Chapter 1

INTRODUCTION

Radiotherapies were discovered in 1895 about a century ago, by Wilhelm Roentgen\(^1\). One year after the discovery, the first attempt to use them for treatment was made. The first documented cure by application of X-rays dates from 1899. Since then, increases in cure rate of patients with (mostly) malignant diseases treated with radiotherapy alone or in combination with surgery and/or chemotherapy can be attributed to the improvements in radiotherapy treatment techniques. A milestone has been the introduction of cobalt-60 radiation and megaelectronvolt radiation in the middle of the previous century.

Radiotherapy is defined as the treatment of malignant (and occasionally non-malignant) diseases by ionizing radiation. The first known reference to cancer or cancer like disease in humans was documented 4500 to 5300 years back. The term cancer, neoplasm, tumor, and malignancy are usually used interchangeably. Radiation used can be either electromagnetic (x-rays, gamma rays) or corpuscular (electrons, protons, neutrons, alpha particles etc). Approximately 40% of people with cancer have radiotherapy as part of their treatment.\(^2\) In radiotherapy, the concern is the transfer of energy to the body by ionizing radiation. The energy transfer takes place depending on the type of radiation beam in use. For a photon beam in use, it could be due to Photoelectric effect, Compton effect or Pair Production depending on the energy of the incident beam. Since the energy used in radiotherapy is in the megaelectronvolt range of 4 to 20 MeV, Compton effect predominates photoelectric effect\(^3\). For an incident charged particle beam such as electron, inelastic collisions with atomic electrons results in
ionization and excitation of atoms thereby transferring energy whereas elastic collisions with atomic nuclei results in no energy transfer. The radiation effect is produced by energy absorbed in living tissue through the process of ionization and excitation of the atoms and molecules that form these tissues. Thus, radiation energy delivered is the “medicine” that is administered and the “radiation dose” is the amount of energy absorbed by tissues. In practical radiotherapy, the radiation dose needs to be controlled with an accuracy of a few percent. Radiobiological and clinical evidence indicates that the dose-effect relations for tumor control are steep\(^4\). For some tumors, a dose variation of a few percent can significantly modify the observed local control rate. The dose-effect relations are even steeper for normal tissue complications. For these two reasons, accurate evaluation of dose from ionizing radiation (dosimetry) is essential. In 1976, the International Commission on Radiation Units and Measurements (ICRU) made the following recommendations: "the available evidence for certain types of tumors points to the need for an accuracy of $\pm$ 5% in the delivery of an absorbed dose in a target volume if the eradication of primary tumor is sought”\(^5-8\). With advancements in technology, Mijnheer et al\(^9\), recommended an accuracy in absorbed dose delivery of 3.5% (one standard deviation, for the dose at the specification point, for radical treatment). This need to accomplish such a high level of accuracy in the amount of “medicine” (radiation dose) delivered, distinguishes radiation therapy from other disciplines of clinical care\(^7-11\).

Radiotherapy departments are equipped with specially designed, radiation producing machines. Aim of the radiotherapy is to deliver a preplanned radiation dose within acceptable margins of uncertainty to the desired site in the patient body. It is inevitable that some radiation dose will be delivered to the body that is not the intended main target of treatment. However the method of irradiation is optimized so that damage to such unintentionally irradiated tissue is minimal\(^15\). The production of radiation, the
technique of tumour localization, the planning of the irradiation, the implementation of the treatment and the estimation of the radiation doses received by different sites in the body are based on principles of physics.

Cancer is one of the most important health care problems faced all over the world and on an average more than half of all cancer patients are treated with radiation therapy. This mode of treatment uses complex technology that involves megaelectronvolt radiation which, if not handled with the greatest of care, could lead to significant patient treatment errors and exposures of patient and the support staff. Recent years have seen a rapid development in radiation therapy technology. One of the prime factors contributing to this rapid development has been the evolution of computer technology and its applications in (i) patient diagnosis, (ii) radiation treatment planning using Treatment Planning Systems (TPSs) which are capable of using data from diagnostic imagers and (iii) radiation dose delivery using relatively simple Cobalt-60 machines or complex Linear Accelerators (LA). These radiation delivery machines now come equipped with Multileaf Collimators (MLCs) for field shaping and modulation of the intensity of radiation output from the machine. The radiation treatment process involves the application of some or all of these technologies to provide the desired dose to the target volume while minimizing exposure to adjacent normal tissues.

Radiation treatment delivery procedure

To understand the use of radiation for treatment, it's necessary to have an insight into the various stages the patient is taken through before radiation is administrated. Once a diagnosis of malignancy is confirmed, the treatment protocol is decided upon. In case, radiation forms part of the protocol, it can be delivered in two ways – (i) Brachytherapy and (ii) Teletherapy.
Brachytherapy (sometimes referred to as curietherapy or endocurie therapy) is a term used to describe the short distance treatment of cancer with radiation from small, encapsulated radionuclide sources. This type of treatment is given by placing sources directly into or near the volume to be treated. The dose is then delivered continuously, either over a short period of time (temporary implants) or over the lifetime of the source till a complete decay of the source occurs (permanent implants). Most common brachytherapy sources emit photons; however, in a few specialized situations beta or neutron emitting sources are also used.

There are two main types of brachytherapy treatment – (i) Intracavitary, in which the sources are placed in body cavities close to the tumour volume and (ii) Interstitial, in which the sources are implanted within the tumour volume.

Intracavitary treatments are always temporary, of short duration, while interstitial treatments may be temporary or permanent. Temporary implants are inserted using either manual or remote afterloading procedures. Other, less common forms of brachytherapy treatments include surface plaque, intraluminal, intraoperative and intravascular source applications; for these treatments either beta or r emitting sources are used. The physical advantage of brachytherapy treatments compared with external beam radiotherapy (Teletherapy) is the improved localized delivery of dose to the target volume of interest. The disadvantage is that brachytherapy can only be used in cases in which the tumour is well localized and relatively small. In a typical radiotherapy centre about 10–20% of all radiotherapy patients are treated with brachytherapy.

In external beam radiotherapy the radiation source is at a certain distance from the patient and the target within the patient is irradiated with an external radiation beam. Most external beam radiotherapy is carried out with photon beams, some with electron beams and a very small fraction with more exotic particles such as protons, heavier ions or neutrons.
Figure 1.1 illustrates the different steps that have to be taken successively in the radiotherapy process (ICRU 1993). The first step involves the confirmation of the presence of a malignancy by review of the histological diagnosis. Further investigations are carried on to define the site and extent of the tumour and its stage according to a recognized staging classification (e.g. TNM, FIGO, AJCCS). This assessment will include clinical examination (e.g. under anaesthetic for cervical, bladder and some other tumors) and imaging by various methods. Following this, final decision to use radiotherapy will be made and the treatment prescription is given, which includes a statement of the aim of the therapy, the definition of volume to be treated, the specification of doses, fractionation and other treatment parameters. Provision must be made for modification of this prescription during treatment planning if necessary. The next step is treatment preparation which involves consideration of the immobilization of the patient (and where possible the tumour with its host organ) and acquisition of anatomical and tumour data for dose planning with the patient in the radiotherapy treatment position. The position chosen for treatment should be comfortable for the patient and easily reproducible. It may be necessary to consider the requirements for computed tomography (CT) imaging using non-radio opaque materials. The same fixation devices (e.g. chest wedges, arm poles, lasers) should be available for localization (whether on the simulator, CT simulator or CT scanner) and for treatment.

Simulator guided radiotherapy is one in which the patient with immobilization is taken on a simulator machine (a machine which stimulates a radiation treatment machine except for the radiation therapy source) wherein the radiation portals are identified. Verification of the portals is done with image intensifier or a X-ray film. These setup parameters are recorded and used during radiation dose delivery. Simulator guided radiotherapy can eliminate exposure to about 30% of normal tissue from the treated volume.
Figure 1.1. Steps in radiotherapy procedure (adapted from ICRU50, 1993)
In a more complex approach, CT scan of the patient is taken with the patient immobilized in the treatment position. The process of determining the volumes for treatment of a malignant disease consists of several distinct steps which have been well described in ICRU Report 50\textsuperscript{20}. This provides clearly defined and unambiguous concepts to ensure a common language between different centers across the world. The gross tumour volume (GTV) is the palpable or visible extent of the malignant tumour and usually corresponds to the site of the cancer where the tumour cell concentration is at its maximum. A margin is then added around the GTV to include direct local sub-clinical microscopic spread. This margin usually has a decreasing malignant cell density towards the periphery where it should reach zero which along with the GTV constitutes the clinical target volume (CTV). If the tumor has been removed prior to radiotherapy, then no GTV can be defined and the volume of sub-clinical disease constitutes a CTV. Following definition of these volumes a margin has to be added around the CTV to account for variation in size and position of tissues relative to the treatment beams which may be due to patient movement, organ movement and variation in daily set-up. This volume is known as the Planning Target Volume (PTV). The various volumes that center around the tumor are marked on CT, MRI, PET or any other imaging technique which is best suitable for that site and tissue of interest. If the localization is done on an image other than CT, special software is used to transfer these to the corresponding region on the CT image taken (image fusion). As part of the imaging process, several reference marks should be placed on the patient. This can be done before imaging by placing radio opaque markers that will show up on the images. These can be used as reference points to the isocenter during planning process. Following delineation of the target volumes, beam arrangements are selected and the dose distribution is computed. Path of the virtual beams aimed on to the delineated tissues and the radiation dose distribution pattern is checked for their acceptance. This will include a choice of beam direction and a choice of collimation (divergent
blocks, asymmetric jaws or MLC’s). The optimum beam arrangement that will provide adequate coverage of the malignant tissues while minimizing the dose to critical normal tissues is to be selected. The dose distribution is then evaluated. Review of the dose distribution will confirm whether the PTV is covered adequately and dosage to normal tissue is being limited to acceptable levels. In addition to this, tools such as Dose Volume Histogram (DVH) can be used to analyze the treatment plan. Having the dose distribution, the physicist can decide on its adequacy and determine whether further addition of beams or modification of beam direction, weighting or shaping are required to improve the treatment plan. Using the iterative process explained above, an optimized radiation treatment plan is developed. This process is repeated until a satisfactory dose distribution is achieved. A three-dimensional dose distribution can be obtained based on localization of the target volume, normal organs and body contour at multiple levels. The beam arrangement for the final dose distribution should then be verified. The monitor units or treatment time is calculated and the patient treatment chart is prepared. Depending on the equipment available, treatment plan may be confirmed using a simulator. The next phase involves treatment delivery where the preplanned immobilization and positioning of the patient is carried out and the parameters are set up according to the treatment plan and the first session of irradiation given. Digital Reconstructed Radiographs (DRR) can be generated from the TPS that can then be used as reference images to compare with megaelectronvolt images obtained during treatment (portal images). Verification should be carried out using portal films and in-vivo dosimetry to check the geometry and dose respectively, during treatment.

The advent of high-energy linear accelerators has given a choice between photon and electron beams to be selected for external beam radiotherapy. Electrons are viable option in treating superficial tumors up to a depth of about 5cm. Electron depth dose characteristics are unique in that
they produce a high skin dose but exhibit a fall off after only a few centimeters. Electron absorption in human tissue is greatly influenced by the presence of air cavities and bone. The dose to underlying tissue is increased when the electron beam passes through an air cavity and is reduced when the beam passes through bone.

External beam therapy, since its introduction, has passed through several stages where in it has incorporated developing technologies to achieve the goal of sparing normal tissue while treating the tumor cells. Soon after the discovery of X-rays, planar radiographs of the human body having the bony landmarks, guided the radiation beams delivered using collimated rectangular fields. Additional blocks were placed daily to match marks on the patient’s skin to treat the two-dimensional projections of the tumor volumes. However, human anatomy and tumor shapes are inherently three-dimensional. By treating a large amount of near by normal tissue, the dose delivered was limited by the tolerance of the normal tissue being irradiated. Additionally, it is not possible to take the three-dimensional structures into consideration because of the limitations of early dose calculations. The reduction in normal tissue irradiation should theoretically improve the therapeutic ratio and allow the tumor target volume to be treated to a higher dose, thereby improving the probability of tumor control.

Three-Dimensional Conformal Radiation Therapy\textsuperscript{21} (3DCRT) is a method of irradiating target volume defined with a set of x-ray beams individually shaped to conform to the two-dimensional Beam’s Eye View (BEV) projection of the target. 3D CRT became feasible with the development of Computed Tomography (CT). The reconstructed images, acquired with patients in the treatment position, provide a model on which geometric and dosimetric computations can be applied. Adequate immobilization devices are provided to the patients to hold their treatment position during imaging and treatment. The development of the Digital
Imaging and Communication in Medicine (DICOM) standard and its various extensions for data exchange has made possible the transfer of image data and treatment plan data across systems over a network.

Intensity-modulated radiation therapy (IMRT) emerged as an advanced version of 3D CRT. The introduction of Multi Leaf Collimator (MLC) (the conventional collimator block replaced by a number of leaves that can be moved independently to produce irregular fields) on linear accelerators in the mid 1990’s lead to the development of different IMRT delivery techniques. Traditional radiation treatment techniques, including three dimensional conformal radiation therapy (3DCRT), do not provide a method for sparing critical structures that push into and are partially or fully surrounded by a target or combination of targets. Intensity-modulated radiation therapy (IMRT) does provide the ability to spare normal tissues that are surrounded by targets with concave surfaces, and this advantage is being exploited to increase tumor dose. Such truly three-dimensionally conformal dose distributions are possible as a result of developments of inverse algorithms for planning, and treatment with accelerators capable of delivering intensity modulated beams using MLC’s.

There are many ways to produce an intensity modulated beam. Physical compensators (high Z material of varying thickness) placed in the beam path are the most straightforward one. But the most popular delivery technique is, however, based on computer-controlled MLC. In this approach, an intensity map is decomposed into a set of MLC-formed apertures. This is the common feature of any IMRT technique, i.e to control the three-dimensional dose distribution through superposition of a large number of independent segmented fields. Depending on the relationship between MLC leaf movements and radiation dose delivery, delivery can generally be divided into (i) step-and-shoot delivery and (ii) dynamic modes. The former is the simplest computer-controlled delivery scheme in which MLC leaf movements
and dose deliveries are done at different instances. In dynamic delivery, leaf movement and dose delivery are done simultaneously.

IMRT planning requires the calculation of a set of parameters for the optimum delivery of a radiation dose to the patient. Although manual forward planning may be possible in some simple cases, computer optimization of the beam parameters is almost always used for IMRT treatment planning because of the large size of permutations and combinations of beam parameters required to arrive at an optimum plan. This is achieved using an inverse treatment planning technique, which derives the optimal beam parameters by starting from a prescribed or desired dose distribution.

Inverse planning uses a computer optimization algorithm to determine the optimal beam parameters that lead to a solution as close as possible to the desired output. Inverse problems can be described as problems in which the output or consequences are known but not the cause. The difference between various treatment planning systems lies in the specifications of the input and output parameters and the criteria used to select the final solution. Specific to Radiation Therapy (RT), the output is generally specified by a desired dose distribution, a set of desired Dose Volume Histogram (DVH), or even the Tumor Control Probability (TCP) and Normal Tissue Complication Probability (NTCP) for the involved structures.

The input parameters to be optimized depend on the delivery scheme. Typically, the number of beams and their incident directions are determined empirically before dose optimization. Each incident beam is discretized into beamlets that form a radiation intensity map. The task of inverse planning is then to optimize the beamlets for their relative weights to create the desired radiation intensity map. These types of optimizations help to conform the radiation dose to irregularly shaped tumor volumes.
Image Guided Radiotherapy (IGRT) has been put into practice as a tool to verify the precision of radiotherapy treatment delivery. This works by integrating an imaging device to the linear accelerator which will help to verify the position of the tumor in relation to the reference image. Necessary corrections to patient positioning can be computed and applied so that the planned treatment is delivered to the tumor region during the entire course of treatment.

Another special technique to deliver high radiation dose in one fraction to small regions is Stereotactic Radio Surgery (SRS). Specially designed collimators are attached to a linear accelerator which deliver a high dose of radiation using multiple arcs to a small volume, usually below 3 cm in diameter (X-Knife). Another delivery method is using Cobalt-60 sources placed around the circumference of a sphere, focused to its center (Gamma Knife).

![Figure 1.2. Comparison of the percentage of the absorbed dose (in water) for different radiotherapy sources. 192-Ir gamma source, 20 MeV electrons, 8 MeV X-Ray beam, p(66)/Be neutrons, monochromatic 200 MeV proton beams (Bragg curve), Spread Out Bragg Peak (SOBP) for a proton spectrum.](image_url)
Proton radiation beams have also been used to deliver curative radiation doses. They are delivered to the tumor in the same manner as photons and electrons. The dose deposited by protons remains relatively constant as they travel through the normal tissues proximal to the target. The kinetic energy of the protons is transferred to the tumors by electrons knocked out of atoms. These electrons ionize DNA, and their biological effectiveness resembles that of mega-electronvolt photons. At the end of the path (range of protons), biological effectiveness increases sharply as the protons slow down and eventually stop producing a Bragg peak. Since the width of the Bragg peak is small and not sufficient enough to match the tumor size, proton beams of varying energy are been used in practice.

The dose absorbed depends on the type of radiation, energy of the beam and the technique used for dose delivery. An accurate measurement of absorbed dose is critical in deciding the outcome of radiation treatment. The deposition of energy in tissues results in damage to DNA and diminishes or eradicates the cell’s ability to replicate indefinitely. Some of the important quantities that need to be understood about a radiation beam, for successful treatment are - the absorbed dose at any point in the beam, intensity pattern of the incident beam at a desired depth and the Percentage of Dose at a Depth (PDD)\textsuperscript{23}.

The absorbed dose at a point represents the amount of energy deposited by a radiation beam to that point. It is measured in joule per kilogram (J/Kg) and its unit is Gray (Gy). The transverse beam profile or fluence map reveals the intensity pattern of radiation that is beamed from the machine or the intensity pattern that has been modified using beam modifying devices kept in the path of the radiation beam. Percentage depth dose is the dose absorbed by tissues at various depths due to radiation interactions as it passes through a medium.
Before a radiation beam is put into clinical use, the above radiation quantities related to an incident beam need to be fully understood. Understanding these quantities requires methods and materials that are accurate, reproducible and in accordance to one of the established protocols (TRS\textsuperscript{24}, TG\textsuperscript{25}, DIN\textsuperscript{26}). These protocols recommend certain devices and procedures for the accurate determination of radiation quantities. These procedures are generally termed as radiation dosimetry and the devices used are called radiation dosimeters. Radiation dosimeters can be defined as devices that show some form of response to incident radiation which can be quantified into a meaningful result.

To evaluate a radiation beam and its properties using a dosimeter, it needs to be positioned against the beam in a reproducible manner and in a medium that could best simulate the environment in which the radiation beam is to be used. Since the intention here is to use a radiation beam to treat patients, the environment that is required is the one which closely simulate body tissue. This substitute medium is called a ‘phantom’. From the early days of radiotherapy, physicist were keen to find a suitable phantom to evaluate radiation properties. First literatures came out with publications of using wax and water as a substitute medium. This was followed by many experiments on various substances and compounds to arrive at an ideal medium as substitute. It was soon realized that no medium or model could cater to all the modalities and requirements. This paved the way for construction of dedicated phantoms for use in radiotherapy dosimetry, imaging and radiation protection.

Literature review

A survey of the published literature shows that there have been numerous tissue substitute phantoms and mathematical models developed that were aimed at doing specific tasks. Ricardo Ochoa et al\textsuperscript{27}
designed a phantom for the quality control of high dose rate Iridium-192 source used in brachytherapy. The phantom was proposed for measuring the strength of Iridium-192 high dose rate sources using TLDs and for verification of the dose calculated by the treatment planning system. Their study could evaluate the accuracy of brachytherapy practice in various radiotherapy departments of Brazil.

Kim et al²⁸, designed a Phantom to simulate a typical Korean male for Radiation Protection studies. The phantom was composed of three tissue equivalent materials: epoxy resin, urethane foam and polyurethane representing bone, lungs and soft tissues, respectively. Here again absorbed doses measured using TLD showed good agreement within 7% of those calculated.

Ibbott et al²⁹ developed an anthropomorphic head and neck phantom for evaluation of intensity modulated radiation therapy. Ibbott et al modified an anthropomorphic head and neck plastic shell to fill water into it. Provision was made to hold ion chamber, TLD and Radiochromic film. This phantom was used to evaluate IMRT plans in 10 hospitals.

McNiven et al³⁰ designed a phantom to analyze TPS's beam display capabilities for irregular fields produced by MLCs. Andrea McNiven et al reported a maximum deviation of 0.7mm from the design. This work was able to analyze the non-dosimetric aspects of TPS.

Lei Dong³¹ designed and fabricated an arc shaped water filled phantom, with an ion chamber, to measure the integral dose for patient-specific quality assurance. This study demonstrated that by deriving a transfer factor (TF) with reference to calibration water phantom, dose in phantom can be measured using an ionisation chamber.

Dee-Ann Radford et al³² designed an Anthropomorphic Intensity Modulated Radiation Therapy Quality Assurance Phantom for the purpose of reviewing IMRT treatment modalities at institutions participating in NCI cooperative clinical trials. The phantom was designed to be water filled with TLD and radiochromic film inserts. It also had inhomogeneities representing
Femur Head, Bladder, Prostate and Rectum. Their study showed that having the provision to use multiple detectors simultaneously could deliver information with a single exposure rather than repeating the measurement with different detectors.

Seaby\textsuperscript{33} et al designed a multi-block phantom for radiotherapy dosimetry applications using epoxy resin. This phantom assembled using variety of building blocks of varying shapes can be used to verify TPS. Blocks in the phantom can be modified to hold any detector of choice.

Swinnen et al\textsuperscript{34} used a multipurpose phantom made of polystyrene for mailed dosimetry checks of therapeutic photon beams: ‘OPERA’ (operational phantom for external radiotherapy audit). This study was conducted to check five irradiation conditions: square fields, asymmetrical fields, wedged beams, oblique incidence and influence of inhomogeneities in the field. The absorbed dose on the central beam axis was measured with TLDs for the first three irradiation conditions and the relative dose distributions were verified with film. Ans Swinnen et al concluded that the ‘OPERA’ phantom can be useful for on- and off-axis verification of the TPS.

Carlos Eduardo de Almeida\textsuperscript{35} designed an anthropomorphic phantom for quality assurance and training in gynecological brachytherapy. The water filled phantom made of PMMA has three PMMA inserts designed to hold a Farmer type ionization chamber of 0.6 cm\textsuperscript{3} positioned at the points to represent the bladder, rectum and point A. Their measured dose was in agreement with the Monte Carlo calculated values. Carlos Eduardo de Almeida concluded that the phantom is suitable for use for the acceptance tests of treatment planning systems and applicators, as educational tool, for dosimetric research problems and for the QA of brachytherapy sources.

Waligo´rski et al\textsuperscript{36} studied on Validation of a Radiotherapy Treatment Planning System using an Anthropomorphic Phantom and Thermoluminescent Detectors. A treatment planning system (TPS) was validated in conditions of simulated radiotherapy (RT) on an anthropomorphic tissue-equivalent phantom. Dose to pre planned point
inside the phantom was calculated using TLDs placed at these points. The study showed that calculated and measured doses ranged between 1.3% and 2.2%.

Jake Van Dyk\cite{37} developed an oval shape acrylic phantom, 20 cm high, 30 cm wide, 12 cm long with 3 openings for Cylindrical Inserts, each opening is 8 cm diameter by 12 cm long. Electron density extension, 12 cm diameter with 5 openings of 2.5 cm diameter. It was designed to accommodate an ion chamber and film for dosimetry. Their study shows that dose computation by a TPS can be evaluated using a phantom containing heterogeneity inserts and different dosimeters.

Cherry et al\cite{38} designed a thorax phantom to evaluate field localization and absolute dose delivery in 3DCRT treatments. The phantom had water-fillable shells that enclose a bone equivalent spinal column, two lung equivalent structures, and a target within the left lung. Doses were evaluated by using treatment inserts containing TLD and radiochromic film dosimeters.

Paliwal et al\cite{39} designed and developed a spiral phantom for IMRT and tomotherapy treatment delivery verification. The phantom made of tissue equivalent material encloses a EDR2 film strip in a spiral fashion. Chamber specific adapters were also developed to do absolute measurements.

Kalef-Ezra J. et al\cite{40} designed and tested a phantom for dosimetric characterization of small radiation fields. The phantom holds 426 TLD chips at 1 mm spacing to measure small circular fields. This study showed that TLD’s could be effectively used to evaluate small fields as used in procedures like SRS and SRT.

Giraud et al\cite{41} designed and developed a phantom made of Plexiglass simulating pelvic anatomy and containing Fricke gel dosimetry. The study concluded that chemical dosimeters can be used to verify complex dose distributions generated by TPS.

Daniel et al\cite{42} modified a commercially available polystyrene phantom to include TLD chips in addition to film and ionization chamber. They measured dose profile of IMRT field at 1 cm interval and it was compared
with the dose profile obtained from TPS. Their study showed that ionisation chamber can be used for point dose measurements at low dose gradient regions and TLD’s gives an profile of radiation dose in an area.

Bhudatt et al\textsuperscript{43} designed and developed solid water pelvic and prostate phantom for imaging, volume rendering, treatment planning, and dosimetry applications and showed that the phantom can evaluate the accuracy and consistency of treatments delivered by institutions participating in national collaborative clinical trials involving 3-D conformal dose escalation.

Reichel\textsuperscript{44} developed an anthropomorphic phantom containing head, chest and abdomen. Organs were represented by shells which can be filled with water or other mixtures and it included a skeleton. A similar phantom was developed by Conway et al\textsuperscript{45}. Their design was a patient equivalent attenuation phantom to evaluate doses during radiological studies. These phantoms were lightweight, transportable, rugged and made from readily available materials (acrylic and aluminium alloy).

Bradley et al\textsuperscript{46} developed a phantom based on SMR(L) [Standard Malaysian Rubber] grade natural rubber and a formulation used for the proprietary rubber phantom-material, Temex. Bradley et al studies on the central axis dose in phantom and its comparison to water values were within 2%. The study concluded that this favorable measured response characteristics combined with the ease of processing and casting the phantom material provide the basis for useful radiotherapy machine calibration and anthropomorphic dosimetry measurements.

Constantinou\textsuperscript{47} designed, manufactured and tested an epoxy resin-based solid substitute for water. His studies showed that this solid water has radiation characteristics very close volumetrically to those of water. When used as a dosimetry phantom for X and r-ray beams in the radiotherapy range, phantom-to-water corrections and density corrections are eliminated. Relative transmission measurements have shown that the transmission through 10 cm of solid water is within 0.2% of that through an
equal thickness of water for X and r-ray. His study shows that the use of epoxy resin based solid substitute material for calibration phantoms achieve the goal of radiotherapy beam calibrations within ±1.0%.

All the above studies were on the design of various types of phantoms with the objective of evaluating a specific therapy machine or procedure. All the above phantoms were made of tissue equivalent substitutes either of solid nature or a few designs were water is filled into it. There were also a few studies on the properties of solid tissue equivalent materials and other alternative approaches to doing an evaluation.

McEwen et al\textsuperscript{48} (2003) studied phantom made of epoxy resin for its properties with electron beams. Their study showed the advantages of using solid phantoms to that of water phantoms. Proper scaling factors were used to correct for the change in fluence due to measurement in phantom medium. McEwen et al concluded that the output can be measured with an uncertainty of 0.12\% for electron beams using epoxy resin phantom. McEwen et al\textsuperscript{49} (2006) also studied on the characteristics of Virtual Water to Water and reported that by having a full understanding of the properties of water substitutes, they can replace water to do absolute measurements. This literature also shows that by use of proper scaling factors good agreement to $<0.2\%$ can be achieved in absolute dose measurement between virtual water and water.

Susanna et al\textsuperscript{50} (2006) studied on Mathematical Phantoms developed by describing size and form of the body and its organs by mathematical expressions. The study also described ‘Voxel Phantoms’ were phantoms based on digital images recorded from scanning real people are used. Susanna et al states that these methods will be the future to individualized dosimetry which avoids all approximations and assumptions in the present day dosimetry phantoms.
Casar$^{51}$ et al did evaluation of water equivalency of Plastic Water$^{TM}$ for high-energy electron beams using IAEA TRS-398 Code of Practice for energies upto 10 MeV. He concluded that the use of water equivalent phantom for electron dosimetry could give results within 1% by applying proper scaling factor.

Venselaar et al$^{52}$ studied on the tolerances for the accuracy of photon beam dose calculations of treatment planning systems. Venselaar et al concluded that though the general aim must be to have good agreement between dose calculation and the actual dose value, e.g. within 2% or 2 mm, current day algorithms and their implementation into commercial treatment planning systems result often in larger deviations. A high accuracy, at present can only be achieved in relatively simple cases. Their study reports that the new set of tolerances and the quantity confidence limit as laid down by AAPM Task group 23 is a proven tool for the acceptance of photon beam dose calculation algorithms of treatment planning systems.

Banjade et al$^{53}$ (2001) did a study of Rhizophora wood phantom for dosimetric purposes using high-energy photon and electron beams. Measurements of percentage depth-dose were made for photons of 6MeV and 5MeV, 12 MeV electron beams. For the 6 MeV photon and 5 MeV electron beams, discrepancies between percentage depth-dose for Rhizophora spp and water, at all depths, are found to be within 2.6 and 2.4% respectively. At 12 MeV electron energies, measured percentage depth-doses in Rhizophora beyond 3.5 cm depth are found to be in significant discord with those for water. The absorbed dose in water measured in Rhizophora at dmax for all three beams produces discrepancies of no more than 1.1% when compared with measurements made in water.

Kron$^{54}$ studied on Applications Of Thermoluminescence Dosimetry In Medicine. He concluded that Thermoluminescence dosimetry (TLD) features many advantages such as small detector size and close tissue equivalence that make it useful in medicine. In radiotherapy, the fact that no cables are required during the measurement allows the use of TLDs inside tissue-
equivalent phantoms to verify radiation doses delivered in treatment techniques.

Traub et al\textsuperscript{55} studied the photon backscatter factor for several irradiation phantoms by calculation and experimental methods and compared it to phantom constructed by International Commission on Radiation Units and Measurements (ICRU). Their calculations and measurements agreed to within uncertainties and demonstrated that the backscatter factors over the magaelectronvolt energy range for a water-filled phantom recommended by the International Organization for Standardization (ISO) and a phantom constructed of tissue-equivalent (RS-1) plastic are nearly the same as that of the ICRU tissue reference phantom. However, the backscatter from a polymethylmethacrylate (PMMA) phantom was up to about 8\% higher. In addition, it was found that a composite phantom of polytetrafluoroethylene (PTFE) and PMMA also produced a backscatter factor quite similar to that of the ICRU tissue reference phantom.

Bohm\textsuperscript{56} studied on the suitability of two polyhedron phantoms for type testing and calibrating individual dosimeters. Their study concluded that an icosahedron phantom with an edge length of 18.64 cm is preferable. It combines the advantage of sphere phantom with those of a cube or slab phantom and allows a no of angles of radiation incidence to be used for a fixed radiation field.

Grosswendt\textsuperscript{57} derived correction coefficients for doing absolute dosimetry in tissue equivalent medium using Monte Carlo methods. In this study calculations were performed for monoenergetic photons in the energy range between 2kev and 1 MeV. The results of the study could be used in calibrating individual dosimeters in terms of dose equivalent quantities in phantoms.

Further to the studies conducted on design and fabrication of various phantoms and tissue substitute materials, there were a few published reports by independent agencies laying down guidelines to the
construction of phantoms and quality assurance tests to be performed on radiotherapy machines and procedures.

ICRU report 44\(^{58}\) on ‘Phantoms and Computational Models in Radiation Therapy, Diagnosis and Protection’ defines different types of phantoms, specific purpose of each one of them, their construction criteria, material selection and tolerance values. ICRU committee reviewed sets of operational radiation dose quantities based on the data collected using anthropomorphic phantoms under various conditions of irradiation.

ICRU report 47\(^{59}\) defined phantom related operational quantities. It recommended 30 cm x 30 cm x 15 cm PMMA phantom which it found to be equivalent to ICRU sphere phantom. Moreover its backscatter properties were close to those of human trunk for photon and neutron irradiation.

Technical Reports Series No.430\(^{60}\) described the commissioning and quality assurance (QA) procedures that should be used with modern TPSs. This report was published after IAEA’s analysis of the radiotherapy accidents involving TPS.

ESTRO\(^{61}\) proposed guidelines to brachytherapy quality assurance, brachytherapy procedures and treatment planning. This report was published as part of the project by ESTRO to device quality assurance procedures for radiotherapy.

IPEM\(^{62}\) provide a reference text to cover quality control procedures that may be used as part of a quality assurance programme in Radiotherapy. The recommendations are based on the results of the survey carried out by the Radiotherapy Topic Group in 1992.

**Conclusion to review of literature and motivation to the present work**

Review of literature for various types of tissue substitute phantoms shows that they can be categorized to (i) phantoms simulating internal body tissue, (ii) phantoms dedicated for evaluating a particular technique of treatment or procedure, (iii) phantoms that are homogenous tissue substitutes designed
for particular radiation detectors, (iv) a generalized phantom which is designed to cater to the objectives with which it was designed. Majority of these phantoms were fabricated using tissue equivalent solids and some with the provision to fill water in its shells. One common observation on the above studies is that phantoms were designed and developed with a specific objective. This limits the use of these phantoms as a general-purpose quality assurance tool in a radiotherapy department. Furthermore the phantom designs in the above studies are inherently tied to a particular type of detector. This limits the possibility of integrating them with any newer detectors being developed. This essentially curtails usage of the phantom along with more modern technologies.

Generalized body phantoms are not recommended for dosimetry of an actual patient. This is because no single phantom can simulate all anatomical areas that are treated by radiation, and no single phantom structure can be an ideal tool to evaluate plans for different anatomical sites. Body phantom when used in radiotherapy dosimetry, can make for difficulties because anatomical variations can cause substantial differences in absorbed doses received by different individuals undergoing identical radiation treatment procedures. Body phantoms range from simple geometry (stacked sheets of tissue substitutes, solid homogenous cubes and cylinders) to complex anthropomorphic phantom having high degree of external and internal realism. During dosimetry, selection and design of the phantom is as important as detector selection. While an anthropomorphic phantom provides realistic looks, they have some significant drawbacks relative to geometric regular phantoms. Anthropomorphic phantoms are often difficult to align to the beam, increasing the spatial uncertainty of measurement. The preparation of film is also more difficult and the presence of internal heterogeneities may make the source of difference between measurement and calculation difficult. Hence standard phantoms are common for dosimetric evaluation in radiotherapy. The standard phantom, which makes use of water or water substitute, finds use in radiotherapy.
depending on the dosimeter to be used and the measurement setup that is required. The selection of dosimeter again depends on the property that needs to be evaluated. In most situations no single dosimetric system will be able to give a full evaluation of the radiation beam under study. Though water is the ideal substitute for tissue, practical problems in using water has forced physicists to think for water equivalent mediums. Compared to solid water equivalent substances, water has its drawbacks. It is difficult to position detectors in water and many a time reproducibility of position is compromised. It is also difficult to contain it to desired orientations. There was keen interest shown to the idea of multipurpose phantoms that could be adapted depending on the quantity to be measured in ICRU discussions on phantom designs\textsuperscript{64}.

\textit{Considering all the literature surveyed on ‘The Tissue Substitute Phantoms Used in Radiotherapy’, we have decided to design and fabricate a tissue equivalent phantom that would serve as a multi utility tool in a radiotherapy department. The design should make the phantom feasible to be used on both Teletherapy machines and Brachytherapy machines. The phantom should be fabricated using a material that is recognized as a tissue equivalent material and whose properties are well documented. In contrary to the existing phantoms, this new design should be evolutionary and be able to address to new developments in treatment procedures. The new phantom should also be capable of accommodating newer detectors that are developed. The over all design should adhere to standards laid down by ICRU 44 and ICRU 48\textsuperscript{65}. All safety requirements on the choice of material and structure need to be satisfied in the new design. The choice of material also should be such that it can be fabricated conforming to the design with acceptable tolerances.}

\textbf{Plan of work}

As a foundation for the new tissue equivalent phantom design, a survey was conducted among a selected group of physicists working in the
field of radiation physics in various radiotherapy centers. The survey collected individualized data on the requirements of a quality assurance tool in a radiotherapy centre. Accordingly a list of objectives was formulated which formed the design guidelines. Also a study was conducted for the choice of radiation dosimetric systems that will help to realize the objectives. Sketches of the phantom parts were done and then using Autocad software machine drawings prepared. Once a suitable material for fabrication of the drawing was finalized, the parts were fabricated within the tolerance levels. The fabricated parts were tested for their adherence to design specifications. Quality assurance was done on parts of the phantom that were designed with specific objectives to check if they function satisfactorily. Finally the fabricated phantom was subjected to functionality tests in a radiotherapy department.

The entire work of design and fabrication of dosimetric phantom for radiotherapy and experimental evaluation is described following five chapters.

Following the introductory chapter, Chapter II provides an insight into the importance of quality assurance in radiotherapy, dosimetry systems used in radiotherapy and role of phantoms in quality assurance. It also looks into the various specifications recommended for the design and fabrication of a tissue equivalent phantom.

The selection of suitable tissue substitute material and design of various parts of the phantom are explained in chapter III.

Chapter IV looks at the fabrication of the phantom parts and the quality assurance done to verify the functionality of the parts.

Chapter V contains experimental setups and their results using the fabricated phantom.

Chapter VI gives a summary of the present study and the future scope of the work.
References:


2  Cancer research UK. Available at: http://www.cancerhelp.org.uk/help/default.asp.


19 Treatment Simulators. Br J Radiology. 1989; Supplement 23


38 Cherry, C P D. Followill, D S, Hanson, W F. Design of a heterogeneous thorax phantom for remote verification of three dimensional conformal therapy, Engineering in medicine and Biology Society 2000; 2: pp 1239-1242.


61 A Practical Guide To Quality Control of Brachytherapy Equipment, European Guidelines For Quality Assurance In Radiotherapy Booklet No. 8, 2004; Belgium.

