How the implementation of an *in-vivo* dosimetry protocol improved the dose delivery accuracy in head and neck radiotherapy

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Doses were measured *in-vivo* at the entrance using semiconductor detectors for patients with head and neck tumors. Regular measurements started on January 1st and continued till June 30th, 2001. Then the evaluation of the discrepancies between the measured and calculated doses was made, and it resulted in the changes in the protocol of *in-vivo* dosimetry made effective by February 1st, 2002. The collection of the measurements was censored by January 30th 2003. The number of patients in the two groups was 285 (1st) and 407 (2nd), respectively.

The results exhibited the not-Gaussian distribution of the measured doses in both groups. The average number of dose checks per patient increased from 4.9 to 6.0 (1st vs. 2nd group). The mean relative difference between the measured and calculated doses was: –1.5% vs. –0.5%, whereas the standard deviation (1 SD) decreased from 6.1% to 5.6%. The Mann-Whitney U test detected a significant difference between the 1st and the 2nd group (p=0.00000), which justified the conclusion that the changes implemented in the protocol improved dose delivery accuracy and reproducibility of irradiation.

Key words: in-vivo dosimetry, semiconductors, head and neck radiotherapy

In many papers a strict relationship between the probability of local tumor control, the normal tissue reactions and the total absorbed dose has been pointed out. The uncertainty of dose delivery in standard radiotherapy should not exceed 3.5% (1 SD) [6, 7, 11, 15, 16]. Recent studies have demonstrated that even higher accuracy can be achieved (1–2%, 1 SD) [2]. The head and neck radiotherapy requires higher accuracy due to close vicinity of the target and critical tissues. The sources of potential errors are numerous and may arise during different steps of the patient preparation and then during the execution of the irradiation [9, 10, 12]. The quality assurance protocol which considers all procedures in the radiotherapy is the key point in improving dose accuracy and then clinical results. The *in-vivo* dosimetry seems to be a crucial part of this protocol [5, 13]. The importance of the *in-vivo* dosimetry has been demonstrated in many papers since the thirties of the last century [8, 9, 10]. Semiconductor and thermoluminescent detectors have been dominantly used for the *in-vivo* dose measurements [1, 14, 17].

The aim of this study was to check if the implementation of an institutional in-vivo dosimetry protocol influenced and improved the dose delivery accuracy and reproducibility of the irradiation in the group of head and neck patients.

Material and methods

For this study, the results of *in-vivo* dose measurements have been evaluated for the head and neck radiotherapy. Dose measurements collected during two periods of time were evaluated. The first group of 285 patients was treated from January 1st till June 30th, 2001, and the second group consisted of 407 patients treated from February 1st, 2002 till January 30th, 2003. Patients were not randomised nor selected in any way. The only criterion of inclusion was the time when radiotherapy was performed, which means that all head and neck patients had the dose checks made and then they were included into this evaluation.

The treatment was performed using linear accelerators (Siemens Mevatron KD2, GE Saturn 43F, Varian Clinac 2300C/D) with photon energy of 6 MV and the Theratron Co-60 unit.

Diode calibration. The semiconductor detectors con-

nected to a Scanditronix DPD-510 multi-channel dosimeter were used for dose measurements in both groups. In the first group, Scanditronix diodes type EDE-s (Co-60) and EDP-10 (4–8 MV) were used. In the second group Sun Nuclear Isorad-p diodes suitable for photons 1–4 MV and 6–12 MV were used. Electron equilibrium was achieved using appropriate build-up caps. The diodes were calibrated at the linacs and at the cobalt unit, respectively. The calibration included standard geometry: field size of 10x10 cm^2 at the source-surface distance (SSD) of 100 cm for the linac and of 80 cm for the Cobalt unit. The detectors were positioned at the front surface of the PMMA phantom on the central axis of the beam (CAX). The calibration dose delivered to the semiconductors was 2 Gy. The time of the exposure was calculated using the treatment planning system CadPlan 3.1.2 for the above conditions. The calibration factor *CF* was calculated for each detector from the following expression:

$$CF = D_{cal}/R_{cal} \quad (1)$$

where D_{cal} was the dose calculated (equal to 2 Gy) and R_{cal} was the dosimeter reading. The *CF* factor was stored in the memory of the electrometer and then automatically added to the readings during the *in-vivo* measurements, producing the doses D_m already expressed in grays.

For the second group of patients the measured doses were also corrected by another factor CF_A which accounted for the individual properties of each accelerator, including certain differences between the photon energies at a fixed megavoltage of 6 MV for different linacs, especially those produced by different manufacturers. The CF_A was defined as the ratio of the reading obtained from the applied machine R_{mA} to the reading of the diode calibrated with the reference accelerator R_{ref}.

$$CF_A = R_{mA}/R_{ref} \quad (2)$$

Finally, the measured *in-vivo* dose D_m was equal to the electrometer reading multiplied by the factors CF and CF_A.

The calibration procedure was scheduled for every two weeks but repeated more frequently whenever the stability of the accelerators was uncertain or unspecified errors in the *in-vivo* measurements occurred.

Dose in-vivo measurements, calculation and the comparison. The *in-vivo* measurements were carried out only for the beam entrance. The appropriate semiconductor detector was positioned on the body of the patient or on the surface of the mask on the central axis of the beam. The exact place of this dosimetrical point was marked during the patient simulation process and then verified before the irradiation using positioning lasers on the accelerator or the cobalt unit.

The reading obtained from the dosimeters was recalculated using calibration factors to the dose on the surface (at the build-up), which had to be recalculated to the dose at the reference point at a reference depth. The recalculation process used tabularised data of the tissue-phantom ratio, field size dependence and tissue inhomogeneity using the density power ratio.

The following parameters were evaluated: N – the mean number of dose checks per patients, R – the mean relative difference (in the groups) between the measured D_m and calculated D_c doses (formula 3), and SD – standard deviation.

$$R = (D_m - D_c)/D_c \times 100\% \quad (3)$$

The calculated dose D_c was specified at the ICRU point at a certain depth on the central axis, which usually corresponded to the centre of the target. It was calculated using the treatment planning system CadPlan 3.1.2. The equivalent tissue-air ratio (EqTAR) algorithm was used [4].

Protocol requirements and changes. The protocol developed for the first group required that doses had to be measured during the first week of the treatment. The next measurement was performed on the request of the physician or whenever modifications of the fields were made.

After the dose data from the first group were evaluated, the second dose check was added in the middle of the radiotherapy course as being mandatory in the protocol, regardless the modifications of the fields. In all fields doses were measured excluding certain specific procedures.

For the first group of patients, the measurements were performed by the physicists and for the next group by the radiographers, who were trained and dedicated to this job. One radiographer per shift per five therapeutic machines was employed.

The action level which included repeated measurement and careful evaluation of the whole procedure was R=7% (relative difference between the measured and calculated doses) for the first group and then it was lowered to 5% for the second group, respectively.

To asses the influence of the significantly excessive doses on the final results all the measured doses larger than those calculated by more than 3 relative standard deviations were excluded. Then the values of the mean Rs and SDs were compared to those obtained for all data.

Results

The total number of measurements was 1346 for the 1st group and 1782 for the 2nd.

The results for the two groups were plotted on the histograms (Fig. 1, Fig. 2) as the frequency distribution of R.

The number of dose checks per patient was N= 4.9 vs. 6.0 (for the 1st vs. 2nd group). The mean difference between the measured and calculated doses (Mean R) was –1.5% vs. –0.5%. The standard deviations (1 SD) were 6.1% vs. 5.6%. The Shapiro-Wilk and Kolmogorov-Smirnov tests showed not-Gaussian distributions (dissymmetry). The Mann-Whitney U test revealed a significant difference between the 1st and the 2nd group (p=0.00000).

Only 10 vs 18 measured doses exceeded the doses calculated by more than 3 relative standard deviations which corresponded to 0.7% vs 1.0% (1st vs. 2nd group). The statistical analysis performed for the measured doses excluding those exceeding 3 SD showed that the mean R was –1.5% vs. 0.4% and SD 5.3% vs. 4.4%. The Mann-Withney U test also revealed a significant difference between the 1st and 2nd group (p=0.00000).

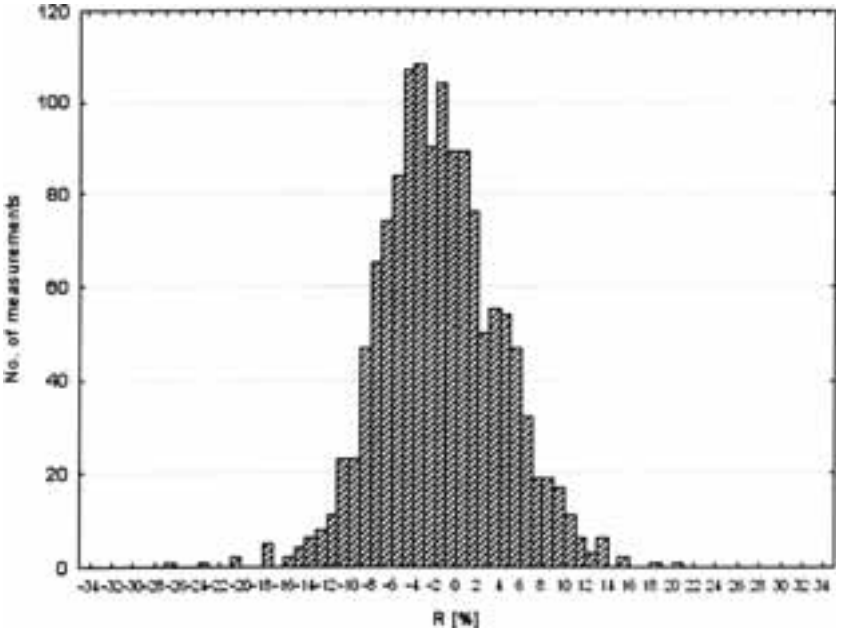

Figure 1. The number of measurements for each range (step of 1%) of differences between the measured and calculated doses for the 1st group.

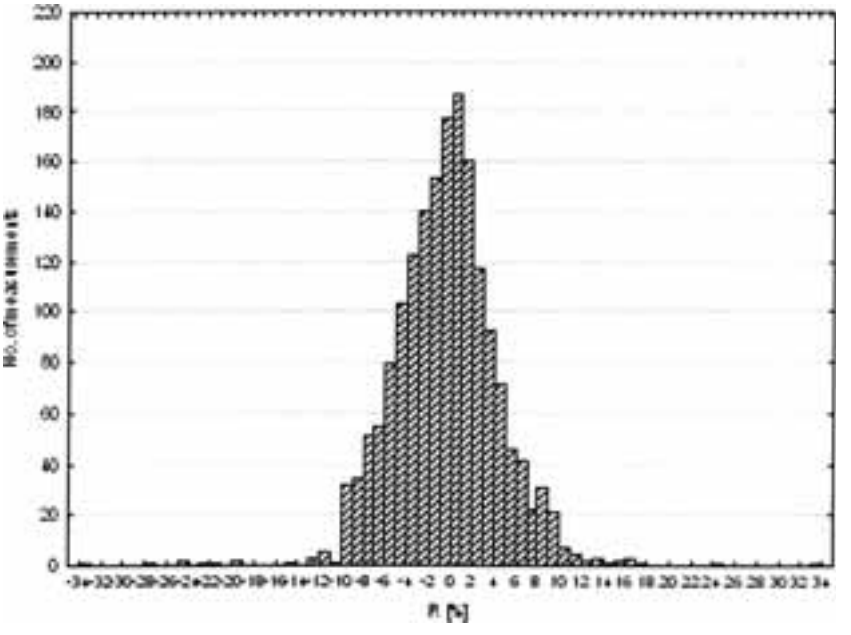

Figure 2. The number of measurements for each range of differences (step of 1%) between the measured and calculated doses for the 2nd group.

Discussion

The fact that the mean R differed from zero in both groups might indicate a systematic error. It might suggest that small underdosage occurred.

The statistically significant decrease in R between the first and the second group (from –1.5% to –0.5%) may indicate the improvement in dose calibration. Therefore, it was rather the imperfect method of dosimetry than inaccuracy of the dose planning resulting in underdosage that was at play. For the second group of patients another correction factor was implemented, which was assumed to improve the dose delivery accuracy and repeatability.

The standard deviation (1 SD) of 6.1% vs. 5.6% in the 1st vs. 2nd group was unsatisfactory, although some improvement was achieved. The standard deviation described the reproducibility of the procedure. The more frequent checks enforced by the changes in the protocol (N= 4.9 vs. 6.0) probably caused that a process of irradiation was more carefully applied. The data of *in-vivo* dosimetry presented in the literature lead to similar results for the mean R values but vary when standard deviation is concerned. The prospective studies made for a strictly defined group of patients showed lower SD values (below 3.5%), while the retrospective evaluation indicated rather higher SD values. Since the number of patients evaluated in the presented study was quite high, it indicated that the results albeit not quite satisfactory were realistic.

The main advantage of the semiconductor detectors used was the instant reading of the dose during the treatment. However, the readings had to be recalculated to the reference point and then interpreted against the previously calculated doses, which is also a time consuming procedure. Nevertheless, the advance in technology made semiconductor detectors a very useful and reliable tool in comparison with thermoluminescent dosimeters. However, the commercially available semiconductor diodes sometimes did

not provide sufficient build-up for *in-vivo* dosimetry [3]. No correction for this effect was considered to be necessary in our study, which might also contribute to the systematic error observed.

Only a very small fraction of the excessive measurements (0.7% vs. 1.0%) showed that the implementation of the *in-vivo* dosimetry protocol created the situation in which significantly erroneous situations were successfully eliminated.

In conclusion, the implementation of the changes in the *in-vivo* dosimetry protocol resulted in:

1. the increase in the workload, in terms of the number of dose checks per patient from 4.9 to 6.0 (1st to 2nd group),

2. the increase in the accuracy of total dose delivery (decrease in the systematic error) in terms of the mean difference between the measured and calculated doses decreasing from –1.5% to –0.5%,

3. the improvement in the reproducibility of irradiation, in terms of the decrease in the standard deviation from 6.1% to 5.6%.

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