A Homogeneous Water-Equivalent Anthropomorphic Phantom for Dosimetric Verification of Radiotherapy Plans

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Abstract

Water is treated as radiological equivalent to human tissue. While this seems justified, there is neither mathematical proof nor sufficient experimental evidence that a water phantom can be treated as equivalent to human tissue. The aim of this work is to simulate and validate a water phantom that is tissue equivalent in terms of the dosimetric characteristics of both water and human tissue. Dynamic, intensity-modulated radiotherapy plans for two head and neck, one brain, one pelvis, and three lung/mediastinum cases were chosen for this study. Using a treatment planning system (TPS) (Eclipse, Varian Medical System, Polo Alto, CA, USA) and Anisotropic Analytic Algorithm in a grid resolution of 5 mm x 5 mm, a patient-equivalent water phantom was calculated from all rays in the isocentric plane as an array of water equivalent depths ($d_{WE}$). These rays were plotted versus isocentric separation and ray-tracing direction. Planar doses were compared between the isocentric plane in the patient computed tomography and the water equivalent phantom using gamma criteria of 2%–2 mm and 3%–3 mm. Except in one lung case, >95% gamma agreement was seen when using 3%–3 mm and >90% pass rate was seen when using 2%–2 mm. For head and neck cases, gamma-fail was restricted to the periphery. For mediastinum cases, gamma-fail was restricted to the lungs. This study demonstrates that a heterogeneous patient can be converted to a water phantom with comparable dosimetric characteristics and disagreements restricted to the lung area for both modulated and open beams. Potential sources of error include the $d_{WE}$ calculation and TPS dose computation.

Keywords: Alent radiological depth, tissue equivalent phantom, water equivalent depth (WED) phantom

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INTRODUCTION

Patient-simulating phantoms have been used since the earliest dosimetric measurements in radiotherapy and radiology. The utility of these phantoms is well established and their use is justified in the standard radiotherapy physics texts.¹ However, in this work, we will revisit the justifications on which the approximation of a patient by a phantom is based. Use of a water phantom as a surrogate for a patient is supported by the following two arguments. First, phantoms are unavoidably essential to commissioning and quality assurance (QA). During the commissioning of a treatment planning system (TPS), beam data are collected using a water phantom, usually one large enough to provide full scattering conditions. These beam data are then used by the TPS to predict dose distributions in actual patients.¹

Patient-specific QA tests (PSQA), including point dose measurement, are performed using a water phantom, after exporting unaltered fluence patterns from the treatment plan. Using this sequence, beam data collection and PSQA tests are also done in water, and TPS dose calculations are done in patient computed tomography (CT) scan.² A QA plan consists of an unaltered fluence pattern from the patient treatment plan which has been calculated in a phantom of uniform density and regular geometry. Although the TPS dose calculation is associated with its corresponding PSQA test through the unaltered fluence pattern, the dose calculation in the patient CT is independent of the calculation in the water phantom.

Second, the radiological properties of water are similar to those of tissue. Water closely approximates the radiation...
absorption and scattering property of soft tissue and muscle. This is the consequence of both having similar atomic number, $Z_{\text{eff}}$. Several investigators, using different methods, calculated the effective atomic number during the period between the early 1940s and the late 1990s.\cite{3-5}

Prasad et al. found almost equal values of $Z_{\text{eff}}$ for muscle and for water, 4.8 and 4.7, respectively. Therefore, it is expected that a water phantom should be a good choice, one that will faithfully mimic human tissue.

However, in clinical radiotherapy, for example, for PSQA, plastic phantoms are often used instead of water. It is assumed that the performance of these plastic phantoms will be similar to that of water. However, available plastic phantoms have physical density and fluence scaling factors that are slightly different than those of water. For example, the PTW solid water phantom has a physical density of 1.029 g/cm$^3$. Other methods, such as Monte Carlo, can be used to generate an identical dose distribution in human tissue and in phantom.

These are the justifications, but there is neither mathematical proof nor sufficient experimental evidence that a phantom can fully simulate a patient. In other words, it is a postulate that a phantom is equivalent to the patient for which it is surrogate. Treating the phantom as equivalent to the patient creates two approximations: first, the deformation from the patient’s anatomy to the phantom’s shape and second, conversion from a heterogeneous patient to a homogeneous phantom. In this work, we attempt to establish the equivalence between the phantom and the patient, using the concept of water equivalent depth ($d_{\text{WE}}$) and its application in a TPS and in PSQA, using the concept of radiological depth.\cite{6}

**Materials and Methods**

A total of seven patients were chosen for this study, each of whom had received radiotherapy treatment in our clinic. These include 3 lung cases, 2 head and neck cases, 1 brain case, and 1 pelvis case. These patients had been planned for treatment with dynamic, intensity-modulated, radiotherapy using the Eclipse V11.0 (Varian Medical System, Palo Alto, CA, USA) TPS. Eclipse calculates dose using the anisotropic analytic algorithm (AAA) with a modified Batho Power Law for its heterogeneity correction. To generate a water-equivalent phantom (WEP), the accuracy of the CT to electron density calibration curve (CT-ED) is essential. To create a CT-ED curve, a Gammex Model 467 ED phantom was scanned using a Philips Bright Speed CT scanner (Philips, The Netherland). CT images were then transferred to the TPS using the DICOM protocol, and a CT-ED calibration file was created for use in dose calculation. To verify the accuracy of our CT-ED calibration curve, the $d_{\text{WE}}$ of a typical radiological path was verified in a CATPhan 500. The TPS calculated value was found to be accurate to within 0.5%.

For an arbitrary continuous heterogeneous medium having a known number of segment ($N$) with segment length ($L$) having a known ED ($\rho_e$), $d_{\text{WE}}$ can be calculated as follows:

$$d_{\text{WE}} = \sum_{j=1}^{N} L(j) \rho_e(j)$$

A WEP was calculated from each patient CT, using the following method. All calculations were based on the isocentric axial plane, and in this plane, the ray traced from surface of the patient through the plane of isocenter was converted to $d_{\text{WE}}$. Figure 1a shows the isocentric axial plane for a typical head and neck patient, with the source to skin distance and the central ray. Next, rays separated by 5 mm, in the plane of isocenter, were each converted to their corresponding radiological depth. Likewise, all the rays separated by 5 mm in the isocentric axial plane were converted to their $d_{\text{WE}}$. This procedure was repeated for all available CT slices and for all available beams. The resulting $d_{\text{WE}}$ per axial slice was plotted as a radial function, maintaining the fixed geometrical relationship between the ray length, ray tracing direction, and plan isocenter. This produced a two-dimensional (2D) reconstructed CT slices reflecting the $d_{\text{WE}}$. This same procedure was then repeated for all axial CT slices, so as to reconstruct the complete 3D water-equivalent volume. These reconstructed 3D volumes are presented in Figure 1b. Further, the sagittal and coronal planes were also reconstructed in TPS. Reconstructed axial, sagittal, and coronal sections of a representative patient-equivalent water phantom are shown in Figure 1c-e.

All the treatment plans were recalculated in the WEP by keeping the heterogeneity correction off as a QA plan using an unaltered fluence pattern. The planar dose was compared between the CT data set isocentric coronal planes and the corresponding coronal plane of the WEP calculated values.

**Results**

Planar doses were compared between the isocentric planes of the patient CT scans and those of the $d_{\text{WE}}$ phantom. The results obtained, using gamma matching analysis, for the seven different cases are presented in Table 1. The gamma index was calculated using threshold values of 2% dose difference and 2 mm distance-to-agreement (2%–2 mm) and using 3%–3 mm. Greater than 95% agreement was seen when using 3%–3 mm for all cases, except for one of the lung patients. Figure 2 shows the gamma analysis results for the two head and neck cases and for the two lung

| Site                | 2 mm 2% | 3 mm 3% |
|---------------------|---------|---------|
| Head and neck (nasopharynx) | 91.1%  | 96.5%   |
| Head and neck (oropharynx) | 97.1%  | 99.9%   |
| Pelvis (cervix)       | 98.6%  | 99.9%   |
| Brain                | 99.9%  | 100.0%  |
| Lung 1               | 84.8%  | 93.0%   |
| Lung 2               | 92.5%  | 96.1%   |
| Lung 3               | 94.0%  | 97.0%   |
mediastinum cases show some gamma fail in the lung area. Gamma-matching analysis using 2%–2 mm shows a lowering of the pass rate. A pass rate >90% was obtained for all the cases except for the lung patient mentioned above. Lung patients showed a lower pass rate than did the other sites. This lower pass rate in lung cases might be due to limitations of the TPS’s AAA dose calculation algorithm.[7] It could also potentially have been due to errors in reconstruction of the WEP. Or, potentially, both of these might have been factors contributing to the mismatch. To understand this result, 5 cm × 5 cm and 10 cm × 10 cm open beams were considered. A flat open beam was selected to simplify testing. The beams were placed anterior in the mediastinum region of the patient CT scan and also in the reconstructed WEP of lung patient 1 as shown in Figure 2. Isocentric coronal planes were then extracted, and gamma-matching analysis was done. Gamma matching was tested for point gamma at ±3 cm from the isocenter position using a test criterion of 2%–2 mm and 3%–3 mm. Gamma-matching results are shown in Figure 3. Failure points for each evaluation criteria are located in the lung at the +3 cm position.

## Discussion

Radiotherapy treatment delivery is validated by PSQA, and there are numerous methods of PSQA. These include point dose measurements, planar dose verification, and 3D dose reconstruction. These are carried out using portal imagers, film, ion chamber arrays, diode arrays, MU calculation software, and other methods.[2,8-12] All of these methodologies are based on phantom measurements; thus, it is necessary to establish the equivalence between heterogeneous patients and homogeneous water phantoms, and this equivalence has not previously been established in the literature. The present study establishes this equivalence.

PSQA is limited by the necessary absence of the patient from the measurement. It is not possible to measure dose distribution directly in patients during radiotherapy treatment; although in recent years, it has become feasible to perform EPID-based transit dosimetry as a part of patient treatment (in vivo dosimetry) or using phantom.[13,14] Transit dosimetry has advantages over other PSQA dosimetric methods, and some investigators describe it as in vivo dosimetry.[14] Transit dosimetry simulates the situation where the patient’s anatomical information is carried by the exit beam to the detector system. Although portal imager-based setup imaging has gained a wide spread application, portal image-based dosimetry has found only limited application. This is due to two reasons.

First, transit dosimetry uses a backprojection algorithm which is not robust for heterogeneity correction. Therefore, it does not hold well for treatment sites, for example, lungs, with tissue heterogeneities.[13-17] Second, the EPID is not water equivalent and can only be converted to water equivalent in an experimental setup.[18,19]
Backprojection dose reconstruction methods are limited, and advanced backprojection software is not commercially available. The Netherlands Cancer Institute developed its own software for calculating transit dosimetry using backprojection.[13,14] In our study, we used a well-established and well-tested TPS. Furthermore, an EPID’s backprojection dose reconstruction system and the TPS use two different calculation algorithms. In our case, only the AAA is available in the TPS and the backprojection algorithm has not been validated against the TPS. As the AAA and backprojection algorithms work independently, if there is a systematic error in either algorithm, this will significantly influence the dose verification. It is not clearly described by other investigators whether their respective algorithms have been validated against each other.[14-16] Wendling et al. converted the patient into a WEP by forcing the density and keeping the patient’s anatomic boundary unaltered. In our method, the patient’s anatomy was deformed based on equivalent radiological path length.[9] Although the anatomical boundary was deformed, all gamma matching was performed in the isocentric plan which was only undeformed plane offering a true dimensional gamma matching. Further introduction of the radiological path length accounts for the heterogeneity correction. Therefore, it can be stated that Wendling et al. dominated the tissue heterogeneity correction, whereas we have corrected for it.

To overcome the problems of dose reconstruction using a backprojection technique in a heterogeneous medium, we have converted the patient into a homogeneous water phantom and reconstructed dose without altering the fluence pattern of the actual treatment plan. However, we could not circumvent the problem of improper dose reconstruction and lower pass rates in the lung area, as reported by other investigators.[14-16] We could not resolve whether this problem was due to limitations in the TPS’s dose calculation algorithm, limitations in the WEP reconstruction, or both of these factors. A lower gamma passing was seen for open as well as intensity-modulated fields in lung cases. We did not account for the noncoplanar beams as ray tracing is difficult in the noncoplanar direction. Although not tested in this study, however if tested, noncoplanar beam will act similarly to that of a planar beam.

At present, described article is suitable for all kinds of target volume like re-entrant target volumes (normal tissue-target-normal tissue-target) or the target volumes that inherently contain inhomogeneities as ray tracing does not differentiate between normal tissue and target volume. We encounter “normal tissue-target-normal tissue-target” situation in case of head and neck cases where the passage of radiation sees the normal structure and target alternatively when looked laterally from gantry 270° and (or) 90° and heterogeneous target volumes in bone metastatic cases. The former situation is called as the re-entrant “normal tissue-target-normal tissue-target” situation and our algorithm works well even under such difficult condition(s). Presented algorithm fails with the metallic implants because of the fact that CT-ED file is not appropriate in the gray scale range provided by metallic implant. However, if it is possible to obtain an accurate CT-ED file in the gray scale range yielded by metallic implant, algorithm will deform the anatomy appropriately to provide accurate radiological distance hence to offer an accurate calculation.

The present work shows that a heterogeneous patient can be converted to a homogeneous water phantom. The conversion was done manually using 2500–3000 points from the TPS. This work is time-consuming and labor intensive and was complete for only seven patients. This work will later be extended to 3D dose reconstruction and verification for patient-specific customized phantom measurements. This study demonstrated the equivalence of the patient and its corresponding water phantom using the concept of $d_{w_{eq}}$. The IMRT planar dose verification was established between the treatment plan made in patient CT scan and patient WEP. However, it may be questioned about the rationale of our study. This study is a step to build an efficient “in vivo” (transit) portal (EPID) dosimetry algorithm which will be more accurate than any of the presently available EPID dosimetry techniques, which will be presented in the subsequent articles.[14-17,20] Table 2 summarizes and compares the physical principle and outcome of the different fundamental transit dosimetry methods. Radiological path length was the most preferred technique for transit dosimetry since its introduction in 2006; which is also the basis of our present work. Literatures review although reveals that radiological path length technique works well in the less heterogeneous medium, it fails in case of large heterogeneities like lung. This is attributed to the incompatibility in terms of the accuracy of the algorithm used in planning system and transit backprojection algorithm. The backprojection algorithm works well in soft tissue or muscle. Even though the discrepancy is not so high in case of bone; back projection algorithm fails in high heterogeneity cases like lung because of the inaccurate dose calculation in TPS. Furthermore, on the top of the effective radiological path-length-based corrections, TPS dose calculation also accounts for the additional corrections such as kernel-based scatter dose calculation for AAA dose calculation engine; for more advanced dose calculation algorithm like Monte Carlo or Boltzmann transport technique, use a ray tracing method. However, in vivo backprojection algorithm considers only the radiological depth, which introduces the discrepancy in the dose matching between the TPS and the backprojection algorithm.

**CONCLUSION**

To establish an efficient in vivo transit dosimetric method, it is essential to establish a good agreement in homogeneous medium at the first point. This is established in the present study by converting patient into a water-equivalent anthropomorphic phantom using an equivalent radiological path length
Arjunan, et al.: Water equivalent anthropomorphic phantom

Table 2: Comparison of the different in vivo dosimetry techniques and its used methods such as radiological path length, collapse cone or Monte Carlo techniques, and their effectiveness of handling the patient’s anatomical heterogeneity

| Author | Reconstruction | Algorithm | Method | Projection type | Results |
|--------|----------------|-----------|--------|-----------------|---------|
| Wendling M, et al. | 2D | In house | Radiological path length | Backprojection to a plane either in phantom or in patient | No volume doses |
| Van Elmpt W3, et al. | 3D | In house | Monte Carlo XVMC | Forward projection to the patient | Good agreement with TPS for all the sites considered. All the EPID measurements are without patient and nothing but pretreatment QA |
| Wendling M, et al. | 3D | In house | Radiological path length | By reconstructing the dose within the patient or phantom volume in multiple planes parallel to the EPID, the 3D dose distribution can be obtained for each beam limited to IMRT | Good agreement for sites other than lung and sites with high heterogeneities. Patient transmitted fluence plane was used for reconstruction |
| Mans A, et al. | 3D | In house | Radiological path length | Same method as Wendling 2009 but extended to VMAT | Good agreement for sites other than lung and sites with high heterogeneities. Patient-transmitted fluence plane was used for reconstruction |
| Wendling M, et al. | 3D | In house | Radiological path length | Backprojection of patient-transmitted fluence to the homogeneous patient | Modification to address the original work of wendling et al. 2006 for heterogeneity. However, the backprojection is not done on the actual patient with heterogeneity |
| Uytven EV, et al. 2015 | 3D | In house | Collapsed cone | Backprojecting the patient-transmitted fluence to arrive an energy fluence which is used as a forward input to patient CT to calculate the dose | Good agreement with TPS for all the sites considered. Monte Carlo was used to convert the initial patient-transmitted fluence to energy fluence |

CT: Computed tomography, 3D: Three dimensional, 2D: Two dimensional, EPID: Electronic portal imaging devices, QA: Quality assurance, TPS: Treatment planning system, IMRT: Intensity-modulated radiotherapy

This will be further implemented in the EPID transit backprojection algorithm for in vivo dose computation more accurately than any presently available techniques.

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Conflicts of interest
There are no conflicts of interest.

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