The dependence of inhomogeneity correction factors on photon beam quality index performed with the Anisotropic Analytical Algorithm

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Abstract

Purpose: The purpose of the study was to investigate the dependence of tissue inhomogeneity correction factors (ICFs) on the photon beam quality index (QI).

Materials and Methods: Heterogeneous phantoms, comprising semi-infinite slabs of the lung (0.10, 0.20, 0.26 and 0.30 g/cm\textsuperscript{3}), adipose tissue (0.92 g/cm\textsuperscript{3}) and bone (1.85 g/cm\textsuperscript{3}) in water, were constructed in the Eclipse treatment planning system. Several calculation models of 6 MV and 15 MV photon beams for quality index (TPR\textsubscript{20,10} = 0.670±k*0.01 and TPR\textsubscript{20,10} = 0.760±k*0.01, k = -3, -2, -1, 0, 1, 2, 3 respectively were built in the Eclipse. The ICFs were calculated with the anisotropic analytical algorithm (AAA) for several beam sizes and points lying at several depths inside of and below inhomogeneities of different thicknesses.

Results: The ICFs increased for lung and adipose tissues with increasing beam quality (TPR\textsubscript{20,10}), while decreased for bone. Calculations with AAA predict that the maximum difference in ICFs of 1.0% and 2.5% for adipose and bone tissues, respectively. For lung tissue, changes of ICFs of a maximum of 9.2% (6 MV) and 13.8% (15 MV). For points where charged particle equilibrium exists, a linear dependence of ICFs on TPR\textsubscript{20,10} was observed. If CPE doesn’t exist, the dependence became more complex. For points inside of the low-density inhomogeneity, the dependence of the ICFs on energy was not linear but the changes of ICFs were smaller than 3.0%. Measurements results carried out with the CIRS phantom were consistent with the calculation results.

Conclusions: A negligible dependence of the ICFs on energy was found for adipose and bone tissue. For lung tissue, in the CPE region, the dependence of ICFs on different beam quality indexes with the same nominal energy may not be neglected, however, this dependence was linear. Where there is no CPE, the dependence of the ICFs on energy was more complicated.

Key words: photon beam; quality index; dose calculation; AAA; inhomogeneity correction factors.

Introduction

To maximize the therapeutic benefit of radiation therapy, it is expected to deliver the prescribed dose with an uncertainty of better than 3.5%\textsuperscript{1}. There are several sources of uncertainties in dose delivery. The accuracy of the dose distribution calculation performed with a treatment planning system is one of the most important tasks. The analysis of uncertainties reviled that a 3.0% accuracy of dose distribution calculation yields to about 5.0% accuracy in dose delivery. The accuracy of dose distribution of 3.0% is difficult to be obtained, especially in the presence of different densities such as lungs, oral cavities, the teeth, nasal passages, sinuses, and bones\textsuperscript{2}. Therefore, the detailed commissioning and Quality Assurance (QA) of each treatment planning system (TPS) is essential. Accuracy of calculation models should be verified before the first clinical application\textsuperscript{3,4}. The most important part of the quality control (QC) is based on a comparison of measured and calculated dose distributions. For inhomogeneous areas, the most common method is to compare the calculated and measured the so-called correction factors (CFs) for inhomogeneities. The inhomogeneity correction factor (ICF) is defined as the ratio of the dose in an inhomogeneous medium and dose at the same point in a homogenous medium\textsuperscript{2}. This task is a very challenging, and time and resource-intensive. Some results of such measurements may be found in the literature\textsuperscript{2,8-11}. For such measurements, the most often block geometry, the geometry of precisely defined areas, blocks are used\textsuperscript{2,12-13}. In order to facilitate the QA procedure for a TPS, it would be convenient if ICFs measured by one user could be used by another. However, a question always remains as to whether the data taken from the literature are applicable to the actual user’s situation or not. In other words, how much the user beam...
differences influence the result of measurements, how much the quality index of a beam influence on the results.

Nowadays, mostly the accelerators produced by two vendors Varian and Elekta are used. One may expect that the spectrum of photon energies of accelerators from the same vendor with the same nominal energy differ a little only. Therefore, one may expect that the ICFs for the same nominal energy generated with the same vendor’s accelerator should not differ very much. If such a conclusion is true, it enables to use the results of other users for testing the treatment planning systems.

To aim of this work was to investigate how much ICFs depends on the beam quality determined by the tissue phantom ratio (TPR\textsuperscript{20,10}). These dependencies were obtained with calculations in the treatment planning systems and also with dosimetric measurements in commercial phantoms.

**Materials and Methods**

**Calculations**

In order to investigate the dependence of ICFs on the quality index (TPR\textsuperscript{20,10}), 6 MV and 15 MV photon energies have been considered. The range of TPR\textsuperscript{20,10} values was as follows:

\begin{equation}
TPR_{20,10}^{6} = 0.67 \pm k \times 0.01
\end{equation}

(6 MV, k is a multiplying factor with the values -3, -2, -1, 0, 1, 2, 3)

\begin{equation}
TPR_{20,10}^{15} = 0.76 \pm k \times 0.01
\end{equation}

(15 MV, k is a multiplying factor with the values -3, -2, -1, 0, 1, 2, 3)

The TPR\textsuperscript{20,10} values were consistent with the data obtained by the Cancer Center-Institute of Oncology for Polish radiotherapy departments. The ICFs were calculated in the Eclipse (Varian Medical System, Palo Alto, California, USA) TPS (version 13.6) with the AAA for 6 MV and 15 MV photon beam energies described by the quality indexes given above.

For each TPR\textsuperscript{20,10}, a separate set of percent depth doses (PDDs) were prepared. The PDDs were calculated with Gerbi’s formulae. The same profiles and output factors were used for all photon beams. These profiles and output factors for 6 MV and 15 MV beams with the TPR\textsuperscript{20,10} of 0.670 and 0.760 respectively were measured in the Cancer Center and Institute of Oncology in Warsaw, Poland. The Gerbi’s formulae enable to calculate the PDDs for the SSD of 100 cm. To ensure the correct data for the Eclipse system, the PDDs obtained for SSD = 100 cm were recalculated for 90 cm SSD using Mayneord’s factor (F).\textsuperscript{16}

The ICFs were calculated for corresponding given beam quality in a water phantom of 40x40x40 cm\textsuperscript{3} in which slabs of inhomogeneities were placed. Three types of slabs with different densities imitating lung tissue, adipose tissue, and bones were used. The calculation geometry is shown in Figure 1. In Table 1, the mass density, CT values (HU), and relative electron density (RED) of these inhomogeneous slabs were given. The build-up water slabs were 3 cm and 5 cm for 6 MV and 15 MV energy respectively.

| Material | Density (g/cm\textsuperscript{3}) | Hounsfield Unit (HU) | Relative Electron density |
|----------|----------------------------------|---------------------|--------------------------|
| Lung     | 0.100                            | -877                | 0.1230                   |
|          | 0.200                            | -778                | 0.2220                   |
|          | 0.260                            | -718                | 0.2861                   |
|          | 0.300                            | -679                | 0.3210                   |
| Adipose tissue | 0.920                   | -122                | 0.8784                   |
| Bone     | 1.850                            | 1488                | 1.7651                   |

Calculations were performed at several depths inside of and below inhomogeneities. Three thicknesses of lungs of 5, 10, and 15 cm were considered. The 5 cm slab thickness of adipose tissue and 3 cm of bone were considered. Calculations were performed for several field sizes of 5x5, 10x10, 15x15, 20x20, 25x25 and 30x30 cm\textsuperscript{2}. In addition, calculations were also performed for three different densities (0.10 g/cm\textsuperscript{3}, 0.20 g/cm\textsuperscript{3} and 0.30 g/cm\textsuperscript{3}) of lung tissue for a field size of 10x10 cm\textsuperscript{2}. The Source Skin Distance (SSD) was always 90 cm.

**Measurements**

ICFs were measured in a CIRS Thorax Phantom, Model-002LFC, (Computerized Imaging Reference Systems (CIRS), Inc, Virginia, USA) with two 6 MV photon beams generated in a Varian TrueBeam accelerator. The first beam was generated with a flattening filter (6 MV, TPR\textsuperscript{20,10} = 0.668) and the second one without a flattening filter (6 MV FFF, TPR\textsuperscript{20,10} = 0.632). The 250 MUs were delivered to measure the dose at points 5, 6, 7 and 8 for beams entered at 0°, 300°, 240° and 270° gantry angles (Figure 2). Two field sizes of 5x5 and 10x10 cm\textsuperscript{2} were used for measurements. The isocenter was kept constant. Point 5 was considered as the default point of measurement.

The dose was measured following the International Atomic Energy Agency (IAEA) Technical Report Series (TRS) 398...
protocol. The PTW (PTW Freiburg GmbH, Germany) Unidose (Sl. 11049) and 0.6 cm$^3$ farmer (Type-30013, Sl. 2698) ionization chamber were used for the measurements. Measurements were repeated three times to assess the reproducibility. All measurements were normalized to the dose measured at the default point for 10 x 10 cm$^2$ field with the gantry at 0°. The normalized doses were treated as the inhomogeneity correction factor. Finally, the measured inhomogeneity correction factor was normalized to higher energy, i.e., 0.668.

**Results and Discussions**

**Calculations**

In Figures 3a and 3b, the ICFs calculated for lung, adipose, and bone inhomogeneities with the AAA method for 6 MV and 15 MV X-rays, are presented. The data were normalized to the QI = 0.670 for 6 MV and to the QI = 0.760 for 15 MV.

For lung, ICFs decreased with increasing QI. For the adipose layer, the ICFs are nearly independent of energy. For the bone layer, ICFs increased with increasing QI. The ICFs changed by less than 2.5% (6 MV) & 2.0% (15 MV) and 1.7% (6 MV) & less than 1.0% (15 MV) for the lung and bone inhomogeneities respectively. An almost linear dependence of ICF on QI was observed.

In Figures 4a and 4b, for lung, calculation of ICFs was performed for several field sizes with the AAA method for 6 MV and 15 MV X-rays, are presented. The data were normalized to the QI = 0.670 for 6 MV and to the QI = 0.760 for 15 MV. For the lung, a very minor dependence of ICFs on field size was observed for 6 MV X-rays (Figure 4a), while a larger dependence was found for 15 MV X-rays (Figure 4b). For smaller the field size, larger difference of ICFs was observed and decreased with increasing the field size. For 6 MV, a linear dependence was observed for all field sizes, however, for 15 MV, the dependence showed a non-linearity in some extent.

In Figures 5a and 5b, the calculation of ICFs performed for several field sizes with the AAA method for 6 MV and 15 MV X-rays for bone inhomogeneity are presented. The data were normalized to the QI = 0.670 for 6 MV and to the QI = 0.760 for 15 MV. For bone, a very small dependence of ICFs on field size was observed as seen in Figures 5a and 5b. The smaller the field size the larger the ICF. For all field sizes, a linear dependence was observed for 6 MV, while for 15 MV, a non-linear dependence was revealed.

In Figures 6a and 6b, illustrated the dependence of ICFs on the three different thicknesses of lung inhomogeneity for a field size of 10x10 cm$^2$. The data were normalized to the QI = 0.670 for 6 MV and to the QI = 0.760 for 15 MV. The ICF depends on the thickness of lung tissue as seen in Figures 6a and 6b. The changes of ICFs is more pronounced for larger thicknesses of the lung slabs. For a 15 cm lung slab, the difference obtained between QI = 0.640 and 0.700 was almost 10.0% for 6 MV while for 15 MV, change of the QI from 0.730 to 0.790 led to change of the ICF of 13.8%.

![Figure 3. ICFs as a function of QI at P$_{dose}$ = 5 cm for 5 cm lung (0.26 g/cc) and adipose (0.92 g/cc), and 3 cm bone (1.85 g/cc) for 10x10 cm$^2$ field size for 6 MV (a) and 15 MV (b) photon energy. The absolute ICFs for the QI of 0.670 were 1.136, 1.024 and 0.923 for Lung, Adipose and Bone respectively. The absolute ICFs for the QI of 0.760 were 1.103, 1.015 and 0.941 for Lung, Adipose and Bone respectively.](image-url)

![Figure 2. CT scan image of the CIRS thorax phantom. Beams and points were measurements were carried out are shown. Each measurement was normalized to dose measured at Point 5.](image-url)
Figure 4. ICFs as a function of QI at P_{below} = 1 cm for 5 cm lung (0.26 g/cc) for 5x5, 10x10, 15x15, 20x20, 25x25 and 30x30 cm² field size for 6 MV (a) and 15 MV (b) photon beams. The absolute ICFs for the QI of 0.670 were 1.158, 1.136, 1.118, 1.108, 1.101 and 1.097 for 5x5, 10x10, 15x15, 20x20, 25x25 and 30x30 cm² field size respectively. The absolute ICFs for the QI of 0.760 were 1.052, 1.073, 1.071, 1.066, 1.062 and 1.059 for 5x5, 10x10, 15x15, 20x20, 25x25 and 30x30 cm² field size respectively.

Figure 5. ICFs as a function of QI at P_{below} = 1 cm for 3 cm bone (1.85 g/cc) for 5x5, 10x10, 15x15, 20x20, 25x25 and 30x30 cm² field sizes for 6 MV (a) and 15 MV (b) photon beams. The absolute ICFs for the QI of 0.670 were 0.916, 0.928, 0.935, 0.941, 1.0.946 and 0.949 for 5x5, 10x10, 15x15, 20x20, 25x25 and 30x30 cm² field size respectively. The absolute ICFs for the QI of 0.760 were 0.945, 0.952, 0.958, 0.959, 0.961 and 0.962 for 5x5, 10x10, 15x15, 20x20, 25x25 and 30x30 cm² field size respectively.

Figure 6. ICFs as a function of QI P_{below} = 1 cm for 5, 10 & 15 cm lung (0.26 g/cm³) for 5x5 cm² field size for 6 MV (a) and 15 MV (b) photon beams. The absolute ICFs for the QI of 0.670 were 1.144, 1.327 and 1.544 for 5, 10, and 15 cm thicknesses of lung inhomogeneity respectively. The absolute ICFs for the QI of 0.760 were 1.052, 1.160 and 1.292 for 5, 10, and 15 cm thicknesses of lung inhomogeneity respectively.
The differences between ICFs obtained for different densities (0.10, 0.20 and 0.30 g/cm³) at normalization QI (0.670 for 6 MV) and (0.760 for 15 MV) were less than 1.0%. For 6 MV beams, the dependence was almost linear. For 15 MV, these dependencies were not linear. It should be noticed that the point 1 cm below inhomogeneities, where calculations were carried out, was in the region where the charged particle equilibrium (CPE) doesn’t exist. The influence of ICFs inside of lung inhomogeneity is demonstrated in Figures 7a and 7b. The data were normalized to the QI = 0.670 for 6 MV and to the QI = 0.760 for 15 MV.

For 6 MV, ICFs increased with increasing the QI at 1 cm below build-up, with very little change with QI at the middle of the lung, and decreased with QI at 1 cm above the lung-water interface as seen in Figure 7a. The maximum difference was less than 1.0%. For 15 MV, ICFs decreased with increasing QI at all depths inside of lung tissue as seen in Figure 7b. The maximum difference of 3.0% of ICFs was obtained at 4 cm (1 cm above the lung-water interface).

**Measurements**

In Figures 8a-d, the normalized ICFs measured with CIRS thorax phantom for 6 MV and 6 MV FFF photon beams for a field size of 10 x 10 cm² are presented. The results obtained from measurements are consistent with the calculation results.

ICFs measured with CIRS phantom differs up to 2.5% at gantry angle 240° (Figure 8a), 1.0% at gantry angle 270° (Figure 8b), less than 1.0% at gantry angle 240° (Figure 8c) and 6.0% at gantry angle 270° (Figure 8d) for points 5, 6, 7 and 8 respectively.

In the previous work, ICFs for several situations were established and it was found that the maximum difference was not more than 5.0% for lung slabs across the considered range of QIs for 6 and 15 MV when calculated with the Batho Power Law method.

| Measurement Point | Gantry Angle | ICFs (0.632 6 MV FFF) | ICFs (0.668 6 MV) |
|-------------------|--------------|-----------------------|-------------------|
| P5                | 0°           | 1.000                 | 1.000             |
| P5                | 300°         | 1.021                 | 1.010             |
| P5                | 270°         | 1.039                 | 1.014             |
| P5                | 240°         | 1.079                 | 1.054             |
| P6                | 270°         | 0.751                 | 0.744             |
| P7                | 240°         | 0.767                 | 0.763             |
| P8                | 270°         | 1.627                 | 1.536             |

In this study, AAA was used to assess the dependence of ICFs on QIs. This algorithm not only correction of attenuation is taken into account but also electron transport. This is especially important in regions where no CPE exists. Calculation with AAA, both for 6 and 15 MV photon energies, ICFs decreased with increasing QI for lung cases. The lung is made up of soft tissue and has an equivalent atomic number similar to soft tissue. It contains alveolar air spaces and is, therefore, less dense than other soft tissues. The reduced density leads to decreased attenuation of the primary radiation beam, which in turn leads to decreased production of scattered radiation. ICFs remained relatively constant with QI for adipose tissue as the density is almost the same as water. While ICFs increased with QI for bone when calculated inside of and beneath heterogeneous slabs. Bone has a higher atomic number than soft tissues and the electron density is also higher. Bone causes increased attenuation for megavoltage beams. For megavoltage beams, the attenuation only occurs to the increased density of bone - there is less attenuation per gram of bone than per gram of water. Pair production increases at higher photon energy levels and bone mineral dose begins to increase again.
Across the considered range of QI, the ICFs changed by less than 2.5%, 1.0% and 1.7% (6 MV) and 2.0%, 1.0% and 1.0% (15 MV) for lung, adipose tissue, and bone heterogeneities respectively, when calculated at 5 cm depth below them for a field size of 10x10 cm$^2$ (Figure 4). An almost linear dependence of ICF on QI was observed.

For all inhomogeneous phantoms (lung, adipose, and bone), the highest differences of ICFs across the beam quality index range was observed for the smallest field size (5x5 cm$^2$) and at 1 cm below the inhomogeneity. The calculation point at 1 cm distance from the lung-soft tissue border is in the region where there is no CPE. Therefore, it may be concluded that, in tissue close to the interface, the ICFs are described by the more complicated function of the energy and distance than for points lying further away from the surface. A change of ICF with QI of about 4.1% (6 MV) and 6.2% (15 MV) for lung (Figure 4), and 1.8% (6 MV) and 2.4% (15 MV) for bone (Figure 5), when calculated below 5 cm of lung tissue and 3 cm bone tissue. These differences decreased with increasing field sizes. This is because of the different contributions of the scattered radiation for the total dose. For a 5 cm lung slab, the ICF variations with QI were 4.1% (6 MV) and 6.2% (15 MV) for a 5x5 cm$^2$ field size, while for a 30x30 cm$^2$ field size were 1.8% (6 MV) and 1.0% (15 MV) for a 30x30 cm$^2$ field size.

The changes were more pronounced if the thickness of heterogeneous slabs increases (Figure 6). Below a 5 cm lung slab, a variation in ICF of 4.1% (6 MV) and 6.2% (15 MV) was obtained across the range of QIs. For a lung slab of thicknesses 10 cm, the variations were 7.3% (6 MV) and 10.4% (15 MV). 10.0% (6 MV) and 13.8 (15 MV) changes were found for 15 cm thickness of the lung slab. The dependence of ICFs on the range of QI was also observed for lung tissue with different densities. The changes of ICFs with QI were higher for lower densities of lung slabs and decreased with increasing slab density. For megavoltage photon beams that rely on inherent scattering for most of their attenuation, the mass attenuation of many tissues is similar. This indicates that just the density of the inhomogeneities is important in the determination of the ICF factor.

For points where there is no CPE, a little more complicated dependence of ICFs over the range of QI was observed. This is for points located in the lung, and bone at 1 cm below build-up, in the middle and 1 cm above the interface of inhomogeneities (Figure 7 and Figure 8).

In this study, the inhomogeneity correction factors were also measured for the 6 MV photon beam energies. These were the beams generated in the Varian TrueBeam accelerator with and without a flattening filter. The QI of the flattening filter beam
was 0.668 and the QI of flattening filter-free beam was 0.632. Measurements were performed for a 10x10 cm² field size. All measurements were performed in the CIRS phantom with lung inserts. It was impossible to measure the correction factor fully according to the definition. All measurements were normalized to the measurements carried out in the homogenous part of the phantom. All these restrictions mean that the results obtained should be treated only as a qualitative confirmation of the results obtained by calculation. The largest difference of up to 6.0% was obtained for the beam at gantry angle 270° (Figure 8d) at point 8. For other points, smaller differences were obtained. For the beam, at gantry angle 240° at point 5, the change was 2.5% (Figure 8a). For point 7, at gantry angle 240°, the difference less than 1.0% (Figure 8c) and for the beam at gantry 270° at point 6, it was 1.0% (Figure 8b).

Measurement results confirmed results obtained from calculations. For the range of energies used in clinical practice, the influence of QIs on the inhomogeneity correction factors is small. It should be emphasized that the difference of QIs for flattening filter-free and flattening filter beams is larger than the difference of QIs of all 6 MV accelerators installed in Poland.

Conclusions

The ICFs measured by one user in a beam with given nominal energy could be used with caution by another beam with the same nominal energy. For energy (beam quality) difference of less than 1.0%, one may expect the difference of ICFs of less than 3.0%.

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