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MEDICAL PHYSICS

IN VIVO DOSIMETRY MEASUREMENTS FOR BREAST RADIATION TREATMENTS

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Abstract. *In vivo* dosimetry is a vital tool in quality assurance program. It is very important to check the dose delivered for each patient undergoing radiation treatment in order to avoid over or under dosing. The aim of this paper is to present a study for quality control protocol using *in vivo* dosimetry for breast cancer radiation treatment. We report one year estimation of the radiation treatments accuracy, expressed as a comparison between delivered and prescribed doses. Dose delivered were measured by *in vivo* dosimetry in order to detect any systematic errors that may have escaped the different check performed in treatment plan preparation, data transfer or patient set-up. The diodes detectors used in this study were tested, calibrated and corrected in order to be accepted for treatment measurements. For each radiation field used in treatment a comparative study between calculated and expected dose on the patient' skin were performed. Depending of the observed discrepancies, the corrective actions and the decision levels were set up. During this study we investigated 634 radiation photon fields, in 1968 measurements, 66 potential errors being detected. Dosimetry checks show that only 3 of them could have induced a variation, over 5% in the dose delivered.

Key words: radiotherapy, quality control, *in vivo* dosimetry**,** diodes.

1. INTRODUCTION

In vivo dosimetry (IVD) represents a direct measurement of the dose delivered to the target volume in the radiotherapy. In order to limit the errors during the radiation treatment, some international and national organizations such as European Society for Therapeutic Radiology and Oncology (ESTRO) [1], American Association of Physicists in Medicine (AAPM) [2], French National Institute of Cancer (INCA) [3] and Romanian National Commission for Nuclear Activity Control (CNCAN) [4] have recommended the implementation of quality control (QC) program to achieve the accuracy and precision of treatments in radiotherapy.

In Romania, the CNCAN recommended since 2004 to verify the radiotherapy treatment delivered using *in vivo* dosimetry, but this technique is not actually used in routine. In France, INCA imposed the implementation of IVD for all measurable beams in the all radiotherapy centers.

The aim of this study was to present a QC program using IVD for breast cancer treatment radiation. The study investigates the accuracy of treatments delivered for 160 breast cancer patients (634 radiation photon fields) treated in the Radiotherapy and Oncology Center, Blois, France, in the period of one year (2010), based on our the previous experience gained in the case of prostate permanent ^{192}Ir implants [5]. Accordingly, the first step of this study was devoted to calibrate the diode's answers [6, 7] while in the second step have investigated any systematic errors in the dose delivery of breast cancer treatments [8–12].

The errors made during the treatment steps can be detected using measurements of entrance dose and compared with the expected one. Therefore, different action levels were defined in this study with action taken to correct the detected errors, by taking into account that the maximal error level was set-up at 5%.

2. MATERIALS AND METHODS

2.1. PHOTON SOURCE AND DETECTORS

All measurements were performed in photon radiation beams generated by Siemens Primus accelerator. *In vivo* detectors used in this study were P10 and P30 hemispherical Apollo diodes with integrated build-up cup (thickness 10 mm and 30 mm respectively) adequate for photon beams energy (6 and 25 MV). Detectors were connected to the Apollo 5 electrometer with resolution of 1 cGy. Diodes calibrations were performed using a slab phantom and the PTW dosimetry system (Unidos E electrometer, Farmer chamber and water phantom).

2.2. DETECTOR CALIBRATION

A large number of stability and linearity tests (about 200) were performed in order to accept the diodes in the clinical filed. Consequently, the stability of response, dose linearity as well as calibration factors were weekly checked as part of quality control process. Moreover, to avoid any influence of dose accumulation in diode response, factor calibration was adjusted weekly.

Finally, the calibration was performed for each diode against the ionization chamber. For this purpose, each diode has been placed on the central axis of the beam on the surface of 30 cm thick slab phantom. For a field size (FS) of 10 cm $\times$ 10 cm defined at the isocenter, the ionization chamber was positioned at a 100 cm

source to surface distance (SSD). In these conditions, we have measured the entrance calibration factor ($F_{cal\ in}$), defined as the ratio between the absorbed dose (D_{in}) and reading diode (M_{in}), in accordance with equation (1).

$$F_{cal\ in} = \frac{D_{in}}{M_{in}}. \tag{1}$$

It is worth mentioning that the absorbed dose was measured by using an ionization chamber placed at the depth corresponding electronic equilibrium, *i.e.* the depth where the absorbed dose reached its maximum.

2.3. CORRECTION FACTORS

Since the diode sensitivity depends on various physical parameters, multiple correction factors (*CF*) were applied: I – energy – CF_{energy} (appropriate detector was used for each beam quality *i.e.* P10 diode was chosen for 6MV photon radiation beams and P30 for 25MV and for that $CF_{energy} = 1$); ii – CF_{SSD} distance between source of radiation and the detector placed on the skin surface (SSD); iii – CF_{gantry}; beam incidence; iv – CF_{FS} – radiation field size; v – CF_w – wedge correction factor; vi – CF_{block} – block tray correction factor; vii – CF_{temp} – temperature correction factor. In order to eliminate the output accelerator variation a large number of tests, as recommended in ref. [1–4] were performed.

Diode's calibrations were performed at 21^0C ± 1^0C, radiation room temperature, while patient's skin temperature was around 31^0C. As diode response depends on temperature, we have always performed a temperature correction.

Taking into account all the influences mentioned above, diode measured dose is better described by the equation (2):

$$\begin{aligned} Dose\ [\mathrm{Gy}] = {} & \text{diode reading} \times F_{cal\ in} \times CF_{energy} \times CF_{SSD} \times \\ & \times CF_{gantry} \times CF_{FS} \times CF_w \times CF_{block} \times CF_{temp}. \end{aligned} \tag{2}$$

2.4. *IN VIVO* DOSIMETRY

In order to detect the systematic errors, 160 breast treatments for 158 patients were screened using *in vivo* dosimetry were performed during one year. From the total number of patients only two of them have bilateral localizations and the rest have unilateral localization. In addition, 131 of patients were treated before the surgical intervention and only 27 patients were radiated after the surgical intervention.

Mammary gland radiation technique consists on classical opposed isocentric tangential wedged low energy photon beams (6 MV) *i.e.* internal to external tangential low energy $\left(Tg_{\text{int}} \middle/ Tg_{ext} \middle|_{L} \right)$.

If the mammary gland was too large, a second set of high energy photon beams (25MV) were added, *i.e.* internal to external tangential high energy $\left(Tg_{\text{int}} \middle/ Tg_{ext} \middle|_{H} \right)$.

Systematically, another two high energy beams were designed to reduce the inhomogeneity of dose distribution in the treated volume, *i.e.* internal / external tangential high energy modified $\left(Tg_{\text{int}} \middle/ Tg_{ext} \middle|_{Hm} \right)$. Generally, it has an asymmetrical and irregular shape performed with multi leaf collimator (MLC), as the last two beams have different inclination angles and SSD with respect to tangential fields.

To perform further *in vivo* dose measurements, immediately after the patient was set up and before treatment starting for all radiation fields, the detectors were carefully fixed on the patient skin in the center of reticular wires projection.

Further, the deviation δ between *in vivo* measured entrance dose (M_d) and expected dose (C_d), as computed by treatment planning system was calculated according to relation:

$$\delta = \frac{M_d - C_d}{C_d}. \tag{3}$$

For breast localization, an action level 5% was applied taking into account the accuracy and reproducibility of measurements. If the discrepancy between the two values was below the action level no other precaution was taken.

In the case the action level was exceeded, it was necessary to perform the first verification process as so: all the parameters of treatment were verified, *i.e.* all treatment planning steps, data transfer, patient set-up, choice of right diode corresponding to energy of radiation field, diode set-up, SSD correction and all CF were strongly verified.

Further, if no error was found during this process, a second action was done: *in vitro* measurement using beams treatment involved, appropriate diode as well as a slab PTW phantom. In case of a result being below the tolerance level, the difference of two measurements *in vivo* and *in vitro* was attributed to the difficulty of positioning the diodes on the patient's skin, to physical movement of the patient during the radiation or to the correction factors.

Finally, in the case the deviation was found again, a third verification action was performed namely, the measurement of the beams involved by using ionization chamber. Once again, if a new difference between diode and ionization chamber

was found, a new possible mistake was investigated in relation to configuration of radiation fields in the treatment planning system.

In conclusion, if these checks were not able to explain the observed deviation, we proceed to acquire a new CT imaging and subsequently, to design a new treatment plan. The same algorithm was applied for all treatment plans.

3. RESULTS

3.1. DETECTOR CALIBRATION

Calibration of each detector was performed against the ionization chamber and in this case the calibration factor was 1. Adjustment of this factor was done weekly in order to avoid the influence of dose accumulation in response of diodes.

The results of diodes response variation with influence of external condition of measurement are summarized in Table 1.

Table 1

The diodes relative response with SSD, field size and beam incidence, physical wedges, temperature and block tray

SSD			**Field size**			**Beam incidence**		
SSD (cm)	P10	P30	FS (cm^2)	P10	P30	angle (^{0})	P10	P30
80	0.688	0.688	3x3	1.063	1.068	-90	0.951	0.950
85	0.786	0.786	5x5	1.041	1.041	-75	0.953	0.952
90	0.880	0.880	7x7	1.020	1.015	-60	0.957	0.954
95	1.000	1.000	10x10	1.000	1.000	-45	1.000	1.000
100	1.158	1.100	12x12	0.990	0.985	-30	1.000	1.000
105	1.222	1.222	15x15	0.982	0.972	-15	1.000	1.000
110	1.375	1.375	17x17	0.978	0.958	0	1.000	1.000
115	1.467	1.467	20x20	0.972	0.949	15	1.000	1.000
120	1.692	1.571	25x25	0.967	0.935	30	1.000	1.000
			30x30	0.963	0.927	45	1.000	1.000
			35x35	0.960	0.923	60	0.957	0.954
			40x40	0.958	0.914	75	0.953	0.952
						90	0.951	950
Physical wedges			**Temperature**			**Block tray**		
Angle (^{0})	P10	P30	T (^{0}C)	P10	P30	Tray	P10	P30
0	1.000	1.000	23	0.994	0.996	plexi	0.945	0.97
15	1.421	1.409	25	0.986	0.983			
30	2.014	2.003	27	0.982	0.977			
45	3.325	3.312	30	0.977	0.972			
60	6.671	6.655	33	0.971	0.968			

3.2. DOSIMETRY MEASUREMENT FOR BREAST RADIATION TREATMENTS

As the numbers of fields involved in treatments depend on chosen radiation technique, in most cases the radiation treatment for breast cancer was performed by using four $Tg_{\text{int}}/Tg_{ext}\big|_{H}$ and $Tg_{\text{int}}/Tg_{ext}\big|_{Hm}$ fields. Occasionally, two additionally photon beams for the excessively large mammary glands were added. The final results describing the distribution of deviation levels are presented in Table 2.

Table 2

Deviation values for *in vivo* measurements

	Deviation levels (± %)						
Field	**0-1**	**1-3**	**3-5**	**5-7**	**7-10**	**10-12**	**12-15**
TG intL	42	63	43	9	1	2	0
TG intH	1	4	5	4	1	0	0
TG intHm	37	32	29	17	22	12	1
TG extL	42	70	30	12	6	0	0
TG extH	2	7	3	2	0	0	0
TG extHm	51	37	21	19	6	1	0

Table 3

Deviation values for *in vitro* measurements

	Deviation levels (± %)				
Field	**0-1**	**1-3**	**3-5**	**5-7**	**7-10**
TG intL	4	2	0	0	0
TG intH	0	2	0	0	0
TG intHm	14	5	3	2	0
TG extL	5	2	1	0	0
TG extH	0	1	1	0	0
TG extHm	15	6	2	1	0

For the measurement where the deviation is greater than tolerance level (*i.e.* 5%), we have performed *in vitro* measurements. Deviation values of deviation are presented in Table 3.

In the third verification action level of quality assurance program, the radiation field with a deviation greater that 5% was investigated.

A global view of the dispersion for the errors between calculated and measured values are presented in Fig. 1.

Statistical tests result showed that for 0.05 level the means are significantly different (*t-test*: t = –6.6914, p = 4.87E-11 and *One way ANOVA-test* F = 1911.8, p = 0). Histogram plot of errors distribution between calculated and measured values fitted with Gaussian function are presented in Fig. 2.

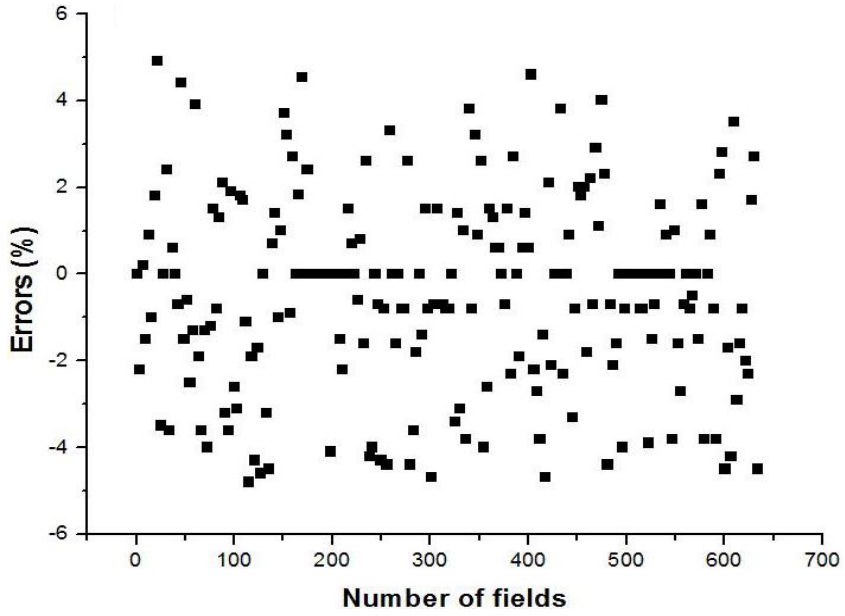

Fig. 1 – Dispersion of errors between calculated and measured values for all fields.

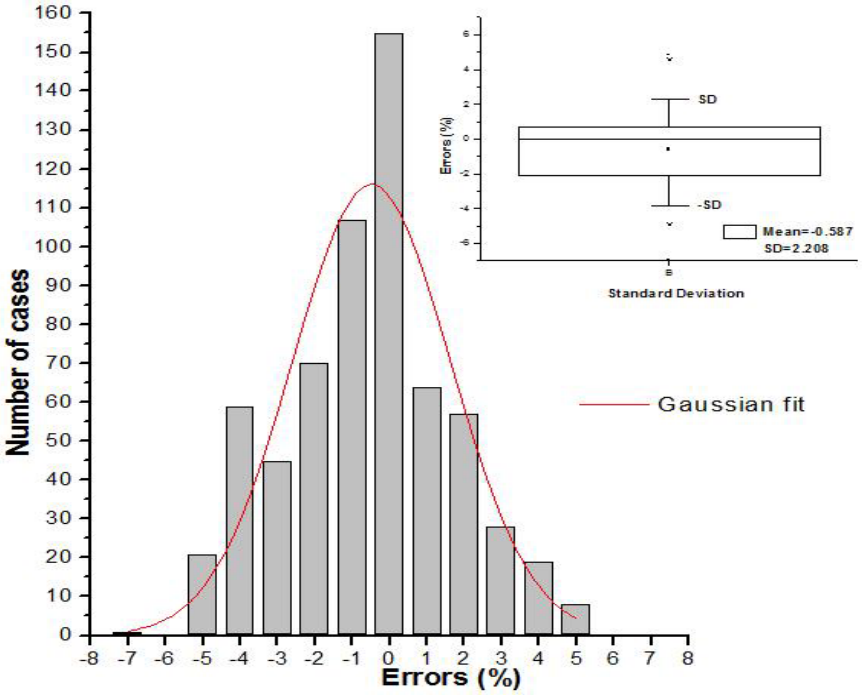

Fig. 2 – Histogram of errors distribution between calculated and measured values.

4. DISCUSSION

4.1. ANALYSIS OF DETECTOR CALIBRATION

As the detector response is strongly dependent on SSD variation, the applied coefficient factors are normalized for the reference SSD *i.e.* 100 cm, by taking into account their linear dependency on SSD variation. The evolution of coefficient factors is quite similar for the 6 MV and 25 MV energy. Field size correction factor was found to have an exponential decrement being more pronounced for the higher energy.

Factor variation with beam incidence is almost the same for the P10 and P30 detectors. Response distribution is symmetrical for the hemispherical detectors. For direct beam incidence angle, the coefficients are negligible but the importance increases for the beam incidence angle above $\pm 60^0$.

The influence of physical wedges is similarly for the two detectors, being directly proportional with the wedge angle.

Detectors response variation with temperature is linear. For *in vitro* measurements, the coefficient factor always is used. In practice the treatment time is short and for this reason the diode placed on the patient's skin cannot reach the thermal equilibrium. Therefore, for *in vivo* measurements, the coefficients at room temperature are used.

Correction factor for the block tray is measured for the plexiglas device. This correction factor can be used for the non breast tumour localization if this is necessary.

4.2. ANALYSIS OF CLINICAL MEASUREMENTS

From a total of 1902 *in vivo* measurements preformed for all 634 investigated radiation photon fields, 115 systematic errors above 5% were found.

Most of them (43%) were attributed to uncorrected SSD during the first process of verification. The most frequently involved fields were Tg int/ ext Hm.

In the second verification process were investigated the remaining 66 (57%) fields. For the 63 radiation beams, where the discrepancies between the calculated and the measured entrance dose are above the tolerance level *in vitro* investigations were done. These errors could be induced by the localization of the radiation beam center in the penumbra zone of exposed radiation beams.

For the 3 systematic errors (0,5%) greater than the tolerance level a new set of patient's imaging were acquired and a new treatment planning was performed.

5. CONCLUSION

A number of serious systematic errors might have escaped to the independent check of the steps involved in treatment process. Therefore the possibility to detect

these errors, using an independent *in vivo* dosimetry system is very useful both for treatment planning and treatment monitoring.

For this purpose, we have defined and implemented in clinical routine of our centre a quality control protocol with three tolerance levels and the appropriate action. By the described protocol, in the case of breast cancer, the random as well the systematic errors in the dose calculation and dose delivery were detected and corrected.

For the 158 patients with breast cancer, the average deviation between the calculated and the measured dose was –0.6 % ± 2.208. In 5% of cases form 634 investigated radiation fields, we traced and corrected systematic errors in dose delivery larger than the action level. Most of them were caused by differences of SSD and placement of detector. In three cases the random errors are presented.

Correction of systematic and random errors detected with *in vivo* dosimetry might lead to improve of accuracy of dose delivered in radiation cancer treatments.

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