the TG111 phantom on the XI system.

centre of the profile the variation was 8%.

A further observation was the variation in film response for the strips as the X-Ray tube heated up. Three successive film strips labelled S1-S3 were measured in the centre of the TG111 phantom and are shown in figure 4.19. No CBCT scans had been acquired before the strip measurements.

The dose in the centre of the film profiles decreased by 10% from S1-S3. Following this, three additional film strips labeled S4-S6 were measured and are shown in figure 4.20. After the tube had warmed up, it was observed that the response of the film remained more steady. However, the average variation between the film profiles with the tube warmed up to the Farmer chamber in the centre of the dose profile increased to 13%.

The 9% and 20% variations in film response for the OBI and XI system are consistent with the literature regarding the high sensitivity of XR-QA2 film attributed to
4.4 Calculating the CTDI using Film Strips

Figure 4.19: Three strips measured in succession on the XI system to evaluate tube warm up on film response. The strips were measured in order S1-S3 with no CBCT scans acquired prior to the measurements. Also shown is the Farmer chamber dose profile.

any minor change in energy spectrum and possible variations in chemical composition of the film even within the same batch [26]. Variations in the output of the film, scanner lamp, as well as the observed variations in film response due to tube warm-up may have also contributed to change in film response.

4.4 Calculating the CTDI using Film Strips

The CTDI was calculated using the film strip profiles for OBI and XI pelvis modes using the method described by Hu et. al. [8]. The sensitive length of the film was 30 cm and along with full scatter from the TG111 phantom overcomes the shortcomings of the $CTDI_{100}$ allowing for an accurate determination of CTDI for CBCT. In order to make comparisons of film calculated $CTDI_{film}$ with the other methodologies, the film
4.4. Calculating the CTDI using Film Strips

Figure 4.20: Three additional strips S4-S6 measured in succession on the XI system after the X-Ray tube had warmed up. Also shown is the Farmer chamber dose profile. Profiles were first normalised relative to the central ionisation chamber measurements as shown in figures 4.21 and 4.22. Since the methodologies were evaluated prior to the change in output of the X-Ray tube on the OBI system, the film profiles were normalised to the initial chamber profile measurements. Error bars indicate the observed variations of ±4.5% and ±10% in film response between the two sets of strip measurements for OBI and XI systems respectively.

Film strips were exposed in the central position and each of the peripheral locations within the phantom for OBI and XI systems as described in section 4.3.2.2. The DLI was computed from the summation of each pixel value $PV_i$ along the strip profile multiplied by the distance between successive pixel values $d$, obtained from the length of the film strip $L_s$ and number of pixels in the strip profile $n$:

$$d = \frac{L_s}{n}$$  \hspace{1cm} (4.9)
4.4. Calculating the CTDI using Film Strips

Figure 4.21: Film profile in the centre of the TG111 phantom for a OBI Pelvis mode CBCT scan and the profile normalised to the central ionisation chamber measurements taken prior to the change in X-Ray tube output. Error bars represent the observed ±4.5% variation in the OBI film profiles.

Figure 4.22: Film profile in the centre of the TG111 phantom for a XI Pelvis mode CBCT scan and the profile normalised to the central ionisation chamber measurement. Error bars represent the observed ±10% variation in the XI film profiles.
4.4. Calculating the CTDI using Film Strips

\[ DLI = \sum_{i=1}^{n} PV_i \ast d \] \hspace{1cm} (4.10)

CTDI\textsubscript{film} was then calculated using the DLIs from the central and peripheral positions within the phantom and z-axis collimation using equations 3.2 and 3.5.

Standard deviation in the CTDI\textsubscript{film} was calculated using the total uncertainty in the calibration curve. This was determined from the standard deviation in the reflectance values for each calibration point, and the root-mean square (RMS) of the curve fit:

\[ \sigma_{\text{total calibration}} = \sqrt{\sum_{i} \sigma_{\text{net}} \Delta R_i^2 + \text{RMS}^2} \] \hspace{1cm} (4.11)

The \( \sigma_{\text{CTDI}_{\text{film}}} \) was then determined by calculating the DLI of standard deviation of the film profile:

\[ \sigma_{DLI} = \sum_{i=1}^{n} \sigma_{\text{dose to pixel}}_i \] \hspace{1cm} (4.12)

\[ \sigma_{\text{CTDI}_{\text{film}}} = \frac{\sigma_{DLI}}{\text{collimation width}} \] \hspace{1cm} (4.13)

A comparison of CTDI\textsubscript{film} to the TG111 method is shown in table 4.4.

| Imaging System | TG111\textsubscript{w} (mGy) | CTDI\textsubscript{film} (mGy) |
|----------------|-----------------------------|-------------------------------|
| OBI            | 21.20 ± 0.01                | 20.47 ± 0.71                  |
| XI             | 22.42 ± 0.01                | 21.28 ± 0.44                  |

Table 4.4: Comparison of TG111\textsubscript{w} with CTDI\textsubscript{film} in mGy for the OBI and XI imaging systems.

The TG111 measured 0.73 mGy higher for the OBI system and 1.14 mGy higher for the XI system compared with CTDI\textsubscript{film}. The small difference between methods is consistent with the under-response of the film observed previously and well within the uncertainty of sensitivity for XR-QA2 film [24, 25, 26, 119]. A small fraction of
the dose profile may have also fallen beyond the 30cm length of the film strip, which would cause a small under estimation of the DLI.

The $CTDI_{film}$ measurements were within 5% of the TG111 which correlates well with the 3|5 agreement between the film and TG111 methods observed by Hu et. al. [8]. The result validate the TG111 method as an accurate representation of CTDI with sufficient detector and phantom length. The result also validates the CBDI as a good approximation to the CTDI using the existing CTDI$_{100}$ phantom and 100mm Pencil ionisation chamber.
Chapter 5

Conclusion

This thesis presents work quantifying the dose output for two commercial kV CBCT imaging systems for use during image guided radiotherapy. The major objectives of this thesis, as outlined by the aims in chapter 1 were:

- Evaluate and compare the current CTDI protocol with three alternate methodologies, the Cone-Beam Dose Index method adapted for wide beam scanning [13], the International Atomic Energy Agency Report No. 5 suggested approach for wide beam dosimetry [11] and the American Association of Physicists in Medicine Task Group 111 methodology[12] [Section 5.1].

- Evaluate and compare methodologies for varying beam width, specifically for clinical settings and compare between the listed protocols [Section 5.2], and

- Compare dose between two Varian imaging systems: the OBI® on the C-Series platform and XI® on Truebeam [Section 5.3].
5.1 Comparison of TG111, CBDI and IAEA methodologies with the current CTDI method

The dose from two commercial kV CBCT systems, the Varian Cliniac® iX On-Board Imager (OBI) and the Varian Truebeam\textsuperscript{TM} X-Ray Imager (XI) were evaluated using four methodologies, the American Association of Physicists in Medicine Task Group 111 approach (TG111)\textsuperscript{[12]}, the International Atomic Energy Agency Report 5 recommendation (IAEA)\textsuperscript{[11]}, the Cone-Beam Dose Index (CBDI)\textsuperscript{[13]} and the current paradigm, the Computed Tomography Dose Index (CTDI)\textsuperscript{[31]}, and the results compared. The methodologies were evaluated on clinically relevant imaging modes for each system. For the OBI system: Pelvis, Thorax, Low Dose, High Dose and Standard Dose Head modes were evaluated. On the XI system: Pelvis, Pelvis Obese, Thorax and Head modes were evaluated.

The TG111 approach measured the highest overall dose across both OBI and XI due to the increased length of the custom phantom. The added length created a maximum scatter contribution to the central measurement and therefore the highest dose. The CBDI measured the second highest dose overall using the existing CTDI phantom and pencil chamber from the CTDI method. The CBDI method was an up-scaling of the CTDI by dividing the integral dose by the sensitive chamber length of 100 mm rather than the collimation width. The dose from this method gave similar results to the TG111. The IAEA approach measured the third highest dose and was significantly lower than CBDI and TG111 due to the lack of scatter and no scaling factor. However the IAEA approach does capture the full beam width for calculating integral dose, which translated into a constant dose for the wide beam widths of CBCT. The CTDI gave the lowest dose of all the methodologies.
5.2 Comparison of dose with varying beam width for TG111, CBDI, IAEA and CTDI, including Integral Dose

Both the TG111 and CBDI methods asymptotically approached a maximum value for increasing beam width, with the TG111 maximum being slightly higher than CBDI. The dose at 40 cm width was 22.45 mGy for TG111 and 20.19 mGy for CBDI on the OBI Pelvis mode. The maximum value occurred when full scatter was achieved on the ionisation chamber. The IAEA method did not increase beyond 15 cm beam width due to the shorter phantom not having sufficient scatter as the beam width was increased. However the IAEA does capture the full dose profile and remains relatively constant from 15 cm out to to 40 cm. The CTDI reached its maximum at 10 cm beam width then decreased as the beam was increased to 40 cm. The CTDI phantom lacks sufficient length to achieve full scatter and the 100 mm sensitive chamber length is not sufficient to capture the full beam profile. This was confirmed by dose profile measurements in the centre of the TG111 phantom for the Pelvis OBI and XI scans. The beam profiles extended beyond the collimated beam width, only dropping to 62% and 58% of the maximum values at ± 10 cm respectively. The underestimation of CTDI became worse as the beam continued to increase. The underestimation was compounded as the divisor in the CTDI, the collimation width, was increased.

The integral dose increased linearly with beam width with higher mAs scans translating to steeper increases. The increase with beam width was greater than for the TG111, CBDI, IAEA and CTDI where dose in the central axis of the phantom is used in the calculations. The result highlights the necessity to limit beam width in CBCT imaging to what is required particularly for scans with a high mAs.

Dose profiles taken with diagnostic Gafchromic\textsuperscript{TM} XR-QA2 film and used to cal-
5.3. Comparing the dose of the OBI and XI kV CBCT imaging systems

culate a weighted CTDI$^w_{film}$ in the TG111 phantom for OBI and XI Pelvis modes. The full scatter and 30 cm length of the film strips overcame the shortcomings of the CTDI$_{100}$. The CT$^w_{film}$ and TG111 values were in good agreement with CT$^w_{film}$ measuring 20.47 ± 0.71 mGy compared with 21.20 ± 0.01 mGy using the TG111 method on the OBI system and 21.28 ± 0.44 mGy and 22.42 ± 0.01 mGy for the XI system for the respective methods. This validated the TG111 as an accurate representation of the CTDI when sufficient detector and phantom lengths are used. The result also confirms the CBDI as a good approximation to the accurate CTDI value using existing dosimetry equipment.

5.3 Comparing the dose of the OBI and XI kV CBCT imaging systems

The overall dose was higher for the OBI system. For Pelvis modes the OBI measured 21.199 ± 0.014 mGy compared with 18.246 ± 0.001 mGy for XI using the TG111 method. For Thorax the OBI measured 6.963 ± 0.001 mGy compared with 5.540 ± 0.000 mGy for XI. Comparing equivalent Head protocols on the two systems, Standard Head OBI with Head XI, the OBI was higher measuring 4.605 ± 0.003 mGy compared with 3.527 ± 0.002 mGy for XI using the CBDI method. The higher OBI values are attributed to its higher photon fluence compared with the XI due to the additional low energy photons in its spectrum.

5.4 Summary and Future Work

Three alternative methods for quantifying kV CBCT dose during image guidance procedures were evaluated and compared to the existing CTDI$_{100}$ method on two commercial imaging systems, the Varian OBI® on the C-Series platform and XI® on
Truebeam. The CTDI$_{100}$ was shown to underestimate the dose due to an insufficient detector length to capture the full dose profile and phantom length to achieve scatter equilibrium. The three alternate methods provided a better approximation with the TG111 method giving the best representation of CTDI when the full profile is collected and full scatter is achieved, which was calculated using Gafchromic® QR-QA2 film strips and a custom built TG111 phantom.

The OBI imaging system delivered similar doses for pelvis mode scanning and higher doses in thorax mode scanning compared with the XI system across all protocols due to a reduction in low energy photons in the XI beam. The CBCT beam was also shown to extend beyond the collimation width in the centre of the PMMA phantom due to beam divergence.

The TG111 method should be applied to give the best estimate of CBCT dose due to the full scatter phantom. However, the CBDI method may be more readily applied clinically due to the use of existing CTDI equipment and measurement methods and will give a close approximation to TG111. A possible approach is to measure TG111 for clinical CBCT modes and scale this to CTDI measurements, which can be more easily taken periodically to evaluate scanner output.

The scope for future work includes:

- Investigating out of field and skin doses from CBCT scanning,
- Determining patient specific dose estimates from CBCT scans, and deriving a relationship between patient specific dose and protocol measurements,
- Using risk models to assess the increased risk of radiation-induced secondary malignancies following CBCT imaging regimes during radiotherapy,
- Incorporating imaging dose calculations into treatment planning systems,
- Investigate optimal image quality requirements to minimise CBCT imaging dose
• Development of a universal CBCT dosimetry protocol, similar to the TG111 for CT scanning, which would be independent of scanner settings, which could facilitate comparisons between imaging systems between clinics.
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