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Intensity Modulated Radiation Therapy Plan (IMRT) Verification Using Indigenous Heterogeneous Phantom

Payal Raina, Rashmi Singh and Mithu Barthakur

Abstract

The dose distribution given by intensity-modulated radiotherapy (IMRT) is highly conformal, compared to conventional radiotherapies; however, due to the presence of the large numbers of fields and irregular shape and size of the treatment segments, the accuracy of IMRT delivery needs to be verified via dose measurement. Different dosimetry techniques are available that measures part of or the whole treatment immediately before a patient is treated and give us the total treatment delivery picture. But the majority of the commercially available phantoms are of homogeneous density, whereas the actual human body is a complex medium of different density patterns. Additionally, the very few heterogeneous phantoms, which are available commercially (i.e., anthropomorphic phantoms) are very costly and are not procured by most of the radiotherapy centers, especially in developing countries. Therefore, an indigenous heterogeneous phantom has been designed to verify the dose distribution prior to patient treatment.

Keywords: heterogeneous phantom, three dimensional conformal therapy, intensity modulated radiation therapy, multi-leaf collimator, Hounsfield unit

1. Introduction

Radiation therapy would not exist without physics. It begins with the discovery of X-rays. This therapy uses ionizing radiation, which is delivered by a linear accelerator. Linear accelerator is a device that uses high-frequency electromagnetic waves to accelerate charged particles, such as electrons to high energies through a linear tube. The high-energy electron beam itself can be used for treating a superficial target, or it can be made to strike a target to produce x-rays for treating a deep-seated target. Radiation therapy works by damaging the DNA of cancerous cells. Photons cause indirect ionization, which happens as a result of the ionization of water, forming free radicals, which then damage. Charged particles, such as electrons, protons, boron, carbon, and neon ions can cause direct damage to target through high-LET (linear energy transfer) [1]. The main focus of physics in radiation therapy is to increase the level of precision

and accuracy of dose delivery to the target volume. From the 1950s to the late 1980s, the approach to radiation therapy was based on a two-dimensional (2D) approach. In 2D radiation therapy, plans were created manually, and a single beam used to be given from one to four directions [2]. Shielding blocks were used to collimate the beam. Advances in imaging technology like Ultrasound (US), Computed Tomography (CT), Magnetic Resonance Imaging (MRI), etc. significantly changed the practice of radiation therapy from the 2D method to a Three Dimensional Conformal Therapy (3DCT), which conforms to the high radiation dose with uniform intensity to tumor. For precise shaping of treatment field to the target volume, Multi-Leaf Collimator (MLC) system was developed in place of shielding blocks [3]. Advanced form of radiation therapy called Intensity Modulated Radiation Therapy (IMRT) has been developed in the mid-1990s and early 2000s. IMRT can provide conformal dose distribution compared with 3DCRT [3]. Intensity-Modulated Arc Therapy (IMAT) uses the Multi-Leaf Collimator (MLC) dynamically to shape the fields, as well as rotate the gantry in the arc therapy mode. Intensity-Modulated Arc Therapy (IMAT) was further improved with the addition of variable gantry rotation speeds and dose rates and was introduced as volumetric-modulated arc therapy (VMAT) in 2007. Brief descriptions of all these techniques are discussed in the following sections.

2. Three-dimensional conformal radiotherapy (3DCRT)

Three-Dimensional Conformal Radiation Therapy (3DCRT) means conformal dose distribution in terms of adequate dose to the tumor and minimum possible dose to normal tissue based on 3D anatomic information. The main distinction between treatment planning of 3DCRT and that of conventional radiation therapy is that treatment planning system optimizes dose distribution in accordance with the clinical objectives using anatomic information. Depending on imaging modality, visible tumor, the suspected tumor spread, patient motion uncertainties, critical structures, and relevant landmarks are outlined slice by slice by the radiation oncologist. This technique, however, fails in achieving conformal dose distribution for patient geometries where Organ at Risks (OARs) are located in close proximity to or are even embedded within complicated tumor shapes. Due to lesser conformity of dose distribution in Three-Dimensional Conformal Radiation Therapy (3DCRT), it may be insufficient to allow adequate escalation of tumor dose and there is a need for further improvement. It is possible only with intensity-modulated radiotherapy.

3. Intensity modulated radiotherapy (IMRT)

Intensity-modulated radiotherapy (IMRT) is an advanced form of 3-D conformal radiation therapy that allows modifying the intensity of the beam by considering each radiation beam as multiple rays or beamlets, and assigning different beam strengths to the individual rays [4]. The radiation intensity is adjusted according to the shape, size, and location of the tumor with the use of computer-controlled, moveable “leaves” called Multi-Leaf Collimator (MLC) systems. It consists of pairs of highly absorbing tungsten leaves that can either block or allow the passage of radiation from the many beams as shown in **Figure 1** to deliver a high dose to the target volume and acceptably low dose to the surrounding normal structures [5].

Figure 1.
 Multileaf collimators (MLCs).

It uses advanced imaging procedures such as Ultrasound (US), Computed Tomography (CT), Magnetic Resonance Imaging (MRI), Positron Emission Tomography (PET), and PET/CT, to achieve precise treatment modality for cancer patients [6]. A computer-controlled multileaf collimator has been programmed in three different ways to deliver IMRT [7].

1. Multi-segmented Static field Delivery: The patient is treated by multiple fields and each field is subdivided into a set of subfields [8]. The subfields are created by the MLC. The radiation beam is turned off when the leaves are moving from one field segment to another and is turned on only when the leaves reach and stop at the designated segment positions [9]. This method of IMRT delivery is also called “step-and-shoot” or “stop-and-shoot” [10].
2. Dynamic Delivery: In this technique, the beam is kept on while the corresponding leaves sweep simultaneously to produce the desired intensity modulation throughout the treatment delivery [11].
3. Intensity-modulated Arc therapy: Yu has developed an Intensity Modulated Arc Therapy (IMAT) technique. IMAT technique is similar to IMRT, which uses the dynamic mode of dose delivery to shape the fields with gantry rotation and the beam is on all the time [12].

4. Volumetric modulated arc therapy (VMAT)

Intensity-Modulated Arc Therapy (IMAT) has been improved with the addition of variable gantry rotation speeds and dose rates and was introduced as Volumetric-Modulated Arc Therapy (VMAT) in 2007 to describe rotational Intensity Modulated Radiotherapy (IMRT) delivered in a “single arc” [13]. VMAT can provide highly conformal dose distributions and can significantly improve the Intensity Modulated Radiotherapy (IMRT) delivery efficiency. The faster treatments reduce the effects of intra-fractional motion on both tumors and organs, and of course, the shorter

treatment times also increase patient comfort. The high plan quality and fast treatment delivery of Volumetric-Modulated Arc Therapy (VMAT) are attractive, and it has been widely applied to many disease sites.

5. IMRT plan verification

In external beam radiation therapy, the energy deposition is three-dimensional in nature. As such particles not only interact with the tumor site but also deposit some of their energy into the adjacent area. Consequently, neighboring normal tissues also receive some amount of radiation dose in this process. Therefore, normal tissue dose tolerance becomes a limiting factor to the success of the treatment. Therefore, a scheduled quality assurance program should be established to verify the plans generated on Treatment Planning System (TPS).

The dose distribution given by Intensity Modulated Radiation Therapy (IMRT) is highly conformal, compared to conventional radiotherapies. But due to the presence of large numbers of fields and irregular shape and size of the treatment segments, the accuracy of Intensity Modulated Radiation Therapy (IMRT) delivery needs to be verified via measurement of dose. Based on the recommendations of the International Atomic Energy Agency (IAEA) published in technical reports series number 277 and 398, there are several techniques to attain accuracy in dosimetry.

Different dosimetry techniques are available to compare the planned dose with delivered dose using an ionization chamber and commercially available phantom, such as slab phantom that measures the point dose at a particular desired reference depth. For reference dosimetry, radiographic film or radiochromic film is placed at a particular depth in slab phantom, and a planned dose is delivered on it. The film quality assurance dosimetry system, for instance, OmniPro IMRT correlates the resultant density of film with the planned dose at each point.

Luminescence dosimetry is also performed using an optically stimulated luminescence (OSL) system and thermoluminescent dosimeters (TLD). It can be also used for in vivo dosimetry in which OSL or TLD are placed on patient's body at reference points for measurement. The electronic portal imaging device is also utilized for reference dosimetry. In addition, many detector-based phantoms are available, for reference dosimetry, such as AccuCheck, Delta 4 phantom.

To evaluate an institution's ability to deliver the planned dose to patients, an indigenous heterogeneous phantom has been designed.

6. Designed phantom

The majority of the commercially available phantoms are of homogeneous density, whereas the actual human body is a complex medium of different density patterns [14]. Additionally, the very few heterogeneous phantoms, which are available commercially (e.g. anthropomorphic phantom) are very costly and are not procured by most radiotherapy centers, especially in developing countries. It is known that human body is composed of fat, tissue, bones, and air cavities having different electron densities that influence the interaction of photon and electron energy deposition affecting the dose delivery to a target volume. Therefore, this study was conducted to develop an indigenous heterogeneous pelvic phantom similar to the patient's anatomy and perform a pre-treatment verification in a realistic clinical scenario to obtain reproducible dosimetry.

A heterogeneous pelvic phantom was designed, shown in **Figure 2**, which was made of wax, a male pelvic bone (**Figure 3**), water, and borax powder. To construct the phantom, male pelvic bone with a density equivalent to that of human pelvic bone was placed in a cylindrical-shaped container. After placing it, a round plastic ball filled with water was placed for bladder. Borax powder with glue and water was placed below the bladder for rectum. Subsequently, molten wax was poured into it and allowed to solidify. After complete solidification of the wax, the outer container was cut and removed. A cavity was prepared at approximately geometrical center of phantom volume, and a 0.6 cm³ ion chamber was kept in the same position till the end

Figure 2.
Designed pelvic phantom.

Figure 3.
Male pelvic and femur bone used in developed phantom.

Figure 4.
CT slice of developed phantom with different parts.

of experiment, **Figure 4**. The three fiducially lead markers were put on two bilateral points, and one anterior point was placed on the surface of the phantom in the same cross-sectional plane to make three reference points [15].

Brivo CT 325 2-slice CT (Wipro GE Healthcare, WI, USA) has been utilized for computed tomography (CT) of the phantom and the CT images were taken at a slice thickness of 3 mm for planning purposes. The CT images were imported into the treatment planning system. The width and height were measured using the length measuring tool available in Treatment Planning System (TPS). The mean width and height were measured as 29 cm and 25 cm in CT images of heterogeneous pelvic phantom, respectively. These geometries of the phantom show that it can accommodate delivered beam field sizes and shapes. It allows the establishment of 3D locations. It is easy to transport, set up, align, and takedown in an accurate and efficient manner.

7. Hounsfield unit (CT number)

Computed Tomography (CT) works on the principle of amount of the X-ray energy absorbed. The amount of X-ray energy absorbed is proportional to the density of the body tissue. The computer generates a grayscale image, where the tissue density is indicated by shades of gray. The Hounsfield Unit (HU) is a relative quantitative measurement of radio density used in the interpretation of computed tomography images. The Hounsfield unit was named after Sir Godfrey Hounsfield, recipient of the Nobel Prize, for the invention of Computed Tomography (CT) [16]. It is proportional to the degree of x-ray attenuation and is defined as:

$$HU_{\text{tissue}} = \left[\frac{(\mu_{\text{tissue}} - \mu_{\text{water}})}{\mu_{\text{water}}} \right] \times 1,000 \quad (1)$$

where μ is the linear attenuation coefficient for water and tissue. On the Hounsfield scale, air is represented by a value of -1000 (black on the grayscale) and bone between $+700$ (cancellous bone) to $+3000$ (dense bone) (white on the grayscale). The linear attenuation coefficient is a function of both electron density and atomic number of the tissue within a pixel.

8. Electron density

The electron density is the measure of the probability of an electron being present at a specific location. It is calculated from its mass density and its atomic composition.

8.1 Comparison of Hounsfield units and relative electron densities of organs

The Hounsfield Unit (HU) and relative electron density of bone, fat, air cavity, bladder, and rectum in Computed Tomography (CT) images of a heterogeneous phantom and an actual patient were measured and has been given in **Table 1**. All the measurements were calculated by using Computed Tomography (CT) scanner console in terms of mean and stander deviation due to density variation in different CT slices. For the actual patient, a CT image of one patient was taken.

According to the results obtained from the Computed Tomography (CT) images of a heterogeneous pelvic phantom, relative electron densities for bone, fat (wax), air cavity, bladder (water), and rectum (borax powder) were 1.632, 0.896, 0.159, 1.037, and 1.051, respectively. On the other hand, relative electron densities for bone, fat, air cavity, bladder, and rectum were 1.335, 0.955, 0.158, 1.039, and 1.054, respectively, in an actual patient Computed Tomography (CT) image.

8.2 Radiation treatment plan creation

Various Intensity Modulated Radiation Therapy (IMRT) plans for prostrate patients were generated on the Monaco planning system. Plans were created with 5, 7, 9, and 12 coplanar 6MV photon beams. Couch and collimator angles were kept

S.No.	Pelvic Organs	Material	In CT images of a heterogeneous phantom		In CT images of an actual patient	
			HU $\pm$ SD	Relative electron density	HU $\pm$ SD	Relative electron density
1	Bone	Male Pelvic Bone	1037 $\pm$ 179	1.632	556 $\pm$ 187	1.335
2	Fat	Wax	-162 ± 45	0.896	-109 ± 108	0.955
3	Air cavity	Air	-846 ± 143	0.159	-847 ± 79	0.158
4	Bladder	Water	-5 ± 5	1.037	-3 ± 8	1.039
5	Rectum	Borax Powder	19 $\pm$ 53	1.051	20 $\pm$ 26	1.054

CT: Computed Tomography; HU: Hounsfield Units; SD: Standard Deviation.

Table 1.
Comparison of Hounsfield units and relative electron densities of organs.

as 0° for all plans. Calculation parameters such as grid spacing, fluence smoothing, and statistical uncertainty were 0.3 cm, medium, and 1% per plan respectively. Furthermore, the Monte Carlo algorithm was used for the plan optimization, and all the plans were generated in step and shoot mode.

8.3 Gamma analysis

The difference between measured and planned dose distribution is evaluated using quantitative evaluation methods. The Quality Assurance (QA) procedures of Treatment Planning System (TPS) narrated by Van Dyk et al. subdivides the dose distribution comparisons into high and low dose gradients regions, each with a different acceptance standard. In regions of low gradient, planned and measured doses are compared directly, with an acceptance tolerance placed on the difference between the measured and calculated doses. On the other hand, in high dose gradient regions, a small spatial error, either in measurement or calculation, results in a large dose difference between measurement and calculation. Therefore, in the region of high dose gradient, the concept of a Distance-To-Agreement (DTA) distribution is used to determine the acceptability of the dose calculation. The Distance-To-Agreement (DTA) is the distance between a measured data point and the nearest point in the calculated dose distribution exhibiting the same dose. The Dose-Difference (DD) and Distance-To-Agreement (DTA) evaluations complement each other when used as determinants of dose distribution calculation quality.

8.4 Pre-treatment verification

Two kinds of phantoms were chosen for absolute dosimetry of plans already done for the treatment. First one is heterogeneous pelvic phantom developed for radiotherapy quality assurance. Second one was Delta4 phantom (Scandidos, Uppsala, Sweden). CT scan of both the phantoms was done and images were transferred to the Monaco planning system.

After the complete optimization of the Intensity Modulated Radiation Therapy (IMRT), the plans were exported to a pelvic phantom and Delta4 phantom for a pre-treatment verification. After position verification, all Intensity Modulated Radiation Therapy (IMRT) plans were delivered by a linear accelerator.

9. Absorbed dose calculation

There are various methods to achieve accuracy in dosimetry and they are based on International Atomic Energy Agency (IAEA) recommendations published in technical reports series number 277 [17] and 398 [18].

In this study, absorbed dose at reference depth was calculated according to the Technical Reports Series No. 398 (TRS398) of the International Atomic Energy Agency (IAEA) [18] using the relation:

$$D = M_Q \times N_{D,W} \times K_{Q,Q_0} \times K_{T,P} \times K_S \times K_{pol} \quad (2)$$

where, M_Q is the electrometer reading (charge), $N_{D,W}$ is the Co^{60} chamber specific factor, $K_{T,P}$ temperature–pressure correction factor, K_{pol} polarity correction factor, K_S ion recombination factor.

9.1 Chamber calibration factor, $N_{D,W}$

The $N_{D,W}$ is the calibration factor in terms of absorbed dose to water for a dosimeter at a reference beam quality Q_0 . The chamber calibration factor $N_{D,W}$ for the ionization chamber (PTW 0.6 cm³; TN 30013–006353) is 5.386×10^7 Gy/C as obtained by BARC, Mumbai.

9.2 Chamber specific factor, K_{Q,Q_0}

The K_{Q,Q_0} is a factor that corrects for the difference between the response of the ion chamber in the reference beam quality Q_0 used for calibrating the chamber and in the actual user beam quality Q . The subscript Q_0 is omitted when the reference quality is Co-60 gamma radiation i.e. K_Q always corresponds to reference quality Co-60. The chamber-specific factor K_{Q,Q_0} is 0.99 for the ionization chamber (PTW 0.6 cm³; TN 30013–006353).

9.3 Temperature pressure correction factor, $K_{T,P}$

The mass of air in the cavity volume is subject to atmospheric variations. The correction factor to be applied to convert the cavity air mass to the reference conditions is given by:

$$K_{T,P} = \frac{(273.2 + T) P_0}{(273.2 + T_0) P} \quad (3)$$

where, P and T are the cavity air pressure and temperature at the time of the measurements, and P_0 and T_0 be the reference values (generally 101.3 kPa and 20°C).

9.4 Polarity factor, K_{pol}

The polarity factor is used to correct the response of an ionization chamber for the effect of change in polarity of the polarizing voltage applied to the chamber. It can be accounted for, by using a correction factor

$$K_{pol} = \frac{|M_+| + |M_-|}{2M} \quad (4)$$

where, M_+ and M_- are the electrometer readings obtained at positive and negative polarity, respectively, and M is the electrometer reading obtained with the polarity used routinely (positive or negative).

9.5 Ion recombination factor k_s

The incomplete collection of charge in an ionization chamber cavity owing to the recombination of ions requires the use of correction factor k_s .

$$k_s = a_0 + a_1 \left(\frac{M1}{M2} \right) + a_2 \left(\frac{M1}{M2} \right)^2 \quad (5)$$

where, $a_0 = 2.337$, $a_1 = -3.636$, $a_2 = 2.299$ and $M1$ and $M2$ are the electrometer readings at the polarizing voltages $V1$ and $V2$, respectively, measured using the same

irradiation conditions. V1 is the normal operating voltage and V2 is a lower voltage; the ratio V1/V2 is equal to two.

In a pelvic phantom, the dose for each plan was measured by PTW UNIDOS E electrometer connected with 0.6 cm³ ion chambers using Eq. 1 according to International Atomic Energy Agency (IAEA) published, Technical Reports Series-398 (TRS 398) protocol. These measured doses were compared with doses planned on the treatment planning system (TPS).

For Delta4 phantom, TPS calculated dose fluence was compared with measured dose fluence using the gamma evaluation method with critically acceptable criteria of 3 mm Distance-To-Agreement (DTA) and 3% Dose-Difference (DD). Before the evaluation of an Intensity Modulated Radiation Therapy (IMRT) plan, two more measurements were done by delivering 100 cGy with a 10 × 10 cm field at gantry angles of 0° and 90° in order to check the phantom for positional corrections and linear accelerator output constancy.

Table 2 shows the planning parameters, including number of fields, segments, and monitor units, and the percentage variation between planned doses and measured doses for each test case using pelvic phantom.

The gamma analysis results of each test case, including Dose-difference (DD), Distance-To-Agreement (DTA), and Gamma Index passing rates, are presented in **Table 3**.

Plan No.	Algorithm	Energy	No. of fields	Measured Dose	Planned Dose	% Variation
P1	Monte Carlo	6 MV	5	190.34	185.8	2.44 (+)
P2	Monte Carlo	6 MV	7	202.15	207.5	2.58(−)
P3	Monte Carlo	6 MV	9	172.46	176.1	2.07(−)
P4	Monte Carlo	6 MV	12	194.57	191.84	1.42(+)
Mean 2.13						
SD 0.52						
MV: Mega Voltage; SD: Standard deviation.						

Table 2.
Percentage variation between planned dose on treatment planning system and measured dose on linear accelerator using heterogeneous pelvic phantom.

Plan No.	Field Number	Segment Number	Monitor Unit	Dose Difference	DTA	Gamma Index
P1	5	14	734.20	79.5%	97.9%	98.4%
P2	7	19	820.31	80.1%	95.1%	97.3%
P3	9	15	775.48	81.4%	94.3%	97.5%
P4	12	14	724.53	80.8%	95.3%	98.8%
DTA: Distance to agreement.						

Table 3.
Result of dose difference, distance to agreement and gamma index using Delta4 phantom.

9.6 Test case P1: intensity modulated radiation therapy (IMRT) plan with 5 coplanar beams

The percentage variation between planned dose and measured dose was noted as 2.44% in an indigenously designed heterogeneous pelvic phantom.

Dose distribution at axial projection on the heterogeneous phantom, Delta4 phantom, and on actual patient CT image are shown in **Figure 5a–c** respectively.

The same plan was verified by using Delta4 phantom. The gamma passing rate for test P1 was 98.4%, whereas the pass percentages of Dose-Difference (DD) and Distance-To-Agreement (DTA) were 79.5% and 97.9%, respectively shown in **Figure 5d**.

9.7 Test case P2: intensity modulated radiation therapy (IMRT) plan with 7 coplanar beams

The percentage variation between planned dose and measured dose was noted as 2.58% in the designed pelvic phantom.

The same plan was verified by using Delta4 phantom. The gamma passing rate for test P2 was 97.3%, whereas the pass percentages of Dose-Difference (DD) and Distance-To-Agreement (DTA) were 80.1% and 94.3%, respectively.

9.8 Test case P3: intensity modulated radiation therapy (IMRT) plan with 9 coplanar beams

Similarly, with 9 coplanar beams, the percentage variation between planned dose and measured dose was noted as 2.07%.

Figure 5.
(a) Dose distribution in heterogeneous phantom, CT slice for test case P1. (b) Dose distribution in Delta4 phantom, CT slice for test case P1. (c) Dose distribution in patient, CT slice for test case P1. (d) Dose distribution, dose deviation, distance to agreement and gamma index of test case P1.

The Dose-Difference (DD) and Distance-To-Agreement (DTA) and Gamma Index were 81.4%, 94.3%, and 97.5% respectively.

9.9 Test case P4: intensity modulated radiation therapy (IMRT) plan with 12 coplanar beams

For the Intensity Modulated Radiation Therapy (IMRT) with 12 coplanar beams, the percentage variation between planned dose and measured dose was noted as 1.42%.

The Dose-Difference (DD) and Distance-To-Agreement (DTA) and Gamma Index were 80.8%, 95.3%, and 98.8%, respectively.

For all the four Intensity Modulated Radiation Therapy (IMRT) plans the percentage variation between the planned dose and measured dose was found to be within the tolerance limit ($< \pm 3\%$) prescribed by International Commission on Radiation Units and Measurements (ICRU 83) [19]. Additionally, Gamma evaluation results are based on the critically acceptable criteria of 3 mm DTA and 3% DD given in **Table 3**.

10. Conclusion

In radiation therapy, Quality Assurance (QA) is an essential aspect to ensure that the most accurate treatments are being delivered to a patient. When a new linear accelerator is commissioned at a hospital, the dosimetric parameters, for example, percentage depth dose, radiation beam symmetry, and flatness, are tested to verify the manufacturer's specifications and recommended guidelines. The machine is then periodically tested on a daily, monthly, and yearly basis to make sure that it remains within the specifications. This ensures that the machine continues to deliver what is indicated by a plan. For simple, traditional treatment methods, the Quality Assurance at the machine level is sufficient because possible errors are considered to be acceptable. However, for Intensity-Modulated Radiotherapy and Volumetric Modulated Arc Therapy planned treatments, the increased complexity along with high dose gradients, result in dosimetric errors and require empirical testing.

The main goal of radiation therapy is to deliver a prescribed dose to a target while minimizing the dose to the surrounding normal tissue. As new techniques are developed to achieve this goal, the treatments become more complex and the importance of having accurate dosimetry methods for both initial systems commissioning and ongoing Quality Assurance (QA) increases. Currently, it is mandatory that a patient-specific quality assurance test be performed prior to each new treatment course. Therefore, this study was conducted to develop an indigenous heterogeneous pelvic phantom similar to patient anatomy and perform a pre-treatment verification in a realistic clinical scenario to obtain reproducible dosimetry. Very few heterogeneous phantoms which are available commercially e.g. anthropomorphic phantom are very costly, and are not procured by most radiotherapy centers, especially in low-budget centers in developing countries [20].

In this study, an indigenous heterogeneous phantom was developed using wax for fat, artificial pelvic bone for pelvic bone, water for bladder, and borax powder with glue for rectum. Hounsfield unit and relative electron density of the phantom for different materials used for mimicking the patient were compared with the actual patient pelvic region. A comparison of Hounsfield Unit and electron density shows that the material used for the construction of phantom is almost equal to the

patient tissue heterogeneity as well as shape and tissue content. Materials used for the construction of phantom were locally available, cost-effective, and strong enough to maintain structural integrity.

In this study, Intensity Modulated Radiotherapy was verified using an indigenous heterogeneous pelvic phantom. For validation of heterogeneous phantom, similar plans were also verified using the Delta4 Phantom. The results obtained for all the studies were found to be within the tolerance limit which is <3% as prescribed by the International Commission of Radiation Protection (ICRU 83). This indicates that the phantom can be used successfully for routine patient-specific verification practices.

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