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Pretreatment verification of dose calculation and delivery by means of measurements with PLEXITOM™ phantom

Paweł Wołowiec[a,*], *Paweł Kukołowicz*[a,b], *Krzysztof Lis*[a]

[a] *Medical Physics Department, Holycross Cancer Centre, Kielce, Poland*
[b] *The Jan Kochanowski University of Humanities and Sciences, Kielce, Poland*

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ABSTRACT

Aim: To validate a pretreatment verification method of dose calculation and dose delivery based on measurements with Metaplex PTW phantom.

Background: The dose-response relationships for local tumor control and radiosensitive tissue complications are strong. It is widely accepted that an accuracy of dose delivery of about 3.5% (one standard deviation) is required in modern radiotherapy. This goal is difficult to achieve. This paper describes our experience with the control of dose delivery and calculations at the ICRU reference point.

Materials and methods: The calculations of dose at the ICRU reference point performed with the treatment planning system CMS XiO were checked by measurements carried out in the PLEXITOM™ phantom.

All measurements were performed with the ion chamber positioned in the phantom, at the central axis of the beam, at depth equivalent to the radiological depth (at gantry zero position). The source-to-phantom surface distance was always set to keep the source-to-detector distance equal to the reference point depth defined in the ICRU Report 50 (generally, 100 cm). The dose was measured according to IAEA TRS 398 report for measurements in solid phantoms. The measurement results were corrected with the actual accelerator's output factor and for the non-full scatter conditions. Measurements were made for 111 patients and 327 fields.

Results: The average differences between measurements and calculations were 0.03% (SD = 1.4%), 0.3% (SD = 1.0%), 0.1% (SD = 1.1%), 0.6% (SD = 1.8%), 0.3% (SD = 1.5%) for all measurements, for total dose, for pelvis, thorax and H&N patients, respectively. Only in 15 cases (4.6%), the difference between the measured and the calculated dose was greater than 3%. For these fields, a detailed analysis was undertaken.

Conclusion: The verification method provides an instantaneous verification of dose calculations before the beginning of a patient's treatment. It allows to detect differences smaller than 3.5%.

* *Corresponding author at*: Medical Physics Department, Holycross Cancer Centre, ul. Artwińskiego 3, 25-734 Kielce, Poland. Tel.: +48 41 367 4416; fax: +48 41 3674 090.
E-mail address: pawel.wolowiec@gmail.com (P. Wołowiec).

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1. Background

The dose–response relationships for local tumour control and radiosensitive tissue complications are strong. It is widely accepted that the accuracy of dose delivery of about 3.5% (one standard deviation) is required in modern radiotherapy.[18] This goal is difficult to achieve.[7,8] Many measures are necessary to minimize the uncertainty in dose delivery during patient treatments.[25] The sources of uncertainties may be divided into four areas: geometrical errors, dosimetry errors, human error (that may lead to both geometrical and dosimetry errors) and, finally, errors that arise directly from equipment.[19,21,23] To minimize geometrical errors, sophisticated measurements of reproducibility of the patient set-up combined with correction strategies are employed.[1,4] Typical human errors include irradiation of an incorrect patient or an incorrect site.[20] These errors are more likely to occur in very busy radiotherapy departments. A good example of an error linked to improper equipment operation is the Saragossa accident.[24] In many accidents human error plays an important role.[22] The uncertainty in dose delivery may be analyzed by reviewing the sequence of steps in the dose delivery chain.[2,17] Alternatively, it may be assessed during treatment using in vivo dosimetry.[5,16]

Systematic and random errors occur in treatment delivery. For many years, "manual" treatments rendered radiotherapy very open to random human errors, such as miss-read or miss-set parameters. Many of these were never noticed or recorded. The number of random treatment error rates decreased considerably when record and verification systems (R&V systems) were introduced.[20] However, even with sophisticated R&V systems, some systemic errors still occur.[6] For example at the Princess Margaret Hospital in Toronto from January 1, 1997, to December 31, 2002 there were 555 errors among 28,136 patient treatments. Eighty-seven errors were directly attributed to incorrect programming of the R&V system.[10] Likewise, at the University of Utah during a 1-year period, 38 errors out of 22,542 external beam treatments administered under their R&V were identified.[20] Most of them arose from incorrect manual transcription of radiotherapy treatment parameters from the planning system to the R&V system. Ideally, all systemic errors should be detected before the start of treatment. The correctness of dose calculations at the prescription point performed with sophisticated treatment planning systems (TPS) is often performed using an independent monitor units (MUs) calculation programme.[15] Calandrino published data from the implementation of an independent control of MU and distribution calculation, together with a check of data reported in the treatment chart.[3] He showed that their system, which was relatively effective in detecting systemic errors before starting the treatment, still missed a quarter to one third of errors. Furthermore, Calandrino's experience confirms the utility of in vivo dosimetry in detecting previously unnoticed systemic errors. This paper details our experience with the control of dose delivery and calculations at the ICRU reference point. The method relies on dose measurements, performed at the prescription point before the start of treatment, using a PLEXITOM^{TM} phantom. We present results for the 111 patients treated with photon beams in our centre.

2. Aim

To validate a pretreatment verification method of dose calculation and dose delivery based on measurements with Metaplex PTW phantom.

3. Materials and methods

The calculations of dose at the ICRU reference point ($ICRU_{Ref}$) performed with the treatment planning system XiO (CMS XiO – Release V4.40.00) were checked by measurements carried out in the PLEXITOM^{TM} phantom (PTW – Freiburg). The calculation algorithm used by TPS was generally FFT (fast Fourier transform) Convolution. Only in the case of the thorax region the calculation algorithm was superposition. The phantom (see Fig. 1) contains two eccentrically mounted rotary acrylic cylinders inside a solid acrylic block. The double rotation provides for quick and precise positioning of a detector along the central beam axis, as well as for the off-axis measurement within a perimeter of 12.2. The phantom is powered by two stepper motors remotely controlled by the TBA CONTROL UNIT (PTW – Freiburg) and by the MEPHYSTO software. The movement control allows for the positioning of an ion chamber with the accuracy of 0.5 mm. The size of the phantom top surface is 19.0 × 11.5 cm. The chamber may be positioned at depths ranging from 1.0 cm to 12.2 cm.

3.1. *Method of dose measurement at the ICRU reference point*

The dose was measured separately for each treatment field. All measurements were performed with the ion chamber ("0.125 ccm flex.", Type/Ser. – No. M31002 – 0594, Manufacturer: PTW – Freiburg, Germany) positioned at the central axis of the beam at the radiological depth and with the UNIDOS electrometer. The radiological depths were obtained from the treatment plan protocols. The phantom density differs from the density of water; therefore, the radiological depth was converted into an equivalent depth in the phantom according to the scaling factor recommended by the manufacturer of the PLEXITOM^{TM} phantom. The source-to-phantom surface distance was always set to keep the source-to-detector distance

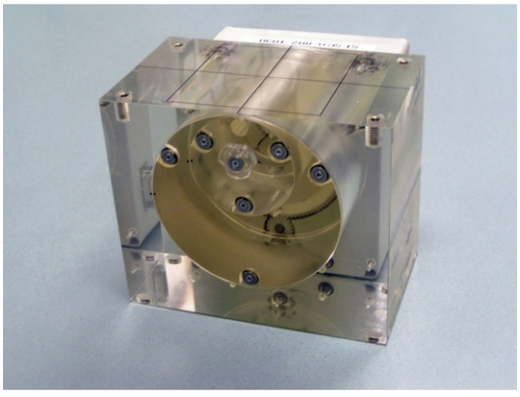

Fig. 1 – PLEXITOM^{TM} phantom.

equal to the reference point depth defined at the ICRU Report 50[13] (in most cases it was 100 cm).

Where the radiological depth was greater than the maximum attainable depth in the phantom (12.2 cm), additional acrylic plates of the thickness of 8 cm were placed on top of the phantom.

The dose was measured according to IAEA TRS 398 report[12] for measurements in solid phantoms. The results were corrected for the actual accelerator's output factor. They were also corrected for the absence of full scatter as the phantom dimensions do not provide full lateral scatter conditions for larger fields. Correction factors (*CF*) for non-full scatter conditions were measured separately for 6 and 15 MV photon beams. The *CF* was defined as follows:

$$CF \quad (d, X, Y) = \frac{D_{water}(d, X, Y)/D_{water}(ref)}{D_{plexitom}(d_{rad}, X, Y)/D_{plexitom}(ref)}, \tag{1}$$

where $D_{water}(d, X, Y)$, $D_{plexitom}(d_{rad}, X, Y)$ are doses measured in the water phantom and the PLEXITOM™ phantom, respectively, for open fields of (X, Y) size, at depth d, for 100 MU and SSD so that the source to chamber distance was 100 cm. $D_{water}(ref)$, $D_{plexitom}(ref)$ are doses measured at reference conditions according to IAEA TRS 398 report[12] (SSD = 90 cm, field size (10 cm, 10 cm), depth 10 cm) in the water phantom and in the PMMA phantom, respectively. In the PMMA phantom, physical depth was 8.8 cm which is equivalent to 10.0 cm in water. Measurements were made for square fields with sides of 5, 10, 15, and 20 cm at four depths 5, 10, 15, and 20 cm. To simplify the procedure, only one *CF* measured at 10 cm depth was used regardless of what the actual radiological depth of the $ICRU_{Ref}$ was.

The dose delivered to the measurement point for a field size (X, Y) at radiological depth d_{rad} was calculated as follows:

$$D(d_{rad}, X, Y) = M_{corr}(d_{rad}, X, Y) \cdot N_{D,w,Q} \cdot k_{Q,Q_0} \cdot \frac{OF(ref)}{M_{corr}(ref) \cdot N_{D,w,Q} \cdot k_{Q,Q_0}} \cdot CF(10, X, Y), \tag{2}$$

where M_{corr} is the measured signal corrected for temperature and pressure. $OF(ref)$ is the reference accelerator dose output (e.g. dose output put into treatment planning system), $CF(10, X, Y)$ is correction factor for non-full scatter conditions at depth of 10 cm. $M_{corr}(ref)$ was measured in the PMMA phantom at 8.8 cm physical depth for 10 cm square field size. $OF(ref)$ was measured in water at 10 cm depth for 10 cm square field size. $M_{corr}(ref)$ and $M_{corr}(d_{rad})$ was measured on the same day.

Because the measurements carried out to determine the correction for accelerator dose output and the verification of patient treatment fields were performed with the same ionization chamber, we can omit $N_{D,w,Q}$ and $k_{Q,Qo}$ in Eq. (2), which gives:

$$D(d_{rad}, X, Y) = M_{corr}(d_{rad}, X, Y) \cdot \frac{OF(10, 10, 10)}{M_{corr}(8.8, 10, 10)} \cdot CF(10, X, Y), \tag{3}$$

where $OF(10, 10, 10)/M_{corr}(8.8, 10, 10)$ is the correction factor for the actual accelerator dose output. The measurement uncertainty of a dose at a radiological depth is smaller than 1.5% (1SD).

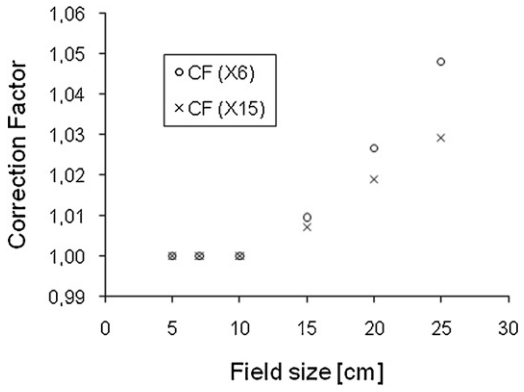

Fig. 2 – Field size correction factors for non full scatter conditions for 6 and 15 MV photon beams determined for d_{rad} = 10 cm.

3.2. *Measurements for patients*

The data include measurements of doses at the $ICRU_{Ref}$ carried out within 6 months for each patient treated with a radical intent. Measurements were made on the Siemens Oncor linear accelerators for each photon treatment field. A total of 327 fields for 111 patients were involved with tumours in the head, neck, lung, thorax, and the pelvic areas. In order to set up the treatment field, the field parameters were loaded from the R&V system (Lantis) to an accelerator. Measurements were always performed when the gantry was set to 0°. The measurement for a single field was compared with the $ICRU_{Ref}$ dose calculated with the treatment planning system. Measurements were not performed if the $ICRU_{Ref}$ was not located on the beam central axis. The difference (Diff) between the measured and calculated dose was calculated using Eq. (4):

$$Diff = \left(\frac{D_{meas}}{D_{calc}} - 1\right) \cdot 100\%. \tag{4}$$

In addition, for patients for whom the measurements for all treatment fields were made, the total dose to $ICRU_{Ref}$ was calculated as a sum of all doses for single fields and compared with the prescribed dose.

4. Results

4.1. *Correction factors for non-full scatter conditions*

Fig. 2 shows correction factors for non-full scatter conditions. Only data at radiological depth of 10 cm are presented. For fields smaller than 15 cm × 15 cm and all depths the correction factors differ from the ones obtained at 10 cm depth by less than ±0.01. Only for field sizes larger than 15 cm × 15 cm and for depths larger than 15 cm do the correction factors differ by ±0.02. Therefore, regardless of field size and depths, correction factors measured at the depth of 10 cm were used.

Table 1 – Results of pretreatment dose verification. The mean and standard deviation (SD) values were calculated from results obtained for patients from one location.

	Head and neck	Thorax	Pelvis	All	Total dose
Number of patients	30	35	46	111	92
Number of fields (measurements)	79	102	146	327	282
Mean (%)	0.3	0.6	0.1	0.3	0.3
Standard deviation (%)	1.5	1.8	1.1	1.4	1.0
Number of deviations >3%	2	11	2	15	0
Min–max difference (%)	−3.6 to +3.2	−4.9 to +5.3	−2.2 to +3.9	−4.9 to +5.3	−2.6 to +2.9

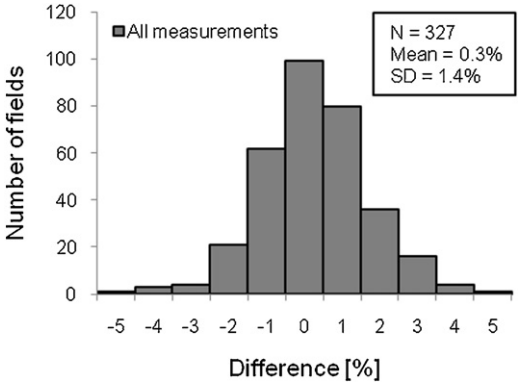

Fig. 3 – Differences between measurements and calculations for all fields.

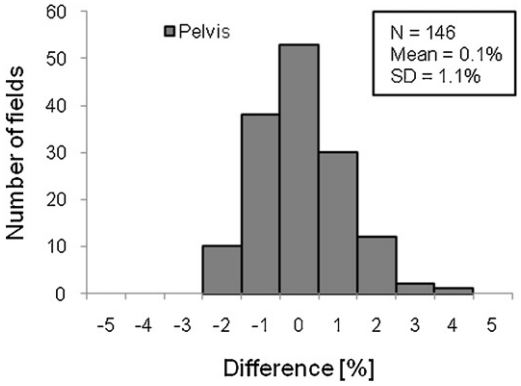

Fig. 4 – Differences between measurements and calculations for pelvis.

For 15 MV photon beams, correction factors are smaller than for 6 MV photon beams. The correction factors are an increasing function of the field sizes. For the largest fields for which measurements were performed, the correction factor is close to 1.05. It would be beneficial to design a larger phantom.

4.2. *Dose measurements at the ICRU reference point*

Table 1 presents information on pre-treatment measurements. The results for head and neck (H&N), thorax and pelvis regions are given separately. In the last column, the comparison of prescribed dose at $ICRU_{Ref}$ and the sum of all the doses measured at this point are compared for each patient. The total dose comparison is presented only for these patients (92 out of 111 patients) for whom the doses for all treatment beams were measured. The largest discrepancies between measured and calculated doses were observed for the thorax region. For 11 out of 102 thorax fields, the difference was larger than 3%. In two other locations, the differences exceeded 3% in the case of only 4 beams (4 out of 225). If the total dose is considered, the discrepancy between the measured and the calculated was always smaller than 3%.

Fig. 3 shows a histogram of differences between measured and calculated doses for all fields. Figs. 4–6 present the differences between measured and calculated doses for the pelvis, thorax and head and neck. The histograms have a Gaussian shape.

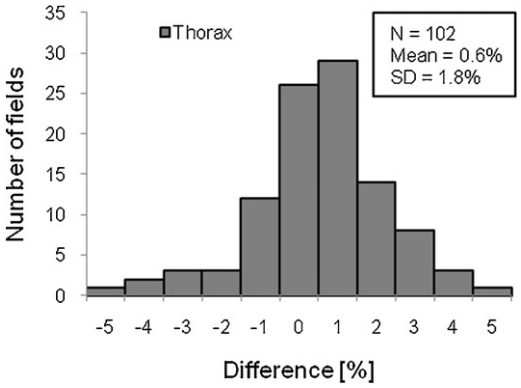

Fig. 5 – Differences between measurements and calculations for thorax.

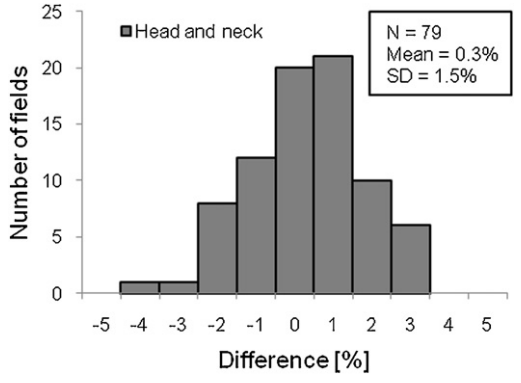

Fig. 6 – Differences between measurements and calculations for head and neck.

5. Discussion

As radiotherapy treatment becomes more sophisticated, its verification becomes more complex. This highly sophisticated

radiotherapy results in the increasing number of possible errors.[3,5,9,14] Errors leading to the administration of a wrong total dose, identified after the start of the treatment, are estimated at about 30% of all detected errors.[3] Thus, elimination of these errors is of crucial importance. In this work, we describe a method to control dose calculation and delivery before the start of therapy.

The results for 111 patients and 327 treatment fields showed a good agreement between measurement and doses calculated with the treatment planning system. The average differences between measurements and calculations were 0.03% (SD = 1.4%), 0.3% (SD = 1.0%), 0.1% (SD = 1.1%), 0.6% (SD = 1.8%), 0.3% (SD = 1.5%) for all measurements, for total dose, for pelvis, thorax and H&N patients, respectively. In only 15 cases out of the 327 fields (4.6%), the difference between the measured and the calculated dose was greater than 3%.

For these 15 fields, a detailed analysis was made in order to identify possible sources of the differences. The largest number (11) was observed in the thorax region. In this location, the dispersion of results is also the largest (−4.9% to +5.3%). It was observed that the larger differences were observed for all these cases were the differences between the physical and radiological depth were large. The largest difference between the radiological and physical depth was 5.9 cm. However, large differences were also obtained in two cases for patients with tangential fields for breast tumours. In these cases, the calculated and measured dose differences were −4.9% and 3.7%. The radiological depths were very small (1.4 cm and 2.2 cm respectively). Tests performed before admission of the TPS XiO system to clinical use showed that the accuracy of system calculations was the poorest at depths close to maximum. In the measurements reported here, a large difference was also obtained for one patient treated in the pelvic region. In this case, the radiological depth for lateral fields was larger than 20 cm. The treatment beams also passed through thick femoral heads and some pelvic bone.

The last two fields, for which differences were greater than 3%, were in the head area. In one case the difference was 3.2% and we did not identify a specific reason that might account for such a large difference. In the other case, there was a lack of lateral scattered radiation (the tumour was located near a skin surface) which could affect the accuracy of calculations.

In some cases, an additional factor influencing the difference between the measured and calculated doses might be oblique incidence of the treatment field. The surface of the absorber was flat, which changed the scattered conditions – the contribution of the scattered dose to calculated and measured dose was a little different.

In many radiotherapy centres, dose delivered from a single field is controlled with in vivo dosimetry. A measurement from the in vivo detector on a patient's skin surface is compared with the calculated dose at depth.[11] This method allows for detection of most large errors made in dose calculations and during dose delivery. However, there are many limitations of this method, one being its rather low sensitivity to smaller differences. As ESTRO notes,[11] most of radiotherapy departments have a 5% fixed tolerance level for in vivo dosimetry. Thus, differences between measured and calculated doses of less than 5% are regarded as acceptable. A second limitation is a rather small specificity of in vivo dosimetry, which, in practice, gives a large number of false positive results. Patient movement (e.g. respiratory motion) may influence the measurement. Also, it may be quite difficult to place a detector accurately in some cases (e.g. presence of hairs).

The action level of in vivo dosimetry is most often of 5%.[11] The dose verification method proposed in this paper allows for a lower action level, namely 3.5%. Another advantage of the method is that a simple analysis of the geometrical situation allows a determination of whether the difference between measurement and calculation results leads to an overestimation or underestimation of the measurement result. Thus, when a measurement exceeds the action level, a relatively simple review of the clinical situation and measuring system will, in many cases, allow one to decide whether to accept the result or to proceed with a more detailed analysis. The approach here does not extend the treatment time, because measurements are made prior to therapy being initiated. The time of a single measurement for one field was, typically, about 5 min. It could be much shorter with improvements to the phantom.

The most important advantage of the method is that it allows for detection of errors before the treatment commences. Its disadvantage is that it does not allow for detection of errors made in SSD determination. However, it should be noted that, although in vivo dosimetry is considered to be a method allowing for detection of SSD errors, in practise only large differences in SSD, those amounting to more than a few centimetres, can be discovered. It is much easier to check the appropriateness of an SSD setting by visual inspection. Due to the fact that the measurement is made for gantry angle 0° it is not possible to check the angular instability of the dose rate. To the best of our knowledge, the angular instability of the dose rate of modern accelerators has never been discovered with in vivo dosimetry. An important limitation of the proposed method is the impossibility to verify the use of the proper immobilization equipment, bolus or plates for blocks.

Other advantages of the method described here arise from the need to take account of the instability of an accelerator. Therefore, this method is an ideal indirect tool for verifying the accuracy of calculations of a treatment planning system at the beam central axis. Our results show that the XiO treatment planning system calculates the central axis dose with a high accuracy. In addition, with this method, if something goes wrong one can repeat measurements in the same dosimetry session. In case of in vivo dosimetry, measurements cannot be repeated until the next fraction. Finally, an extremely valuable feature of the proposed method is the ability to verify if a given dose differs significantly from the prescribed dose, even where in vivo measurement is not possible, e.g. PA fields.

If not permanently, it may be used as a quality control method at the beginning of the clinical use of a new treatment planning system.

6. Conclusions

Dose verification measurements at radiological depth by means of a PLEXITOM™ phantom can be recommended as a powerful tool to improve safety of radiotherapy, especially in centres where the treatment is carried out under control of

a R&V system. Being a sensitive method, it allows detection of differences smaller than 3.5%. The method provides instantaneous verification of dose calculations before the beginning of treatment. The greatest advantage of the dose verification with the PLEXITOM™ phantom is that it allows verification of dose calculation and delivery directly to the ICRU reference point. Additionally, the method provides an indirect verification of the accuracy of calculations performed with a treatment planning system. Our results show that XiO system calculates doses on the central axis with high accuracy. The disadvantage of the proposed method is that it does not allow for verification errors resulting from mistakes in determining the SSD or mistakes resulting from an improper use of immobilization tools and boluses.

Conflict of interest

There were no financial and/or personal relationship with other organizations and people that could influence this work.

Financial disclosure

None declared.

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