ADAPTATION OF COMPUTED RADIOGRAPHIC SYSTEM FOR TREATMENT SETUP VERIFICATION IN EXTERNAL BEAM RADIOTHERAPY.

BY

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THIS THESIS IS SUBMITTED TO THE GRADUATE SCHOOL OF NUCLEAR AND ALLIED SCIENCES, UNIVERSITY OF GHANA, LEGON

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DECLARATION

This thesis is a result of a work undertaken by Francis Sandy in the Medical Physics department, School of Nuclear and Allied Science, University of Ghana under the supervision of Mr. Samuel Nii Adu Tagoe, Dr. Stephen Inkoom and Professor John H. Amuasi.

I, Francis Sandy, declare that except for references cited in this work, the contents of this research was done by me and that this thesis work has not been submitted by any one for academic examination towards any qualification in the University of Ghana or elsewhere.

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<table>
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<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>ALARA</td>
<td>As low as reasonable achievable</td>
</tr>
<tr>
<td>CR</td>
<td>Computed Radiography</td>
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<tr>
<td>CNR</td>
<td>Contrast-to-noise ratio</td>
</tr>
<tr>
<td>CT</td>
<td>Computed Tomography</td>
</tr>
<tr>
<td>CBCT</td>
<td>Cone-beam computed tomography</td>
</tr>
<tr>
<td>CRI</td>
<td>Computed Radiographic Imaging</td>
</tr>
<tr>
<td>DRRs</td>
<td>Digitally Reconstructed Radiographs</td>
</tr>
<tr>
<td>EBRT</td>
<td>External Beam Radiotherapy</td>
</tr>
<tr>
<td>EPID</td>
<td>Electronic Portal Imaging Device</td>
</tr>
<tr>
<td>ESF</td>
<td>Edge Spread Function</td>
</tr>
<tr>
<td>HV</td>
<td>High Voltage</td>
</tr>
<tr>
<td>ICRU</td>
<td>International Commission on Radiation Units and Measurement</td>
</tr>
<tr>
<td>IGRT</td>
<td>Image Guidance Radiation Therapy</td>
</tr>
<tr>
<td>IP</td>
<td>Imaging Plate</td>
</tr>
<tr>
<td>IEC</td>
<td>International Electrotechnical Commission</td>
</tr>
<tr>
<td>IAEA</td>
<td>International Atomic Energy Agencies</td>
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<tr>
<td>KVCBCT</td>
<td>Kilovoltage Cone-Beam Computed Tomography</td>
</tr>
<tr>
<td>Abbreviation</td>
<td>Description</td>
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<tr>
<td>--------------</td>
<td>--------------------------------------------------</td>
</tr>
<tr>
<td>KBTH</td>
<td>Korle-Bu Teaching Hospital</td>
</tr>
<tr>
<td>LED</td>
<td>Light-Emitting Diode</td>
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<tr>
<td>MTF</td>
<td>Modulation Transfer Function</td>
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<tr>
<td>MVCBCT</td>
<td>Megavoltage Cone-Beam Computed Tomography</td>
</tr>
<tr>
<td>MeV</td>
<td>Mega-electron Volts</td>
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<tr>
<td>MR</td>
<td>Magnetic Resonance</td>
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<tr>
<td>MLC</td>
<td>Multileaf Collimator</td>
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<tr>
<td>MW</td>
<td>Motorized Wedge</td>
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<tr>
<td>NTCP</td>
<td>Normal Tissues Complication Probability</td>
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<tr>
<td>PACS</td>
<td>Picture Archival Communication System</td>
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<tr>
<td>PSP</td>
<td>Photostimulable Phosphor Plate</td>
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<td>PMT</td>
<td>Photomultiplier Tube</td>
</tr>
<tr>
<td>PSL</td>
<td>Photostimulated Luminescence</td>
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<tr>
<td>PV</td>
<td>Pixel Values</td>
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<tr>
<td>PTV</td>
<td>Planning Target Volume</td>
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<tr>
<td>PVC</td>
<td>Polyvinyl Chloride</td>
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<tr>
<td>PDDs</td>
<td>Percentage Depth Dose</td>
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<tr>
<td>PMMA</td>
<td>Poly-Methyl Methacrylate</td>
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<tr>
<td>Abbreviation</td>
<td>Description</td>
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<td>--------------</td>
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<tr>
<td>QC</td>
<td>Quality Control</td>
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<tr>
<td>ROI</td>
<td>Region of Interest</td>
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<tr>
<td>SNR</td>
<td>Signal-To-Noise Ratio</td>
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<tr>
<td>SF</td>
<td>Screen Film</td>
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<tr>
<td>TRS</td>
<td>Technical Series Report</td>
</tr>
<tr>
<td>TCP</td>
<td>Tumour Control Probability</td>
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<tr>
<td>VEPIDs</td>
<td>Video-Camera Electronic Portal Imaging Device</td>
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ABSTRACT

In radiation therapy, the traditional way of providing accuracy of megavoltage radiation treatment fields, is by the use of portal radiographs. In these radiographs the film detector is positioned at the exist point of the beam beside the patient and details of its anatomical parts located within the field borders are taken. There is poor visualization of landmarks in portal films due to the high energy of photons used for treatment. Due to this drawback, an electronic portal imaging device (EPID) is now being assembled with modern therapy machine. This device uses kilovoltage energies for treatment setup verification, which provides clear anatomical landmarks of patient for treatment. The importance for field edges verification is to:

- Verify the shape of the radiation beam.
- To establish a coordinate system common to both reference and portal image in which to account for deviation in patient setup.

The aim of this study was to assess the possibility of using computed radiography (CR) system instead of the conventional film in the use for treatment setup verification with the telecobalt machine at the radiotherapy department of the Korle-Bu Teaching Hospital. In diagnostic radiology, the use of CR has been long in existence to obtain digital radiographic images which help in providing picture archival communication system (PACS). The film is substituted with a photostimulable phosphor plate (PSP) in CR system. Energies deposited by the X-ray beam on the PSP are read by the scanner laser beam in the CR reader. After the PSP has been scanned by the reader, it can also be ready for reuse by exposing it to light for about forty seconds for the erasure of any remaining
image on the plate. A phantom was constructed and fabricated from a perspex which compensates for the human tissue. The phantom was irradiated using cobalt-60 source in which both the CR and the radiographic film were used as detectors to detect the images of the built phantom. These images were scanned using the scan maker 9800 XL. This thesis defines the use of the current image quality matrices such as Modulation Transfer Function (MTF), Contrast to noise ratio (CNR) and Signal to noise ratio (SNR) which show a practical easy-to-use software based program Image J for measurable assessment of digital (CR) and conventional detector system (film).

From the results obtained, the CR demonstrated superior values in CNR and SNR than that of the radiographic film, which shows better quality of images on CR. Notwithstanding these qualities in CR, the film also showed a better resolution than that of the CR.

Therefore digital treatment portal and simulation images, can easily be obtained with CR system which offers a base for a picture archival communication system in radiation oncology. Simulation and digital portal images will ease verification of treatment setup.
CHAPTER ONE

INTRODUCTION

This section of the study deals with background, statement of research problems and justification, objectives of the study, scope and delimitation and the organization of the thesis.

1.1 Background

The major aim of radiation therapy is to deliver accurate prescribed radiation dose to a defined target volume in order to damage the life intimidating tumor. The contradictory goal is the avoidance of damage to the patient’s normal, tissue within the irradiated region. (Taylor & Francis, 2007). Over exposing normal tissues to radiation dose beyond their thresholds may result in the loss of functionality of the tissues involved or affliction of secondary cancer for an individual. It is therefore imperative to choose irradiation geometries that will maximize radiation dose to the intended target volume, while concurrently reducing radiation doses to normal tissues in close proximity to the target volume. This is done in other to achieve favorable treatment outcome during external beam radiotherapy (Buzdar et al., 2013). The determining factor is how well both goals can be achieved. In external beam radiotherapy (EBRT), the total dose which is to be administered to a patient is divided into smaller segments (or fractions) and delivered over many days to weeks. This process is called fractionation. This is done to give the normal tissues time to repair sublethal damage, since normal tissues are faster in repairing sublethal damage than a tumour. Thus prolonging the radiation dose delivery and
choosing the optimal prescribed dose per fraction will improve tumour control probability (TCP) and reduce normal tissues complication probability (NTCP). The time-dose constraints are considered a lot by clinicians when prescribing radiation dose to a patient undergoing EBRT. Also, the radiation dose deposited at any point within the irradiated region of a patient must be accurately known. Dose distribution within the patient is calculated based on dosimetric functions and or data obtained in a full scatter water phantom and mathematical algorithms that can be used to explain the interaction of radiation with a medium. Owing to the complexities of the dose calculation procedures, determination of the dose distribution within a patient has now become the preserve of specialized computers known as treatment planning systems (TPS). The process of simulating the treatment delivery with a treatment planning system is to realize the intent of treatment prior to the real treatment delivered, (known as treatment planning). Treatment planning procedures have to be in agreement with the International Commission on Radiation Units and Measurement (Monti et al., 1995) (Stroom & Heijmen, 2002). Treatment planning involves the most appropriate way to irradiate the patient. Treatment planning is the first and one of the most important parts of the radiation delivery process. It outlines course of action for the weeks of individual treatment session. It begins immediately after the radiotherapy decision is made and chosen as the treatment modality. Unlike pharmaceuticals, radiation is not metabolized. Once dose of radiation has been delivered, it cannot be removed. Therefore, it is most important that the initial planning should be done accurately. Radiation kills cancer cells by depositing energy within the cell. Malignant cells that are not struck by radiation will not be affected by treatment, thus resulting in spreading and eventual death of the patient.
Similarly, when a large number of normal tissue cells are hit by high levels of radiation, they are most likely to die, which result in complication of the patient. Therefore patient treatment designed to treat tumor must be accurate to avoid unnecessary irradiation of normal tissue. Treatment planning procedures are as follows:

- Selecting the right location and immobilization technique so that treatment will be the same throughout the treatment phase.
- Classifying the position, shape of the tumor, and the neighboring organs surrounding the tumour.
- Choosing an appropriate beam arrangement
- Assessing the distribution of dose around the planned target.
- Calculations are done based on the treatment machine settings to deliver the required total dose. (Khan, 2007)

Figure 1.1 below gives details of steps in treatment planning. From the figure, treatment verification is one of the key component in treatment planning system. Treatment verification is a way of assessing localized treatment beam relative to patient’s anatomy. In EBRT, dosimetric verification and geometric localization are done to enhance accuracy and confidence in dose delivery. Dosimetric verifications are normally done by measuring the dose at certain points inside (natural cavity) or outside the patient. These measured doses are then compared with the calculated and predicted doses at the same points by the treatment planning system. Geometric localization is enhanced by acquiring a portal image of patient within the course of treatment. Treatment fields earmarked the treatment of patient by creating radiographic images with teletherapy machine. This is done to ascertain the relationship between the treatment radiation fields
and the anatomical landmarks within the irradiated region of the patient. The radiographic images acquired with teletherapy machine are compared with those obtained during treatment simulation. These procedures help in establishing uncertainties associated with patient setup and chance to rectify these uncertainties prior to treatment delivery. The process of verifying the localization of an intended target volume being irradiated is known as treatment verification. Treatment verification plays a key part in administration of patients getting radiation treatment. Verification of patients, as a component of quality assurance program is planned to identify mistakes and errors that happen during the course of treatment.
than predicted and are bound to take place in all stages of planning and treatment. These faults are shared in two classes: those related to prescription, computation, and specification of target doses and those that happen during the delivery stage. This takes account of treatment machine or patient alignment relative to the treatment machine. Portal imaging, in which radiographs are generated with transmitted megavoltage beams through patient are received on image receptors. The receptors are placed vertically to the direction of propagation beam and is used to verify errors (Shalev, 1995), such as; patient motion, proper alignment of treatment fields on the patient. Radiotherapy treatment beam is mostly used for portal imaging in therapy to ensure that prescribed irradiation treatment within anatomical regions and surrounding normal tissues are spared. The commonly used image receptor in portal imaging is a radiographic detector, positioned in lead covered cassette. Verification of field arrangement, using portal imaging can increase accuracy by identifying localization errors. It has also been reported that a significant reduction in localization error could be achieved with increasing frequency of portal imaging. Identifying localization errors with portal imaging is greatly influenced by our ability to see clearly certain anatomical landmarks (such as bones) and their spatial resolutions to a radiation fields on a portal image. Most portal images suffer from extremely poor contrast. (Roehrig, 1990) This poor contrast is due to the megavoltage energy range beams which are used to acquire the images. As the energy increases from the diagnostic range i.e. from 30 – 150 kV, to therapeutic range i.e. 1 – 10 MV the mode of interaction of photons with biological material changes from photoelectric absorption to Compton scattering, which leads to the decrease in differential absorption of the tissues. Additionally, the size of the irradiation source (geometrical status) and voluntary
or involuntary movement on the part of the patient are all causes of image degradation. The traditional way of proving the accuracy of megavoltage radiation treatment fields is by the use of portal radiography. The film detector is positioned at the exit point of the beam beside the patient and details of its anatomical parts located within the field borders are taken. Portal radiographs are divided into three types (i) a localization radiograph is a portal radiograph produced by a short exposure and then comparing it to treatment time on daily basis required for that field of treatment. This portal film is used to regulate patient positioning and field boundaries prior to the delivery of the daily treatment. (ii) a verification radiography is formed when the image receptors are exposed to the field during the whole treatment sessions. Insensitive detectors such as slow films are used in this portal radiographs. (iii) the last one is the double exposure radiography, in which small exposure is done with an open field followed by another exposure in the real treatment field. It has the benefit of viewing anatomical structures beyond the field borders, which is significant in classifying patient setup, but most often use is restricted by the radiation acceptance of exposed normal tissues (Shalev, 1995). Portal films are emulsion-type film placed in a light-tight enclosure. Intensifying screens are not used in radiotherapy films as compared to those used in diagnostic radiology. Instead, a thin single layer of copper or aluminum metal is placed on the top surface of the film (beam entry side) which offers an electronic buildup that helps to raise the efficiency of the film.

Due to the draw-back of poor visualization of landmarks in portal film systems, an electronic portal imaging device is now being assembled with modem radiotherapy machines (linear accelerators) for treatment. Electronic portal imaging device (EPID) is used to measure the transmitted X-ray intensity that goes out through a patient from a
radiation port during a treatment phase. The radiation signal is converted electronically into a two dimensional (2D) digital radiographic image to verify the true beam location in relation to patient’s anatomy. Advantages of EPID are that images can be acquired and viewed prior to treatment delivery. This research work was undertaken at the Korle-Bu Teaching Hospital in Ghana which has both linear accelerator and cobalt teletherapy units for radiotherapy treatment. Unfortunately during this research work the linear accelerator was installed but not commissioned and which had the EPI device as a package. Therefore, at Korle-Bu Teaching Hospital where this study was done, radiographic film system was used to verify radiotherapy treatment. Fuji Corporation was the first body to introduce computed radiography (CR) which gives the opportunity for the first practical digital X-ray system to be seen by the radiologist. The CR system was explored to obtain digital portal images for treatment verification at the hospital. The aim of the CR system, was to digitize X-ray images from storage phosphor plates. CR system consists of a cassette containing imaging detector, image reader/ digitizer, a workstation, software, monitors, and a printer. There are two types of CR readers :(i) single-plate readers. In this reader, cassettes are loaded manually and read separately. (ii) the other reader is the multiple-plate readers, which holds multiple plates-up to eight, and these plates are stacked and loaded automatically. Imaging detectors are placed in a radiographic cassette and images are obtained using the X-ray system. Therefore, the use of the CR system for portal image verification at the Korle-Bu-Teaching Hospital would be of importance since they are planning to go digital, and this work will emphasize more on the use of CR system to verify patient treatment. This intervention would help to obtain better image quality, and enhance accurate treatment of patients which is the goal of radiation therapy.
1.2 Statement of Research Problem and Justification

Computed radiographic system (CR) has been used in diagnostic X-ray facilities as a substitute for radiographic films to minimize cost and reduce exposure to hazardous chemicals used for film processing, which is also labour intensive. With the CR system, images are able to be stored in a digital format making it possible to address issues relating to long term storage stability and retrieval of films. The radiotherapy department of the Korle-Bu-Teaching Hospital is planning to digitize its patient records, and has proposed the use of CR system for treatment verification with the telecobalt machine. The proposal was also as a result of the deplorable condition of the film processor within the facility, and also issues relating to procurement of films. Since the beam energy of the therapy level is by far higher than that of the diagnostic level, there is need to assess the possibility of using the CR system and how to enhance the quality of image obtained with the CR system.

1.3 Objectives

The main aim of this study was to assess the possibility of using CR system instead of the conventional screen films in use for treatment verification with the telecobalt machine at the radiotherapy department of the Korle-Bu-Teaching Hospital.

1.4 Specific Objectives

The specific objective of the study was:
➢ To construct a phantom from which CR (digital detector) and conventional film detectors images would be assessed.

➢ To quantitatively determine the quality of images obtained from CR and film using the image quality parameters such as: signal to noise ratio (SNR), contrast to noise ratio (CNR) and Relative Modulation Transfer Function (RMTF).

1.5 Scope and Delimitation

This project work was carried out at the Korle-Bu-Teaching Hospital in Accra, Ghana. This work covered the use of a constructed phantom which replicated patient geometries. The CR cassettes would be used for treatment verification with the phantom after being irradiated with telecobalt machine. Test objects having densities similar to tissue of human body will be placed in the phantom to determine their contrast level. The CR reader would be used to read the image on the CR cassette. Quantitative analysis would be done on the images acquired on both CR and radiographic film system using Image J software.

1.6 Thesis Organization

This research write up covers five chapters. The first chapter focuses on the introduction of the study. Second chapter reviews existing literature relating to research problem of your study. Third chapter focuses on materials and methodology of your study. The fourth chapter looks at the results and discussions, while the fifth chapter provides the
conclusion and recommendations from the overall findings. The reference section contains all the relevant works cited throughout the project work.
CHAPTER TWO

LITERATURE REVIEW

2.0 Introduction

In this section relevant literature on the subject is reviewed. These include the physics of computed radiography, historical background of CR system, megavoltage imaging, radiographic films, electronic portal imaging device, and papers relating to the study.

2.1. The physics of computed radiography

The use of computed radiography (CR) is presently built on photostimulable phosphors, BaFBr: Eu2+ commonly identified as storage phosphor (Sonoda et al., 1983), (Rowlands, 2002). Phosphors are mostly in powder form and are obtained from barium fluorohalide family (Rowlands, 2002), which are deposited on the substrate to form an imaging plate. Imaging plate (IP) used in the CR system is positioned in a light-tight enclosure. This plates are exposed to an x-ray source and are read out using the raster scanner which has a laser that releases the photostimulated luminescence (PSL). The light guide collects the blue PSL light and the light is detected using photomultiplier tube (PMT). The signal collected from the PMT is digitized and images are formed (Fujita et al., 1989). The huge dynamic range, digital nature, easy convenience and unique image quality of CR system has led to an extensive approval in diagnostic radiology. The sequence of events engaged in producing a PSL signal begins as shown in figure 2.1
Upon exposure of the PSP to an X-ray source, energy is transferred to the electron which results in the excitation of electrons into a metastable condition. Mostly, half of these electrons go immediately to their ground state resulting in the quick release of light. Over time the 50% of the metastable electrons will go back to their ground state. This sometimes causes fading in the latent image which therefore requires the IP to be read immediately after exposure. The signal formed on the imaging plate (IP) are significantly degraded after eight hours of exposure of the CR cassette (Bushong, 2013).

The other stage in the process is stimulation, which requires an infrared light with diameter 50 to 100 µm to be focused on the PSP. The beam of infrared light penetrates, and spreads on the PSP. The amount of spread increases with the PSP thickness, thus stimulating a lot of metastable electrons to return to their ground state thus emitting a lot
of light for the formation of latent image. The third step is the detection (reading) of the stimulated emission. The laser beam makes the metastable electrons in the PSP to go back to their original state thus leading to the emission of a shorter wavelength light. Results leading to the loss of signal are as follows:

- Scattered emitted light
- The collection proficiency of the photo detector.

The signal acquisition at the stimulation stage of PSL does not totally change all the metastable electrons to their original state, some still remain in that higher state. Ghosting normally appears on subsequent use of imaging plate, if residual latent images are not removed. Residual latent images are removed by exposing the phosphor to high strong white light from the storage of a designed fluorescent lamps. At the stimulation cycle of PSP, result of no latent image would be formed when the dwell time of the laser beam is much longer at each position on the PSP. Background radiation usually fogged images due to the high sensitivity of the PSP. The laser light used to stimulate the PSP is monochromatic. HeNe gas laser was used as stimulating source, but has been replaced with a solid state laser. Solid-state laser produces longer wavelength light and therefore is less likely to interfere with emitted light. Even so, optical filters are necessary to allow only emitted light that reaches the photo detector while blocking the intense stimulated light.
2.2 Historical background

The original innovation of CR system was done by Kodak (Luckey & Rochester, 1975) who considered keeping an X-ray image in a phosphor screen. The conceptualization of application was done by Fuji (“Thermoluminescence of solids,” 1980) to manufacture the leading imagers in the medical settings. Fuji, being the leading designer of CR in the early years, used BaFBr: Eu phosphor and a cassette-based method. At the same time, Agfa and Kodak undertook a work based on x-ray imaging on phosphor screen and intended to make an improvement on the same method done by Fuji, but unfortunately they were inhibited from publicizing their work based on patent right issues and fear of destroying the foundation they had on screen-film. At that time, effect of storage on screen-film applications were observed when prints not needed were done over, i.e. a ghost image of previous exposure on the screen that appears on a succeeding film shown in the same cassette. The outcome of storage was based on the principle of thermal induced luminescence of exposed materials, i.e. thermoluminescence. There are histories on photoluminescence and thermoluminescence that can be traced back (Mckeever, 1985) and forwarded to medicine in current day applications e.g. radiation dosimetry, and archaeological dating.

2.3 Megavoltage imaging in radiotherapy

Megavoltage imaging was initially developed to make image guided radiotherapy (IGRT) possible, and made use of imaging practices on daily basis.
2.3.1 Cone Beam Computed Tomography (CBCT)

Conventionally, CT scanners have round ring detectors, moving opposite an x-ray tube. Moreover, CT scans also have detectors imbedded circular ring of the flat panel. The geometry used by this type of CT scan is called cone-beam computed tomography (CBCT). Multiple planer projection images are obtained using cone-beam CT, as the source moves about the patients through 180 degree and more. Patient anatomical structures in three dimensions, sagittal, cross-sectional and coronal planes are reconstructed based on the data provided by the multidirectional images of the CBCT. Volumetric images are reconstructed using an algorithm called filtered back-projection (Murphy et al., 2007). CBCT systems are commercially available as accessories to linear accelerators. The CBCT is connected to the gantry of the accelerator with which volumetric image data are acquired under normal treatment conditions, thus enabling localization of intended target volume and critical structures before each treatment. Application of the system is done using kilovoltage x-ray source or megavoltage beams.

2.3.2 Kilovoltage Cone Beam Computed Tomography (KVCBCT)

Kilovoltage cone-beam CT (KVCBCT) systems are made by connecting a conventional X-ray tube to a retractable arm at 90 degrees to the therapy beam position. Flat panel of X-ray detectors are attached to the X-ray tube, thus providing very good images which are suitable to be used in cone-beam CT and also in 2-D radiography and fluoroscopy. The advantages of a KVCBCT are its capacity to:
Have quality contrast and submillimeter spatial resolution due to the volumetric CT images produced by the system.

- Obtain images in therapy room conditions.
- Also uses two-dimensional methods of radiographic and fluoroscopic procedures to confirm the accuracy in portal imaging, patient movement, and ensuring position and dosimetric adjustment before and during treatments.

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**2.3.3 Megavoltage Cone Beam Computed Tomography (MVCBCT)**

In megavoltage cone-beam CT (MVCBCT) system, X-ray beam and EPID of the linear accelerator are mounted in opposite direction to the source. EPIDs with Si flat panel detectors are highly sensitive to permit speedy acquisition of multiple, low dose images as the gantry is revolving through 180 degrees or more. After these, multidirectional two-dimensional images, volumetric CT images are reconstructed (Gray et al., 2005).

The MVCBCT produces quality images for bony anatomical structures and, in some cases, soft tissue targets as well. MVCBCT is an important instrument for on-line verification of patient, and identification of implanted metal markers if used for patient’s setup.

Although KVCBCT has better image quality (resolution and contrast), MVCBCT has the following potential advantage over KVCBCT:

- Less exposure to artifacts due to high-Z objects such as metallic markers in the target, metallic hip implants, and dental fillings.
In MVCBCT, extrapolation of attenuation coefficients from kilovoltage to megavoltage energies is of no importance in term of dosimetric corrections.

2.3.4 Radio Frequency Beacons (RFB)

In RF beacons system, small RF transponders can be implanted in the patient instead of x-ray fiducials. Location of these beacons can be determined electrically by the system in real time (occurring ten times in every seconds). Real-time monitoring of location of target is allowed by the system during therapy, and it also gives the chance to stop the exposure if the target beacon moves beyond the predetermined acceptance.

2.3.5 Magnetic Resonance Device (MRD)

Few new equipment using MRI have been established for IGRT treatment. Scanning of patients are done in the treatment location, patients are then moved straight beneath the accelerator for treatment. Another device is also developed (view ray), that incorporate a rotating gantry system fixed with three cobalt -60 sources equally spread about the ring. Each of these rings has its own multi leaf collimator (MLC) system which is connected to a 0.3 T permanent magnet of MRI imaging system. Several MR- Linac units have presently been established, each with the objective of delivering sophisticated photon treatment using 6 MV photons while at the same time executing 1.5 T MR imaging.

2.3.6 Video Surface Imaging (VSI)

This system uses video camera methods for projection onto patients skin so that the 3D shape of the patient can be determined in real time. Images acquired can be compared to
the expected model of the patient surface as a way of identifying positional or motional differences between the current patient surface and the model.

2.3.7 Electronic Portal Imaging Device (EPID)

If the portal images are in digital form then they can also be processed to improve image quality. The devices that produce digital portal images are called Electronic Portal Imaging Devices (EPIDs). Due to the delays caused in film processing, took advantage of real time portal imaging for treatment. In the past years, great effort has been devoted to developing new equipment, i.e., electronic portal imaging devices. These include scanning linear arrays of diodes (Taborsky et al., 1981) and scintillating crystals (Morton et al., 1991), video-camera based systems (Munro et al., 1990) (Baily et al., 1980), matrix ion-chamber systems (Meertens et al., 1985), an Flat- panel arrays (Zhao & Rowlands, 1992) (Antonuk et al., 1990)

2.3.8 Video-Camera Based EPIDs

Video-camera electronic portal imaging devices (VEPIDs) consist of a metal plate with the underside coated with a fluorescent phosphor screen to produce visible photons. A 45° mirror deflects some of the light onto a video camera. The major advantages of VEPIDs are that:

- Radiations exiting the patient have the ability to produce a signal in the EPID, i.e. the detector views the entire area of the irradiated screen
- The spatial resolution can be high, depending upon the thickness of the screen.
The system can generate an image quickly. Images can also be acquired rapidly, at up to 30 frames per second. However a major limitation to the design is that, only a very small fraction of light is captured by the lens (0.01-0.1%) and is focussed onto the camera (Munro et al., 1990). Because of this, the image quality can be reduced in two ways:

- When none of the light produced by the interaction of the photon with X-ray detectors reaches the TV camera, then no measurable signal is produced.
- If the noise produced by the preamplifier and the electronics of the system is much greater than the signal produced on the TV camera.

However from figure 2.2, the development of this device has concentrated on improving the light collection efficiency by increasing the output of the phosphor, increasing the aperture of the lens and finally increasing the detection efficiency of the camera. Moreover, the thicker screen will increase the quality of images, but suffers from the loss of spatial resolution and blemishes. Also increasing the lens size to increase the light collection efficiency will decrease spatial resolution causing non-uniform brightness and image distortion.
2.3.9 Matrix- Ion Chamber Array

Figure 2.3 shows a type of EPID system that was first developed by (Meertens et al., 1985) and (Meertens et al., 1988). The system employs ion transport in a liquid. The EPID consists of: (i) camera cassette and (ii) control unit.

The camera cassette consists of a 256-channel electrometer system, control electronics and a 256-channel high voltage switch system. The detector consists of two sets of 256 strip electrodes perpendicular to each other. The electrodes lie on the inner surface of two printed circuit boards, parted by a 0.8 mm gap filled with liquid film (Iso-octane, spectroscopically pure Merck), which serve as the ionization medium. A set of the electrodes is linked to a high-voltage supply, and the next is connected to a sensitive electrometer. A small ionisation chamber is positioned at the cross-point of the electrodes. There is a 1mm steel plate in front of the upper board which acts as the main radiation build-up material. The liquid is ionised when it is irradiated. In order to acquire an image, the ionization matrix should be scanned row by row, by successively switching the high voltage (HV) to different voltage electrodes and measuring the currents in all the
256 column electrodes. Since the analog to digital A/D conversion time is ~ 4µs, a single row can be read-out in 1ms. The outputs of the electrometers are multiplexed to a single amplifier and digitized in the control unit.

When the HV is switched to a row, a transient pulse is induced in the electrometers. Thus each row cycle consists of a wait-time for this pulse followed by readout of the 256 electrometers. The HV is the latest faster model of this detector which is 500V. The HV cycle time is based on two cycles of the accelerator 360 Hz timing pulses is 5.6 ms, comprising of a waiting time of 3ms, and the remainder is signal readout time. The current in each ionization chamber is measured ten times, and the values are averaged. The total frame acquisition time is therefore ~1.4 s (which is a result of multiplication of 256 electrometers by 5.6 ms pulse time). For the older models the total frame acquisition time is ~3s (Boyer et al., 1992). The signal readout is synchronised with the pulse timing of the accelerator to obtain smooth images without artifacts due to radiation pulse variation. Several frames can be averaged depending on the time available, and the dose rate delivered. Based on the long read out time of the device, it is only possible to measure the dose rate instead of measuring the integrated dose directly. The relation between output pixel values PV, that is ionization current, and dose rate is given by (Essers et al., 1995) as follows:

\[ PV = G(D) = a \times D^{1/2} + b \times D, \quad \text{---------2.1} \]

Where \(a\) and \(b\) are parameters with units \((\text{min/cGy})^{1/2}\) and \(\text{min/cGy}\), respectively. \(G\) is the gray level and is a function of dose rate.
The electronic portal imaging device (EPID), has several possible advantages over conventional X-ray film for portal imaging. Images acquired using EPID are available instantly and can be easily used for interactive adjustment of patient field positioning throughout radiotherapy. The images obtained are digital, which helps image processing, contrast enhancement and image matching. Moreover, because of the digital archiving nature of the EPID, a lot of space is available which permits recollection of images over a network system. The disadvantage of the EPID system is that they have huge or bulky devices that are of limited practicality with disappointing image quality. There is a tendency of changes with the introduction of more modern technology, such as
amorphous silicon-based devices, but there still remain issues with image quality. (Langmack, 2001)

Lastly we would also made mention of the computed radiography (CR) as an imaging detector that is used in the therapy department for patient verification. The CR system is designed to acquire and digitize X-ray images from image storage phosphor plates. The CR system consists of several components such as: an image reader/digitizer, cassettes containing imaging receptors (photostimulable-phosphor plates), a computer console, software, monitors, and a printer. The system has two plate readers. The first one is a single-plate readers (each cassette is loaded manually and read separately) and the next is the multiple-plate readers (multiple plate –up to -10- can be stacked and loaded automatically). In CR systems, images are obtained using X-rays in which imaging plates are inserted in a radiographic table cassette holder. When the imaging plate is exposed to an X-rays, electrons in the phosphor plate are excited to higher energy state, thus forming a latent image. The laser spot found within the reader scan the phosphor. Energy is absorbed by the trapped electrons and light is emitted as they go back to their original state. The light guide collects the light and transmits it to a photomultiplier tube, which produces an analog electrical signal that is amplified, and converted to digital signal, and stored digitally. After use the plate can be reused after it has been exposed to an erasing light that removes residual radiation. Images captured on the imaging plate by the X-ray source are extracted by the image reader. In the reader the cassette is slotted manually or automatically. Digital images acquired are downloaded unto the image-processing system, usually a computer workstation, for display and manipulation of images within 30-120 seconds. The plate is erased for reuse. Mostly damages to imaging Plates are due
to reckless handling and are expensive to replace. Modern CR readers have brushes or fans that instantly clear dust off plates, which helps to prevent scratches. The performance of the imaging plates can be deteriorated by cracks or scratches which thus reduce the quality of the plate for imaging. The light guide should always be cleaned as part of routine maintenance (quality of images are affected by dirt accumulation).

### 2.3.10 Scanning Linear Arrays (Solid state system).

This system uses a linear array of 255 diodes with a centre-to-center spacing of 2 mm that scans the image in 2 mm increments using a stepping motor. A 1.1 mm thick lead plate covering the diode array acts as build up layer. As the array only covers a small portion of the field the doses required to form an image are large. Spatial resolution is lower, due to large diode spacing than a finer detail system such as fluoroscopy camera devices. (Morton et al., 1991) produced a linear array of scintillating crystals (ZnWO4), each 5x5x25 mm in size. The crystals are arranged in a double row, 64 crystals per row with each row offset by half the crystal width to reduce the sampling interval. The X-rays interact and create high-energy electrons that pass through the volume of these crystals resulting in the creation of visible photons that are detected by photodiodes. As the crystals are optically transparent they can be long (25 mm) and hence the detection efficiency for the X-ray photons is very high, ~ 50% for 6 MV. The high quantum efficiency of the system allows the image to be produced with high subject contrast. However the spatial resolution is low.
2.3.11 Flat-Panel Device

Flat-panel based X-ray imaging is a new technology used to meaningfully increase the quality of on-line portal imagers. There are two main types of flat-panel solid state imaging device being developed for megavoltage imaging. These are amorphous silicon photodiode array (Street et al., 1992), and amorphous selenium photoconductor arrays (Zhao & Rowlands, 1992).

2.4 Radiographic Films

Figure 2.4 shows a cross section of a radiographic film system. During the early 1960s, at the height of the nuclear weapons testing, X-ray film manufacturers took extraordinary precautions to ensure that contamination from radioactive fallout did not invade their manufacturing environment, which could seriously fog the film. A radiographic film has two parts: (figure 2.4) the base and the emulsion

![Cross section of radiographic film](image)

*Figure 2.4: Cross section of radiographic film*
Emulsion coated on both sides in an X-ray film is called double emulsion film. The adhesive layer is made of a thin coating material found between the emulsion and the base, and its function is to provide uniform adhesion between the base and the emulsion. The adhesive layer maintains proper contact between the emulsion and the base during use. The emulsion is protected by a protecting covering of gelatin called overcoat. The overcoat protects the emulsion from pressure, scratches, and contamination during handling, processing, and storage.

2.4.1 The Base

The foundation of a radiographic film is the base. The function of the base is to provide a hard structure onto which the emulsion can be coated. The radiographic film can easily be handled due to the flexibility and fracture resistant nature of the base. The base of a radiographic film keeps the size and shape of the film during processing and use thus preventing image distortion. Originally, radiographic film base was made of a glass plate which was referred to as X-ray plate. High-quality glass in the World War 1 became extremely unavailable while medical application of X-rays, by the military, was increasing every day. The next substitute material used at that time was cellulose nitrate, which became the standard base. But this cellulose nitrate had one major deficiency: it was highly flammable. During the 1920s and early 1930s, due to improper and handling of some X-ray film files resulted in severe hospital fires. In the-1920s, film with a safety base cellulose triacetate, was introduced. Cellulose triacetate has a property similar to that of cellulose nitrate but not as flammable as cellulose nitrate. In the early 1960s, a
polyester base was introduced. Polyester is more resistant and stronger than cellulose triacetate, allowing easier transport through automatic processors. Polyester base are more used as film base.

2.4.2 Emulsion

The heart of the radiographic film is the emulsion. The emulsion is made of material with which the light photon, X-ray of the radiographic intensifying screen interacts. The emulsion consist of a homogeneous mixture of galatin and silver halide crystals. The galatin is clear, so it transmits lights, and it is highly porous for processing chemicals that penetrate the crystals of silver halide. The main function of the galatin is to give mechanical support by holding silver halide crystals uniformly together in place.

The most active component of the radiographic emulsion is the silver halide crystal. Ninety eight percent of halides in the emulsion is silver bromide and two percent is silver iodide. The atomic numbers of these halides in the emulsion are relatively higher (Z_{Br} = 35, Z_{Ag} = 47, Z_{I} = 53) compared to that of galatin and base (for both, Z~7). The formation of latent image on the radiographs rapidly increases when materials with high-Z atoms interact with X-ray and light photons. The silver halide crystals of the emulsion have different shapes, of which some are tabular, cubic, octahedral, polyhedral, or irregular in shape. The crystals are formed by dissolving metallic silver (Ag) in nitric acid (HNO₃). Light-sensitive silver bromide (AgBr) crystals are made by mixing silver nitrate with potassium bromide (KBr). The reaction is as fellow:

\[
AgNO_3 + KBr \rightarrow AgBr + KNO_3.
\]
From equation (2.2) silver bromide is precipitated while the potassium nitrate, which is soluble, is washed away. Radiographic film is manufactured in total darkness. From the moment the emulsion ingredients are brought together until final packaging, no light is present.

This radiographic film is still used for verification of patients in radiotherapy department. Nevertheless, their wide use is declining every day due to the facts of fixed dose latitude, fixed non-linear grey scale response, and limited potential for reducing dose to patient. The image contrast of these films remains unchanged once they have been processed. In spite of this, films are expensive, labour intensive, uses dangerous chemicals for processing, and finally the retrieval of films after been kept long is difficult. The compatibility of picture archiving and communication systems (PACS) in radiographic film system is poor.

2.5 Image Quality Assessment with Image J software.

Image quality assessment remains challenging in the field of image processing. According to the International Atomic Energy Agency (IAEA: TRS-457, 2007), optimization processes which involve balancing radiation dose and image quality do not always lead to reduction in radiation dose and it is essential to emphasize that image quality must always be sufficient to meet clinical requirements. Assessment of image quality plays a major role in the various imaging processing applications. It is meant to quantify a visual quality or anatomical amount of distortion or degradation in a given image. These degradations (artifacts) are inevitably part of any digital image processing
chain from acquisition, processing and transmission of images (Kud, 2012). Efforts have been made which seek to develop various image quality matrices that correlate very well with the perceived quality measurement, but only limited success has been achieved. (Zhou & Bovik, 2002) Assessment of image quality subjectively or by utilizing physically-based matrices is still in practice in modern radiation therapy. Notwithstanding that, a method for objective task-based image quality (IQ) assessment in radiation therapy was proposed by (Barrett & Kupinski, 2013). However details regarding its implementation remain to be investigated and the methodology has yet to find widespread application in the field of radiation oncology. This study shows a comprehensive implementation and evaluation of the matrices (SNR, CNR and RMTF) image quality assessment methodology. According to (Mohammadi & Ebrahimi-Moghadam, 2014) suggested that the need for evaluating the reliability has increased due to the increasing demand for image-based applications. According to these authors, this is based on the fact that assessing image quality is of importance for several image processing applications where the aim of image quality assessment methods is to instantly evaluate the quality of images in accordance with human judgements. In the field of radiation therapy, oncologist depends greatly on imaging modalities for quality image. From initial consultation to treatment completion, images are used to direct nearly every step of the patient encounter. Image guidance radiation therapy (IGRT) has become a standard of care for many treatment situations in radiation therapy (Thapa & Molloy, 2015). During simulation in IGRT, the CT scan obtained provides virtually full details regarding patient-specified, localized subject contrast. This information helps to establish an imaging goal and to select patient-specific imaging acquisition parameters that can
improve the similarity between reference and daily set-up images. (Thapa & Molloy, 2015). Generation of digitally reconstructed radiographs (DRRs) is one of the vital steps in treatment planning processes. These radiographs are used as reference images and processing tools which give full anatomical detail of the patients (Thapa & Molloy, 2015). The quality of a medical image is determined by the parameters such as: the characteristics of the equipment, and the imaging methods selected by the operator. The assessment of image quality is a composite of at least five factors: contrast, blur, noise, artifacts, and distortion (Sprawls, 1996). However, quality image is important for patient treatment in radiation therapy. Generally image quality can be evaluated mostly by using two approaches: objective and subjective methods (Kud, 2012). Subjective image quality assessment is by visual inspection of 2-D images by human observers. Objective image quality assessment uses mathematical models to calculate the quality of an image accurately.

Image J software is an imaging analysis programme created by the National Institute of Health, UAS, that performs image quality assessment objectively. The programme calculates the different parameters (area, standard deviation etc.) depending on the set measurements needed. Signal-to-noise ratio, (SNR) Contrast to noise ratio,(CNR) and Relative Modulation Transfer Function (RMTF) can be calculated using image J software, by selecting one or more specific regions of interest (ROI) in the image.
2.5.1 Calculation of Signal-to Noise ratio (SNR)

The signal to noise ratio (SNR) expresses the ratio between signal and noise in a large scale object. Using image J software, region of interest (ROI) is drawn on each of the step to determine their mean, standard deviation and other parameters. A study was carried out by (Roehrig, 1990) to determine the SNR of the CR and conventional film. The SNR was determined by taking the difference in the mean gray value of the objects and its background by the quadratic sum of the standard deviation of the background as a measure of noise.

\[
\text{SNR} = \frac{\text{signal}}{\text{noise}} = \frac{\text{mean object (ROI)} - \text{mean background (ROI)}}{\sqrt{\text{std object} + \text{std background}}} \quad \text{(2.3)}
\]

Where (std object) is the standard deviation of the object, (std background) is the standard deviation of the background, (mean object) is the mean gray value of the object within the region of interest and (mean background) is the mean gray value of the background with the ROI. According to the author, the result obtained from this study shows that SNR for CR system was meaningfully more effective than conventional films. Irradiation at different dose levels i.e. (9, 6, 3 and 1 rad) were done on both detectors. The SNR value for film at dose of 9 rad used was 0.766. Most of the SNR for the computed radiography system were significantly higher. At a dose of 9 rads the value was 1.13, at a dose 6 rad the value was 1.11 and at a dose of 3 rad the value was 1.03. Only at a dose of 1 rad was the SNR smaller than that of the film, which shows value of 0.732.
Figure 2.9: Portal image QC phantom obtained with BEAMVIEW Portal imaging system. The ROIs used for QC test marked on the image

The CNR for the QC phantom was calculated by using the formula

\[ \text{CNR} = \frac{l_1 - l_2}{\delta} = 2.5 \]

Where \( l_1 \) and \( l_2 \) are the average gray values of the brightest and darkest ROI and \( \delta \) is the random image noise. In their study, the authors determined CNR by comparing the energy of linac 6 MV and 23 MV.
Figure 2.10: Plot of CNR for the BEAMVIEW portal image system

Figure 2.10 shows a plot of CNR values recorded on a daily basis for BEAMVIEW\textsuperscript{plus} portal imaging system. During the 25 day calibration period, the mean values of CNR were 15.9 ± 1.2 and 26.4 ± 3.2 for 6 and 23 MV beams respectively. The dotted and dashed lines represent ± 3 standard deviations determined during the calibration period of 25 days. According to the author it shows that the CNR for 23 MV is higher than that of 6 MV, which is due to the higher dose rate (300 cGy/min) at 23 MV compared to that of 6 MV (200 cGy/min).

2.5.3 Calculation of Relative Modulation Transfer Function (RMTF)

A study was carried out by (Rajapakshe et al., 1996) to determine the relative modulation transfer function. The square wave modulation transfer function (SWMTF) of the
imaging system was defined as shown in equation 2.6, since the modulation of the bar pattern is in a form of a wave.

$$\text{SWMTF} = \frac{\Delta E(f)}{\Delta E_0}$$ \hspace{2cm} 2.6

Where $\Delta E_0$ and $\Delta E(f)$ are the modulation of input to and output of the system. According to this work they were not interested in an absolute measurement of the SWMTF, but only in the day-to-day variation in the system resolution, which use a relative measure (RMTF) of the SWMTF by calculating

$$\text{RMTF} (f) = \frac{\Delta E(f)}{\Delta E(f_1)}$$ \hspace{2cm} 2.7

Where $\Delta E(f_1)$ is the output modulation for the lowest frequency. Based on this study the output modulation is usually difficult to be obtain from noisy image, therefore,(Droege & Morin, 2011) suggested using the relationship between a signal amplitude and its variance ($M^2$) within an ROI containing the bar pattern, and the above relation can be rewritten as:

$$\text{RMTF} (f) = \frac{M(f)}{M(f_1)}$$ \hspace{2cm} 2.8

In the presence of random image noise, $M(f)$ can be obtained by

$$M^2 (f) = \delta_m^2 (f) - \delta^2 (f)$$ \hspace{2cm} 2.9

Where $\delta_m^2 (f)$ and $\delta^2(f)$ are the measured total variance and the variance due to random noise respectively. The total variance $\delta_m^2 (f)$ is obtained by measuring the pixels of the
ROI corresponding to frequency f. In order to measure the random noise in an image according to the author a pair of similar images are subtracted, and the standard deviation is obtained from the difference, thus avoiding contributions from fixed pattern noise. The variance of the subtracted ROI (\( \delta_{sub}^2 \)) will be

\[
\delta_{sub}^2 = \delta_1^2 + \delta_2^2
\]

Where \( \delta_1^2 \) and \( \delta_2^2 \) are the random noise variances of the ROIs for each image. The author therefore assume that these variances are equal and hence

\[
\delta_1 = \delta = \delta_{sub} / \sqrt{2}
\]

This work was comparing the resolution of a linac system with dual energies 6 MV and 23 MV. From the figure 2.11 below, the dotted and dashed lines represents the \( \pm 3 \) standard deviations determined during the calibration period and extrapolated to the test period for 6 and 23 MV respectively. Based on the author the system resolution of 6 MV is superior to that at 23 MV, this result is due to the larger physical beam penumbra at higher energies (Munro et al., 1990) and increase transmission through the bar patterns by the higher energy photons.
2.6. Quality Assurance of External Beam Radiotherapy

2.6.1 Quality Assurance

In radiotherapy, quality assurance is all measures that ensure reliability of medical prescription, and safe completion of that prescription, as regards the dose to the target volume, together with negligible dose to normal tissue, minimal exposure of personal and adequate patient monitoring aimed at determining the end result of the treatment. It must be stressed again that quality assurance in radiotherapy is connected to all aspects of radiotherapy process and all groups of staff should be involved in proceedings, since quality activities are interdependent.
2.6.2 Quality Control

Quality control is a tool used to regulate a process through which quality performance is measured with existing standards. Actions necessary to maintain the performance of quality are kept within the speculated standards. Quality control is one part of overall quality assurance, and is concerned with operational techniques and activities used:

- To ensure that quality requirements are in place.
- To adjust and correct performance if the requirements are not to standard.

2.6.3. Need for quality assurance in radiotherapy

High accuracy assessment in clinical setting in therapy is important in yielding a desired result of tumour control rates that is consistent in keeping complication rates within standard levels. Quality assurance measures in radiotherapy can be characterized as follows:

- Uncertainties and errors in dosimetry, treatment planning, equipment performance, and treatment delivery are minimized when quality assurance are in place and enforced. Quality assurance thus helps to improve dosimetric, geometric accuracy and the precision of dose delivery. With this, radiotherapy results will be improved (treatment outcomes), thus increasing tumour control rates and reducing complication and recurrence rates.
- Quality assurance as a tool does not only reduce the likelihood of errors and accident occurring, but it also provides the opportunity for recognizing and rectifying problems that tend to occur during patient treatment. It is not only for
incidents that are big but also for those that have higher possibility of minor incidents.

- Quality assurance also provides the opportunity of comparing true results obtained at different radiotherapy centres, thus ensuring quality and accurate dosimetry and treatment delivery.
- In modern radiotherapy technology, complex treatments can only be achieved when there is a high level of consistency and accuracy in the treatment.

The objective of radiotherapy is to ensure that patient safety regarding exposure to normal tissue is kept as low as reasonably achievable (ALARA) in accordance with dose delivered to the planning target volume (PTV). The aim of providing quality in radiotherapy treatment is to ensure patient safety and evasion of accidental exposures. The safety of patient is therefore key in quality assurance of radiotherapy treatments.

### 2.6.4 Verification of radiation field shape in radiotherapy

Detection of radiation field edges is required for two purposes: (i). to verify the shape of the radiation beam, (ii). to establish a coordinate system common to both reference and portal image in which to account for deviation in patient setup. Several image processing methods have been reported which extract the edges of radiation field from portal images, e. g (Bijhold et al., 1991b). (Eilertsen et al., 1994). (Petrascu et al., 2000). With the advance of MLC technology, methods are being developed to accurately detect the leaf positions, e.g. (Zhou & Verhey, 1994) (Eilertsen, 1997) Although MLC systems generally have their own feed-back control systems, there is an increasing interest in
independent online verification of proper leaf positioning using portal imaging, e.g. (James et al., 2000). Other methods for verification of beam shaping elements are based on geometric moments (Bijhold et al., 1992) chamfer matching (Gilhuijs & van Herk, 1993). The increasing use of small beam intensity modulated fields (e.g. by use of MLC technology) provides an interesting challenge for the verification of patient setup from portal images. The modulation process produces portal images with limited feedback mechanism for gated radiation therapy in order to cope effectively with organ motion during treatment. The largest challenge, however, is to narrow the existing gap between image processing research and the daily clinical routine.

2.6.5 Verification of patient setup

Inaccuracies in any steps, tolerances in mechanical components, and anatomical changes in the patient, movement of internal organs, incorrect immobilization, and human mistakes can lead to differences between the prescribed volumetric dose and the dose that is delivered over a period (Gilhuij et al., 1996), (Bansal et al., 1999). Instances of curve features are edges or ridges in the projection of bony anatomy or generalized gray value media axes. Template features are usually rectangular regions of pixel values.
The methods described figure 2.12 above are based on image arrangement in two dimensions (2D). Since the position and orientation of a patient are three-dimensional concepts, the comparison of portal with reference images in 2-D sometimes yields insufficient information to establish a complete correction of the setup in 3-D. For instance, patients rotating along the axes that are not perpendicular to the imaging plane cannot be quantified. Moreover, analysis in 2-D only provides accurate results of patient translations when the errors in rotational are small. For example, out-of-plane rotations of the pelvis larger than 2 degrees in prostrate treatment may cause dosimetrically significant misinterpretation of the position of the isocenter inside the patient (Hanley et al., 1995), (Bijhold et al., 1991a) (Remeijer et al., 2000). Thirdly, geometrical degeneracy in the alignment of rotation symmetrical objects like the femoral heads may cause ambiguous alignment in 2-D from a single view.

Figure 2.12: Picture illustrating systematic and random error in field placement
2.7 Accidents in radiotherapy

Patient with disease treated using therapy are faced with twofold of risk:

- Firstly, there is a possibility of failure to control the original disease, which, when it becomes malignant, is finally lethal to the patient
- There is also risk from increased radiation exposure to normal tissue.

In radiotherapy wrong administration or an accident is serious if results are due to overdose or an underdose. Overdoses are of general concern in conventional radiation protection legislation and protocols. There is an accident or wrong administration when the difference between prescribed and delivered dose are not in line as proposed. From the general aim of an accuracy approaching 5% (95% CL), about twice this seems to be an accepted limit for the definition of an accidental exposure (i.e. a 10% difference). Moreover dose applied accidentally outside the planned target volume may lead to bigger complications. In 2000 the IAEA investigated series of accidental exposures in radiotherapy, (Thwaites et al., 2003) and tended to draw lessons for the prevention of such occurrences. Criteria for classifying radiological accidents are:

- Direct causes of wrong administration
- Classification of possible dangers
- Preventability of misadministration.
Table 2.1 below shows some examples of direct causes of misadministration in external beam radiotherapy cataloged and analyzed in the IAEA report.

**Table 2.1: Causes of misadministration in external beam therapy in the 2000 IAEA Report** (Thwaites et al., 2003).

<table>
<thead>
<tr>
<th>Cause</th>
<th>Number of accidents</th>
</tr>
</thead>
<tbody>
<tr>
<td>Calculation error of exposure time or dose</td>
<td>15</td>
</tr>
<tr>
<td>Inadequate review of patient chart</td>
<td>9</td>
</tr>
<tr>
<td>Error in anatomical areas to be treated</td>
<td>8</td>
</tr>
<tr>
<td>Error in identifying the correct patient</td>
<td>4</td>
</tr>
<tr>
<td>Error involving lack of or misuse of a wedge</td>
<td>4</td>
</tr>
<tr>
<td>Error in calibration of Co-60 source</td>
<td>3</td>
</tr>
<tr>
<td>Transcription error of prescribed dose</td>
<td>3</td>
</tr>
<tr>
<td>Decommissioning of teletherapy source error</td>
<td>2</td>
</tr>
<tr>
<td>Human error during simulation</td>
<td>2</td>
</tr>
<tr>
<td>Error in commissioning of TPS</td>
<td>2</td>
</tr>
<tr>
<td>Technologist misread the treatment time or MU</td>
<td>2</td>
</tr>
<tr>
<td>Malfunction of accelerator</td>
<td>1</td>
</tr>
<tr>
<td>Treatment unit mechanical failure</td>
<td>1</td>
</tr>
<tr>
<td>Accelerator control software error</td>
<td>1</td>
</tr>
<tr>
<td>Wrong repair followed by human error</td>
<td>1</td>
</tr>
</tbody>
</table>
At every stage of treatment, errors can be done by every staff member involved in the process. Most accidental exposures in therapy are due to the failure in the application of adequate quality assurance programme. Mostly errors caused by human beings are leading to lack of knowledge, complacency and overconfidence.

Errors relating to human beings will always occur in any organization and in any activity. Therefore one of the aims of applied quality assurance programme is to reduce the amount of incidence and to classify them as early as possible in order to reduce the consequences.
CHAPTER THREE

MATERIALS AND METHOD

This chapter outlines in details the experimental method and materials employed in this study.

3.1. Study Area

This research was conducted in two major departments of the Korle-Bu Teaching Hospital in Ghana. These units were the radiotherapy and radiology departments of the hospital. This hospital was chosen because of the availability of CR system and the only teaching hospital using Cobalt machine for cancer treatment within Accra.

3.2. Theratron Equinox 100 Cobalt 60 teletherapy Machine.

Portal image quality assessments were performed for an Equinox 100 cobalt 60 teletherapy machine (Best Theratronics, Canada). The telecobalt unit was manufactured in April, 2013. The treatment head of the telecobalt machine is mounted isocentrically with source axial distance of 100 cm. In the treatment head is contained a double encapsulated cobalt 60 source having initial total source activity of 399.0 TBq measured on August 1st, 2013, by source manufacturer, Nordion Inc., Canada. This gives the teletherapy machine a reference beam output in water at a maximum depth dose (0.5 cm) of 189.49 cGy/min on December 12, 2013), measured after installation of the telecobalt
machine, based on International Atomic Energy Agency (IAEA) technical series report (TRS) 398 protocol. (Andreo et al., 2006). Percentage depth dose for the reference field size of 10 cm x 10 cm for a depth of 10 cm in water (PDD\textsubscript{10}), which was used as a beam quality specifier for megavoltage beam (Tagoe et al., 2017), is 58.36 % for the telecobalt machine. High activity encapsulated cobalt 60 source within the treatment head of the teletherapy machine has a diameter 2 cm and length of 4 cm, and is classified as C-146 teletherapy source capsules by Canadian Nuclear Safety Commission (Effects, 2016). The cobalt 60 source was formed by exposing stable cobalt 59 with neutrons in a reactor. The cobalt 60 source decays to nickel 60 with the emission of beta particles (E\textsubscript{max} = 0.32 MeV) and two photons energies 1.17 MeV and 1.33 MeV, given the cobalt 60 source an average energy of 1.25 MeV. The source is embedded in a source drawer mechanism which uses a pneumatic system to bring the source in and out of treatment position. Within the treatment head are asymmetrical collimators that allow the jaws, which define the shape of beam to move independently of each other, providing more freedom in treatment planning. Attached to the collimator system is an accessory holder with block tray code interlock to prevent the use of wrong accessory for treatment. The distance of radiation source to the accessory holder (or block tray) is 59.3 cm. The system has two timers: one is activated to sense the source when fully exposed and the other is activated to sense when the source when returns to its safe position. A door interlock device is also linked to the machine in a way that:

- It is possible to return the beam when the door is open.
- Also if the door is open when the beam is on, the source will automatically returns to its safe position.
When the unit is not in use, the source will be in the off position thus preventing radiation dose from exiting the treatment head. The source is slid or rotated into the treatment position when dose delivery is required. Dose delivered, is measured by treatment time as opposed to monitor unit in a linac machine. The source exposure system is designed in a way that under suitable faulty circumstance, such as mains power failure or any control system failure, the source exposure mechanism will return the source to its safe state. According to the International Electrotechnical Commission (IEC) standard a manual beam control system is available that directly operates on the source exposure device that is used in an emergency to return system to its beam-off position.

The telecobalt machine is configured to have features of a modern medical linear accelerator with the exception of a multileaf collimator system and an electronic portal imager. The field size that can be set on the machine ranges from 1x1 to 43 x 43 cm$^2$ (defines at the machine isocenter). The machine features a new motorized wedge (MW) system, which allows one to treat with any wedge angle ranging from 0 to 60 degrees. This is made possible with a fixed 60 degrees physical wedge permanently positioned in the treatment head of the machine, which can be brought automatically in and out of the path of the radiation beam during treatment delivery, such that combinations of time weighted beams with and without the wedge filter create the dosimetric effects of the required wedge. A picture of the telecobalt unit is shown in Figure 3.1
3.3. Computed Radiography Cassette

The CR cassette used for this study was available at the Korle-Bu Teaching Hospital in the radiology department. The model of the CR cassette was Fuji film with dimension 35.4 x 43.0 cm, serial number (CCWSL 17-189081) and was clinically accepted for use in 2012. The CR cassette used has rectangular plastic container which holds the imaging plate (IP). The function of the cassette is to:

- hold the IP and protect it from damage
- Exclude dusts and dirt from the sensitive imaging plate.

The plate is coated with europium-activated Barium Fluoro-Halide (BaFX: Eu$^{2+}$). The data are kept in the PSP imaging plate as electrons, in a semi-stable higher state. Image
details are obtained by scanning the plate with laser beam. The CR cassettes are extremely sensitive to background radiation and scatter. If the CR cassette is out of use for days, it should be inserted into the reader for re-erasure to prevent fogging on the image. The CR cassettes get deteriorated with usage over a period of two years especially when not properly handled. This effect of damage on the cassette for not closing tightly on the imaging plate (IP) will lead to poor quality image. Figure 3.2 shows samples of different dimensions of the IP cassette used in CR systems.

![Figure 3.2: Fujifilm IP Cassette type CC](image.png)

**Figure 3.2: Fujifilm IP Cassette type CC**

### 3.4. Computed Radiography Reader

Figure 3.3 shows the CR reader that was employed for this study and was found at the radiology department of the Korle-Bu Teaching Hospital. The model is CR-IR 359, with serial No. 26126981. The reader was manufactured by Shimadzu Corporation, Japan in January 2012, and installed for clinical use in 2012. Averagely, this CR reader serves fifty (50) patients per day. It has two insertion openings for cassette of different sizes. The reader is used to extract the image that has been stored on the imaging plate. The cassettes were inserted manually into the reader. The reader scans the latent image that
has been stored on the phosphor plate with a laser spot. Energy is absorbed by the electrons upon interaction with the laser spot and lights are emitted as they go back to their original level. The light guide collects the light and transfer it to the photomultiplier tube, which produces an analog electrical signal that is amplified, and converted to a digital signal which is kept digitally. After 45 seconds digital image were acquired and downloaded to a computer workstation for display and manipulation. The images obtained were then printed on a film for examination. The plate was erased for reuse after exposure to white light.

Figure 3.3: CR reader at Korle-Bu Teaching Hospital

3.5. Radiotherapy Film Cassette.

Figure 3.4 show the radiographic film cassette used by the radiotherapy department of the Hospital for setup verification. The film cassette specification used for this study was
Curix Universal AGFA-GEVAERT of size $34 \times 43$ cm in dimension. The film cassette is a rectangular rigid holder that contains the film. The top cover facing the beam entrance is made of material with low atomic number such as plastic which helps to minimize the attenuation of the X-ray beam. The back cover is made of metal to minimize back scatter. Its functions are:

- Hold the film and protect it from damaged.
- Exclude all light from entering the cassette and thus fogging the film.
- Maintain close and uniform contact of the film.
- Prevent the film from accumulating dust and dirt

The features of the film cassette used are as follow:

- Lightweight to facilitate easy handling and carrying.
- Strong and rigid to withstand physical damage from daily wear and tear.
- The internal surface have adequate layer of lead foil attached to reduce the risk of back scatter

The cassette containing the film is irradiated with an X-ray beam. The film is removed from the cassette and loaded manually into the automatic processing system at the hospital. A lot of steps are involved in the processing. This include wetting, development, fixing, washing and drying. From this processes, the film is then ready to be used.
3.6. Scanner (Microtek Scan Maker 9800XL)

There are many scanners with high quality, which are used to scan coloured films. Figure 3.5 shows Microtek Scan Maker 9800XL which was used to scan the films of various sizes. It is flat bed with a large format A3 scanner with super and versatile function specifically designed for proper image quality. It weighs 12 kg with dimensions $62.7 \times 37.6 \times 13$ cm (W× D× H).

With the worm-up free energy saving Light-Emitting Diode (LED) source, the scanner echoes to go green. The scanner has an optical dynamic range of 3.7 Dmax which deliver accurate colour, superb shadows and sharp images. It has a smart-touch button (scan) located at the front panel of the scanner which provides a quick and easy way to capture image that can automatically save as files or sent to another application. The tabloid size scan bed of this scanner allow oversize scanning of original contact sheets, large pieces of art, mechanical blue prints and X-rays.
With a true optical resolution of 1600dpi scanning, the Scan Maker 9800XL can capture a greater amount of details from the format A3 scanner areas, allowing scan to be printed with high image quality.

*Figure 3. 5: Scanner (Microtek scan maker 9800XL)*

### 3.7. Image J Software

The Image J is a Java-based image processing program created at the National Institute of Health (NIH). This software was designed with an open architecture that provides extensibility via java plugins and records macros. User written plugins make it possible to solve many image processing and analysis problems. Figure 3.6 shown below shows an open view of the software interface that was used for the study.
3.8. Film Processor

The film processor that was used for this study is shown in figure 3.7, which is Kodak Medical X-ray Processor 102, with serial No. 117510-0903-7107 and was made in Germany for CARESTREAM HEALTH INC. Rochester. NY 14608. This processor was installed for clinical use in 2008 at the Korle-Bu Teaching Hospital. The amount of film processed by the system is approximately 35 per day depending on the cases available for treatment. The processor is placed in a dark room at the hospital. Image acquisition involves several steps: The film is manually removed from the cassette and inserted in the processor. In the processor the processing sequence involves the wetting of the film to swell the emulsion, so that subsequent chemical baths can reach all the part of the emulsion uniformly. The developing stage is very short and highly critical. The film is rinsed in an acid solution designed to stop the developing process and remove excess developer chemicals from the emulsion. The next stage is fixing which removes remaining silver halide from emulsion and hardens galatin. The final stage is washing and drying in other to removes excess chemicals and water for viewing of radiograph. The time duration of the process is approximately two minutes.
3.9. Phantom

The constructed phantom was made of plexiglas of thickness 1.8 cm with dimension 40 x 40 x 17 cm$^3$. Various objects such as steel, cerrobend, plexiglas having densities similar to tissues (bone and soft tissue) found within the human body were placed in the phantom to give contrast level.

Two step wedges made with cerrobend and plexiglas materials were placed at the two opposing side of the phantom which were used to determine signal to noise ratio and contrast level as well. The cerrobend is eutectic alloy 50% bismuth, 27% lead, 13.3% tin, and 10% cardium by weight. Bar patterns made of steel of different thickness ranging from 4, 3, 2, and 1 mm with spacer made with plexiglas material were used to determine the resolution of the imaging system.

Patient irradiation geometries were replicated by the use of the constructed phantom. The thickness of this phantom replicated the anatomical structure of the abdomen of human.
The phantom was irradiated with Theraton equinox-100 cobalt 60 unit at the hospital. The experimental setup was done by placing the phantom on the couch of the treatment machine employing source to surface distance (SSD) technique of 100 cm. Both the CR and the radiographic film cassette were mounted on the cassette holder one at a time for every irradiation. The cassette on the holder was level accurately using a sprit or alcohol level in order to maintain equal balance of the cassette on the holder. The cassette was then placed under the couch with detector to couch distance 14.5 cm, and source to detector distance 131.6 cm.

The phantom was irradiated with CR as detector with beam having a field size of 30 x 30 cm and at various treatment time ranging from 0.02 to 0.4 min. Immediately after each irradiation the CR cassette obtained from the radiological department of the Korle-Bu Teaching hospital was taken back to be read with the CR reader.

The CR cassette with the imaging plate was manually inserted into the reader for image acquisition. After 45 seconds, the reader which was connected to a computer, displayed an image on the screen. The digitized image was printed on a film for scanning.

A similar procedure was repeated using a radiographic film as a detector. The film cassette was then taken to the dark room for processing using an automatic film processor, in order to acquire the image. Both the CR and the radiographic images were acquired and scanned using the scan maker 9800XL model at the teaching hospital. Images were finally analyzed using Image J software. Figure 3.8 shows the experimental setup with the constructed phantom.
3.10. Image Quality Assessment

A sample of images were acquired at treatment time ranging from 0.02, 0.2, 0.3, 0.4 minutes for examination done on each CR and radiographic film system employed. Image quality was assessed using Image J software (version 1.48). These images were assessed by measuring the SNR, CNR and RMTF which determine the spatial resolution of the imaging system.

In calculating the contrast to noise ratio (CNR), regions of interest (ROI) were marked on each of the step wedge design from the image of the phantom constructed, and a background ROI was also marked. The mean gray values for each of the steps and the
background results were obtained by the image J software. The formula used to estimate CNR is found in the literature review with equation (2.4).

The results obtained from this formula were then used to compare the CNR for CR detector and the screen film used at the therapy department for portal imaging.

The SNR was determined by drawing ROIs on each of the six steps of the phantom. SNR was determined by taking the difference in the mean gray value of the object and its background by the quadratic sum of the standard deviation of the background as a measured noise. The SNR was then calculated by using the formula found in chapter two with equation (2.3)

The relative modulation transfer function (RMTF) which determines the resolution of an imaging system was determined using equation (2.8), (2.9), (2.10), (2.11).

The images for both CR and radiographic film from which these quality assessments were made are show in figures 3.9 and 3.10 below.
<table>
<thead>
<tr>
<th>No</th>
<th>Area of ROI</th>
<th>Mean Gray Values of ROI</th>
<th>Standard Deviation (std)</th>
<th>Min Gray Values</th>
<th>Max Gray Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>10000</td>
<td>152.66</td>
<td>3.29</td>
<td>143</td>
<td>161</td>
</tr>
<tr>
<td>2</td>
<td>10000</td>
<td>142.95</td>
<td>2.89</td>
<td>131</td>
<td>151</td>
</tr>
<tr>
<td>3</td>
<td>10000</td>
<td>114.59</td>
<td>3.12</td>
<td>105</td>
<td>122</td>
</tr>
<tr>
<td>4</td>
<td>10000</td>
<td>62.27</td>
<td>3.36</td>
<td>51</td>
<td>76</td>
</tr>
<tr>
<td>5</td>
<td>10000</td>
<td>15.57</td>
<td>1.91</td>
<td>10</td>
<td>20</td>
</tr>
<tr>
<td>6</td>
<td>10000</td>
<td>8.08</td>
<td>0.28</td>
<td>8</td>
<td>9</td>
</tr>
</tbody>
</table>

**Table 4.2: Calculated results of SNR for CR obtained from scanner data of irradiation time of 0.1 min**

<table>
<thead>
<tr>
<th>Step Thickness (cm)</th>
<th>Transmission Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>6.18</td>
<td>0.04</td>
</tr>
<tr>
<td>5.07</td>
<td>0.07</td>
</tr>
<tr>
<td>3.82</td>
<td>0.14</td>
</tr>
<tr>
<td>2.63</td>
<td>0.26</td>
</tr>
<tr>
<td>1.23</td>
<td>0.54</td>
</tr>
</tbody>
</table>
Table 4.1 shows measured values obtained from the scanned phantom using image J. A region of interest (ROI) was drawn on each of the step found in the phantom and measured values such as the area, standard deviation, mean, minimum and maximum gray values seen in the table were generated by the software.

Table 4.2 was deducted form table 4.1 which helps in the calculation of signal to noise ratio of CR. In this table, the SNR, transmission values for each of the steps with various thickness were calculated as shown in the table. The calculated SNR is high (76.42) when the step thickness is high (6.18 cm) thus showing a very low transmission value (0.04). Also form table 4.2 its clear that the SNR value decreases as the step thickness decreases while the transmission value increases with the decrease in both the step and SNR.

![SNR For CR (0.1min)](image)

**Figure 4.1: Graph of SNR for CR of irradiation time of 0.1 min**
Table 4.3: Result of scanned data using image J software (FILM) of 0.1 min

<table>
<thead>
<tr>
<th>No</th>
<th>Area of ROI</th>
<th>Standard deviation (std)</th>
<th>Mean of ROI</th>
<th>Min Value</th>
<th>Gray Value</th>
<th>Max Value</th>
<th>Gray Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>10000</td>
<td>1.38</td>
<td>199.59</td>
<td>183</td>
<td>203</td>
<td></td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>10000</td>
<td>0.97</td>
<td>198.29</td>
<td>180</td>
<td>203</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>10000</td>
<td>1.52</td>
<td>192.28</td>
<td>172</td>
<td>197</td>
<td></td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>10000</td>
<td>1.43</td>
<td>181.69</td>
<td>167</td>
<td>186</td>
<td></td>
<td></td>
</tr>
<tr>
<td>5</td>
<td>10000</td>
<td>1.84</td>
<td>166.14</td>
<td>153</td>
<td>173</td>
<td></td>
<td></td>
</tr>
<tr>
<td>6</td>
<td>10000</td>
<td>1.63</td>
<td>144.03</td>
<td>129</td>
<td>150</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 4.4: Calculated results of SNR for Film obtained from scanner data of irradiation time of 0.1 min

<table>
<thead>
<tr>
<th>Mean(M) - Background (B)</th>
<th>Std object + std of Background</th>
<th>Std of object + Std of B</th>
<th>SNR = Mean - background / Std object + Std of B</th>
<th>Step Thickness (cm)</th>
<th>Transmission Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>55.56</td>
<td>3.02</td>
<td>1.73</td>
<td>31.96</td>
<td>6.18</td>
<td>0.04</td>
</tr>
<tr>
<td>54.26</td>
<td>2.61</td>
<td>1.61</td>
<td>33.56</td>
<td>5.07</td>
<td>0.07</td>
</tr>
<tr>
<td>48.24</td>
<td>1.52</td>
<td>1.23</td>
<td>39.06</td>
<td>3.82</td>
<td>0.14</td>
</tr>
<tr>
<td>37.66</td>
<td>3.07</td>
<td>1.75</td>
<td>21.46</td>
<td>2.63</td>
<td>0.26</td>
</tr>
<tr>
<td>22.11</td>
<td>3.48</td>
<td>1.86</td>
<td>11.85</td>
<td>1.23</td>
<td>0.54</td>
</tr>
</tbody>
</table>
Table 4.3 shows measured values obtained from scanned phantom using image J. A region of interest (ROI) was drawn on each of the steps found in the phantom and measured values such as the area, standard deviation, mean, minimum and maximum gray values seen in the table were generated by the software.

Table 4.4 was deducted from table 4.3 which helps in the calculation of signal to noise ratio of film. In this table, the SNR, transmission values for each of the steps with various thicknesses were calculated as shown in the table. The calculated values for SNR are not consistent with the step thickness for film as that shown for CR in table 4.2. The transmission values increase as the step thickness decreases.

Figure 4.2: Graph of SNR for Film of irradiation time of 0.1 min
Figure 4.1 shows a graph of signal to noise ratio versus transmission values of the CR detector when using a step wedge found in the phantom of 0.1 minute treatment time. The graph shows a decrease in SNR value from 76.42 to 5.04 exhibiting a decrease from 31.4% to 2.1%, with an increase in the transmission values from 0.04 to 0.54 as the step thickness decreases from (6.18 cm) to (1.23 cm). On the other hand, the graph of Figure 4.2 shows a conventional film detector with an increase in SNR value from 31.96 to 39.06 exhibiting an increment from 23.2% to 28.3%. The graph shows a decrease in SNR from 39.06 to 11.85 exhibiting a decrease of 28.3% to 8.6%.

### 4.2 Comparing SNR of the two detectors (CR and Film) of irradiation time 0.1 min.

#### Table 4.5: SNR values for both detectors (CR and FILM)

<table>
<thead>
<tr>
<th>SNR (CR)</th>
<th>SNR (FILM)</th>
<th>STEP THICKNESS (cm)</th>
<th>TRANSMISSION VALUES</th>
</tr>
</thead>
<tbody>
<tr>
<td>76.42</td>
<td>31.96</td>
<td>6.18</td>
<td>0.04</td>
</tr>
<tr>
<td>75.65</td>
<td>33.56</td>
<td>5.07</td>
<td>0.07</td>
</tr>
<tr>
<td>57.71</td>
<td>39.06</td>
<td>3.82</td>
<td>0.14</td>
</tr>
<tr>
<td>28.37</td>
<td>21.46</td>
<td>2.63</td>
<td>0.26</td>
</tr>
<tr>
<td>5.04</td>
<td>11.85</td>
<td>1.23</td>
<td>0.54</td>
</tr>
</tbody>
</table>

Table 4.5 shows calculated values of signal to noise ratio, step thickness and transmission for the two detectors (CR and film) used in the experiments.
Figure 4.3. *Graph showing the two detectors SNR (CR and film)*

In figure 4.3 at step five where the step thickness is high (6.18 cm), with the lowest transmission value of 0.04, CR shows a superior SNR value (76.42) exhibiting 31.4% more than that of the film (31.96) exhibiting 23.2%. From the graph, a decrease in steps in CR, the SNR values decreases from (76.42) to (5.04) with an increase in the transmission values from (0.04) to (0.54). With the film on the other hand at step thickness of 6.18 cm it shows a SNR value of (31.96), a decrease in steps, it shows a gradual increase of (39.06), immediately after this point there is a decrease in SNR to (11.85). At the lower steps (1.23 cm) of the two detectors the film shows a higher value in SNR of 11.85 (8.6%) to that of CR of value 5.04 (2.1%).
4.3 Contrast to Noise Ratio for both CR and Radiographic Film at irradiation time 0.2min

Table 4. 6: Result of CNR from scanned data using image J (CR) 0.2 min

<table>
<thead>
<tr>
<th>No</th>
<th>Area of ROI</th>
<th>Standard deviation (std)</th>
<th>Mean Values of ROI</th>
<th>Min Gray Values</th>
<th>Max Gray Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>10000</td>
<td>5.30</td>
<td>126.49</td>
<td>114</td>
<td>149</td>
</tr>
<tr>
<td>2</td>
<td>10000</td>
<td>4.96</td>
<td>108.16</td>
<td>94</td>
<td>120</td>
</tr>
<tr>
<td>3</td>
<td>10000</td>
<td>4.50</td>
<td>83.29</td>
<td>63</td>
<td>95</td>
</tr>
<tr>
<td>4</td>
<td>10000</td>
<td>2.81</td>
<td>29.16</td>
<td>21</td>
<td>39</td>
</tr>
<tr>
<td>5</td>
<td>10000</td>
<td>0.79</td>
<td>9.87</td>
<td>8</td>
<td>12</td>
</tr>
<tr>
<td>6</td>
<td>10000</td>
<td>0.50</td>
<td>8.50</td>
<td>7</td>
<td>9</td>
</tr>
</tbody>
</table>

Table 4. 7: Calculated result of CNR for CR acquire from scanner data of irradiation time of 0.2 min

\[
CNR = \frac{SA - SB}{\text{Std of background}}
\]

<table>
<thead>
<tr>
<th>Signal in A- Signal in B (SA-SB)</th>
<th>CNR</th>
<th>Step thickness (cm)</th>
<th>Transmission Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>18.33</td>
<td>36.59</td>
<td>6.18</td>
<td>0.04</td>
</tr>
<tr>
<td>24.87</td>
<td>49.64</td>
<td>5.07</td>
<td>0.07</td>
</tr>
</tbody>
</table>
Table 4.6 shows measured values obtained from scanned phantom when evaluating for CNR in CR using the software. Regions of interest (ROI) on each of the steps give measured values for each parameters: area, standard deviation, mean, maximum and minimum gray values seen in the table were generated by the image J software.

Table 4.7 was obtained from table 4.6 which helps in the calculation of CNR of the CR. The transmission values for each of the step with various thickness were calculated as shown in the table. From the table the CNR values for CR shows some fluctuations as the step thickness decreases.

<p>| | | | | |</p>
<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>54.12</td>
<td>108.03</td>
<td>3.82</td>
<td>0.14</td>
<td></td>
</tr>
<tr>
<td>19.29</td>
<td>38.51</td>
<td>2.63</td>
<td>0.26</td>
<td></td>
</tr>
</tbody>
</table>

**Figure 4. 4: Graph of CNR for CR of 0.2 min irradiation time.**

### Table 4.8: Result of CNR from scanned phantom for film of 0.2 min

<table>
<thead>
<tr>
<th>No</th>
<th>Area of ROI</th>
<th>Standard deviation (std)</th>
<th>Mean Values of ROI</th>
<th>Min Gray Values</th>
<th>Max Gray Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>10000</td>
<td>2.23</td>
<td>179.87</td>
<td>157</td>
<td>188</td>
</tr>
<tr>
<td>2</td>
<td>10000</td>
<td>1.51</td>
<td>179.11</td>
<td>167</td>
<td>184</td>
</tr>
<tr>
<td>3</td>
<td>10000</td>
<td>1.56</td>
<td>170.41</td>
<td>156</td>
<td>176</td>
</tr>
<tr>
<td>4</td>
<td>10000</td>
<td>1.58</td>
<td>158.67</td>
<td>149</td>
<td>165</td>
</tr>
<tr>
<td>5</td>
<td>10000</td>
<td>2.61</td>
<td>132.05</td>
<td>111</td>
<td>141</td>
</tr>
<tr>
<td>6</td>
<td>10000</td>
<td>2.64</td>
<td>102.85</td>
<td>82</td>
<td>113</td>
</tr>
</tbody>
</table>

### Table 4.9: Calculated result of CNR for Film acquire from scanner data of irradiation time of 0.2 min

\[
CNR = \frac{SA - SB}{Std \ of \ background}
\]

<table>
<thead>
<tr>
<th>Step thickness (cm)</th>
<th>Transmission Values (SA - SB)</th>
</tr>
</thead>
<tbody>
<tr>
<td>6.18</td>
<td>0.04</td>
</tr>
<tr>
<td>5.07</td>
<td>0.07</td>
</tr>
<tr>
<td>3.82</td>
<td>0.14</td>
</tr>
<tr>
<td>2.63</td>
<td>0.26</td>
</tr>
</tbody>
</table>

70
Table 4.8 shows measured values obtained from scanned phantom when evaluating for CNR in film using the software. Region of interest (ROI) on each of the steps gives measured values for each parameters: area, standard deviation, mean, maximum and minimum gray values seen in the table were generated by the image J software.

Table 4.9 was obtained from table 4.8 which helps in the calculation of CNR of the Film. In the table the CNR, transmission values for each of the steps with various thicknesses were calculated and results shown in the table. From the table the CNR values for Film shows an increase as the step thickness decreases from 6.18 cm to 2.63 cm while the transmission values increases.

![Graph of CNR for Film of 0.2 min](image)

**Figure 4.5: Graph of CNR for Film of 0.2 min**

Figure 4.4 shows a graph of contrast to noise ratio versus transmission values of the CR detector when using a step wedge found in the phantom of 0.2 minutes treatment time. The graph therefore shows an increase in CNR value from 36.59 (15.7 %) to 108.03 (46.4
% with an increase in transmission value from 0.04 to 0.14, with a decrease in step thickness from 6.18 cm to 3.82 cm. After this pick there is sharp fallout in the CNR from 108.03 exhibiting (46.4%) to 38.51 exhibiting (16.5%). On the other hand, the graph of Figure 4.5 shows a conventional Film detector. In this graph the CNR value increases from 0.28 to 10.08 exhibiting an increase from (1.6%) to (55.6%) with an increase also in the transmission value from 0.04 to 0.268 while the step thickness decreases from 6.18 cm to 2.63 cm.

4.4 Comparing CNR of the two detectors (CR and Radiographic Film) of 0.2 min

Table 4.10: Showing CNR values for CR digital detector and Film.

<table>
<thead>
<tr>
<th>Step thickness (cm)</th>
<th>Transmission</th>
<th>CNR for CR</th>
<th>CNR for Film</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Values</td>
<td></td>
<td></td>
</tr>
<tr>
<td>6.18</td>
<td>0.04</td>
<td>36.59</td>
<td>0.28</td>
</tr>
<tr>
<td>5.07</td>
<td>0.07</td>
<td>49.64</td>
<td>3.29</td>
</tr>
<tr>
<td>3.82</td>
<td>0.14</td>
<td>108.03</td>
<td>4.44</td>
</tr>
<tr>
<td>2.63</td>
<td>0.26</td>
<td>38.51</td>
<td>10.08</td>
</tr>
</tbody>
</table>

Table 4.10 shows compared calculated values for the two detectors (CR and Film). From the table the calculated values of CNR in the CR are far much higher than that for the film with the same transmission and step thickness values.
4.5 Spatial Resolution

Table 4.11: showing RMTF for CR digital detector of 0.1 min irradiation time

<table>
<thead>
<tr>
<th>No</th>
<th>Area (ROI)</th>
<th>Mean (ROI)</th>
<th>Standard deviation (M)</th>
<th>Frequency (1lp/mm)</th>
<th>$\delta^2 = \delta_{sub}^2 / \sqrt{2}$</th>
<th>Variance $(M)^2$</th>
<th>RMTF $= M^2 - \delta^2 / 124.9$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>10000</td>
<td>96.57</td>
<td>11.17</td>
<td>0.12</td>
<td>16.34</td>
<td>124.94</td>
<td>0.8</td>
</tr>
<tr>
<td>2</td>
<td>10000</td>
<td>50.74</td>
<td>16.35</td>
<td>0.16</td>
<td>16.34</td>
<td>267.35</td>
<td>2.0</td>
</tr>
<tr>
<td>3</td>
<td>10000</td>
<td>17.52</td>
<td>6.59</td>
<td>0.25</td>
<td>16.34</td>
<td>43.44</td>
<td>0.2</td>
</tr>
<tr>
<td>4</td>
<td>10000</td>
<td>13.29</td>
<td>1.78</td>
<td>0.5</td>
<td>16.34</td>
<td>3.18</td>
<td>0.1</td>
</tr>
</tbody>
</table>

Table 4.11 shows the parameters: standard deviation, mean, variance and frequencies values on the bar pattern in the phantom which were used to calculate the relative modulation function, of the CR system to determine the resolution of the imaging unit used as a function of the frequency.
Figure 4. 9: Graph of RMTF for CR of 0.1 min

Table 4. 12: Showing RMTF for Film of 0.1 min

<table>
<thead>
<tr>
<th>No</th>
<th>Area (ROI)</th>
<th>Mean (ROI)</th>
<th>Standard deviation (M)</th>
<th>Frequency (1lp/mm)</th>
<th>( \delta^2 = \delta_{sub}^2 / \sqrt{2} )</th>
<th>Variance ( (M)^2 )</th>
<th>RMTF = ( M^2 ) - ( \delta^2 )</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>10000</td>
<td>191.77</td>
<td>3.251</td>
<td>0.12</td>
<td>7.04</td>
<td>10.56</td>
<td>0.3</td>
</tr>
<tr>
<td>2</td>
<td>10000</td>
<td>180.10</td>
<td>5.023</td>
<td>0.16</td>
<td>7.04</td>
<td>25.23</td>
<td>1.7</td>
</tr>
<tr>
<td>3</td>
<td>10000</td>
<td>170.58</td>
<td>4.71</td>
<td>0.25</td>
<td>7.04</td>
<td>22.18</td>
<td>1.4</td>
</tr>
<tr>
<td>4</td>
<td>10000</td>
<td>167.84</td>
<td>1.554</td>
<td>0.5</td>
<td>7.04</td>
<td>2.41</td>
<td>0.4</td>
</tr>
</tbody>
</table>

Table 4.12 shows the parameters as: standard deviation, mean, variance and frequencies values on the bar pattern in the phantom that were used to calculate the relative modulation function, of the film system to determine the resolution of the imaging unit used as a function of the frequency.
Figure 4.10: Graph of RMTF for Film detector

Figure 4.11 and 4.12 shows the graphical representation of RMTF for CR and Film. From the graph in figure 4.11 the CR shows a higher RMTF value of 2.0 at a lower frequency of 0.12 lp/mm, and a fall in RMTF at higher frequencies. Also in fig 4.12, the film shows a higher RMTF value of 1.7 at a frequency of 0.16 lp/mm and also a fall in RMTF at higher frequencies.
4.6 Comparing the RMTF for the two detectors (CR and Film) at 0.1 Min

Table 4.13: Shows RMTF values for CR digital detector and Film

<table>
<thead>
<tr>
<th>Frequency (1lp/mm)</th>
<th>RMTF (CR)</th>
<th>RMTF (FILM)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.12</td>
<td>0.8</td>
<td>0.3</td>
</tr>
<tr>
<td>0.16</td>
<td>2</td>
<td>1.7</td>
</tr>
<tr>
<td>0.25</td>
<td>0.2</td>
<td>1.4</td>
</tr>
<tr>
<td>0.5</td>
<td>0.1</td>
<td>0.4</td>
</tr>
</tbody>
</table>

Figure 4.13 shows the RMTF of both the CR and Film drawn on the same axis. From this graph at lower frequency of 0.12 lp/mm to 0.16 lp/mm the RMTF of the CR is higher than that of the film from 25.8% to 7.8% and 64.5% to 44.7%. As the frequency increases
from 0.25 lp/mm to 0.5 lp/mm, there was a drastic reduction in the RMTF of the CR to that of the film from 6.4% to 36.8%. The resolution of the CR at 50% of RMTF was 0.2 lp/mm and at 50% of the film RMTF the resolution was 0.4 lp/mm at higher frequency. At 50% RMTF the frequency (lp/mm) of the CR is 0.2lp/mm while that of the Film is 0.4 lp/mm indicating better resolution. This result is in accordance with work done by (Konstantinidis, 2011) (Williams et al., 2007), (Lu, Nickoloff, So, & Dutta, 2003), and (Bansal, 2006).
4.7 summary of findings

*Table 4.14: Shows the findings of the research in accordance with the specific objectives*

<table>
<thead>
<tr>
<th>Specific Objectives</th>
<th>Findings based on the detectors (CR and Film)</th>
</tr>
</thead>
<tbody>
<tr>
<td>To quantitatively determine the quality of images obtained from CR and Films using the image quality parameters</td>
<td>The calculated values of SNR for both detectors shows that the CR system have a higher SNR than the film.</td>
</tr>
<tr>
<td>Signal to noise ratio (SNR)</td>
<td>It also shows that CNR value is higher for CR system than film</td>
</tr>
<tr>
<td>Contrast to noise ratio (CNR)</td>
<td>The spatial resolution is however marginally higher for film system than CR system.</td>
</tr>
<tr>
<td>Relative modulation transfer function (RMTF)</td>
<td>Another very important finding of the work was that at low doses anatomical landmarks are clearly visible on the CR system, while nothing is seen on the film. At higher doses, images on the CR system become darker while images on the film now become clear. This implies that high doses are needed for the film to be useful while low doses are enough for the CR system thereby sparing patients of the deleterious effects of high doses.</td>
</tr>
<tr>
<td>Visibility response of images on detectors (CR and Film) at low and high doses or treatment time.</td>
<td></td>
</tr>
</tbody>
</table>

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1.8 Contribution to knowledge.

The study

- Establishes the superiority of image quality with the CR to that acquired with film as portal imager.

- Establishes the suitability of the CR system for routine use for patient positioning verification at radiotherapy department of the Korle-Bu Teaching Hospital

- Involved the construction of a phantom for studying radiation dose distribution and assessing the positioning of patients. This is a phantom that the resident Medical Physicist could be using on daily basis for verifying the patient positioning in telecobalt therapy set ups
CHAPTER FIVE

CONCLUSION AND RECOMMENDATION

This chapter provides the conclusion drawn from this study and recommendations on the use of CR detector for treatment setup verification at the Korle-Bu Teaching Hospital in Ghana.

5.1 CONCLUSION

Based on the results obtained from the study using CNR image quality evaluator shows that at irradiation time of 0.2 minutes, the CR shows an average value of 58.2 while that of the conventional film exhibit a value of 4.52 at the same irradiation time. The above result indicates that CR digital detector shows a superior image quality than that of the film.

The SNR average values at irradiation time of 0.1 min was also calculated as a means of quantifying the level of noise in the images acquired from both conventional film and CR digital detector used. The average value for CR was 48.6, while the film value was 27.6. This result clearly shows that images acquired on the CR show less noise to that of the film, which thus exhibit superior image quality to that of the film.

Notwithstanding this, the RMTF values for both the film and CR detectors were evaluated to determine their spatial resolution. The value of RMTF at 0.1 min irradiation
time for film was 0.95 while the CR was 0.78. From this results, the film shows a better spatial resolution ability than that of the CR system.

Furthermore, irradiation of treatment time of 0.02 minutes was delivered on the constructed phantom, with which the two detectors were used to acquire images. At this treatment time, image of the phantom was visible on the CR digital detector while no image of the phantom was seen on the film detector. This result clearly shows that at lower dose, anatomical land marks images of the phantom can be clearly displayed on the CR detector. Also with the CR, images acquired becomes darker as the irradiation time start increasing from 0.2 min and above, indicating that quality images are acquired with low doses whilst with the film good images are acquired when the treatment time is high.

The research indicated that patient going through radiotherapy treatment using conventional portal film for imaging are been induced with more dose than using CR system for portal imaging.

This clearly shows that using CR to acquire portal images would allow the radiation oncologist to take advantage of computerized image-processing techniques and Picture Archival Communication (PAC) technology with minimal expenditure of time and money.

5.2 RECOMMENDATION

At the Korle-Bu Teaching Hospital were this research was undertaken, conventional film for portal imaging is still in use for patient setup verification. From the results of this
study, it is recommended that the use of CR system be incorporated into the radiotherapy department of the Korle-Bu Teaching Hospital for portal imaging in the verification of patient setup for treatment. The use of CR system for patient verification at the therapy department would help to minimize patient dose during setup verification. Since the hospital is also planning to go digital, the use of CR at the radiotherapy department would help to improve the quality of patient treatment.

Also for further research, it would be better if the findings of this work could be applied to some of the patients visiting the center for treatment.

It would be better also for policy makers, regulators of ionizing radiation to look into these recommendations seriously in other to ensure the safety of patient’s undertaken treatment at the center.
REFERENCES


Desai et al. (2010). Practical Evaluation of Image Quality in Computed


imaging. 2nd ed, (August), 1996.


