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วิทยานิพนธ์นี้เป็นส่วนหนึ่งของการศึกษาตามหลักสูตรปริญญาวิทยาศาสตรมหาบัณฑิต
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DETERMINATION OF EFFECTIVE DOSES IN IMAGE-GUIDED RADATION THERAPY SYSTEM

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A Thesis Submitted in Partial Fulfillment of the Requirements
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Thesis Title	DETERMINATION OF EFFECTIVE DOSES IN IMAGE-GUIDED RADATION THERAPY SYSTEM
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ยิน ยิน เปียว : การหาค่าปริมาณรังสียังผลจากระบบภาพทางรังสีรักษา (DETERMINATION OF EFFECTIVE DOSES IN IMAGE-GUIDED RADATION THERAPY SYSTEM) อ.ที่ปรึกษาวิทยานิพนธ์หลัก: รศ. ศิวลี สุริยาปี, อ.ที่ปรึกษาวิทยานิพนธ์ร่วม: ทวีป แสงแห่งธรรม, 56 หน้า.

การศึกษาครั้งนี้ต้องการหาปริมาณรังสีจากการทำ ระบบภาพนำวิถี (IGRT) ในส่วนศีรษะ ทรวงอก และอุ้งเชิงกราน ทำการวัดรังสี จากเครื่องจำลองการรักษา Acuity และ on-board imager ที่ผิวทางเข้า (entrance surface air kerma; ESAK) จากนั้นแปลงค่า ESAK เป็นค่าปริมาณรังสียังผล (effective dose) ในส่วนปริมาณรังสีที่อวัยวะจาก EPID ได้จากวาดอวัยวะบนภาพเอกซเรย์คอมพิวเตอร์จากแฟนทอม Rando และคำนวณปริมาณรังสีโดยใช้เครื่องวางแผน การรักษา Eclipse ปริมาณรังสีที่อวัยวะสามารถแปลงเป็นปริมาณรังสียังผล โดยการแก้ค่าความไวต่อรังสีของแต่ละอวัยวะ (tissue weighting factor) ตามมาตรฐาน ICRP 103 ในส่วนของ ระบบภาพ 3 มิติ ทำการวัดปริมาณรังสี วัดค่า CTDI และ CBDI ในแฟนทอม โดยใช้หัววัดไอออไนเซชันที่มีความยาว 100 มม. ร่วมกับ เครื่องอ่านรังสี Accu-Pro จากเครื่องเอกซเรย์คอมพิวเตอร์ (CT) ยี่ห้อ Philips รุ่น Brilliant Big Bore และจาก เครื่องเอกซเรย์คอมพิวเตอร์แบบรูปกรวย (CBCT) ที่ติดอยู่กับ เครื่องฉายรังสียี่ห้อ Varian รุ่น TrueBeam ทำการคำนวณหาค่าปริมาณรังสียังผลโดยใช้ซอฟแวร์ ImPACT นอกจากนี้ ความถูกต้องของการคำนวณจาก ImPACT ยังถูกตรวจสอบจากการวัดด้วย TLD ในแฟนทอม Rando บริเวณอุ้งเชิงกราน จากการทดลองพบว่า ปริมาณรังสียังผล จากเครื่องจำลองการรักษา Acuity จากการถ่ายภาพบริเวณศีรษะ ทรวงอก และอุ้งเชิงกราน มีค่าเท่ากับ 0.03, 0.31 และ 0.43 mSv และจากเครื่องเอกซเรย์คอมพิวเตอร์มีค่าเท่ากับ 3.3, 13 และ 7.2 mSv ตามลำดับ ในส่วนของปริมาณรังสี ที่ได้จากระบบภาพสำหรับ การตรวจสอบตำแหน่งก่อนการฉายรังสี พบว่าค่าปริมาณรังสียังผลจากการตั้งโปรโตคอล บริเวณศีรษะ ทรวงอก และอุ้งเชิงกราน จากระบบภาพ OBI มีค่าเท่ากับ 0.018, 0.12 และ 0.16 mSv จากระบบภาพ EPID มีค่าเท่ากับ 3.54, 6.24 และ 4.86 mSv จากระบบภาพ CBCT ที่คำนวณจาก ซอฟแวร์ ImPACT มีค่าเท่ากับ 0.14, 2.14 และ 4.99 mSv ตามลำดับ โดยค่าปริมาณรังสียังผลที่วัดจาก TLD ในโปรโตคอลอุ้งเชิงกรานวัดได้ 5.17 mSv โดยสรุป ในส่วนระบบภาพแบบ 2 มิติ EPID ให้ปริมาณรังสีสูงที่สุด เนื่องจากใช้รังสี พลังงานระดับล้านโวลต์ ในการสร้างภาพ ขณะที่ระบบภาพ 3 มิติ CT ให้ปริมาณรังสีสูงกว่า CBCT และยังพบว่าค่าปริมาณรังสีที่วัดได้จาก TLD ต่างจากค่าที่ได้จากการคำนวณโดย ImPACT ถึง 35% ทั้งนี้อาจมีผลจากความแปรปรวนจากการวัดรังสีด้วย TLD ถ้ามีการ ฉายรังสี 30 ครั้ง ปริมาณรังสียังผลรวม (3D CT, ตรวจสอบตำแหน่งแบบ 2D และ CBCT จำนวน 6 ครั้ง) มีค่าเท่ากับ 4.17, 27.71 และ 37.57 mSv สำหรับการถ่ายภาพบริเวณศีรษะ ทรวงอก และอุ้งเชิงกราน ตามลำดับ ดังนั้นการใช้งานระบบภาพนำวิถี ควรคำนึงถึงปริมาณรังสีที่ผู้ป่วยจะได้รับเพิ่มเติมด้วย

ภาควิชา รังสีวิทยา	ลายมือชื่อนิสิต .....
สาขาวิชา ฉายาเวชศาสตร์	ลายมือชื่อ อ.ที่ปรึกษาหลัก .....
ปีการศึกษา 2558	ลายมือชื่อ อ.ที่ปรึกษาร่วม .....

5774070130 : MAJOR MEDICAL IMAGING
KEYWORDS: IGRT / EFFECTIVE DOSE / IMPACT/ ENTRANCE SURFACE AIR KERMA/ TREAMENT PLANNING SYSTEM

YIN YIN PYONE: DETERMINATION OF EFFECTIVE DOSES IN IMAGE-GUIDED RADATION THERAPY SYSTEM. ADVISOR: ASSOC. PROF. SIVALEE SURIYAPEE, CO-ADVISOR: TAWEAP SANGHANGTHUM, 56 pp.

This study determined the imaging doses starting from the planning to the verification process involved in IGRT for clinical protocols of head, chest and pelvis regions. The dose calculation from Varian Acuity simulator and on-board image were performed to obtain the entrance surface air kerma (ESAK). The product of ESAK with the conversion coefficient was resulted as the effective dose. The organ doses from electronic portal imaging device (EPID) were calculated by Varian Eclipse treatment planning system. The tissue weighting factors were applied according to the ICRP 103 for calculating the total effective dose. For the 3D imaging system, computed tomography dose index (CTDI) and cone-beam CT dose index (CBDI) measurements were performed using a 100 mm pencil ionization chamber with Accu-Pro dosimeter and CTDI phantoms. Then, effective doses were calculated by ImPACT software. Moreover, the organ and effective doses from CBCT pelvic protocol was measured by using the thermoluminescent dosimeters inserted in Rando phantom to verify the accuracy of ImPACT calculation. The total effective doses from planning imaging modalities for head, chest and pelvic protocols were 0.03, 0.31 and 0.43 mSv, respectively, from the 2D conventional simulator and 3.3, 13 and 7.2 mSv, from the 3D CT. From the verification imaging modalities, the total effective doses for head, chest and pelvis protocols were 0.018, 0.12 and 0.16 mSv from OBI system and 3.54, 6.24 and 4.86 mSv, respectively from EPID. In kV CBCT, the total effective doses were 0.14 mSv for head, 2.4 mSv for chest and 4.99 mSv for pelvis from the ImPACT calculation and 5.17 mSv for pelvis protocol from TLD measurement. Among the 2D imaging modalities, EPID effective doses were higher than the others because of using the treatment megavoltage beam. In 3D, planning CT doses were higher than those from the CBCT. For CBCT, ImPACT dose calculation result showed 35 % of variation from TLD measurement and it can be applied with acceptable result. Based on 30 fractions of treatment course, total effective doses (3D CT, 2D setup verification and 6 times CBCT) of head, thorax and pelvis were 4.17, 27.71and 37.57mSv, respectively. Therefore, IGRT should be applied carefully to optimize the dose.

Department:	Radiology	Student's Signature	.....
Field of Study:	Medical Imaging	Advisor's Signature	.....
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LIST OF ABBREVIATIONS

Abbreviation	Terms
2D	Two dimensional
3D	Three dimensional
AAPM	American Association of Physics in Medicine
CBCT	Cone-beam computed tomography
CBDI	Cone-beam dose index
CT	Computed tomography
CTDI	Computed tomography dose index
DICOM	Digital Imaging and Communications in Medicine
E	Effective dose
ECC	The element correction coefficient
EPID	Electronic portal imaging device
FOV	Field of view
H_T	Equivalent dose in a tissue or organ
HVL	Half value layer
ICRP	International Commission on Radiological Protection
ICRU	International Commission on Radiation Units and Measurements
IGRT	Image-guided radiation therapy
ImPACT	Imaging Performance Assessment of CT
IMRT	Intensity modulated radiation therapy
kV	Kilovoltage
kV CBCT	Kilovoltage cone-beam computed tomography
kVp	Kilovoltage peak

mA	Milliampere
mAs	Milliampere second
mGy	Milligray
mSv	MilliSievert
MV	Megavoltage
MV CBCT	Megavoltage cone-beam CT
OBI	On-board imager
R	Roentgen
RCF	The reader calibration factor
TLD	Thermoluminescent dosimeter
TPS	Treatment Planning System
VMAT	Volumetric modulated arc therapy
SSD	Source to surface distance
W_R	Radiation weighting factor
W_T	Tissue weighting factor

CHAPTER 1

INTRODUCTION

1.1 Background and Rationale

Radiation therapy is one of the most common treatment methods for cancer. The aim of radiotherapy is to give the required dose to the target volume as precisely as possible and to minimize doses to surrounding healthy tissues. The precise radiation therapy offers the reduce severity and risk of therapy-induced complications and increase both quality and probability of success.

New treatment modalities and advanced delivery techniques such as intensity- modulated radiation therapy (IMRT) and volumetric modulated arc therapy (VMAT) can increase the radiation dose to cancer cells. To get the full benefit, the position of the target volume is needed to be accurate and the magnitudes of both external and internal motion of the patient are minimized.[1] Image-guided radiation therapy (IGRT) is introduced for imaging, volumetric imaging, marker localization, patient surface tracking to improve the precision of radiation therapy delivery. Therefore, various IGRT techniques have been introduced to allow the 3D visualization of the position of the target volume and organ at risk for dose calculation in simulation process and for patient setup verification in the treatment process, Figure 1.1. [2]

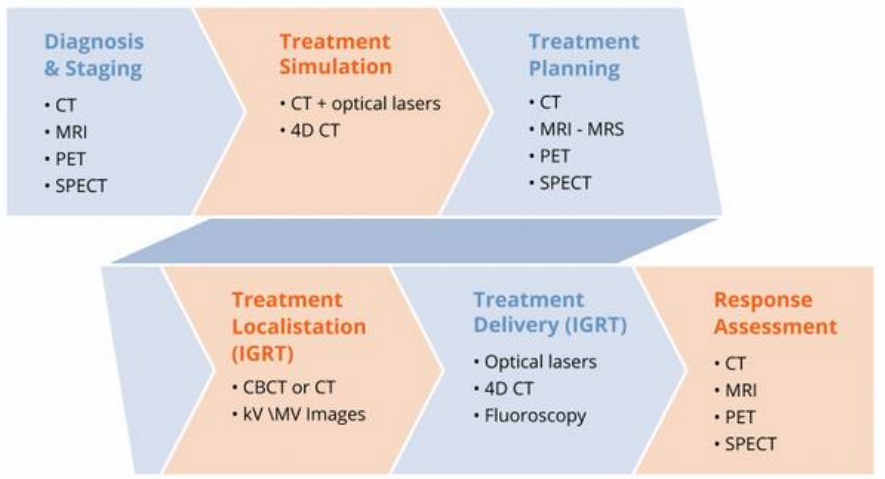


Figure 1. 1The imaging phases involved in radiation therapy process

So, the use of IGRT is to support for safer application of advanced treatment techniques and enables additional margin reduction. Although the use

of imaging modalities in the radiation therapy process can achieve the great benefit, on the other hand, any other use of ionizing radiation can result in additional dose contributions to the integral patient dose. Like the concomitant dose produced by external linac head leakage and scatter, internal direct and scattered therapy dose outside the target volume, as well as nontherapeutic doses from imaging for planning and delivery. Even though radiation therapy patients are already being exposed to very high and localized doses of radiation, the additional radiation from imaging has an associated risk and should be kept low. These additional doses can increase the risk of radiation (stochastic effects) and can increase the severity of side effects from the cancer therapy treatment.[1] As pointed out by the AAPM task group 75 [3] an overview of imaging dose should be evaluated and should find out the way to optimize the dose.

This study was designed to determine the effective doses from 2D conventional simulator, 3D planning computed tomography (3D CT), on-board imaging (OBI) device, portal imaging device and kilovoltage cone-beam computed tomography (kV CBCT) involved in IGRT process.

1.2 Objective

The objective of this research was to determine the organ and effective doses in Image-Guided Radiation Therapy System.

CHAPTER 2

REVIEW OF RELATED LITERATURE

2.1 Theory

2.1.1 Image Guided Radiation Therapy (IGRT) [1,3]

Image-guided radiation therapy (IGRT) is the use of frequent imaging during a course of radiation therapy for the purpose of improving the precision and accuracy of the delivery of treatment. It involves multiple imaging procedures for simulation, planning, setup, and intrafraction monitoring. Imaging modalities are used to reduce the targeting uncertainties in the process of radiotherapy planning and treatment delivery. There are several sources of uncertainty and error in radiotherapy planning and treatment process. These include but may not be limited to uncertainty in target volume delineation, unknown extent of the microscopic tumor, organ positional variation within the patient, and set-up errors. Image-guided radiation therapy aims at reducing geometrical uncertainties by evaluating the patient geometry at treatment and either changing the patient position or adapting the treatment plan with respect to anatomical changes that occur during a course of radiotherapy. The ability to acquire the image in the treatment room immediately prior to irradiation facilitates many possibilities to generate a more accurate image of the tumor's extent and coordinates in 3D space. Verification imaging that obtained before, during, or after treatment can record the patient's position at the time of radiotherapy. The conformation of the dose around the tumor achieves greater healthy tissue sparing, which facilitates the safe implementation of short-course or hypo-fractionated regimens. In IGRT, linear accelerators are equipped with special imaging technology (such as, cone-beam CT) that allow the physician to image the tumor immediately before or even during the time radiation is delivered, while the patient is positioned on the treatment table. These images are then compared to the reference images taken during simulation by using specialized computer software. Any necessary adjustments are then made to the patient's position and/or radiation beams in order to more precisely target radiation at the tumor and avoid healthy surrounding tissue.

IGRT uses of many different imaging techniques and is ranging from portal imaging, diagnostic x-ray imaging, or in-room CT for both conventional and cone-beam with kilovoltage and megavoltage and following regimens as simple as a single setup image or as complex as intrafraction tumor tracking.

(A) Portal Imaging

Portal imaging has progressed from the use of film as the imaging detector, through screen/camera imagers and liquid ionization chambers, to solid-state flat-panel detectors. The electronic portal imaging device (EPID), shown in Figure 2.1, is emerging as a new standard detector for portal imaging in IGRT. A flat panel detector, based on Amorphous-Silicon Flat Panel technology, is mounted on a linear accelerator. In clinical, different image acquisition modes are available: single exposure acquired a single image for a short period of time at the beginning of the treatment and more than one image can be made during the one fraction of treatment. In double exposure mode, a single exposure image is taken and followed by a "larger open field" image to give more information about patient anatomy. EPID also allows on-line fluoroscopy to be acquired during treatment and very useful to investigate the movements of internal organs. The images can be used to analyze the portal images and compare them to treatment planning images for verification and to verify the field placement, characterized by the isocenter relative to anatomical structures of the patient during the treatment. Because of using the megavoltage treatment beam, EPID still has limitations seen in normal portal imaging (the poor contrast between tissues due to Compton Interactions). The advantage of EPID is that new software is developed. The images can be saved in DICOM file format and it can be reviewed at any time.

Figure 2. 1The electronic portal imaging device (EPID) based linear accelerator from Varian

(B) Gantry-mounted kV Imaging

Gantry-mounted kV imaging is the main feature of new medical accelerators specifically for IGRT, mainly due to its in-room imaging capability at the patient treatment position and the diagnosis x-ray quality. There are two commercially available products, Elekta's Synergy X-Ray Volume Imaging (XVI), shown in Figure 2.2 and Varian's On-Board Imager

(OBI) with similar mechanical arrangements and operations. The kV imaging systems are mounted on the gantry orthogonal to the MV treatment and portal imaging system, nominally sharing the same isocenter of the treatment radiation sources. For these devices, their isocenter must be checked in addition to the primary beam and it might be increased the workload needed for quality assurance. But, this imaging provides better image quality and lesser patient dose than the MV electronic portal imaging.

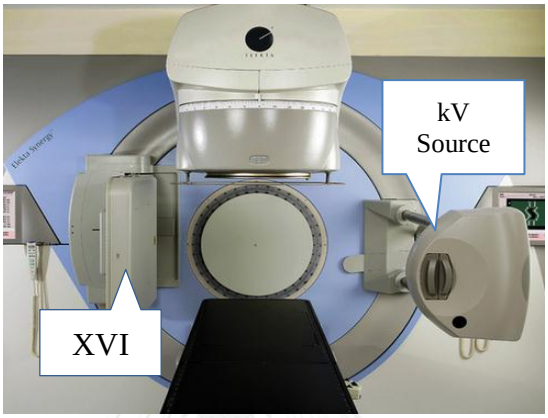


Figure 2. 2 The Elekta Synergy X-Ray Volume Imaging (XVI)

(C) Computed Tomography

Computed tomography (CT) commonly used for radiotherapy consists of a flat-couch and external laser beams to provide the same position as in the treatment machine. It can provide the volumetric 3D imaging and multiple cross-sectional CT slices to be inputted into the treatment planning. The CT is the primary imaging device for radiation therapy due to the accurate spatial information in both external and internal anatomy. It has the bigger bore size than diagnosis CT to allow the immobilization devices and other special patient monitoring device to provide for better treatment. The DRR images provided by CT were used to check the patient positioning on the day of treatment by matching with 2D verification images. The CT images contain the information that the electron densities needed for dose computation during the planning process can be measured. The 4D CT scan also called the breathing-correlated CT which is comprised a large number of individual CT scans obtained at various phases of the respiratory cycle. A small box with infrared reflective markers and a CCD camera at the bottom of the treatment couch is used to monitor the respiration by capturing the movement of the box using the camera. The images are sorted according to the phase of the breathing cycle. This technique can be used to capture the motion of the target volume and the surrounding organs at risk to allow for dose escalation and dose hypo-fractionation.

CT-on-rails/In-room CT, shown in Figure 2.3, is a conventional CT scanner that placed in the treatment room, either on the same couch axis as the linac gantry, or on an orthogonal axis to the gantry. A single couch serves the CT scanner and the beam delivery system. The couch is first moved into alignment with the CT scanner to acquire a pretreatment CT. The CT scanner is mounted on rails so that it, rather than the couch, moves in the axial direction relative to the patient to collect a volumetric scan. The CT data set can be also used without any transformation for treatment planning and dose calculation. Therefore, it is good for target point verification, correction of setup errors and inter-fraction target deviations (with diagnosis quality image) due to organ motion, as well as for recalculation of the actually given dose.

Figure 2. 3 The CT-on-rail installed in the treatment room

(D) Cone-beam CT

Cone-beam CT uses either the therapy beam itself (MV CBCT) with EPID, shown in Figure 2.4, or a diagnostic kV source and a flat-panel detector (kV CBCT), mounted on opposite sides of the patient and orthogonal to the LINAC beam. This technique eliminates the need of using CT scanner in the treatment room. There are two reconstruction modes for CBCT: in full-fan CBCT reconstruction, the detector is centered at the beam axis, therefore, the detector size becomes a limitation for FOV. In order to increase the FOV, the detector center can be shifted from the beam axis so that one side of the anatomy can be fully scanned by the projections with the initial 180-degree gantry rotation and the other side of the anatomy can be completely sampled by the projections with the second 180-degree gantry rotation. The CBCT reconstruction using two clipped sets of data is commonly referred to as half-fan CBCT. The CBCT provides 3D images of patient with acceptable contrast and can be registered to the reference planning CT data set, by means of automatic or manual image registration for calculation of the target position relative to the planned reference position.

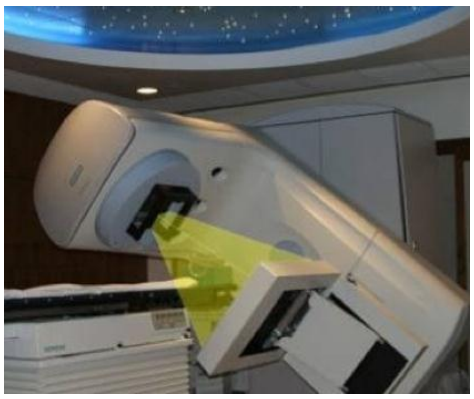


Figure 2. 4 The MV CBCT using the treatment beam and EPID

Moreover, the target motion can also be managed or reduced, for example, by applying the respiratory gating technology to deliver the treatment only at a certain phase of the breathing cycle. This enables improved tumor control and reduction in the toxicity from treatment.

2.1.2 Imaging Dose [3, 4, 13]

2.1.2.1 Basic Dosimetric Quantities

The total imaging radiation dose experienced by a patient can include multiple CT scans for planning, pretreatment fluoroscopic studies to analyze tumor motion, and a series of interfraction and intrafraction images for target localization. The delivery of this dose can be spread out over several weeks during conventional radiotherapy or confined to a short time for hypo-fractionated radiotherapy and radiosurgery. Since the more adapt in IGRT, the imaging modalities are widely used in the radiation therapy process to improve the patient positioning accuracy and therapeutic dose conformity. However, these additional doses can increase the risk of radiation-induced stochastic effect mainly in the young patient and even can cause or increase the severity of side effects from the cancer therapy treatment. The total concomitant dose is increasing steadily with the introduction of more imaging procedures to the treatment process; therefore, it is important to assess all the doses involved in image-guided radiation therapy as the cumulative imaging dose in order to optimize the dose.

(A) Absorbed Dose

In radiation biology, radiology, and radiological protection the absorbed dose, D, is the basic physical dose quantity and is used for all types of ionizing radiation and any irradiation geometry. It is defined as the equation [2.1], the quotient of mean energy, $d\bar{\varepsilon}$, imparted by ionizing radiation

in a volume element and the mass, dm, of the matter in that volume, that is D. The SI unit of absorbed dose is J kg-1 and its special name is gray (Gy).

$$D = \frac{d\bar{\varepsilon}}{dm} \quad [2.1]$$

Absorbed dose is derived from the mean value of the stochastic quantity of energy imparted, ε, and does not reflect the random fluctuations of the interaction events in tissue. While it is defined at any point in matter, its value is obtained as an average over a mass element dm and hence over many atoms or molecules of matter. In principle, absorbed dose is a measurable quantity and primary standards exist to determine its value. The definition of absorbed dose has the scientific rigour required for a basic physical quantity. It implicitly takes account of the radiation field as well as all of its interactions with matter inside and outside the specified volume.

(B) Organ Dose

The organ dose is a quantity defined in relation to the probability of stochastic effects (mainly cancer induction) as the absorbed dose averaged over tissue or organ (D_T), shown in equation [2.2], the quotient of the total energy imparted to the tissue or organ ($\bar{\varepsilon}_T$) and the total mass of the tissue or organ (m_T). The unit is the Joule per kilogram (Jkg^{-1}) and is given the special name Gray (Gy).

$$D_T = \frac{\bar{\varepsilon}_T}{m_T} \quad [2.2]$$

(C) Equivalent Dose

The equivalent dose is a dose quantity used in radiological protection to represent the stochastic health effects of low levels of ionizing radiation on the human body. It is based on the physical quantity absorbed dose but takes into account the biological effectiveness of the radiation, which is dependent on the radiation type and energy. The equivalent dose (H_T) is calculated as in equation [2.3].

$$H_T = \sum(W_R * D_{T,R}) \quad [2.3]$$

Where, $D_{T,R}$ is the mean absorbed dose from radiation R deposited in body tissue or organ T, W_R is the radiation weighting factor which is dependent on the type and energy of the radiation.

The unit is the Joule per kilogram (Jkg^{-1}) and is given the special name Sievert (Sv).

(D) Effective Dose

Effective dose is a quantity defined in International Commission on Radiological Protection (ICRP) Publication 60 and 103 as a weighted sum of equivalent doses to all relevant tissues and organ with the purpose "to indicate the combination of different doses to several different tissues in a way that is likely to correlate well with the total of the stochastic effects". Its calculation is based on the application of tissue –weighting factors (W_T) on the equivalent doses (H_T) absorbed by the various radiosensitive organs of the human body. The unit is the Joule per kilogram (J kg^{-1}) and is given the special name Sievert (Sv). The effective dose (E) is calculated as in equation [2.4]

$$E = \sum(W_T * H_T) \qquad [2.4]$$

Where, W_T is the tissue weighting factor for tissue and $\sum W_T = 1$.

The effective dose has been introduced by ICRP as the primary radiological protection quantity to be used for setting and controlling dose limits in the regulatory context and for enabling the practical implementation of the optimization principle.

The ICRP 103 recommendations update the radiation and tissue weighting factors in the quantities equivalent and effective dose and update the radiation detriment, shown in Table 2.1, based on the latest available scientific information of the biology and physics of radiation exposure.

Table 2. 1Organ and tissue weighting factors publication 103 recommended for ICRP

Tissue/ Organ	ICRP 103 (W_T)
Red bone-marrow	0.12
Colon	0.12
Lung	0.12
Stomach	0.12
Breast	0.12
Gonads	0.08
Bladder	0.04
Liver	0.04
Esophagus	0.04
Thyroid	0.04
Skin	0.01
Bone surface	0.01
Brain	0.01
Salivary gland	0.01
Reminder organs *	0.12

*Remainder organ of extra thoracic region, lymphatic nodes and oral mucosa, are approximated by thyroid, muscle and brain.

(E) Stochastic Effects

These effects result from an alteration in the genome of a cell, which, in the case of cancer gives rise to a clone of uncontrolled, rapidly dividing cells. Various mechanisms might be involved in mutation of the DNA. These may include the activation of an oncogeny (a cancer-causing gene), the inactivation of a tumor-suppressor gene or the loss of function of a repair-mutator gene. These can give rise to carcinogenesis and hereditary effects. In this effect, the severity of radiation damage is not related to the dose and the probability of occurrence increases with increasing radiation dose, e.g., development of cancer. There is no threshold for stochastic effects.

2.1.2.1 Specific Dosimetric Quantities for Evaluating the Imaging Dose

Different imaging technique may deliver different quality of radiation distribution than the treatment beam.

In planar imaging the dose to the patient is greatest at the skin surface nearest to the source and falls off progressively as the radiation transits the body to the image detector, the maximum dose is at the skin's surface. Therefore, entrance skin dose is one standard measure of diagnostic imaging dose. [3]

(A) Incident Air Kerma (K_i) [4, 13]

The incident air kerma (K_i) is the sum of kinetic energy of all charged particles liberated per unit mass. It is the kerma to air from an incident x-ray beam measured on the central beam axis at the position of the patient or phantom surface. Only the radiation incident on the patient and not the backscattered radiation are included. The name for the unit of kerma is gray (Gy) and can be calculated by the equation:

$$K_i = M_Q \, N_{K,Q0} \, k_Q \, k_{TP} \qquad [2.5]$$

Where, K_i = incident air kerma, M_Q = mean value of the dose meter reading, $N_{K,Q0}$ = dosimeter calibration coefficient, k_Q = radiation quality factor and k_{TP} = temperature , pressure correction factor.

(B) Entrance Surface Air Kerma (ESAK) [13]

The entrance surface air kerma (K_E) is the kerma to air measured on the central beam axis at the position of the patient or phantom surface. The radiation incident on the patient or phantom and the backscattered radiation are included. It is related to the incident air kerma by the backscatter factor, B, therefore: it can be calculated as in equation [2.6].

$$K_E = K_i \, B \qquad [2.6]$$

The unit for kerma is gray (Gy).

(C) CT Dose Index (CTDI)

In axial imaging, 3D CT, the dose is distributed nearly uniformly throughout the imaged volume to produce 3D images of uniform cross-sectional quality. Therefore, CT dose quantity, CT dose index ($CTDI_{100}$), being a measure of internally deposited dose in a phantom, does include scatter. [3] For kilovoltage CT, there has air kerma on the axis of rotation in mGy, with or without scatter included, i.e., CTDI in air ($CTDI_{air}$) and in a phantom ($CTDI_w$).

The CT dose index is a standardized measurement of radiation output of a CT scanner which allows the user to compare radiation output of different CT scanner. It was defined as the integral of the dose profile, D (z), from a single axial scan along a line perpendicular to the tomographic plane (z-axis) divided by the nominal slice thickness (T), shown in equation [2.7]:

$$CTDI = \frac{1}{T} \int_{-\infty}^{+\infty} D\,(z)\ .dz \qquad [2.7]$$

For a multi-slice scanner with N simultaneously acquired slices of nominal thickness T (nominal width of irradiated beam NT), the following equation [2.8] is used;

$$CTDI_{100} = \frac{1}{T}\int_{-50}^{+50} D\,(z)\,.\,dz \qquad [2.8]$$

$CTDI_{100}$ is measured by 100 mm pencil type ionization chamber, both in free air and within standard CTDI phantoms having the diameter of 16 cm and 32 diameters.

CTDI air is the CTDI value measured with the pencil ionization chamber position in free air.

CTDI center is the CTDI value measured with the pencil ionization chamber position in center of the phantom.

CTDI peripheral are defined as the CTDI value measured with the pencil ionization chamber within the four positions (12 o'clock, 3 o'clock, 6 o'clock, 9 o'clock) of the phantom.

The weighted CTDI is the combined value of values of CTDI measured at the center and periphery of a standard CTDI phantom and used for approximating the average dose over a single slice and is defined by the equation [2.9],

$$\mathrm{CTDI}_{100}^{\mathrm{weighted}} = \frac{1}{3}\mathrm{CTDI}_{100}^{\mathrm{center}} + \frac{2}{3}\mathrm{CTDI}_{100}^{\mathrm{periphery}} \qquad [2.9]$$

The normalized CTDI is the CTDI value normalized by the 100mAs (Absorbed dose/ 100mAs). The unit for all the CTDI values is Gray (Gy).

(D) Cone-beam Dose Index (CBDI) [7]

For CBCT, the term CTDI is changed to cone-beam dose index: CBDI. The CBCT acquires a volumetric image in a single rotation instead of sequential in conventional CT; there is no need to consider the contribution of scatter from adjacent slices. Therefore, the use of pencil ionization chamber in CBCT provides a measure of the average dose over the central 100 mm of the longitudinal FOV and which is slightly underestimate the dose at the center of the dose profile (maximum at the center and decrease to the boundaries). The $\mathrm{CBDI}_{100}^{\mathrm{air}}$ is the CBDI measurement performed in air and, $\mathrm{CBDI}_{100}^{\mathrm{center}}$ and $\mathrm{CBDI}_{100}^{\mathrm{periphery}}$ represent the center and the average of the four peripheral dose measurements performed in CTDI phantom. The weighted CBDI ($\mathrm{CBDI}_{100}^{\mathrm{weighted}}$), is derived to represent the average volumetric dose in the CTDI phantom and calculated as equation [2.10]:

$$CBDI_{100}^{weighted} = \frac{1}{3} CBDI_{100}^{center} + \frac{2}{3} CBDI_{100}^{periphery} \quad [2.10]$$

2.1.2.2 The ImPACT CT Patient Dosimetry Calculator [14]

This spreadsheet is a tool for calculating patient organ and effective doses from CT scanner examinations. It uses the Monte Carlo dose data set from the irradiation of a mathematical phantom by a range of CT scanners. The measured CTDI values, type of scanner, range of the scan, the kVp and mAs are used as input to estimate the organ dose. The International Commission on Radiological Protection (ICRP) Publication 60 or 103 which have different organ weighting factor can be applied to estimate the effective dose. The dosimetry data obtained from ImPACT calculation spreadsheet are based on a mathematical phantom that represents the human body with simple geometric objects. The measured CTDI/CBDI values, the exposure parameters (kVp, mAs, pitch and scan range) are needed to use the ImPACT calculation spreadsheet to estimate the organ dose. The CT Dosimetry calculation program consists of 12 worksheets:

1. Introduction: Provides an introduction and instructions for use
2. Scan Calculation: The data entry and results sheet
3. Paediatric: Information on relative doses to adult and paediatric patients
4. Phantom: Allows interactive selection of the scan range used for dose calculation using a diagram of the phantom used to generate SR250
5. Scanners: Provides data on CT scanner models, including CTDI in air and phantom, as well as the scanner matching data
6. Match Data: Gives data required to perform the scanner matching in the Scanners worksheet
7. Collimation: Lists relative CTDI values at different collimations for a range of CT scanners. These values are more useful for multi-slice scanners, as the CTDI can vary considerably over the range of available collimations
8. Monte Carlo Data: Contains the unformatted SR250 data set.
9. Doses: Contains the formatted dose data from the SR250 data set that is currently loaded.
10. Dose Calculations: Performs the organ dose calculations, and calculation of remainder organ doses etc.
11. Selections: Provides data for the drop down selection boxes in the Scan Calculation worksheet, and performs calculations for 'remainder' organ doses

12. Version: Details changes made in each version, from version 0.99e onwards

In the case of modern CT scanner and Cone-beam CT, the ImPACT factor is used to match with the scanner that is included in the software. The ImPACT factor can be derived by the equation [2.11];

$$\text{ImPACT factor} = a*\frac{CBDI_{100}^{center}}{CBDI_{100}^{air}} + b*\frac{CBDI_{100}^{periphery}}{CBDI_{100}^{air}} + c \qquad [2.11]$$

Where, $CBDI_{100}^{air}$, $CBDI_{100}^{center}$ and $CBDI_{100}^{periphery}$ are the Cone-beam Dose Index measured in air and measured in center and periphery positions of phantom, respectively.

a, b, c are empirical factors developed by the ImPACT group.

2.1.2.3 The Thermoluminescent Dosimetry [16, 19]

Thermoluminescent detectors (TLD) are based on the phenomenon of visible photons released by heat and this characteristic was used for the purpose of dosimetry. The TLD-100H, lithium fluoride (LiF) crystal activated with magnesium, copper, and phosphate (LiF:Mg,Cu,P) were used to measure the organ dose from cone-beam CT pelvis protocol. They have the properties of high sensitivity, low fading rate, and high energy response. The preparation TLD includes many steps starting from the activation and annealing of detectors to make the reproducibility for accurate dosimetry. The sensitivity, linearity, the minimal detectable dose and the energy dependency must be determined before doing the dose measurement.

2.1.2.3.1 Calibration of Thermoluminescent Dosimeter

The calibration of thermoluminescent dosimeters was performed to produce the consistent and accurate reading in dosimetrically meaningful unit. The calibration process involved the following 3 steps:

(A) Generate Calibration Dosimeter

In this process, an element correction coefficient (ECC) is generated by using a set of dosimeters, typically 1-2 % of the total population to be calibration dosimeters. They are identified and segregated from the field dosimeters.

All dosimeters are annealed to clear all of their residual exposure. The time duration between the annealing and exposing should be the same for all dosimeters. After being exposed to the known radiation dose, the charge integral value (Q_i) in nanocouloumb (nC) of each dosimeter (i) is read out and recorded. Then, the average charge integral ($\bar{Q}$) of all dosimeters is

calculated and the element correction coefficient (ECC_i) for individual dosimeter I (I = 1,2,3,…,n) is computed by dividing the average charge integral by the individual charge (Q_i) as in equation [2.12]:

$$ECC_i = \frac{\bar{Q}}{Q_i} \quad [2.12]$$

(B) Calibration of Thermoluminescent Dosimeter Reader

Groups of dosimeter about 1-2 % of dosimeters which have ECC_i value close to 1 are chosen to be calibration dosimeters. The calibration dosimeters are exposed to known amount of radiation dose (D) in grays and read by thermoluminescent dosimeter reader. As Q_i is the reading for the dosimeter I, the corrected charge integral (Q_{ci}) of the dosimeter is calculated by equation [2.13]:

$$Q_{ci} = Q_i \cdot ECC_i \quad [2.13]$$

Then the reader calibration factor (RCF), used for convert TLD signal to radiation dose, and is calculated from the equation [2.14]:

$$RCF = \frac{\overline{Q_c}}{D} \quad [2.14]$$

When $\overline{Q_c}$ is the average corrected charge integral (Q_{ci}) and calculated by equation [2.15]:

$$\overline{Q_c} = \frac{1}{n} \sum_{i=1}^{n} Q_{ci} \quad [2.15]$$

(C) Calibration of Field Dosimeter

The rest of the dosimeters are used as field dosimeters. They are exposed by the known radiation dose of L ray and read by the thermoluminescent dosimeter reader. The calibration value of the element correction coefficient for the individual dosimeter (ECC_i) is then calculated by the equation [2.16]:

$$ECC_{ci} = \frac{RCF \,.\, L}{Q_i} \quad [2.16]$$

2.1.2.3.2 Determination of Unknown Radiation Dose

The field dosimeters represented as *i* are used to measure the unknown radiation dose. The unknown dose D in gray is calculated by using ECC_{ci} and Q_i (the reading of field dosimeter i) from the equation [2.17]:

$$D = \frac{Q_i \ * ECC_{ci}}{RCF} \quad [2.17]$$

2.2 Literature Review

Hälga R. A. et al. [1] studied the absorbed dose measurements in an anthropomorphic phantom from typical imaging procedures involved in an image-guided radiation therapy treatment for a complete course of treatment. The series of measurements were performed from several imaging irradiations of the imaging modalities in conjunction with treatment machines from the manufacturers Accuray, Elekta, Siemens, and Varian. The imaging techniques investigated on these treatment machines were kilovoltage and megavoltage cone beam computed tomography, megavoltage fan beam computed tomography, kilovoltage planar imaging, stereoscopic kilovoltage imaging, and megavoltage portal imaging. The absorb dose measurements were performed using the individually calibrated thermoluminescent dosimeters (TLDs) placed inside the Alderson-Rando phantom at 184 locations and the resulting effective dose was calculated. The results of effective doses additional to the treatment stray dose of outside the treated volume were less than 1mSv for conventional 3D planning CT, while a 4D planning CT resulted in a 10 times larger dose, less than 0.4 mSv and 1.4 mSv for a daily setup of the patient with two planar kilovoltage images or with a fan beam CT at the TomoTherapy. Using kilovoltage or megavoltage radiation to obtain cone-beam computed tomography (total course of treatment taken once per fraction) scans led to an additional dose of 8–46 mSv. For treatment verification images performed once per week using double exposure technique, an additional effective dose of up to 18 mSv was measured. Depending on the imaging scheme applied, image-guided radiation therapy can be administered without increasing the dose outside of the treated volume compared to therapies without image guidance.

Cheng H. C. Y.et al., [6] investigated organ absorbed dose, total effective dose, and image quality of the CBCT system for the head-and-neck and pelvic regions. The absorbed dose of different organs was measured in a female anthropomorphic phantom with thermoluminescent dosimeters (TLD) and the total effective dose was estimated according to International Commission on Radiological Protection (ICRP) Publication 103 and comparison was made between the new and old protocols, version 1.4.13 and version 1.4.11. The image quality of the head-and-neck protocols, the low-contrast resolution and CNR for the new protocol was poorer than those of the old protocol. However, both protocols demonstrated the same high-contrast spatial resolution. The total effective doses from the new and old head scanning protocols were 1.65 and 9.39mSv; half-fan pelvic scanning protocols were 1.7 and 8.2 mSv, respectively. The total effective doses from new version were much lower than the old version. It was due to lower exposure parameters for new version scanning protocols. The additional secondary cancer risk from daily CBCT might be up to 2.8%. The new Varian CBCT provided volumetric

information for image guidance with acceptable image quality and lower radiation dose.

Hyer D.E. et al, [7] showed a practical method for estimating organ doses from kilovoltage cone-beam CT (kV CBCT) in the Elekta x-ray volumetric imager software version 4.0 and the Varian on-board imager software version 1.4.13 at three clinically relevant scan sites: Head, chest, and pelvis. The measurements were performed by using an in-house adult male anthropomorphic phantom and fiber optic coupled dosimetry system. The results were used to evaluate the accuracy of estimated organ dose by the ImPACT patient dose calculator. A 100 mm pencil chamber and standard CT dose index (CTDI) phantoms were used to measure the cone-beam dose index (CBDI). Then, weighted CBDI (CBDIw) was calculated from these measurements to represent the average volumetric dose in the CTDI phantom. The measured CBDI values were also used as inputs for the ImPACT calculator to estimate organ doses. All CBDI dose measurements were performed on both the Elekta XVI and Varian OBI at three protocols: Head, chest, and pelvis. The head, chest, and pelvis protocols result CBDIweighted values of 0.98, 16.62, and 24.13 mGy for the XVI system and 5.17, 6.14, and 21.57mGy for the OBI system, respectively. Organ doses estimated with the ImPACT CT dose calculator showed a large range of variation from the previously measured organ doses because of difference in phantoms used for calculation and measurement, and becoming its limitations for use with CBCT. The organ dose conversion coefficients result in related CBDIweighted values to organ doses previously measured using the same clinical protocols. These coefficients allow for the quick estimation of organ doses from routine measurements performed using standard CTDI phantoms and pencil chambers.

CHAPTER 3

RESEARCH METHODOLOGY

3.1 Research Design

This study is an observational analytical study.

3.2 Research Question

The objective of this study is to determine the organ and effective doses in Image-Guided Radiation Therapy system.

3.3 Research Design Model

The research design model of this study was separated into two parts for imaging modalities used in planning process, Figure 3.1, and those involved in verification process, Figure 3.2.

3.3.1 Research Design Model for Planning Imaging Modalities

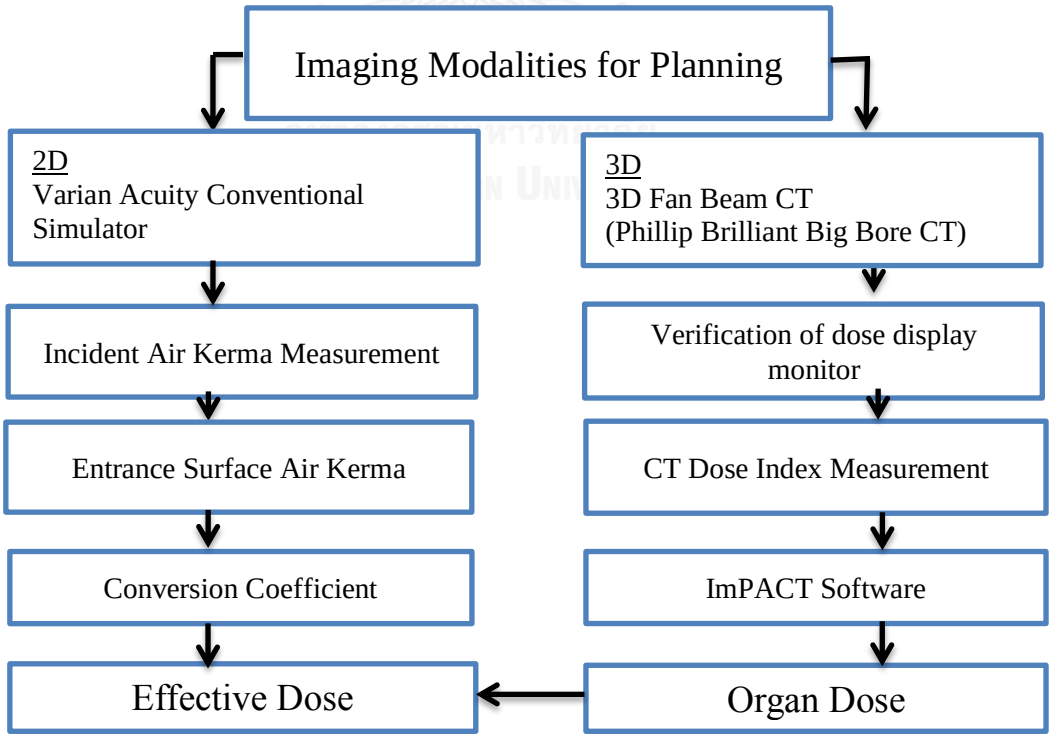


Figure 3. 1 Research design model for planning imaging modalities

3.3.2 Research Design Model for Imaging Modalities Involved in Verification Process

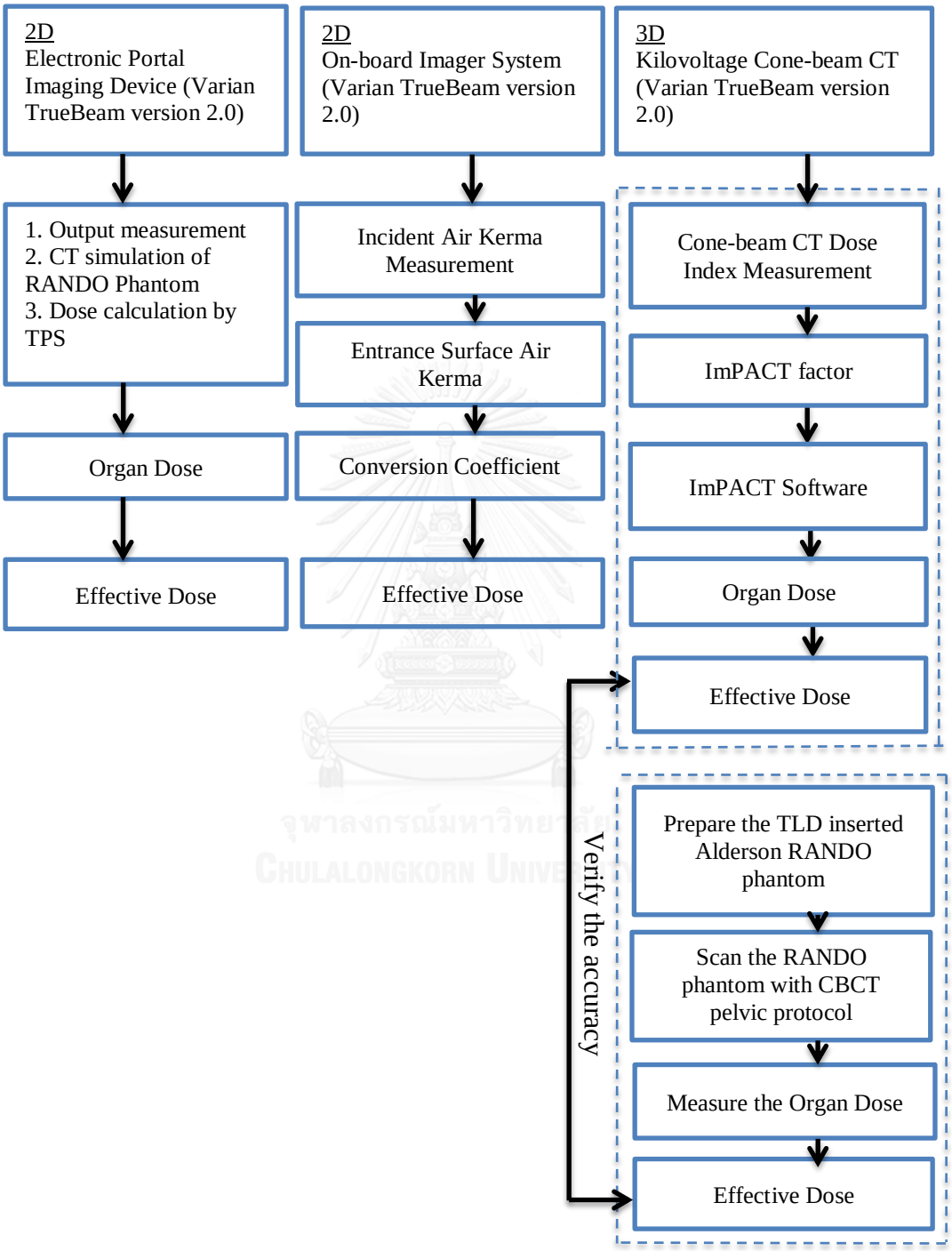


Figure 3. 2 Research design model for imaging modalities involved in treatment verification

3.4 Conceptual Framework

The effective dose of the patient from imaging modalities used in IGRT is affected by many factors such as CTDI / CBDI measured value, exposure parameters, Entrance Surface Air Kerma (ESAK), radiation types and tissue weighting factors, shown in Figure 3.3.

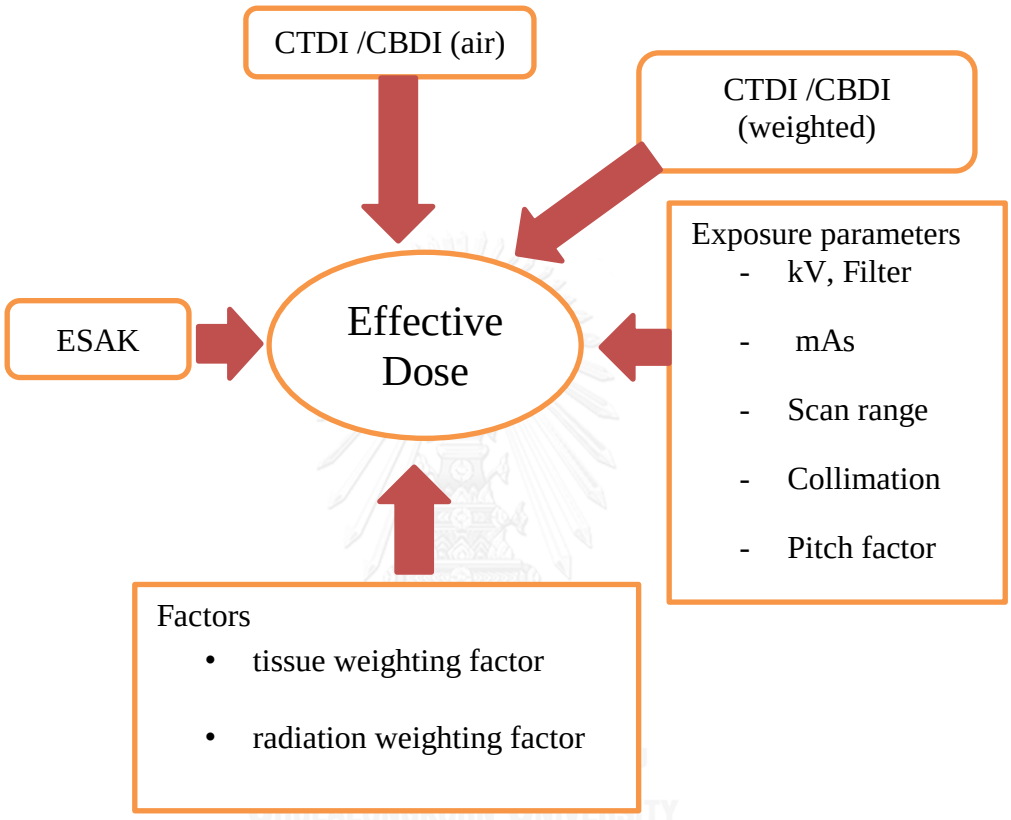


Figure 3. 3 Conceptual Framework

3.5 Materials

The materials used in this study are from the Department of Therapeutic Radiology and Oncology, King Chulalongkorn Memorial Hospital.

3.5.1 Imaging Modalities for Planning

3.5.1.1 Acuity Simulator (Varian Medical System)

The Acuity planning, simulation and verification system (Varian medical Systems, Palo Alto, USA) shown in Figure 3.4, consists of kilovoltage x-ray source and detector. It features a rotating gantry that moves in exactly the same geometry as a rotating gantry on a megavoltage radiation therapy unit. It

has the console for kV, mA and the timer setting and also control that allows the couch and gantry position to be moved remotely. It contains laser patient positioning and marking system and allows the beam direction and treatment field to be determined to include the target volume and spare normal structures excessive radiation. It gives the 2D reference images for planning.

Figure 3. 4 Acuity simulator (Varian Medical System)

3.5.1.2 Philips Brilliance Big Bore CT

The Philips Brilliance Big Bore CT scanner (Philips Health Care, Guildford, UK) shown in Figure 3.5 is a 16 slices multi-detector CT scanner. It has an 85 cm bore and true 60 cm scan field of view with the kVp range of 90-120-140. This 85 cm bore size can be used to scan patients with immobilization devices, patient monitoring, intravenous delivery devices, respiratory devices and other apparatus without compromising image quality or positioning. Moreover, treatment planning can be performed at the time of virtual simulation by contouring the organs and volume of interest and determining the isocenter. It allows for the generation of DRRs for patient positioning in case of 2D matching and the 3D CT images are used to verify the position with cone-beam CT images.

Figure 3. 5 Philips Brilliance Big Bore CT (85 cm bore)

3.5.2 Imaging Modalities for Verification

3.5.2.1 Electronic Portal Imaging Device (EPID)

The Electronic Portal Imaging Device (EPID) shown in Figure 3.6 from the TrueBeam linear accelerator (Varian medical Systems, Palo Alto, USA) is performed for treatment localization and verification. It has the different acquisition modes; single exposure mode and double exposure mode. In double exposure mode is commonly selected that, one image is taken by using the planned field size and the second is a larger open field image to give more information about patient anatomy. This technique uses the megavoltage treatment beam to acquire the 2D images (usually AP and Lateral). These images are used to determine the position of target in relation to the bony landmarks and also used to check patient setup before each treatment. It can be used for online analysis by determining the unacceptable discrepancies between the portal image and reference image (simulator or DRR image) and make a decision about continuation of treatment.

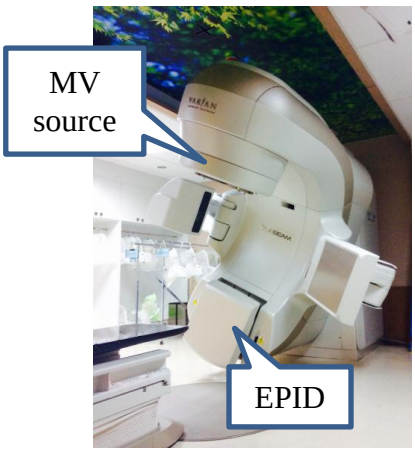


Figure 3. 6 Electronic Portal Imaging Device from Truebeam (Version 2.0)

3.5.2.2 Kilovoltage On-Board Imaging

The On-Board Imager (OBI) system attached to the TrueBeam linear accelerator (Varian medical Systems, Palo Alto, USA) shown in Figure 3.7 is designed to correct for motion and setup errors in patients undergoing radiation therapy. The OBI system provides three imaging modes: radiographic and fluoroscopic image acquisition and three-dimensional cone-beam computed tomography (CBCT) acquisition. It is placed on the gantry at a 90 degree offset to the primary beam. It has the similar detection system to electronic portal imaging device (EPID).It has the benefit of using kilovoltage x-ray which provides improve contrast over standard portal images. These diagnostic quality images can be used to compare with digital reconstruction radiographs (DRRs) originating in CT simulation according to bony structure with 2D/2D match (to confirm a match of treatment setup to the planned treatment).

3.5.2.3 Cone-beam CT Mounted on the Treatment Machine

The TrueBeam Cone-beam CT system (Varian medical Systems, Palo Alto, USA) shown in Figure 3.7 includes the kilovoltage x-ray source and on-board imager, is attached on the treatment machine. It provides 3D images of patient. This CBCT verification image is registered to the reference planning CT data set, by means of automatic or manual image registration for calculation of the target position relative to the planned reference position. These images are used for monitoring of anatomical deformations such as weight loss and tumor regression and patient positioning at the time of treatment and target localization throughout the treatment course.

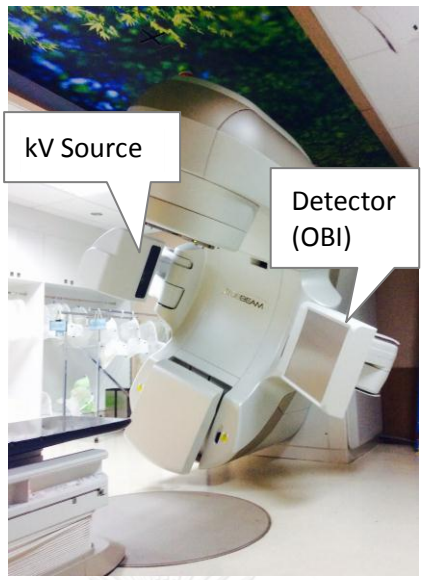


Figure 3. 7 The OBI and Cone-beam CT attached to the Truebeam (Version 2.0) linear accelerator

3.5.3 The 6 CC Ionization Chamber

The 6 cc ionization chamber (Radcal corporation, Monrovia, CA), shown in Figure 3.8, is a wide dynamic range (10 μR to 59 kR) chamber with many dose and dose rate application. The calibration accuracy is within ±4% using X-rays at 60kVp (2.8 mmAL HVL) and ±5% energy dependence (30 keV to 1.33 MeV range) with build-up material.

Figure 3. 8 A 6 cc ionization chamber

3.5.4 The 100 mm Pencil Ionization Chamber

The pencil ionization chamber (Radcal corporation, Monrovia, CA) shown in Figure 3.9 was used in this study for computed tomography (CT) and cone-beam CT dosimetric measurements. It is designed specifically for CT X-ray beam measurements, either free-in-air or mounted in a head or body phantom. The chamber has the excellent energy and partial volume response as well as uniformity along its entire 100 mm active length. The minimum detectable dose is 20 μR with ±4% accuracy and ±5% energy dependence, 3 to 20 HVL range.

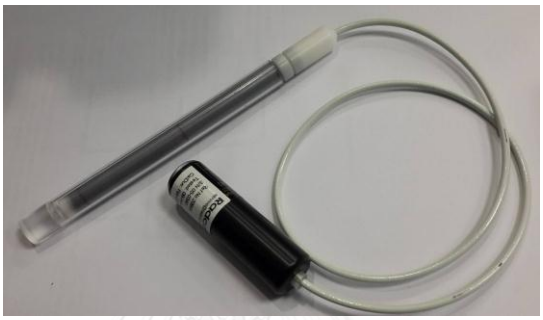


Figure 3. 9 A 100 mm pencil ionization chamber

3.5.5 ACCU- PRO Dosimeter

The ACCU-PRO dosimeter (Radcal corporation, Monrovia, CA), shown in Figure 3.10, was used in this study. It can measure kVp, Time, Dose, Dose Rate, mA/mAs, HVL and filtration detection.

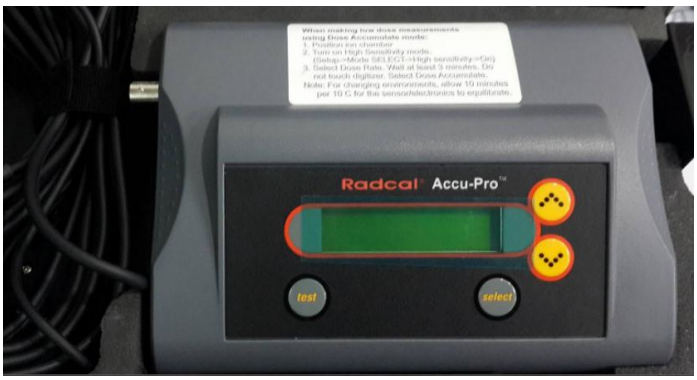


Figure 3. 10 Accu pro dosimeter

3.5.6 CTDI Polymethyl Methacrylate Acrylic (PMMA) Phantoms

The CTDI head and abdomen phantom are shown in Figure 3.11. It is made from (PMMA) polymethyl methacrylate and includes a 16 cm diameter head phantom and 32 cm diameter body phantom. The phantom is constructed with a center hole and eight peripheral holes (four in the head phantom and four in the body phantom) for CT ion chamber placement. It also includes PMMA inserts for peripheral holes.

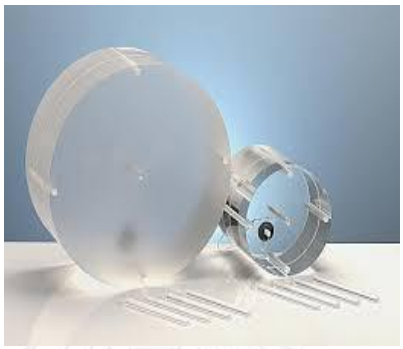


Figure 3. 11 CTDI Phantoms

3.5.7 The ImPACT CT Patient Dosimetry Calculator (version 1.0.2)

In this study, the ImPACT (Imaging Performance Assessment of CT scanners) (ImPACT, Knightsbridge Wing, St George's Hospital, London) CT patient dosimetry spreadsheet was used to calculate the organ and effective doses from Fan beam CT and cone-beam CT. This spreadsheet is a tool for calculating patient organ and effective doses from CT scanner examinations. It makes use of the NRPB Monte Carlo dose data sets produced in report SR250. To calculate doses using CTDosimetry.xls, the user must enter a number of parameters relating to the scanner and the scan series as shown in Figure 3.12. This spreadsheet is distributed freely but remains copyright ImPACT.

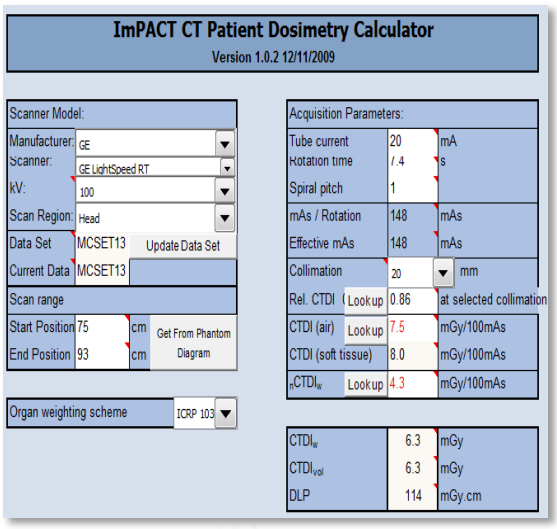


Figure 3. 12 The ImPACT Software (version 1.0.2)

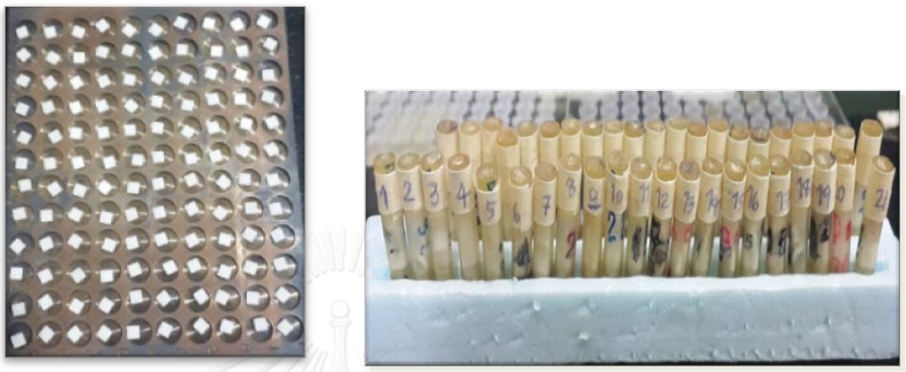
3.5.8 The Female Alderson RANDO Phantom

The female Alderson RANDO phantom (RSD, Long Beach, CA) which is shown in Figure 3.13 is constructed with a natural human skeleton cast inside the tissue equivalent material with a density of 0.985 g/cm^3 and effective atomic number of 7.3 based on the International Commission on Radiation units and Measurement (ICRU) Report No. 44. The phantom is transected-horizontally into 2.5cm thick slices and shaped into a human torso. Each slice has matrix of 0.5 cm diameter holes spaced 3 cm apart which are plugged with bone, soft tissue or lung tissue equivalent pins which can be replaced by TLD holder tubes.

Figure 3. 13 A female Alderson RANDO phantom

3.5.9 Thermoluminescent Dosimeters (TLD)

In this study, the TLD 100-H, chip type dosimeters shown in Figure 3.14 (a) were used for dose measurement. Each chip has a dimension of 3.2 x 3.2 x 0.89 mm^3 and has a nominal density of 2.64 g/cm^3 and effective atomic number of 8.2. And Figure 3.14 (b) showed the three pieces of TLD chips those placed into the plastic tube, to put inside the phantom.



(a) (b)

Figure 3. 14 (a) The TLD chips and (b) the plastic tubes with TLD

3.5.10 The Harshaw Model 5500 Automatic TLD Reader

The Harshaw model 5500 automatic TLD reader (Thermo Fisher Scientific, E.L, Germany) shown in Figure 3.15 is a personal computer driven, tabletop instrument for thermoluminescent dosimeter measurement. This reader is capable of reading 50 dosimeters per loading and accommodates TLD chips, rods and cubes in a variety of sizes. It uses hot nitrogen gas heating with a closed loop feedback system that produces linearly ramped temperatures accurate within ± 1°C to 400°C. Nitrogen is routed through the PMT chamber to eliminate condensation.

Figure 3. 15 The Harshaw model 5500 automatic TLD reader

3.5.11 The PTW –TLDO Oven (Annealing Oven)

The TLD annealing oven from PTW –TLDO, shown in Figure 3.16, was used to anneal the TLDs to eliminate the residual TL signals. The temperature accuracy of annealing oven is 0.5% ± 1 digit. The annealing oven was controlled by THELDO program which is used to create and process the heating program.

Figure 3. 16 PTW TLDO Oven

3.5.12 Eclipse Treatment Planning System Version 11.0.31

Eclipse treatment planning system version 11.0.31 (Varian medical Systems, CA, USA) shown in Figure 3.17 is an integrated and comprehensive treatment planning system supporting radiation treatment modalities like photon, including FFF beams, proton and electron external beams treatment options, low dose rate brachytherapy and Cobalt. It provides the two photon dose calculation algorithms, Analytical Anisotropic Algorithm (AAA) and the new algorithm Acuros XB and electron Monte Carlo. It can support dosimetrists, physicists, and physicians efficiently to create, select and verify the best treatment plans for the patients.

Figure 3. 17 The Varian Eclipse treatment planning system

3.6 Method

All the exposure parameters employed in our study were the average of clinical patient data.

3.6.1 Effective Dose Calculation from Imaging Modalities Used for Planning

3.6.1.1 Half-value Layer (HVL), Measurement from Acuity Conventional Simulator

The HVL measurement was performed by using a 6 cc ionization chamber, Accu-Pro dosimeter, and aluminum (Al) attenuators. The measurement was done as followed:

1. Set the ionization chamber at isocenter in the x-ray beam and connect the chamber to the dosimeter.
2. Collimate the field size to obtain the narrow beam geometry (just cover the chamber) so as to minimize the interruption from scatter radiation and select the exposure parameter.
3. Expose the chamber without Al attenuators and record the measured value.
4. Insert the Al attenuator, exposed the chamber and record the measured value.
5. Repeat the step (4) until the measured value is less than 50% of the value from step (3).
6. Calculate the HVL.

3.6.1.2 Effective Dose Calculation from Acuity Conventional Simulator (2D)

In 2D imaging system, the effective doses from Varian Acuity conventional simulator, electronic portal imaging device (EPID) which is attached to the TrueBeam (version 2.0) linear accelerator was calculated for each clinical protocol of the head, chest and pelvic regions.

For effective dose calculation from Acuity Simulator, the incident air kerma measurement were performed by placing a 6 cc ionization chamber with support stand at a 20cm distance from the couch to avoid back scatter as displayed in Figure 3.18 (a). Then, it was connected to the Accu Pro dosimeter and exposed with the selected exposure parameters according the clinical

protocols in Table 3.1. The dose from dosimeter was recorded and the measurements were repeated about five times to calculate the mean and SD value. By using the Accu Pro dosimeter, the temperature and pressure were automatically corrected by itself. Then, the entrance surface air kerma was calculated by multiplying the incident air kerma with the backscatter factor. The effective doses for each clinical procedure for head, chest and pelvis region were directly obtained from the entrance skin dose in miligray multiply with the conversion coefficient. [3] The conversion coefficients were selected according to the field size, kVp and HVL for each procedure.

(a) (b)

Figure 3. 18 (a) Chamber setup (b) field size opened just to cover the chamber to avoid backscatter for Incident Air Kerma measurement

Table 3. 1The clinical acquisition parameters from Acuity conventional simulator

Parameters	Head		Chest		Pelvis	
	AP	LAT	AP	LAT	AP	LAT
kV	85	85	95	95	110	120
mA	80	80	80	80	80	80
mAs	9.9	20	10	20	18	20
Field size (cm^2)	15x15	15x15	20x20	20x20	20x20	20x20

3.6.1.3 The Organ and Effective Dose Calculation from Computed Tomography System (3D)

The Computed Tomography Dose Index (CTDI) measurements were performed for effective dose calculation of Phillip Brilliance Big Bore CT scanner. For this measurement, the 100 mm pencil ionization chamber with

PMMA CTDI phantoms 16 cm in diameter for head protocol and 32 cm in diameter for chest and pelvis protocol were used, Figure 3.19.

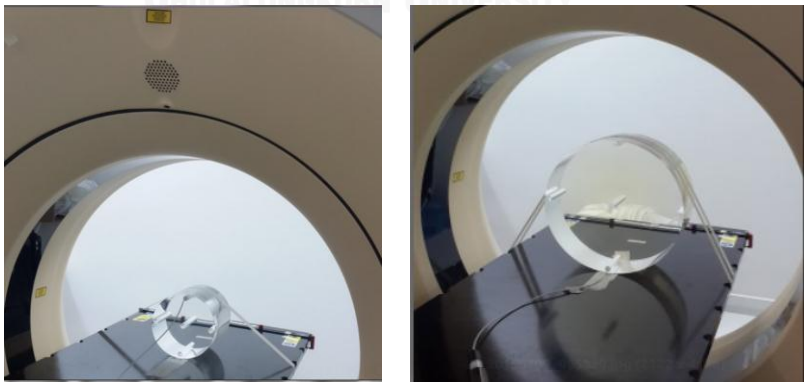
1) Firstly, CTDI in air measurement was performed at the isocenter by placing the ionization chamber in air with the support stem at 20 cm distance above the treatment couch. Then, it was connected with the Accu Pro dosimeter for dose reading and exposed for five times with selected protocol, shown in Table 3.2. The mean dose was taken from five times reading and the normalized CTDI air was calculated.

2) Secondly, the CTDI center and peripheries measurements were performed by using CTDI phantom. For CTDI center, the chamber was inserted at the center of the corresponding phantom depending on the selected protocol. The other peripheral holes were filled with PMMA inserts. Then, chamber was connected with the dosimeter for dose reading and exposed for five times with selected protocol. The mean dose was taken from five times reading and the normalized CTDI center was calculated.

For CTDI peripheries, the chamber was placed at the twelve, three, six and nine o'clock positions. For each position, we followed the above measurement steps (CTDI ^{center}). After that, normalized CTDI ^{periphery} was calculated for each position.

3) Thirdly, the CTDI weighted was calculated by using the equation [2.9].

4) Finally, CTDI ^{air}, CTDI^{weighted}, type of CT unit, total mAs and scan range of each protocol were entered into the ImPACT software to calculate the organ and effective doses.



(a) (b)

Figure 3. 19 The 100 cc ionization chamber at the center of (a) the head phantom and (b) body phantom.

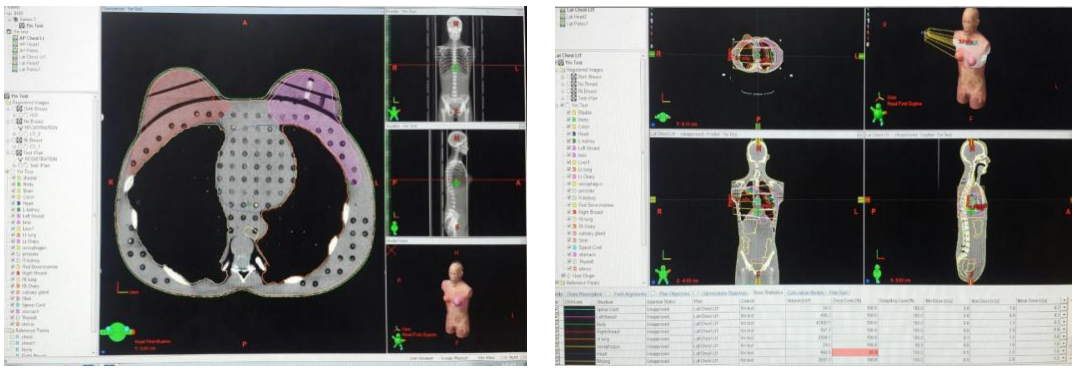
Table 3. 2 Acquisition parameters for planning CT

Parameter	Head	Chest	Pelvis
kVp	120	120	120
mA	263	406	325
Collimation (mm)	16 x 1.5	16 x 1.5	16 x 1.5
Rotation Time (second)	0.75	0.75	0.75
Pitch	0.563	0.938	0.813
Scan Length (cm)	24	43	34
$CTDI_{air}$ (mGy)	15.23	15.24	15.24
$^{n}CTDI_{weighted}$ (mGy)	12.41	8.05	8.05
$CTDI_{vol}$ (mGy)	32.2	27.9	27.9
DLP (mGy.cm)	790	990	781

3.6.2 Effective Dose Calculation from Imaging Modalities Used for Verification

3.6.2.1 Effective Dose Calculation from Portal Imaging (2D)

The effective dose calculation was performed for electronic portal imaging device (EPID) that is attached to TrueBeam (version 2.0) linear accelerator. The organ doses were calculated by using the Eclipse treatment planning system. Firstly, the RANDO phantom was prepared to perform the CT simulation and the images were imported to the Eclipse treatment planning system. The organs were created by contouring the structures according to the human anatomy CT atlas. Then, as a double exposure technique, the field size of 10 x 10 cm^2 and 20 x 20 cm^2 were created and the monitor unit of 1 MU was set for each field. After that, the organ doses were calculated by using the Eclipse Treatment planning system. The setting of field size and monitor units, and dose calculation process were repeated for each clinical protocol (Head AP, Lateral, Chest AP, lateral, and Pelvis AP, Lateral) of portal imaging procedure. Finally, the effective doses were calculated by the equation [2.4]. The organ contouring and field size setting in TPS are shown in Figure 3.20.



(a) (b)

Figure 3. 20 (a) Creating the organ from CT images of RANDO phantom and (b) dose calculation by Eclipse treatment planning system

3.6.2.2 Half-value Layer (HVL) Measurement from On-board Imager (OBI) System

The HVL measurement from OBI system which is attached to the TrueBeam (version 2.0) linear accelerator was performed as the same steps in those performed in Acuity conventional simulator. The setting of HVL measurement was performed as shown in Figure 3 21.

Figure 3. 21 The setup for HVL measurement of OBI system

3.6.2.3 Effective Dose Calculation from OBI System (2D)

For on-board imager, the KV source and detector were set in vertical position. The 6 cc ionization chamber was set at the isocenter with support stand as presented in Figure 3.22 and connected to the Accu-Pro

dosimeter for incident air kerma measurement. Then, the chamber was exposed for each OBI clinical protocol with the automatic setting up exposure parameter in Table 3.3. The incident air kerma and backscatter were multiplied to obtain the entrance surface air kerma. The effective doses were calculated by multiplying the entrance surface air kerma with the conversion coefficient.

Figure 3. 22 The setup for incident air kerma measurement from OBI system

Table 3. 3 The clinical acquisition parameters from OBI

Parameters	Head		Chest		Pelvis	
	AP	LAT	AP	LAT	AP	LAT
kV	85	70	80	100	75	100
mAs	5	5	5	5	5	10
Field size (cm^2)	26x20	26x20	26x20	26x20	26x20	26x20

3.6.2.4 The Organ and Effective Dose Calculation from Cone-beam CT (3D)

The Cone-beam CT Dose Index (CBDI) measurements were performed for effective dose calculation of cone-beam CT system attached to the TrueBeam (version 2.0) linear accelerator. Figure 3.23 (a) shows the measurement setup that the 100 mm pencil ionization chamber in air and (b) with PMMA CTDI phantoms, 16 cm in diameter for head protocol and 32 cm in diameter for chest and pelvis protocol.

1) Firstly, CBDI in air measurement was performed at the isocenter by placing the ionization chamber in air with the support stem at 20 cm distance above the treatment couch. Then, it was connected with the dosimeter for dose reading

and exposed for five times with selected clinical protocol, shown in Table 3.4. The mean dose was taken from five times reading and the normalized $CBDI^{air}$ was calculated.

2) Secondly, CBDI center and peripheries measurements were performed by placing the chamber at the center and twelve, three, six and nine o'clock positions of the corresponding phantom and the exposure parameters were selected for each protocol. Then, chamber was connected with the dosimeter for dose reading and exposed for five times with selected protocol. The mean dose was taken from five times reading and the normalized $CBDI^{center}$ and normalized $CBDI^{periphery}$ were calculated.

3) Thirdly, the $CBDI^{weighted}$ was calculated by using the equation [2.9].

4) Finally, CBDI air, CBDI weighted, total mAs and scan range of each protocol were entered into the ImPACT software to calculate the organ and effective doses.

As ImPACT software is designed for the specific CT scanner, for cone-beam CT, ImPACT factor was used to match with ImPACT software to select the suitable reference CT unit. The ImPACT factor was calculated from the equation [2.11].

(a) (b)

Figure 3. 23 The 100cc pencil chamber (a) at the isocenter in air and (b) in the PMMA body phantom for CBDI measurement

Table 3. 4 Acquisition parameters for cone-beam CT TrueBeam version 2.0

Parameters	Standard Head	Low Dose Thorax	Pelvis
kV	100	125	125
mA	20	20	80
mAs	148	266	693
Fan Type	Full	Half	Half
Trajectory	Half	Full	Full

3.6.2.5 The Organ and Effective Dose Measurement Using Thermoluminescence Dosimeter (TLD)

To verify the accuracy of ImPACT dose calculation program, the organ and effective dose measurements were performed in Alderson RANDO phantom by using the TLD for cone-beam CT pelvic protocol as shown in Figure 3.24.

This measurement involved 200 chips of calibrated TLD 100-H, and the three chips of TLD were loaded to the plastic tubes. These plastic tubes were inserted in each section of the corresponding organs in RANDO phantom with guidance from a human anatomy CT atlas. The organ selection was based on the ICRP 103 recommendation.

The inserted TLD in RANDO phantom was scanned using the cone-beam CT pelvic protocol and the scanning setup is shown in Figure 3.27. After scanning, the signals from TLD were read and converted to the organ dose, D, by the equation [2.17]. Then, the effective dose for CBCT pelvic protocol was calculated according to the equation [2.4].

Figure 3. 24 The setup of RANDO phantom to scan the CBCT pelvic protocol

3.7 Outcome to be Measured

The outcomes for the determination of effective dose calculation were;

- CTDI / CBDI in air and weighted values (mGy) from the 3D and 4D imaging.
- Incident Air Kerma and Entrance Surface Air Kerma (mGy) from 2D imaging.
- Organ equivalent doses (mSv) from all of the imaging modalities used in Image Guided Radiation Therapy System.

3.8 Sample Size

For effective dose calculation, the sample size was determined by using following equation,

$$n = \left[\frac{Z_{\alpha/2} * \sigma^2}{d^2}\right]$$

Where, Z is the reliability coefficient of normal distribution.

For 95 % confident level, $\alpha = 0.05$, $Z_{\alpha/2} = 1.96$, variance of difference, $\sigma = 0.11$ (from previous study) and acceptable error, $d = 0.1$.

Therefore, $n = 4.648$. So, we performed the measurement 5 times in each imaging modality.

3.9 Data Analysis

The doses were collected as in mean and standard deviation. The measured and calculated organ and effective doses were presented in the form of table.

3.10 Expected Benefit and Application

This study, determined the effective doses from all of the imaging modalities involved in IGRT process and also shown which machine gives lower effective dose to the patient. Moreover, we can determine that how much extra-radiation dose patient will get after treatment.

3.11 Ethical Consideration

Although this study was performed in phantom, the research proposal was submitted and approved by Ethics Committee of Faculty of Medicine, Chulalongkorn University shown in Figure 3.23.

INSTITUTIONAL REVIEW BOARD
Faculty of Medicine, Chulalongkorn University
1873 Rama 4 Road, Patumwan, Bangkok 10330, Thailand, Tel 662-256-4493

Certificate of Exemption

The Institutional Review Board of the Faculty of Medicine, Chulalongkorn University, Bangkok, Thailand, has exempted the following study in compliance with the International guidelines for human research protection as Declaration of Helsinki, The Belmont Report, CIOMS Guideline, International Conference on Harmonization in Good Clinical Practice (ICH-GCP) and 45CFR 46.101(b)

Study Title : Determination of Effective Doses in Image Guided Radiation Therapy System: Phantom Study

Study Code : -

Principal Investigator : Miss Yin Yin Pyone

Study Center : Department of Radiology, Faculty of Medicine, Chulalongkorn University.

Document Reviewed :

1. Protocol Version 1.0 Date July 27, 2015
2. Protocol synopsis Version 1.0 Date July 27, 2015
3. Principle investigator's CV Version 1.0 Date July 27, 2015

Signature: ______ (Emeritus Professor Tada Sueblinvong MD) Chairperson The Institutional Review Board

Signature: ______ (Assistant Professor Prapapan Rajatapiti MD, PhD) Member and Secretary The Institutional Review Board

Date of Exemption : August 24, 2015

Note 1. No Continuing review required
2. Submit notification of final report when finish

Figure 3. 25 Certificate of Approval from Ethic Committee of Faculty of Medicine, Chulalongkorn University

CHAPTER 4

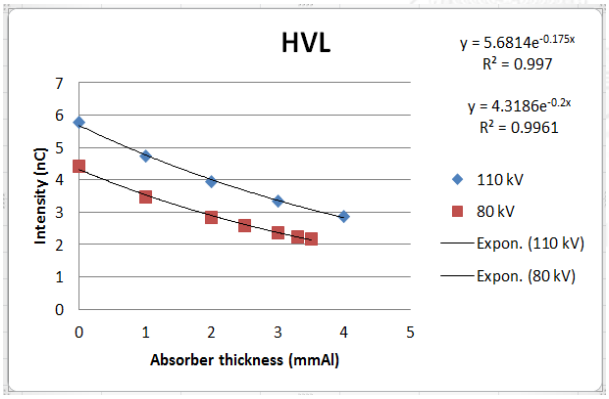
RESULTS

4.1 Effective Dose Calculation from Imaging Modalities Used for Planning

The effective dose calculations from 2D imaging system were performed by measuring the incident air kerma as the AAPM Task Group 75 recommendation [5]. The conversion coefficients were determined according to the exposure parameters and Half Value Layer of the machine. For 3D imaging, the measured CTDI values were used to calculate the effective doses by employing the ImPACT program.

4.1.1 The HVL Measurement from Acuity Simulator

The HVL in terms of millimeter aluminum (mmAl) was determined for the radiographic mode of Acuity Simulator. The measured HVL value and calculation graph are shown in Figure 4.1.



(a)

kVp setting	HVL (mm AL)
80	3.41
110	3.96

(b)

Figure 4. 1 (a) the HVL calculation graph and [b] the HVL value for Acuity Simulator

4.1.2 The Effective Doses Calculation from Acuity Simulator

The acquisition parameters and calculated effective doses from acuity simulator are shown in Table 4.1. The entrance skin doses in lateral projections were higher than the antero-posterior projections because of high exposure setting for the thicker anatomic structures. The effective doses for AP and

lateral projections were 0.008 and 0.013 mSv for head, 0.137 and 0.128 mSv for chest and 0.315 and 0.255 mSv for pelvic regions, respectively. The total effective dose in head, chest and pelvis AP and Lat: were 0.021, 0.27 and 0.57 mSv, respectively.

Table 4. 1 Effective doses and acquisition parameters for each protocol of Acuity Simulator

Parameters	Head		Chest		Pelvis	
	AP	Lateral	AP	Lateral	AP	Lateral
kV	85	85	95	100	110	120
mAs	9.9	20	10	20	18	20
Field size (cm^2)	15x15	15x15	20x20	20x20	20x20	20x20
Entrance skin dose (mGy)	0.87	1.49	1.14	2.55	2.42	3.64
Conversion coefficient	0.009	0.009	0.12	0.05	0.13	0.07
Effective dose (mSv)	0.008	0.013	0.137	0.128	0.315	0.255
Total effective dose (mSv)	**0.021**		**0.27**		**0.57**	

4.1.3 The Effective Doses Calculation from Philips Brilliance Big Bore CT

In planning 3D CT, the CTDI were measured with 100cc ionization chamber in both head and body phantom for head, chest and pelvic clinical protocols. The normalized $CTDI_{100}^{air}$, $CTDI_{100}^{periphery}$, $CTDI_{100}^{center}$, and $CTDI_{100}^{w}$, are shown in Table 4.2 .

Table 4. 2 The normalized measured CTDI values in air, at the center and the periphery position of head and body phantoms

Protocol	The normalized CTDI (mGy/mAs)			
	$CTDI_{100}^{air}$	$CTDI_{100}^{center}$	$CTDI_{100}^{periphery}$	$CTDI_{100}^{W}$
Head	15.23	10.28	13.48	12.41
Body	15.24	3.83	10.16	8.05

The organ equivalent dose and effective doses from the planning fan beam CT and are shown in table 4.3 for head, chest and pelvis region. The dose calculations were performed based on the departmental routine of the planning CT. For effective dose calculation, the organ and tissue weighting factors were

applied as recommended by ICRP 103 [6]. From the ImPACT dose calculation, the thyroid, esophagus and bladder received the highest organ equivalent doses of 39, 32 and 27 mSv, for head, chest and pelvis, respectively. The highest total effective dose of 16 mSv was observed in chest and followed by pelvis with 6.8 mSv and 3.5 mSv from head protocols.

Table 4. 3 The organ equivalent and effective doses from Philips Brilliant Big Bore CT

Organs	**W_T**	**Organ Equivalent Dose, (H_T), (mSv)**		
		Head	**Chest**	**Pelvis**
Gonads	0.08	0.000043	0.61	25
Bone Marrow	0.12	3.3	11	7
Colon	0.12	0.0012	5.1	17
Lung	0.12	1	29	0.058
Stomach	0.12	0.032	26	1.2
Bladder	0.04	0	0.28	27
Breast	0.12	0.16	22	0.044
Liver	0.04	0.052	25	0.75
Esophagus (Thymus)	0.04	0.84	32	0.0069
Thyroid	0.04	39	4.9	0.00023
Skin	0.01	4.2	7.7	6.2
Bone Surface	0.01	13	19	10
Brain	0.01	27	0.23	0.000054
Salivary Glands (Brain)	0.01	27	0.23	0.000054
Remainder	0.12	5.6	17	4.9
Total Effective Dose (mSv)		**3.5**	**16**	**6.8**

4.2 Effective Dose Calculation from Imaging Modalities for Verification

In the treatment verification imaging modalities, the 2D imaging system effective dose calculations were performed by measuring the incident air kerma as the AAPM Task Group 75 recommendation for OBI system and the Eclipse treatment planning system was used to calculate the doses from portal imaging. For OBI, the conversion coefficients were determined according to the exposure parameters and Half Value Layer of the machine. For 3D imaging, the measured CBDI values were used to calculate the effective doses by employing the ImPACT program.

4.2.1 The Organ and Effective Doses Calculation from Portal Imaging

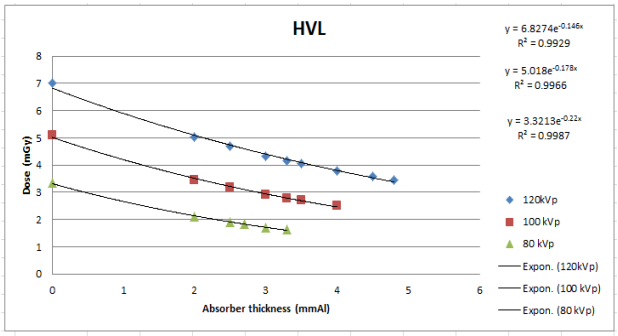
The portal images were performed with 1 MU per image and the total of 2 MU was obtained by double exposure technique for each AP and Lat; protocol of head, chest and pelvis regions. The organ and effective doses calculated by the Eclipse TPS are shown in Table 4.4. The effective doses from chest AP and lateral showed the highest value of 3.55 and 3.23 mSv while the other two received 2.73 and 2.37 mSv from pelvis and 2.03 and 1.85 mSv from head. The total effective doses received from a total 4 MU (double exposures technique for both AP and lateral projections) of head, chest and pelvis were 3.88, 6.78 and 5.1 mSv, respectively.

Table 4. 4 The organ and effective doses calculation from EPID from TrueBeam version 2.0 linear accelerator

Tissue/Organ	Tisssue weighting factor	Organ Equivalent Dose (mSv)					
		Head		Chest		Pelvis	
		AP	LAT	AP	LAT	AP	LAT
Red bone marrow	0.12	4	4	1	1	2	4
Colon	0.12					4	3
Lung	0.12			8	10.5		
Stomach	0.12			2	2		
Breast	0.12			7.5	3.5		
Gonads	0.08					16.5	12.5
Bladder	0.04					17	13
Liver	0.04						
Oesophagus	0.04	1	1	9	10		
Thyroid	0.04	11	10				
Skin	0.01	1	1	1	1	1	1
Bone surface	0.01						
Brain	0.01	6	7				
Salivary gland	0.01	16	19				
Remainder organs	0.12	7	5.5	8	6.5		
Total effective dose (mSv) for 2 MU		**2.03**	**1.85**	**3.55**	**3.23**	**2.73**	**2.37**
Total effective dose (mSv) for AP & LAT		**3.88**		**6.78**		**5.1**	

4.2.2 The HVL Measurement from OBI System

The HVL in terms of millimeter aluminum (mmAl) was determined for the radiographic mode of Varian OBI system. The measured HVL value and calculation graph is shown in Figure 4.2.



(a)

kVp setting	HVL (mm AL)
80	3.15
100	3.89
120	4.74

(b)

Figure 4. 2 (a) the HVL calculation graph and (b) the measured HVL value for OBI

4.2.3 The Effective Doses Calculation from On-board Imager System

The clinical acquisition parameters and calculated effective doses from OBI system are shown in Table 4.5. The lower entrance skin dose was found in head lateral than that of the AP protocol because the lower exposure is needed for the thinner anatomy of the head lateral position than the AP projection. The total effective doses for head, chest and pelvis AP and Lateral; were 0.02, 0.12 and 0.17 mSv, respectively.

Table 4. 5 Effective doses and acquisition parameters for each protocol of OBI attached to TrueBeam version 2.0

Parameters	Head		Chest		Pelvis	
	AP	Lateral	AP	Lateral	AP	Lateral
kV	85	70	80	100	75	100
mAs	5	5	5	5	5	10
Field size (cm^2)	26x20	26x20	26x20	26x20	26x20	26x20
Entrance skin dose (mGy)	0.61	0.4	0.54	0.84	0.47	1.69
Conversion coefficient	0.02	0.02	0.13	0.06	0.1	0.07
Effective dose (mSv)	0.012	0.008	0.07	0.05	0.047	0.118
Total effective dose (mSv)	**0.02**		**0.12**		**0.17**	

4.2.4 Effective Dose Calculation from 3D Imaging by Using ImPACT Software

For cone-beam CT system, the CBDI were measured with 100cc ionization chamber in both head and body phantom for head, chest and pelvic clinical protocols of kV CBCT attached to the TrueBeam (version 2.0) linear accelerator. The normalized $CBDI_{100}^{air}$, $CBDI_{100}^{center}$, and $CBDI_{100}^{w}$, are shown in Table 4.6.

Table 4. 6 The normalized measured CBDI values in air, at the center and the periphery position of head and body phantoms

Protocol	The normalized CBDI (mGy/mAs)			
	$CBDI_{100}^{air}$	$CBDI_{100}^{center}$	$CBDI_{100}^{periphery}$	$CBDI_{100}^{W}$
Head	3.39	2.7	2.19	2.35
Body	6.11	1.38	2.04	1.82

The organ equivalent dose and effective doses from the cone-beam CT are shown in table 4.7 for the head, chest and pelvis regions. The dose calculations were performed based on manufacturer protocol of the Varian cone-beam CT system. For effective dose calculation, the organ and tissue weighting factors were applied as recommended by ICRP 103 [6]. From the ImPACT dose calculation showed that the highest organ equivalent doses were observed in salivary glands, esophagus and bladder with 2.5, 10 and 34 mSv for head, chest and pelvis protocols, respectively. The effective dose from pelvis protocol illustrated the highest with 4.99 mSv and, 2.40 and 0.14 mSv from chest and head protocols.

Table 4. 7 Organ equivalent dose and effective doses from Cone-beam CT attached to the Varian TrueBeam Linear Accelerator.

Organs	W_T	Organ Equivalent Dose, (H_T), (mSv)		
		Head	**Chest**	**Pelvis**
Gonads	0.08	<0.01	<0.01	19
Bone Marrow	0.12	0.22	1.6	4
Colon	0.12	0.000032	0.017	9.3
Lung	0.12	0.015	5.7	0.01
Stomach	0.12	0.00058	0.29	0.2
Bladder	0.04	0	0.0023	34
Breast	0.12	0.0037	6.6	0.02
Liver	0.04	0.00088	0.44	0.13
Esophagus (Thymus)	0.04	0.013	10	0
Thyroid	0.04	0.42	0.76	0
Skin	0.01	0.23	1	4.4
Bone Surface	0.01	0.95	3.1	6.1
Brain	0.01	2.5	0.039	0
Salivary Glands (Brain)	0.01	2.5	0.039	0
Remainder	0.12	0.25	1.4	3.1
Total Effective Dose (mSv)		**0.14**	**2.40**	**4.99**

4.2.5 Effective Dose Measurement from CBCT Pelvic Protocol with TLD in RANDO Phantom

The organ equivalent doses and effective dose were measured for CBCT pelvic protocol with TLD inserted in RANDO phantom. The reading data is shown in appendix. The measurement results and the comparison with the calculated results from ImPACT are shown in Table 4.8. From the TLD measurement, the bladder received the highest organ equivalent dose of 23.47 mSv and the total effective dose was 5.17 mSv.

4.2.6 Comparison between Measured and Calculated Organ Equivalent Doses and Effective Dose

The comparison between measured and calculated doses was done for cone-beam CT pelvic protocol to verify the accuracy of ImPACT dose calculation software and the results are shown in Table 4.8. The large variation of the measured organ doses from the ImPACT dose calculation was observed. However, the maximum organ equivalent dose was found in bladder of 23.47 mSv from TLD measurement while it was shown 34 mSv in ImPACT calculation. The measured effective dose by TLD was 5.17 mSv, while the calculated dose by ImPACT was 4.99 mSv. In the case of gonadal organs, the measurement showed 2.81 mSv and it was shown 19 mSv in calculation. The

percentage of variation of effective dose difference between ImPACT calculation and TLD measurement was 3.5 percent.

Table 4. 8 Comparison of organ equivalent and effective doses between TLD measurement and ImPACT calculation for CBCT pelvic protocol

Organ	Tissue weighting factor W_T	Organ Equivalent Dose ,H_T, (mSv)	
		TLD	ImPACT
Gonads	0.08	2.81	19.00
Bone Marrow	0.12	3.05	4.00
Colon	0.12	20.92	9.30
Lung	0.12	0.36	0.01
Stomach	0.12	0.99	0.20
Bladder	0.04	23.47	34.00
Breast	0.12	0.31	0.02
Liver	0.04	1.83	0.13
Oesophagus (Thymus)	0.04	0.12	0.00
Thyroid	0.04	0.13	0.00
Skin	0.01	2.37	4.40
Bone Surface	0.01	6.06	6.10
Brain	0.01	0.71	0.00
Salivary Glands (Brain)	0.01	0.11	0.00
Remainder	0.12	6.27	3.10
Total Effective dose (mSv)		5.17	4.99

CHAPTER 5

DISCUSSION AND CONCLUSION

5.1 Discussion

The image-guided radiation therapy becomes very popular after newly advanced treatment techniques are released. Therefore, the total concomitant dose is increasing steadily with the introduction of more imaging procedures to the treatment process. The summary organ and effective doses from the common imaging modalities involved in image-guided radiation therapy process are shown in, Table 4.9.

Table 5. 1 Effective Doses from each imaging modalities

Type of imaging	Head	Chest	Pelvis
2D setup (Acuity) (kV)	0.021 mSv	0.270 mSv	0.570 mSv
3D Planning CT	3.500 mSv	16.00 mSv	6.800 mSv
2D On-board imaging (kV)	0.020 mSv	0.120 mSv	0.170 mSv
2D Portal Imaging (MV)	3.880 mSv	6.780 mSv	5.100 mSv
3D Cone-beam CT (kV)	0.140 mSv	2.40 mSv	4.990 mSv

5.1.1 Imaging Doses from Planning Imaging Modalities

5.1.1.1 Two-dimensional Imaging Doses

The effective doses from the Acuity 2D setup imaging were very low compared with the volumetric 3D imaging of planning CT. The maximum total dose was observed at the pelvis protocol of 0.57 mSv and followed by chest and head protocols because of the largest setting up parameters in pelvic region. However, it is about twelve times lower than the 3D CT pelvis effective dose.

5.1.1.2 Three-dimensional Imaging Doses

In 3D imaging, the total effective doses from the planning CT are higher than those from the verification cone-beam CT and the highest total effective dose was observed in chest protocol of 16 mSv. The highly radiation

sensitive organs, the lung and breast tissue acquired 29 mSv and 22 mSv, gonads and colon acquired 25 mSv and 17 mSv from chest and pelvis protocols, respectively. The thyroid, esophagus, and bladder obtained the highest organ equivalent dose of 39 mSv, 32 mSv and 27 mSv for the head, chest and pelvis protocols, respectively. The effective dose from planning CT chest protocol was higher than that of the pelvis protocol because the scan length for chest protocol was longer than the pelvis protocol to cover the metastasis area, as shown in table [3.4]. Even though, the dose from chest protocol is high, the total effective doses from this study are within the acceptable range and have good agreement with the Mettler F. A. et al. study [18] who reported the effective doses from different CT scanner and set the range of doses (0.9-4 mSv, 4-18 mSv and 3.5-25 mSv for head, chest and pelvic protocols respectively).

5.1.2 Imaging Doses from Verification Imaging Modalities

5.1.2.1 Two-dimensional Imaging Doses

Among the 2D imaging modalities, the total effective doses (AP and LAT) from OBI system were the lowest in comparison with the other two 2D imaging modalities for all regions. Because of using the megavoltage treatment beam and double exposure technique, the effective doses from the portal imaging were very high in compared with the other modalities. The maximum total dose was observed at the chest protocol of 6.78 mSv and followed by pelvis and head protocols, because, this region includes many high radiation sensitive organs (such as lung, stomach and breast) than the other two regions.

5.1.2.2 Three-dimensional Imaging Doses

(A) Dose Calculation by ImPACT Software

In cone-beam CT imaging, salivary gland, esophagus, and bladder received the highest organ equivalent dose of 2.5, 10 and 34 mSv for the standard head, low dose thorax, and pelvis protocols, respectively. However, in the high radiation sensitive organ, the lungs acquired 5.7 mSv in low dose thorax, gonads and colon acquired 19 and 9.3 mSv from pelvis protocols. The effective doses from TrueBeam CBCT are lower than those from the Clinac iX (version 1.4.13) machine (0.3mSv, 3.6mSv and 6 mSv for head chest and pelvic regions) which has been reported by the Tanawat T. et al. study [19]. This is because of difference in acquisition parameters and advanced manufacturer configuration. The effective doses from CBCT were lower than that of the planning CT; but, to achieve the precise and accuracy treatment, CBCT has to perform every week to every day which increasing the amount of radiation dose to the patient.

(B) Dose Measurement By TLD and Verification of ImPACT Software in Dose Calculation

According to the TLD dose measurement for CBCT pelvic protocol, the highest organ dose was found in bladder of 23.47 mSv and radiation sensitive organs, colons and gonads, received the organ equivalent dose of 20.92 and 2.81mSv respectively. The results for organ equivalent and effective doses were lower than in compared to the former study which measured the organ and effective doses in Clinac iX version 1.4.13. Tanawat T. et al. study [19] reported the bladder dose of 35.39 mSv and 25.81mSv and 10.09mSv for colon and gonads and effective dose of 7.31mSv.

The verification of ImPACT software for organ and effective dose calculation of CBCT was performed by TLD measurement. The difference between the calculation and measurement is from the reason that the ImPACT software is designed for the CT scanner and fan beam geometry and the CTDI weighted is calculate from the 360 degree scanning geometry. In the case of CBCT, it may have some difference in 200 degree scanning. Moreover, the different application of phantoms for calculation and measurement also can involve as one parameter of variation because ImPACT calculation is based on mathematical phantom while TLD measurement were performed in Rando phantom (tissue equivalent). From this comparison between TLD measurement and ImPACT dose calculation, we observed that the organ equivalent doses were in both higher and lower trend, but mostly the measurement showed the higher dose and the effective dose from measurement was higher than the calculation. The percentage of variation of calculation from the measurement was 3.5%. It implied that the calculation from the ImPACT software can be applied to calculate the CTDI and CBDI with reasonable results.

5.2 Conclusions

Although the doses from the planning setup and verification images are smaller than the therapeutic dose, they play in very important part in the treatment process. Moreover, in advanced treatment technique such as IMRT and VMAT, the number of MU required for treatment delivery is increased. Therefore, the primary beam leakage and scattered doses are also increased and then leads to increase the concomitant (extra-target) dose. To achieve the precise and accuracy treatment, the application of imaging modalities in radiotherapy process can be varied from every week to every day depending on the size and location of the target, choice of treatment technique.

If we assume for a 30 fractions of the treatment course, at least, one scan of 3D planning CT, one exposed of 2D setup verification, and 6 times CBCT have been performed. Therefore, the total effective doses of the head,

thorax, and pelvis regions are 4.36, 30.67 and 37.31 mSv, respectively. For small target which is much closed to critical organs, the CBCT may need to be performed every day before every treatment fraction and the total effective doses (3D CT, 2D setup verification and 30 times CBCT) are 7.72 mSv for head, 88.27 mSv for chest and 157.07 mSv for pelvic regions. So, daily image guidance should be used with certain caution where the treatment dose reaches limits for organ at risk or children.

To improve the therapeutic dose conformity, there are increasingly use of imaging in RT process. In IGRT the beam alignment information derived from the images for targeting of tumor is less dependent on the image quality and more dependent on the imaging frequency. AAPM Task Group 75 stated that all medical use of ionizing radiation can give a significant radiation dose to the patient. [ALARA: As Low As Reasonably Achievable]. So, there should have the way to reduce the effective dose without reducing the imaging frequency. Therefore, IGRT should be administered with significant parameters to maintain a balance between the imaging dose and therapeutic goal.

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APPENDIX

Table I. The averaged doses for each organ from TLD measurements

Organs			TLD Rod no.	Charge (nC)	Average dose (mGy)
Gonads			39	13.79717	2.8095
Bone Marrow					3.0479
	Head		2	0.40420	
	Chest		3	2.34237	
	Pelvis		4	42.15717	
Colon			5	102.75683	20.9242
Lung					0.3615
	Right		6	1.85153	
	Left		7	1.69877	
Stomach			8	4.86597	0.9908
Bladder			9	115.23617	23.4654
Breast			R40 L41	3.01780	0.3073
Liver			10	8.99383	1.8314
Oesophagus (Thymus)			11	0.57183	0.1164
Thyroid			12	0.63553	0.1294
Skin					2.3749
	AP		42	13.40583	
	PA		43	14.31417	
	Right		44	9.49707	
	Left		*7	9.43417	
Bone Surface					6.0597
	head		13	0.40409	
	chest		14	1.48037	
	pelvis		15	87.39083	
Brain			16	3.49646	0.7120
Salivary Glands (Brain)					0.1148
	Rt		17	0.54778	
	Lt		18	0.57960	
Remainder Organs					
	Adrenals				2.6569
		Right	19	12.25383	
		Left	20	13.84217	
	Small Intestine				12.5099
		Upper	21	24.19083	

		Lower	22	98.67883	
	Kidney				4.0122
		Right	23	20.15517	
		Left	24	19.25150	
	Pancreas		25	14.02350	2.8556
	Spleen		26	7.06093	1.4378
	Thymus		27	1.08590	0.2211
	Muscle		28	76.85117	15.6491
	Gall Bladder		29	14.69717	2.9928
	Heart		30	1.96853	0.4008
	ET region (Thyroid)		31	0.67240	0.1369
	Lymph nodes (Muscle)				12.3793
		Right	32	66.19750	
		Left	33	55.38983	
	Oral mucosa (Brain)				0.0874
		Right	34	0.45860	
		Left	35	0.39967	
Other organs of interest					
	Eye lenses		R*8 L*9	0.73544	0.0749
	Ovaries				20.2972
		Rt	36	112.61617	
		Lt	37	86.73883	
	Uterus		38	94.16583	19.1749
	Prostate		1		20.8803

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