Assessment of four dose calculation algorithms using IAEA-TECDOC-1583 with medium dependency correction factor (Kmed) application

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ABSTRACT

Purpose: This study discusses the measurement of dose in clinical commissioning tests described in IAEA-TECDOC-1583. It explores the application of Monte Carlo (MC) modelled medium dependency correction factors (Kmed) for accurate dose measurement in bone and lung materials using the CIRS phantom.

Methods: BEAMnrc codes simulate radiation sources and model radiation transport for 6 MV and 15 MV photon beams. CT images of the CIRS phantom are converted to an MC compatible phantom. The PTW 30013 farmer chamber measures doses within modeled CIRS phantom. Kmed are determined by averaging values from four central voxels within the sensitive volume of the farmer chamber. Kmed is calculated for Dm and Dw algorithm types in bone and lung media for both photon beams.

Results: Average modelled correction factors for Dw calculations using the farmer chamber are 0.976 (±0.1 %) for 6 MV and 0.979 (±0.1 %) for 15 MV in bone media. Correspondingly, correction factors for Dw, calculations are 0.99 (±0.3 %) and 0.992 (±0.4 %), respectively. For lung media, average correction factors for Dw calculations are 1.02 (±0.3 %) for 6 MV and 1.022 (±0.4 %) for 15 MV. Correspondingly, correction factors for Dw calculations are 1.01 (±0.3 %) and 1.012 (±0.2 %), respectively.

Conclusions: This study highlights the significant impact of applying Kmed on dose differences between measurement and calculation during the dose audit process.

1. Introduction

Dose computation verification is an important part of acceptance testing and commissioning procedures of treatment planning systems (TPSs) [1]. The International Atomic Energy Agency (IAEA) has recommended a set of clinically practical tests (IAEA-TECDOC-1583) based on technical report series (TRS-430) for the assessment of TPS dose calculation algorithms [2].

The IAEA-TECDOC-1583 has recommended clinical commissioning test cases covering a wide range of typical clinical situations. These are structured in a way that dose in specific points is checked for single beams, then standard multiple field techniques are used, and finally complex multi field arrangements are applied [2]. These tests are primarily aimed at confirming that the absorbed dose as calculated by the TPS is in agreement with that determined by measurement. The clinical commissioning tests described in this report are based on the use of the CIRS Thorax phantom Model 002LFC as shown in Fig. 1.a. The phantom has a body made of plastic water as well as lung and bone equivalent material sections with holes to hold interchangeable rod inserts that allow insertion of radiation detectors. The inserts are distributed within the phantom as follows: two inserts in each lung are filled with a lung equivalent material, one insert at the location of the vertebral body is filled with a bone equivalent material, and five inserts located in the mediastinum region are filled with tissue equivalent material (ten inserts in total). A schematic diagram can be seen in Fig. 1.b. Use of an ionization chamber is recommended for absorbed dose measurements. A chamber is placed in the corresponding plug, and this plug is fully inserted into the selected hole of the phantom. According to TECDOC-1583, all doses refer to absorbed dose to water regardless of the measurement region of the phantom (lump or bone) [2].

Additional considerations should be taken in to account when carrying out dose measurements in the bone and lung material parts of the CIRS phantom. Ionization chambers are calibrated at the standard laboratories reference beam of quality Co-60 and for measurements in

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water, thus they are applicable only in these conditions and require additional considerations when measured in other materials [3-5]. Therefore, absorbed dose measurements in non-water equivalent media such as lung and bone materials with a detector calibrated in a water phantom will be inaccurate. This is because the secondary electrons from the surrounding non-water equivalent media will differ from those under calibration conditions in water. In other words, the perturbation correction factor for an air-filled detector in water will not be the same as those in other media [4,5]. Thus the non-corrected ionization readings should not be used directly for bone and lung dosimetry with ion chambers [4].

In this study, we measured and calculated the specific correction factors that should be applied to detector readings for TECDOC-1583 tests in order to directly measure dose in bone and lung materials to allow comparison to TPS algorithms that calculate either dose-to-water in water (\(D_{w,w}\)) or dose-to-medium in medium (\(D_{m,m}\)).

To calculate the required corrections for the detector readings, it is important to understand how the absorbed dose is reported by the TPS dose calculation algorithms that are commercially available today. There are three separate quantities used to specify the dose:

- \(D_{w,w}\): This type of algorithm calculates dose-to-water in water and all material are considered as water of varying density. In the \(D_{m,m}\) case where bony and air materials are present, the TPS treats these regions as high density water and low density water respectively [6].
- \(D_{m,m}\): This type of algorithm calculates dose-to-medium in medium and the TPS accounts for radiation transport through a number of patient like materials and the density of such materials [6].
- \(D_{w,m}\): For \(D_{w,m}\) calculations, the TPS first calculates \(D_{m,m}\) and then converts the dose to \(D_{w,m}\) using stopping power ratios from Monte Carlo algorithms [6].

This study aims to investigate the measurement of dose in clinical commissioning cases outlined in TECDOC-1583. It focuses on the application of Monte Carlo (MC) modelled correction factors for dose measurement in bone and lung materials of the CIRS phantom. Furthermore, the study compares all measurements with four different dose calculation algorithms utilized by two different TPSs.

2. Materials and methods

2.1. Theoretical consideration

Since ionization chambers are calibrated for measurement of absorbed dose-to-water in water, discrepancies between measured doses and planned doses in non-water equivalent materials like bone and lung can be expected. The relative electron density values for lung, bone, and soft tissue in a semi-anthropomorphic 002LFC CIRS Thorax phantom (Norfolk, USA) as specified by the manufacturer are 0.207, 1.506, and 1.04, respectively, relative to water. As electron density differences between water and soft tissue are negligible, in this study, only correction factors for dose measurements performed in bone and lung materials were calculated.

Andreo [5] recommended avoiding the conversion from \(D_{m,m}\) to \(D_{w,m}\) via stopping power ratios due to the uncertainties involved. Also, the number of TPSs that use dose calculation algorithms which calculate \(D_{w,m}\) are limited. Based on these two reasons, the \(D_{w,m}\) reporting scenario has not been investigated. Diagrams of the eight test cases of TECDOC-1583 are shown in Fig. 2.

In \(D_{w,m}\) dose calculation algorithms for tests that compare measured and planned doses in lung (point 9 of test 1, point 6 of test 4, and point 7 of tests 5 and 6) and bone (point 10 of tests 1, 4, and 6), the dose calculation algorithm works by assigning different densities of water to the different materials. For lung measurements, low-density water is assigned, while for bone measurements, high-density water is assigned. However, during the measurement process, the ionization chamber is surrounded by low-Z lung or high-Z bone material, respectively. This mismatched condition deviates from the calibration conditions of the detector. To account for these discrepancies, we propose the use of correction factors (\(K_{med}\) for PTW 30013 farmer 0.6 cc ionization chamber (PTW Freiburg, Germany) when measuring in bone and lung equivalent-material as per Equations (1) and (2).

In \(D_{w,m}\) scenario for bone material:

\[
D_{HDW,w} = M_{bone} N_{D,w} K_{med,DHDL,w} \tag{1}
\]

In \(D_{w,m}\) scenario for lung material:

\[
D_{LDW,w} = M_{lung} N_{D,w} K_{med,DLDL,w} \tag{2}
\]

In the \(D_{w,w}\) algorithm scenario, doses in the bone and lung materials are considered as dose in high density water (\(D_{HDW,w}\)) and dose in low density water (\(D_{LW,w}\)) which are calculated by multiplying the charge readings of detector in bone (\(M_{bone}\)) and lung (\(M_{lung}\)), with the detector calibration coefficient in terms of absorbed dose to water in a Co-60 beam (\(N_{D,w}\)) and a medium dependent dose correction factor for bone (\(K_{med,DHDL,w}\)) and lung (\(K_{med,DLDL,w}\)).

In the case of \(D_{m,m}\) dose calculation algorithms for tests that compare measured and planned doses in lung (point 9 of test 1, point 6 test 4, and point 7 of tests 5 and 6) and bone (point 10 of tests 1, 4, and 6), the dose calculation algorithms calculate the dose to bone and dose to lung directly. However, the detector is calibrated in terms of absorbed dose-to-water and therefore, requires correction. In this scenario to account for this discrepancy, Equations (3) and (4) contains the correction factor (\(K_{med}\)).

Fig. 1. A Thorax Phantom (CIRS Model 002LFC). Fig. 1.b. Labelling of holes and the recommended arrangement of the certified electron density reference plugs for the CT scan. Plugs 1, 2, 3, 4 and 5 – soft tissue substitute, plugs 6, 7, 8 and 9 – lung substitute and plug 10- bone substitute.
In $D_{m,m}$ scenario for bone material:

$$D_{\text{bone},\text{bone}} = M_{\text{bone}} \cdot N_{D_{\text{w}},w} \cdot K_{\text{med}_{\text{bone}},\text{bone}}$$

(3)

In $D_{m,m}$ scenario for lung material:

$$D_{\text{lung},\text{lung}} = M_{\text{lung}} \cdot N_{D_{\text{w}},w} \cdot K_{\text{med}_{\text{lung}},\text{lung}}$$

(4)

For the $D_{m,m}$ algorithm scenario, the quantity of interest is the dose in the bone material ($D_{\text{bone},\text{bone}}$) and dose in the lung material ($D_{\text{lung},\text{lung}}$). This is calculated through multiplying the charge readings of detector in bone ($M_{\text{bone}}$) and lung ($M_{\text{lung}}$), the detector calibration coefficient in terms of absorbed dose to water in a Co-60 beam ($N_{D_{\text{w}},w}$) and a medium dependent dose correction factor for bone ($K_{\text{med}_{\text{bone}},\text{bone}}$) and lung ($K_{\text{med}_{\text{lung}},\text{lung}}$).

Eight situations have been modeled by MC simulation to determine $K_{\text{med}}$ for each material (bone and lung) and each algorithm scenario ($D_{w,w}$ and $D_{m,m}$). All the different modeled situations are summarized in Table 1.

Fig. 2. Eight clinical test cases of TECDOC-1583 in anthropomorphic 002LFC CIRS thorax phantom with measurement holds.

Table 1

<table>
<thead>
<tr>
<th>Algorithms scenarios</th>
<th>Materials</th>
<th>Equations</th>
<th>MC Modeled Scenarios</th>
</tr>
</thead>
<tbody>
<tr>
<td>$D_{w,w}$</td>
<td>Bone</td>
<td>$K_{\text{med}_{\text{bone}},\text{bone}}$</td>
<td>$D_{\text{bone},\text{bone}}$ : Absorbed dose to water in a water phantom</td>
</tr>
<tr>
<td>$D_{w,w}$</td>
<td>Lung</td>
<td>$K_{\text{med}_{\text{lung}},\text{lung}}$</td>
<td>$D_{\text{lung},\text{lung}}$ : Absorbed dose to water in a water phantom</td>
</tr>
<tr>
<td>$D_{m,m}$</td>
<td>Bone</td>
<td>$K_{\text{med}_{\text{bone}},\text{bone}}$</td>
<td>$D_{\text{bone},\text{bone}}$ : TPS calculated dose to high density water in $D_{w,w}$ scenario</td>
</tr>
<tr>
<td>$D_{m,m}$</td>
<td>Lung</td>
<td>$K_{\text{med}_{\text{lung}},\text{lung}}$</td>
<td>$D_{\text{lung},\text{lung}}$ : TPS calculated dose to low density water in $D_{w,w}$ scenario</td>
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</tbody>
</table>

2.2 Monte Carlo modeling

In this study, we applied BEAMnrc and DOSXYZnrc codes to carry out simulation of the radiation source and modelling of radiation transport. Both these codes are based on an electron gamma shower user code (EGSnrc) developed by the National Research Council of Canada (NRC) [6,7]. The Monte Carlo geometry of a Varian iX linear accelerator was built based on the technical data provided by Varian (Varian Medical System Inc., Palo Alto, CA, USA) using the BEAMnrc user code. 6 MV and 15 MV photon beam energies were simulated and applied to 4 out of the 8 tests (tests 1, 4, 5 and 6) recommended by the TECDOC-1583 (see Fig. 2) [2].

In the BEAMnrc simulations, the global cut-off energies for electron and photon transport were set at ECUT (electron cut-off energy) = 521 keV and PCUT (photons cut-off energy) = 10 keV, respectively. In this experiment, $1.0 \times 10^5$ initial electrons incident on the target contains $5.1 \times 10^7$ particles set for MC calculations.

To decrease the simulation time and uncertainty, the variance reduction technique of directional Bremsstrahlung splitting (DBS) was adopted and the splitting number was set to 1000. The DBS field radius was equal to the corresponding radiation field at a distance of 100 cm from the source. Electron range rejection was applied with an ESAVE (energy save) value available in BEAMnrc, this is the threshold energy of the electrons at which range rejection is considered (1 MeV in the target and 2 MeV in other components). The phase space file, which represents the beam output of the accelerator, was scored below the jaws for each of the fields of TECDOC-1583’s cases at a corresponding SSD of each evaluated tests. To model the incident electron parameters in the treatment head, the BEAMnrc source ISOURCE = 19 (elliptical beam with Gaussian distributions in X and Y) was used.

In the next step of the simulation, the phase space file derived from the BEAMnrc code was used as input data in the DOSXYZnrc code. A total of 424 computer tomography (CT) slices were obtained of the 002LFC CIRS Thorax phantom (Norfolk, USA) with image resolution of $512 \times 512$ pixels and slice thickness of 1 mm. The DICOM CT scan images were converted to a MC compatible phantom using the stand-alone code, CTCREATE, which converts the patient’s CT data to the desired dimension, material type, and mass density [7]. To convert CT data into material and mass density, the conventional CTCREATE/ DOSXYZnrc CT ramp file was utilized in the simulation. For further details regarding the threshold of material definition, please refer to the DOSXYZnrc user’s manual [7]. Images were segmented into three different voxel intensities (size of each voxel: $0.2 \times 0.2 \times 0.2$ cm$^3$), including bone ($\rho = 1.60$ g/cm$^3$), lung ($\rho = 0.21$ g/cm$^3$), and soft tissue ($\rho = 1.06$ g/cm$^3$). The density and material of the bone and lung cubes
were varied to obtain each of the scenarios detailed in Equations (1) and (2) for both 6 MV and 15 MV photon beams. For each point of measurement in the modeled CIRS phantom (as shown in Fig. 1.b), a PTW 30013 Farmer chamber with a sensitive volume of 0.6 cc (PTW Freiburg, Germany) was used. The detector geometry was modeled in detail based on the manufacturer’s specifications [8]. Additionally, a code “egs_chamber,” was utilized to incorporate the detailed geometry of the PTW 30013 farmer chamber within the Monte Carlo simulations. The PTW 30013 Farmer chamber was simulated in detail including areas such as its cavity volume, cavity length, cavity radius, central electrode (aluminum), and wall material (PMMA). After the simulation was completed, the DOSXYZnrc code produced a 3Ddose file. This file applies units of dose per particle incident on the target (Gy per electron). The simulated results were converted to absolute dose according to the method proposed by Popescu et al. [9].

According to the material composition of components supplied by the manufacturer, we employed the ESTAR program to create density correction files for the CIRS lung and bone to generate the PEGS4 electron stopping power data files for the simulations. For Dw,m simulations, the mass density of the CIRS bone and Lung was 1.60 g/cm$^3$, respectively. For Dw,w calculations, (where the voxels are simulated as water with a density of bone), the mass density of water was considered as 0.21 g/cm$^3$, designated for dose scoring. (To calculate the dose in scenarios involving a detector, four central voxels within the chamber volume are utilized, each possessing a volume of 0.2 cm$^3$, designated for dose scoring.).

Table 2

<table>
<thead>
<tr>
<th>Algorithms</th>
<th>TPS</th>
<th>Dose calculation type</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anisotropic Analytical Algorithms (AAA)</td>
<td>Eclipse</td>
<td>Dw,m</td>
</tr>
<tr>
<td>Collapsed Cone Convolution (CCC)</td>
<td>RayStation</td>
<td>Dw,w</td>
</tr>
<tr>
<td>Monte Carlo (MC)</td>
<td>RayStation</td>
<td>Dw,w</td>
</tr>
</tbody>
</table>

3. Results

3.1. Material validation

Furthermore, to validate the doses produced by the MC model in the CIRS phantom, doses simulated in soft tissue for four fields of which two pass through lung materials and one pass through bone material (point 5 of test 4) in EGSnrc MC has been compared to measurements with PTW 30013 Farmer 0.6 cc detector (see Fig. 4). The difference in the simulated dose by MC vs measured dose in the CIRS phantom with a 0.6 cc Farmer chamber was found to be within 0.22 % and 0.46 % for 6 MV and 15 MV respectively. The differences in the simulated dose by MC vs evaluated dose calculation algorithms in this study (AAA, AXB, CCC and MC) are summarized in Table 3. The type A uncertainty in the Monte Carlo models was 0.4 %.

2.3. IAEA-TECDOC-1583 dosimetry audit

The TECDOC-1583 audit was performed on the CIRS thorax phantom. The phantom was CT scanned before the audit measurements were carried out and a dosimetric plan was created for each case according to the TECDOC-1583 protocol. For dose measurements, the PTW 30013 farmer 0.6 cc detector was used. The detector is calibrated for absolute dosimetry using a normalized dose to water approach (Dw,w) at the PTW standard laboratory. For the Farmer chamber dose measurement, the modelled Kmed correction factors were applied to the measured data and plan dose were compared to corrected and uncorrected measured dose. Four different algorithms from two different TPS, Eclipse (Varian Medical System, Palo Alto, CA, USA) and RayStation (RaySearch Laboratories, Stockholm, Sweden), were evaluated in this study. Table 2 lists algorithms and calculation methods based on the manufacturer’s specifications. All dose calculations in this manuscript, utilizing AAA, AXB, CCC and MC dose calculation algorithms, were performed using a 0.2 cm dose calculation grid size. The report of calculated doses by each algorithm in all tests has been based on the average of four voxels along the central axis.

Fig. 3. Simplified schematic of MC simulation to calculate the medium dependency correction factor (Kmed) for bone and lung in different dose calculation algorithms types (Dw,w and Dw,m). a: Absorbed dose measured by detector in water phantom, b: Absorbed dose to water in a water phantom, c: TPS calculated dose to high density water in Dw,w scenario, d: TPS calculated dose to Low density water in Dw,w scenario, e: Absorbed dose measured by detector in lung (Audit measurement condition for lung), f: TPS calculated dose to lung in Dw,m scenario, g: Absorbed dose measured by detector in bone (Audit measurement condition for bone), h: TPS calculated dose to bone in Dw,m scenario.(To calculate the dose in scenarios involving a detector, four central voxels within the chamber volume are utilized, each possessing a volume of 0.2 cm$^3$, designated for dose scoring.).
3.2. Algorithm correction factors for the PTW 30013 Farmer

The results of the calculated medium dependency correction factors for the PTW 30013 Farmer (K_{med} PTW 30013) type chamber by MC simulation are summarized in Table 4 for the selected tests of TECDOCE1583. For the Farmer chamber, the correction factors were determined from the average of 4 central voxels in the sensitive volume of the Farmer chamber. In both media (bone and lung) and for both energies (6 MV and 15 MV) medium dependency correction factors for \( D_{m,m} \) type algorithms were greater than medium dependency correction factors for \( D_{w,w} \) types algorithms.

Fig. 5 shows the percentage dose difference between the measured...
dose with the PTW 30013 Farmer ionization chamber and the calculated dose by the AXB, AAA, CCC and MC dose calculation algorithms for 6 MV photons for the selected tests of TECDO-1583. Results are shown in Table 4, before and after applying the $K_{med}$ PTW 30013 to the measured dose in lung and bone materials. Table 5 shows for 6 MV photons the mean percentage dose differences between the measured and calculated doses by the AXB, AAA, CCC and MC dose calculation algorithms, before and after applying $K_{med}$ PTW 30013 for lung material (point 9 of test 1, point 6 of case test, and point 7 of test 5 and 6) and bone material (point 10 of tests 1, 4 and 6). Note that before applying $K_{med}$ PTW 30013 to the AAA dose calculation algorithm in 6 MV photon beam measurements, the percentage dose difference between measured and calculated dose for Point 6 of field 3 and field 4 in test 4 were 4.4 % and 3.2 % respectively. These differences are greater than the acceptable tolerance level defined by TECDO-1583 for this test, which is 3 %. After applying $K_{med}$ PTW 30013, the percentage dose difference reduced to 2.9 % and 1.9 % for Point 6 of field 3 and field 4 in test 4 respectively. This is now within the tolerance value set by TECDO-1583. All other tests of TECDO-1583 evaluated in this study passed before and after applying the medium dependency factors for the AAA dose calculation algorithm. For AXB dose calculation algorithms in 6 MV photon beam, before applying $K_{med}$ PTW 30013, the percentage dose difference between measured and calculated dose for Point 10 of field 1 in test 1 and for point 10 of field 1 in test 4, failed by $-3.8$ and $-4.0$ % respectively. Again, these differences are greater than the acceptable tolerance level defined by TECDO-1583, which is 3 %. After applying $K_{med}$ PTW 30013, the percentage dose difference reduced to $-2.0$ % and $-1.9$ % for the same points in the same tests respectively, bringing the difference to within the TECDO-1583 tolerance level. All other evaluated cases of TECDO-1583 passed before and after applying medium dependency correction factors for the AXB dose calculation algorithm. In MC and CCC dose calculation algorithms all evaluated cases of TECDO-1583 passed before and after applying $K_{med}$ PTW 30013.

Table 5 shows for 6 MV the mean percentage dose differences between the measured and calculated doses by the AXB, AAA, CCC and MC dose calculation algorithms before and after applying $K_{med}$ PTW 30013 for lung material (point 10 of field 1 in test 1) and bone material (point 10 of tests 1, 4 and 6). Before implementing the $K_{med}$ PTW 30013 correction, the AAA dose calculation algorithm for 15 MV photons resulted in a percentage dose difference of $-3.1$ % between the measured and calculated doses at Point 10 of field 1 in test 1. This discrepancy exceeds the acceptable tolerance level of 3 % specified by TECDO-1583 for this test. After applying $K_{med}$ PTW 30013 to the measurement, the percentage dose difference reduced to $-2.0$ % for this

<table>
<thead>
<tr>
<th>Material</th>
<th>AAA Measured</th>
<th>AAA Corrected</th>
<th>AXB Measured</th>
<th>AXB Corrected</th>
<th>CCC Measured</th>
<th>CCC Corrected</th>
<th>MC Measured</th>
<th>MC Corrected</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lung</td>
<td>$2.11 \pm 0.2 %$</td>
<td>$1.25 \pm 1.65 %$</td>
<td>$0.22 \pm 0.98 %$</td>
<td>$-0.34 \pm 1.66 %$</td>
<td>$-0.55 \pm 2.15 %$</td>
<td>$-1.17 \pm 1.43 %$</td>
<td>$-0.17 \pm 2.43 %$</td>
<td>$-1.04 \pm 1.66 %$</td>
</tr>
<tr>
<td>Bone</td>
<td>$-0.43 \pm 0.87 %$</td>
<td>$0.10 \pm 0.77 %$</td>
<td>$-1.70 \pm 1.9 %$</td>
<td>$-0.73 \pm 1.23 %$</td>
<td>$-1.03 \pm 0.97 %$</td>
<td>$-0.25 \pm 1.05 %$</td>
<td>$-0.85 \pm 1.45 %$</td>
<td>$0.01 \pm 0.89 %$</td>
</tr>
</tbody>
</table>

Fig. 5. Percentage dose difference between measured dose with the PTW 30013 Farmer ionization chamber and calculated dose by the Anisotropic Analytical Algorithm (AAA), Acuros (AXB), collapsed cone convolution (CCC) and Monte Carlo (MC) dose calculation algorithms before (measured) and after (corrected) applying the medium dependency correction $K_{med}$ in 6 MV photons for tests 1, 4, 5, and 6 of TECDO-1583. The red lines within graph space indicate acceptance tolerance level for each test performed based on TECDO-1583 recommendation. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)
point, bringing the difference within the tolerance level. All other evaluated tests of TECDOC-1583 passed before and after applying $K_{\text{med}}$ PTW 30013 for the AAA dose calculation algorithm. Before incorporating the $K_{\text{med}}$ PTW 30013 into the measurement, the AXB dose calculation algorithm for 15 MV photons exhibited a percentage dose difference of $-3.1\%$ between the measured and calculated doses at Point 10 of field 1 in test 1. This discrepancy exceeded the acceptable tolerance level of 3 % defined by TECDOC-1583. After applying $K_{\text{med}}$ PTW 30013 to the measurement, the percentage dose difference reduced to $-1.15\%$ for this point, bringing the difference within the tolerance level. All other evaluated tests of TECDOC-1583 passed before and after applying $K_{\text{med}}$ PTW 30013 for the AXB dose calculation algorithm. Before the application of $K_{\text{med}}$ PTW 30013, the CCC dose calculation algorithm for 15 MV photons resulted in a percentage dose difference of 4 % between the measured and calculated doses at Point 6 of field 4 in test 4. This discrepancy exceeded the acceptable tolerance level of 3 %. After applying $K_{\text{med}}$ PTW 30013 to the measurement, the percentage dose difference reduced to 2.5 % for this point, bringing the difference within the tolerance level. All other evaluated tests of TECDOC-1583 passed before and after applying $K_{\text{med}}$ PTW 30013 for the MC dose calculation algorithm.

4. Discussion

The TECDOC-1583 report was prepared by the IAEA in 2008 for dosimetric audit of TPSs [2] and has been widely used for evaluating the accuracy of dose calculation algorithms in heterogeneous media by various studies and national audit committees [10–13]. TECDOC-1583 recommends using an ionization chamber for dose measurements in the 002LFC CIRS Thorax phantom. However, ionization chambers are usually calibrated in terms of absorbed dose-to-water, while the CIRS phantom has lung-equivalent media ($\rho = 0.21$ g/cm$^3$) and bone-equivalent media ($\rho = 1.60$ g/cm$^3$). Dose measurement in these media with an ionization chamber calibrated in water can cause inaccurate dose measurement and eventually lead to incorrect results for dosimetric audit based on TECDOC-1583.

In this study, a specific correction factor $K_{\text{med}}$ PTW 30013 for lung and bone measurement points of the CIRS phantom was calculated with Monte Carlo (MC) simulation for tests 1, 4, 5, and 6 of TECDOC-1583. For lung measurements in the CIRS phantom, the calculated $K_{\text{med}}$ PTW 30013 for ionization measurements were greater than unity for both 6 MV and 15 MV energies, which could be due to the rise in primary TERMA (total energy per unit mass (TERMA)) transmission through the lung materials in comparison to water materials [4]. This finding is
consistent with measurements reported by Zhu et al. [14] and Mauceri et al. [15]. For 6 MV photons, the minimum and maximum calculated $K_{\text{med}}$ PTW 30013 for lung media were 1.007 and 1.025 for point 9 of test 1 in $D_{\text{w},w}$ type dose calculation algorithms and for point 7 of test 6 in $D_{\text{m},m}$ type dose calculation algorithms, respectively. The minimum and maximum calculated $K_{\text{med}}$ PTW 30013 for 15 MV photons in lung media were 1.01 and 1.029, respectively, for the same minimum and maximum points and the same dose calculation algorithms type. The results of this study show that $D_{\text{m},m}$ type dose calculation algorithms generally need a higher $K_{\text{med}}$ PTW 30013 for lung media in comparison to $D_{\text{w},w}$ type dose calculation algorithms. To justify the need for higher correction factors in $D_{\text{m},m}$ algorithms compared to $D_{\text{w},w}$ algorithms, $D_{\text{m},m}$ algorithms mainly account for density, aligning with water, while $D_{\text{m},m}$ algorithms also take into account the atomic number (Z) of materials. This makes $D_{\text{m},m}$ algorithms more comprehensive in evaluating radiation interactions. Given that lung tissue has a different density and Z compared to water, $D_{\text{m},m}$ algorithms need a higher correction factor to account for these complexities [4]. No significant difference between 6 MV and 15 MV photon beams was observed for lung media’s $K_{\text{med}}$ PTW 30013.

For bone measurements in the CIRS phantom, the calculated $K_{\text{med}}$ PTW 30013 for ionization measurements were less than unity for both 6 MV and 15 MV energies, which could be due to the build-up, build-down, and rebuild-up effects [4]. This finding is consistent with measurements reported by Zhu et al. [14], Mauceri et al. [15] and Shaw et al. [16]. For 6 MV photons, the minimum and maximum calculated $K_{\text{med}}$ PTW 30013 for bone media were 0.992 and 0.968 for point 9 of test 4 (field4) in $D_{\text{w},w}$ type dose calculation algorithms and for point 10 of test 4 (field4) in $D_{\text{m},m}$ type dose calculation algorithms, respectively. The minimum and maximum calculated $K_{\text{med}}$ PTW 30013 for 15 MV beam in bone media were 0.994 and 0.970, respectively, for the same minimum and maximum points and the same dose calculation algorithms type. The results of this study show that $D_{\text{m},m}$ type dose calculation algorithms generally need a higher $K_{\text{med}}$ PTW 30013 for bone media in comparison to $D_{\text{w},w}$ type dose calculation algorithms. No significant difference between 6 MV and 15 MV photon beams was observed for bone media’s $K_{\text{med}}$ PTW 30013.

There were no significant differences observed in $K_{\text{med}}$ PTW 30013 between 6 MV and 15 MV photon beams for lung and bone media, implying minimal dependence of $K_{\text{med}}$ on energy fluctuations. The similarity in material properties (density and composition) at measurement points within the CIRS phantom for both energy levels might have led to insignificant variations in the correction factor. Another possible explanation is that the calculation of $K_{\text{med}}$ PTW 30013 involved relative dose measurements between ionization chamber readings and Monte Carlo calculated doses. Relative dosimetry, by its fundamental nature, can naturally alleviate energy-dependent effects, prioritizing proportional differences over absolute energy-dependent values. This inherent independence of Farmer as ionization chamber from energy variations highlights the strength and suitability of the $K_{\text{med}}$ factor across diverse photon energies [8].

After applying $K_{\text{med}}$ PTW 30013 for lung measurements with 15 MV beams, the mean percentage dose difference between measurements and calculations decreases from $0.37 \pm 1.03 \%$ to $-0.11 \pm 2.19 \%$, $0.65 \pm 1.05 \%$ to $-0.58 \pm 0.72 \%$ and $-0.71 \pm 2.09 \%$ to $0.38 \pm 2.02 \%$ for AAA ($D_{\text{w},w}$), AXB ($D_{\text{m},m}$) and MC ($D_{\text{m},m}$) dose calculation algorithms, respectively. The mean percentage dose differences in the same cases showed slight increase from $-0.42 \pm 2.28 \%$ to $-0.46 \pm 2.24 \%$ for CCC dose calculation algorithm.

For bone measurements, after applying the $K_{\text{med}}$ PTW 30013 in 6 MV photon beams, the mean percentage dose differences between the measurements and calculations decreased from $-0.43 \pm 0.87 \%$ to $0.1 \pm 0.77 \%$, $-1.7 \pm 1.9 \%$ to $0.73 \pm 1.23 \%$, $-1.03 \pm 0.97 \%$ to $-0.25 \pm 1.05 \%$, and $-0.85 \pm 1.45 \%$ to $0.01 \pm 0.89 \%$ for AAA ($D_{\text{w},w}$), AXB ($D_{\text{m},m}$), CCC ($D_{\text{w},w}$), and MC ($D_{\text{m},m}$) dose calculation algorithms, respectively. For 15 MV photon beams, the mean percentage dose difference decreases from $-1.06 \pm 2.04 \%$ to $0.71 \pm 1.29 \%$, $-1.55 \pm 1.55 \%$ to $-0.97 \pm 0.18 \%$, $-1.6 \pm 1.9 \%$ to $-1.31 \pm 0.99 \%$ and $-1.61 \pm 0.82 \%$ to $-0.31 \pm 1.39 \%$ for AAA ($D_{\text{w},w}$), AXB ($D_{\text{m},m}$), CCC ($D_{\text{w},w}$), and MC ($D_{\text{m},m}$) dose calculation algorithms, respectively. All dose calculation algorithms showed a lower mean percentage dose difference after the application of the bone media’s $K_{\text{med}}$ PTW 30013 to measurement for both energies (6 MV and 15 MV). The decrease in mean percentage dose difference was more significant for $D_{\text{m},m}$ type dose calculation algorithms in bone media.

According to TECDOC-1583, all designed tests in this report should pass through the dose calculation algorithm in the best possible situation. Using the calculated $K_{\text{med}}$ PTW 30013 in this study converted the results that failed to pass status in some tests, including: Point 6 of field 3 and field 4 in test 4 for the AAA dose calculation algorithm with 6 MV photon beams, Point 10 of field 1 in test 1 and Point 10 of field 1 in test 4 for the AXB dose calculation algorithm with 6 MV photon beams, Point 10 of field 1 in test 1 for the AAA, AXB and CCC dose calculation algorithm with 15 MV photon beams, Point 6 of field 4 in test 4 for the CCC dose calculation algorithm with 15 MV photon beams, and Point 10 of field 3 in test 4 for the MC dose calculation algorithm with 15 MV photon beams.

Modern radiotherapy uses a number of different dose calculation algorithms and reporting modes [17]. However, comparing dose calculation algorithms can be difficult if differences in measurements in bone and lung media are not properly addressed. This paper emphasizes the importance of addressing these differences when evaluating dose calculation algorithms in heterogeneous media. Monte Carlo simulations showed large $K_{\text{med}}$ when measurements of dose were made in lung and bone. The results of this study show that applying $K_{\text{med}}$ has a significant impact on the absolute dose differences between measurement and calculation during the dose audit process. The impact of the medium correction factor is also relevant for clinical treatment plans. Consideration should be given to verification measurements where bone and lung media are present in the target volumes or organs at risk. For end-to-end testing and patient-specific quality assurance (PSQA) where the phantom includes regions of lung like and bony-like material, we recommend calculating algorithm-specific corrections for detectors to improve measurement accuracy.

For institutions lacking access to Monte Carlo simulations or adequate computing resources, alternative strategies must be considered when encountering discrepancies beyond the acceptance threshold between measurements and calculations. In such instances, utilizing empirical correction factors verified on prior experimental data, utilizing simplified modeling approaches validated against benchmark measurements could offer pragmatic solutions for dosimetric audits in heterogeneous media [14–16]. To facilitate wider applicability in clinics conducting audits, future investigations might explore broader methodologies to accommodate these chamber-specific corrections across different phantoms, detectors and scenarios. Establishing standardized protocols or reference data for correction factors in varied clinical setups could aid in ensuring consistent and accurate dose estimations across...
5. Conclusion

In conclusion, this study investigated the impact of applying a medium correction factor, $K_{med}$ PTW 30013, on dose measurement accuracy in lung and bone media when evaluating dose calculation algorithms in heterogeneous media. The results showed that the $K_{med}$ values for lung measurements were generally greater than unity, while for bone measurements, they were less than unity. The application of $K_{med}$ led to changes in the mean percentage dose difference between measurement and calculation, bringing them closer to the real and correct values. The study also highlighted the importance of properly addressing the differences in measurements in bone and lung media when comparing dose calculation algorithms. Monte Carlo simulations played a crucial role in determining the correction factors. The findings emphasize the need for considering $K_{med}$ in the dose audit process and for verifying measurements in the presence of bone and lung media in treatment plans. Overall, algorithm-specific corrections for detectors are recommended to improve measurement accuracy in end-to-end testing and patient-specific quality assurance.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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