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REVIEW PAPER

In vivo dose verification from back projection of a transit dose measurement on the central axis of photon beams

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KEYWORDS
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Abstract *Purpose:* In vivo dose verification is used to prevent major deviations between the prescribed dose and the dose really delivered to the patient. This work presents a quick and simple alternative method for verification of dose delivered to the patient using photon beams. During the treatment session, a transit dose is measured with the EPID and the dose in the patient is estimated from back projection of the portal dose.
Methods and Materials: The formalism for dose calculation is described. It is independent of the detector and has been validated for different beam energies using an ionization chamber (IC). Central axis doses estimated by this formalism were compared with measured dose. Subsequently, the IC was replaced by the EPID appropriately calibrated. The feasibility of the method and its applicability in clinical use has been evaluated on 3 8 patients treated with conformal therapy for various localizations.
Results: Ratios between stated and measured doses are reported. They are within the accepted tolerance of classical in vivo dosimetry (SD of 3.5%).
Conclusions: The proposed method for in vivo dose verification is very simple to implement and to use in clinics. Measurements can be repeated during several sessions giving the opportunity to built new strategies for the validation by statistical evaluation of the data. The trending of in vivo dose along the treatment becomes also possible. The number of checkable beams is also increased by this method.
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Introduction

External radiation therapy is a treatment with significant risks of secondary effects. But its efficiency has been proved for a long time. To ensure the treatment quality,

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irradiations have to be imperatively delivered at the right place with the right intensity. This is why all steps of the treatment have to be secured by quality processes. In this way, actions and criteria for validation are defined from the simulation to the treatment end. The vigilance of the personal still is important but objective indicators could be useful and complementary. They become necessary when the techniques get more and more complex. However, in vivo dose verification remains the ultimate step to secure the treatment before a significant fraction of the total dose is already given. In vivo dose verification is classically performed by placing dosimeters, such as diodes, thermoluminescent dosimeters, or metal oxide semiconductor field effect transistors MOSFETs, on the patient's skin or inside patients to derive the dose at specific points within the patient [1—6]. In clinical practice, placing detectors on the patient's skin at the entrance and/or exit surface of the beam is not always easy or feasible and requires some extra set up time in the treatment room. This explains why in vivo dosimetry remains limited to the first session of treatment and also why it is still not implemented in every radiotherapy center.

aSi Electronic Portal Imaging Device (EPID) appears to be an interesting measurement device for several practical reasons [7—10]. The dose response could be considered as linear within the range of this application (at this point no correction was applied in this study), it is very easy to use, no additional set up is necessary and all the field incidences can be measured directly. It therefore appears as an interesting alternative for in vivo dose measurements although one has to consider the formalism of dose reconstruction.

The properties and adaptation requirements for the use of amorphous silicon (aSi) EPID as dosimeters have been evaluated by several authors [8—11]. EPID dosimetry is now available in a large majority of radiotherapy center. It can be used for pretreatment Quality Assurance (QA) of Intensity Modulated Radiation Radiotherapy (IMRT) [11]. EPID may also be used for machines QA [12]. For in vivo dosimetry, no commercial solution exists but several teams have proposed different methods which answer at different requirements. Two approaches may be considered. With the "forward approach," the measured portal dose image is compared to a predicted dose at the plane of the EPID [13—20]. In the "backward approach," portal dose images are used to reconstruct the dose within the patient or phantom [21—30] which could be compared with the calculated one. All these methods are promising but still under development and their validation will probably go beyond the objective of in vivo dose verification making adaptive radiotherapy possible when they will be commercially available. In addition, full 3D dose reconstruction within the patient could be accessed. It gives lot of information but it requires time for calculation after the treatment session, sophisticated home made software and additional human resources for measurements and analysis.

Before that, in vivo dose verification at the point of specification remains the ultimate end to end check of the treatment preparation and is of major importance for the safety of the treatments [25,26]. For these reasons, it will become mandatory in our country within a few months [31] like it is already in some other countries. The aim of this work was to propose a simple and easy method for in vivo verification. This method is based on the back projection of the transit signal acquired during the treatment session. The method is independent of the detector allowing flexibility for users. Like "classical methods" it gives a fast answer immediately after the treatment session while the patient is still on the treatment couch. It does not need extra time for set up. Some of the limitations of the classical methods (posterior field incidences) are reduced making daily in vivo dose verification feasible.

The formalism for dose reconstruction has been validated with ionization chamber measurements for different high-energy photon beam configurations. The use of EPID for transit dose measurements has been evaluated allowing in vivo dose estimation within the patient from the portal dose images. First results on patients treated using conformal beams on two machines with photon energies ranging from 4 to 20 MV are presented and the results are encouraging.

Methods, material and notations

Methods and material

This study was performed using three Varian machines (Varian Medical Systems, Palo Alto, CA), a Clinac 2100C/S with 4 MV and 10 MV photon beams and two Clinac 2300EX with 6 MV and 20 MV photon beams. EPID dose measurements were done using two types of aSi portal imagers (aS500 and aS1000, Varian Medical Systems) in a specific dosimetric acquisition mode (respectively synchronous and asynchronous to the dose rate) and after an appropriate dose calibration. The portal dose image (PDI) is converted in Calibration Units (CU) by means of a calibration realized under reference conditions in the dosimetric workspace of the PortalVision software (respectively IAS2 and IAS3 from Varian MS). It has to be noted that these equipments are not essential in this work, provided one is able to convert the EPID signal into dose or to measure a transit dose with any appropriate calibrated detector. In this work, dose measured with an ionization chamber (when necessary) were performed using an IC 15 (Scanditronix-Wellhöfer) ionization chamber connected to a Keithley electrometer. The Treatment Planning System (TPS) used as reference for patient dose specification was Eclipse using Pencil Beam algorithm (Varian MS).

Notation rules

For clarity and simplification we adopted the following rules of notation for dose and transit dose measurements (see Fig. 1).

$D^t_{W_{d_{max}}}(d_{meas}, S, SDD)$ is the transit dose measured in a water phantom (or water equivalent material W) for a field size S at the depth of d_{max} for the considered beam energy. The measurement is performed with an ionization chamber in the presence of an absorber (or attenuator, the two terms will be used indifferently in this paper) of thickness t and made of water equivalent material. The detector is placed at depth d_{meas} (i.e. distance from the entrance face of the absorber to the detector) at a source

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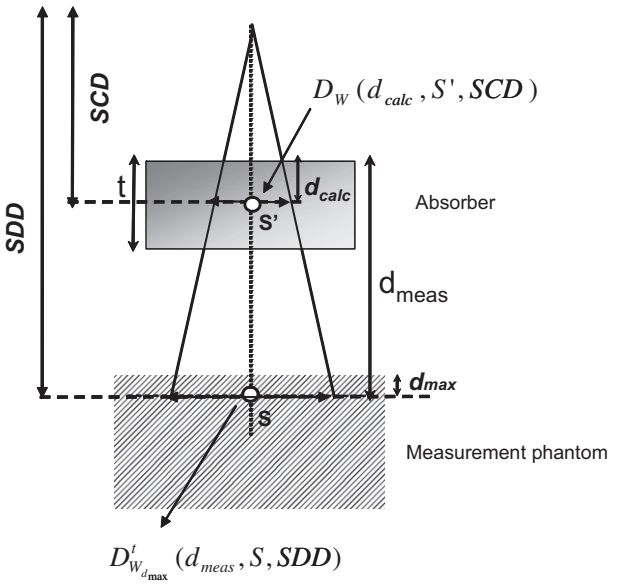

Figure 1 Set up for the transit dose measurements using an ionization chamber with the rules of notations.

to detector distance (SDD). The field size S is defined at the source to detector distance. Following these rules of notation, the transit dose measured with an aSi EPID in the same geometrical conditions would therefore be noted $D^t_{aSi}(d_{meas}, S, SDD)$ (see Fig. 2 step 1). This dose in the aSi material is given in calibration units (CU) by the manufacturer.

$D_W(d_{calc}, S', SCD)$ is the dose calculated at depth d_{calc} in the absorber (water equivalent material) for a field size S' at a source to calculation distance SCD corresponding to the field S at the SDD. The term calculated is used because this dose is calculated using the proposed formalism (see below) from a measured transit dose. It has to be noted that if the depth d_{meas} is smaller than the thickness $t + d_{max}$ of the absorber than $D^t_{W_{d_{max}}}(d_{meas}, S, SDD)$ becomes $D_W(d_{meas}, S, SDD)$. In this work, all dose measurements or calculations were done on the central axis of the beam.

Proposed formalism for open field on the central axis

By analogy of the ESTRO formalism for dose calculation [32,33], we propose a similar formalism for dose verification from transit dosimetry. In this "backward approach," reported in Fig. 2, the transit dose is measured with the EPID and used to compute the dose delivered in the patient in 4 steps.

Step 1: The aSi EPID response is converted in dose at the maximum depth in water
Step 2: The dose at the maximum depth without phantom is obtained by applying the transit Tissue Maximum Ratio.
Step 3: To determine the dose at the distance of interest (called Source Calculation Distance in this paper SCD) from the dose at the source to detector distance we use the inverse square law.
Step 4: The dose on the beam central axis in the phantom at the point of interest is calculated using the Tissue Maximum Ratio.

Steps 3 and 4 are very straightforward and will not be discussed in this paper.

Step 1: from transit dose in the EPID to transit dose in water

One of the main problems when using EPID for transit dosimetry is the non equivalence to water of this type of

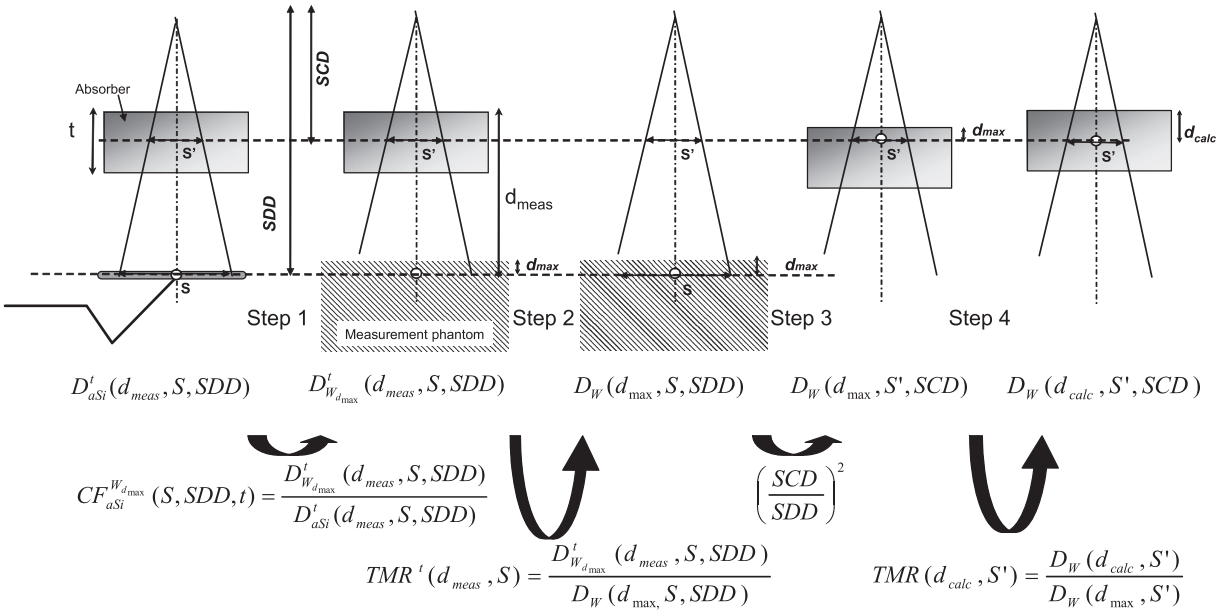

Figure 2 Proposed formalism for back projection of a transit dose measured with an EPID to determine the dose within the patient. Step 2–4 can be used with any type of detector.

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detector. We have demonstrated, in a previous work, that in the absence of absorber, the dose measured in the EPID material can be correlated to an equivalent depth of measurement in water. This water equivalent depth of measurement is independent of the field size (measurable with the EPID) but increases linearly with beam quality index. For all the beam energies included in this study, it was always found shallower than the depth of maximum in water [30,34]. Moreover, further investigations showed that if we add a water equivalent absorber in the beam, the quality index increases with the absorber thickness especially for low energy beams suggesting that the water equivalent depth of measurement is strongly depending on the absorber thickness due to the beam hardening and phantom scatter variation. To take into account these two phenomena we defined a conversion factor as follows:

$$CF_{aSi}^{W_{d_{max}}}(S, SDD, t) = \frac{D^{t}_{W_{d_{max}}}(d_{meas}, S, SDD)}{D^{t}_{aSi}(d_{meas}, S, SDD)} \left(\text{Gy CU}^{-1}\right) \quad (1)$$

This quantity is the ratio between the transit dose (Gy) measured with the ionization chamber in a water equivalent phantom at the depth of maximum of the non attenuated (i.e. without absorber) beam and the transit dose (in CU) measured with the EPID for a given attenuator thickness t and field size S. This conversion factor has been measured for square fields ranging from 2 × 2 to 20 × 20 cm^2 and absorber thickness of 0, 15, 20, 25, 30 and 40 cm for all the beam energies included in this study. These values are normalized to $CF_{aSi}^{W_{d_{max}}}(S=10,\ SDD,\ t=0)$ corresponding to the geometrical condition of dose calibration of the aSi EPID recommended by the manufacturer [11]. Table 1 reports these conversion factors for the 6 and 20 MV beams measured at SDD 150 cm. One can observe that for low energy and large absorber thickness this factor is not negligible and must be taken into account when converting aSi dose to dose in water at d_{max}. It can be noted that if one uses a detector directly calibrated in dose to water, step one becomes useless and one can proceed directly with step two in the proposed formalism.

Step 2: from transit dose in the phantom to dose without the absorber. Definition of TMR^t

The method presented here uses a set of transmission functions which are taking into account the air gap, for the scatter component, between the patient and the detector. According to their use, they have been defined (Eq. (2)) as a transit Tissue Maximum Ratio TMR^t. This quantity, similar to the Tissue Maximum Ratio (TMR), is measured in the presence of an absorber of finite thickness t (Fig. 2 step 2) as opposed to the TMR where the phantom dimensions are infinite. It is the ratio between two doses measured in a phantom at d_{max}. The numerator is the dose measured in the presence of an absorber of thickness t and the denominator is the dose measured in the same conditions without absorber. Like the TMR, the field size is defined at the source to detector distance SDD which was chosen to be 150 cm for this study. The depth of measurement d_{meas} is defined as the distance from the front face of the absorber to the detector.

Table 1 Conversion coefficient between measured dose on the EPID (CU) to the equivalent dose (Gy) in a water phantom at the depth of maximum.

Beam energy	Absorber thickness t (cm)	Square field size (cm)				
		2	5	10	15	20
4 MV	0	1.072	1.028	1.000	0.969	0.961
	15	1.307	1.267	1.216	1.148	1.084
	20	1.368	1.314	1.263	1.178	1.100
	25	1.406	1.360	1.291	1.196	1.108
	30	1.442	1.378	1.307	1.205	1.120
	40	1.520	1.445	1.331	1.222	1.148
6 MV	0	1.049	1.024	1.000	0.978	0.967
	15	1.150	1.124	1.097	1.056	1.015
	20	1.169	1.149	1.115	1.069	1.016
	25	1.197	1.164	1.126	1.078	1.021
	30	1.219	1.187	1.126	1.077	1.032
	40	1.210	1.177	1.157	1.097	1.037
10 MV	0	1.031	1.030	1.000	0.971	0.954
	15	1.052	1.051	1.009	0.967	0.939
	20	1.053	1.051	1.012	0.963	0.937
	25	1.057	1.051	1.009	0.959	0.934
	30	1.057	1.046	1.001	0.951	0.931
	40	1.063	1.045	0.993	0.938	0.913
20 MV	0	1.018	1.040	1.000	0.961	0.935
	15	1.033	1.047	0.985	0.932	0.892
	20	1.044	1.047	0.985	0.933	0.888
	25	1.045	1.049	0.976	0.919	0.883
	30	1.050	1.050	0.976	0.915	0.872
	40	1.039	1.035	0.940	0.883	0.853

$$TMR^{t}(d_{meas}, S) = \frac{D^{t}_{W_{d_{max}}}(d_{meas}, S)}{D_{W}(d_{max}, S)} \quad (2)$$

It has to be noted that when the depth d_{meas} is smaller than the absorber thickness t, the TMR^t becomes a classical TMR. TMR^t depends on 3 parameters: field size S, depth of measurement d_{meas} (function to the air gap) and thickness t of the absorber. Consequently measurements of this quantity for one beam energy would require a large amount of measurement data. To reduce the quantity of measurements, we studied the variation of this parameter and we found that when d_{meas} is larger than $t + d_{max}$ its variation is well fitted by the following expression ($R^2 = 1000$ with a maximum deviation of 0.002):

$$\text{For } d_{meas} > t + d_{max}\ TMR^{t}(d_{meas}, S) = a(S,t)\log(d_{meas}) + b(S,t) \quad (3)$$

Consequently, for $d_{meas} > t + d_{max}$ measurements may be considerably reduced. Since the expression of the fitting curve shows a linear variation of the TMR^t as a function of $\log(d_{meas})$ only two points of measurement are needed for a given beam configuration (i.e. field size and phantom thickness t). The four available beam energies were investigated at SDD = 150 cm. All the geometrical configurations with phantom thickness of 10, 15, 20, 30 and 40 cm and field sizes of 5 × 5, 10 × 10, 15 × 15, 20 × 20 and 30 × 30 cm^2 where explored.

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Fig. 3 reports the variation of TMR^t for a given beam (4, 6, 10 and 20 MV), a field size of 15 × 15 cm^2 and different absorber thickness. As one would expects, the TMR^t decreases with absorber thickness. For $d_{meas} < t + d_{max}$ the solid line is the classical TMR. For $d_{meas} > t + d_{max}$ the dots represent the measured points and the lines are the fitting curves. One can observe the very good agreement between the interpolated points and the measurements.

Using the presented formalism, the general expression for dose calculation in the patient from a measured transit dose with an EPID is given by the following expression:

$$D(d_{calc}, S', SCD) = D^t_{aSi}(d_{meas}, S, SDD) \cdot CF^{W_{d_{max}}}_{aSi}(S, SDD, t) \cdot \frac{1}{TMR^t(d_{meas}, S)} \cdot \left(\frac{SDD}{SCD}\right)^2 \cdot TMR(d_{calc}, S') \quad (4)$$

This formalism has been developed and written in C++ and implemented in a Personal Computer. However its application is very simple and can easily be performed by hand calculation.

Validation measurements

Treatment plan verification in phantom

In the first step we tested the formalism in real irradiation conditions by replacing the patient with a polystyrene phantom. We took treatment plans of 13 patients starting their conformal treatment for different cancer localizations on the three machines included in this study. These plans were transferred on a phantom in our TPS and the dose at the isocenter was computed for each beam. After that, we treated the phantom as if it was the patient and transit dose were measured using the aSi EPID (steps 1–4 in Fig. 2). Doses at the isocenter in the center of the phantom were calculated from the measured transit doses by applying the formalism and were compared to the measured doses using an ionization chamber. In the presence of a complex field, the formalism is applied considering the equivalent square field S^*. This value is computed by our treatment planning system. In total, 46 beams with energies ranging from 4 to 20 MV were evaluated and Fig. 4 reports the results. The distribution of ratios between the reconstructed and prescribed dose shows a mean value of 0.998 and a standard deviation of 0.0264 suggesting that the method is accurate and do not introduce a systematic error when applied. We have decided then to proceed with real patients.

In vivo dose verifications

Preliminary feasibility study

Between April to July 2008, 18 patients starting their conformal treatment on the three accelerators were included in this study, transit dose measurements were performed and doses at the isocenter were estimated with the formalism for each beam. These measurements were repeated during several days (up to 15 days for some patients) to evaluate the reproducibility of the results.

As opposed to the phantom, patients are not homogenous. Consequently, in the presence of heterogeneity on the central axis, the equivalent water depth $d*$ and the equivalent water thickness $t*$ are taken into account. These data are used to determine all the quantities needed in the formalism $D^t_{aSi}(d_{meas}, S, SDD)$ is therefore replaced by $D^{t^*}_{aSi}(d^*_{meas}, S^*, SDD)$.

Overall, 54 beams were checked in either SAD or SSD technique with the four energies available. The dose image acquisitions were programmed every day in the record and verify system. Nothing was changed concerning the patient set up and verification of positioning and it was asked to the technicians to proceed as usual. Dose images were automatically acquired without any extra set up time for the technicians. Taking advantage of the method we were able to repeat measurements several days (between 4 to more than 15 consecutive treatment sessions for some beams) for all the beams to evaluate the reproducibility and accuracy of our method in treatment conditions.

Fig. 5 reports the distribution of the ratios between the dose computed with the formalism using the transit dose measured on the EPID and the prescribed dose. These ratios are calculated for each beam every time it was checked. One can see a very good agreement between the prescribed and calculated doses with a mean ratio value of 0.998 and a standard deviation of 0.035. Fig. 6 reports the mean value, over the number of session every individual beam was checked, of the ratios between the reconstructed dose and the prescribed dose with the standard deviation. The results are reported for all the beam energies available in our department (4, 6, 10 and 20 MV). The global average of these mean ratios has been found to be 0.999 with a standard deviation of 0.027.

This preliminary study on 18 patients treated for various localizations demonstrated the feasibility of using the proposed method in clinical routine. The second step was the definition of the actions levels for each localization according to the usual deviations observed with this method.

Clinical routine implementation: study on prostate treatments

Between February and December 2009, in vivo dosimetry was implemented for prostate cancer treated in conformational radiotherapy in clinical conditions. The tasks were distributed and performed between the physics team and the technicians. Prostate is treated using five beams at Curie institute (180°, 90° 270°, 50 and 310°) and the patient is lying in prone position on the couch. 20 patients were included in the study. First, for every patient, dose verification was performed during the first five fractions. In a second step, dose verification was then performed every seven fraction till the end of the treatment. It has to be noticed that even if dose verification was not performed every day, it was asked to the technicians to acquire portal images every day. These images may be used as

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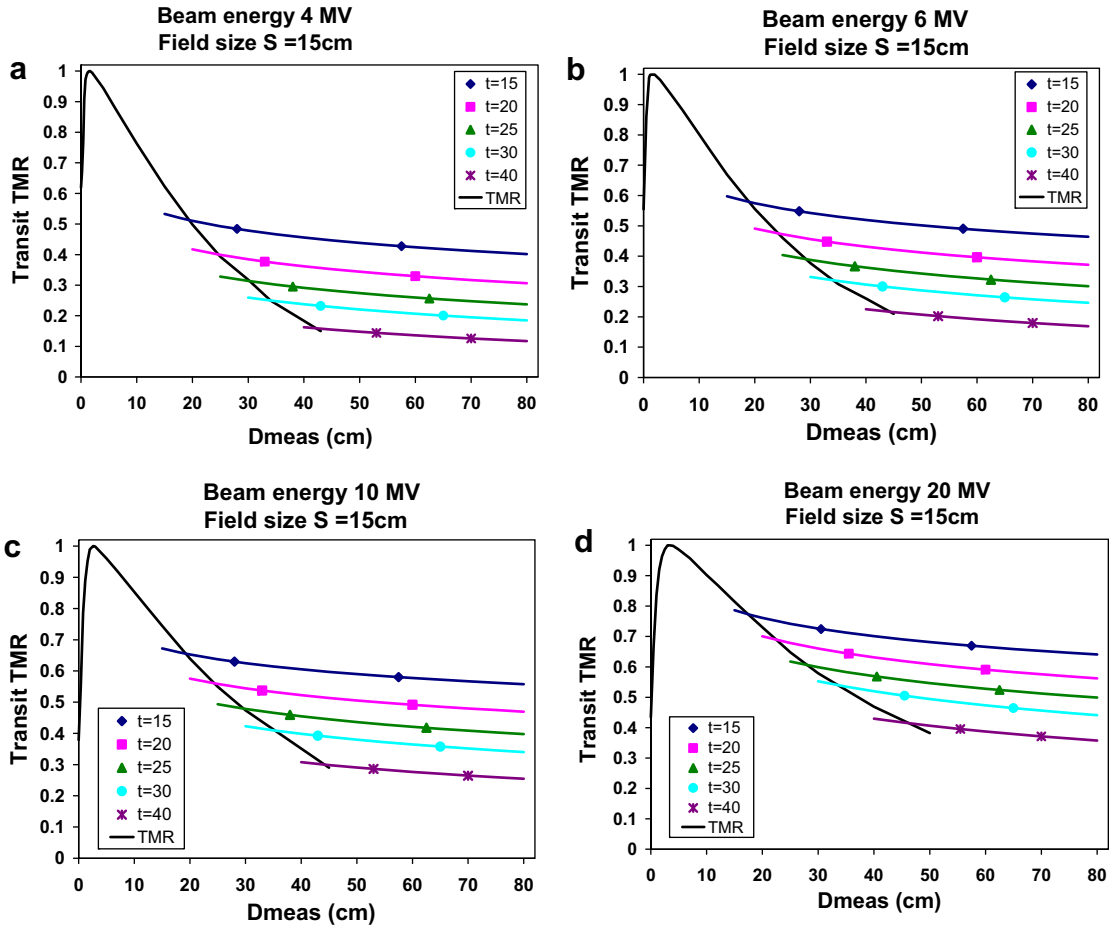

Figure 3 Transit tissue maximum ratio for a 4, 6, 10 and 20 MV beam as a function of absorber thickness t for a field size $S = 15$ cm. The points are measured and the lines are the interpolated curves from equation (3).

supplementary data for decisions in case of a large discrepancy occurred during programmed in vivo check.

Reconstructed doses were compared to the dose given by our TPS. It has to be noted that this reference dose is computed by the TPS with its own uncertainty. The mean over all the checked fractions was calculated. The results giving the mean over all the fractions for all the patients treated for prostate cancer (20 MV beams) are reported in

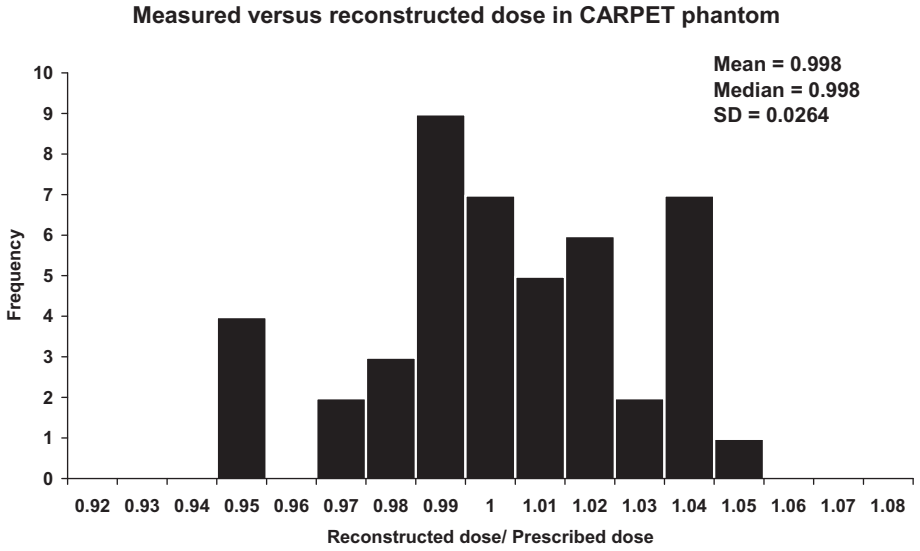

Figure 4 Distribution of the ratios between the dose measured in the phantom and the dose reconstructed from transit dose measured with the EPID.

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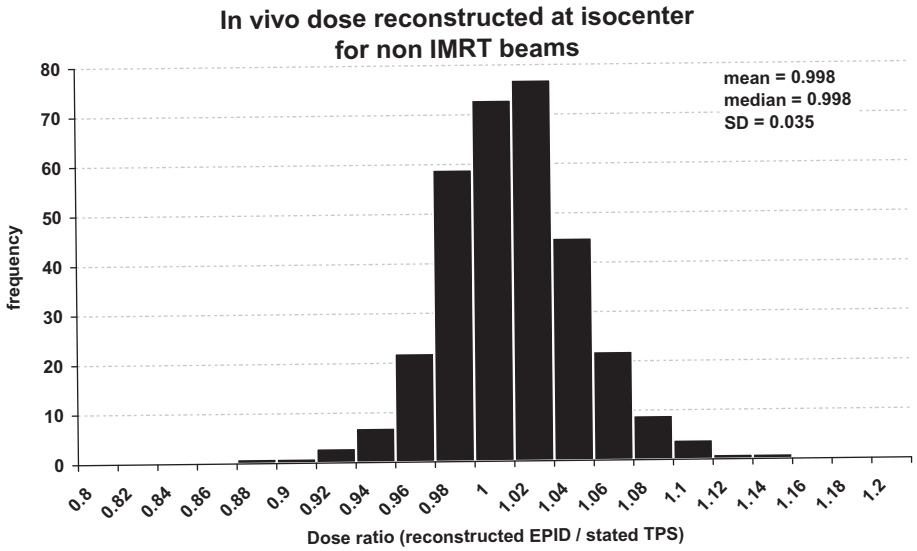

Figure 5 Distribution of the ratios between the prescribed dose in the patient and the dose reconstructed from transit dose measured with the EPID for non IMRT beams.

Fig. 7. According to these results, new tolerances and actions levels have been established (Table 2) which are better reflecting our clinical practice.

Discussion

The method proposed here has been developed to be easily implementable in every radiotherapy center with no need of supplementary tools or human resources. It has to be noted that this method is available for each calibrated dosimeter. To reduce the amount of data measurements needed for the commissioning of the method, standard references (for example a and b in Eq. (3)) could be defined using ionization measurements according to the beam quality indexes.

The phantom measurements allow us to say that the difference observed between the results with ionization chamber and EPID were acceptable and allowed us to validate the use of EPID as dosimeters. The overall uncertainty was in the same order of magnitude that ionometric measurements. If we separate data by beam energy, we obtain the same kind of results and we can conclude that the method can be applied with the same accuracy and precision in the entire range of energies used in radiotherapy.

The preliminary study on 18 patients treated for randomized localization allowed us to say that the results are more than acceptable for in vivo dose verification [2,4]. When a patient treatment is checked over several fractions, we can say that when discrepancies are observed the first day between the prescribed and the

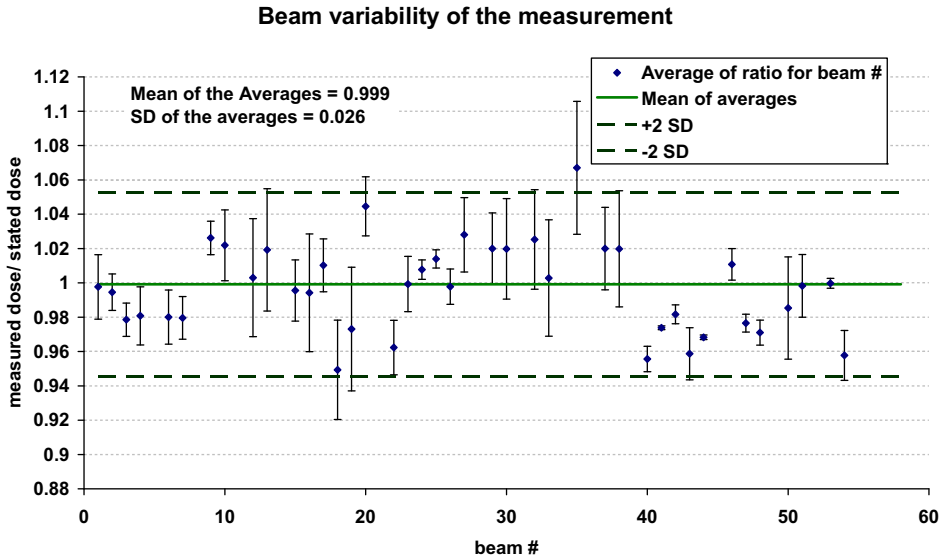

Figure 6 Mean ratios between the prescribed dose in the patient and the dose reconstructed from transit dose measured with the EPID for repeated dose reconstruction during consecutive treatment sessions. Results for 4, 6, 10 and 20 MV beams are presented.

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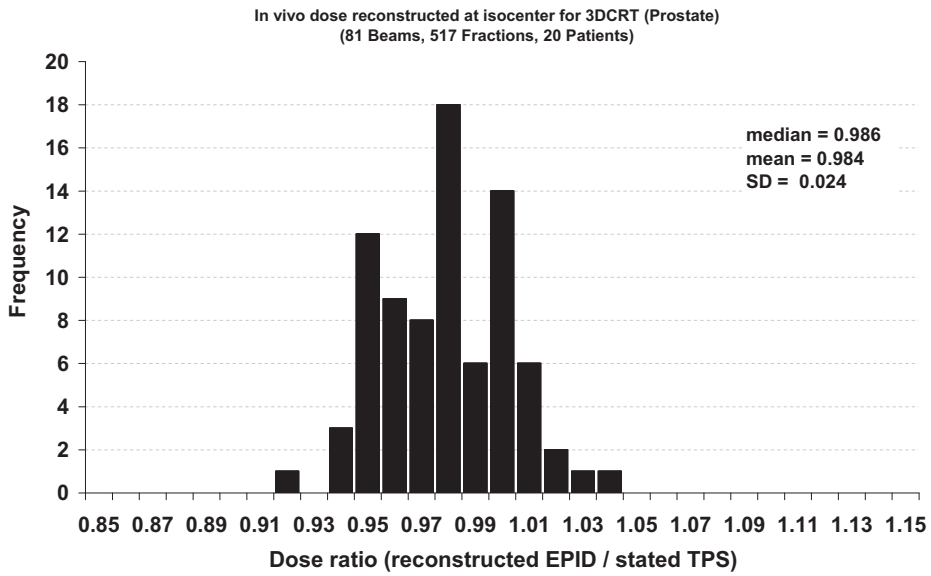

Figure 7 Distribution of the means ratios between the prescribed dose in the patient and the dose reconstructed from transit dose measured with the EPID for repeated prostate treatment verifications. The results presented are for 20 MV beams.

reconstructed dose, they are usually confirmed and reproducible the following days. However, repeating the measurement several days reduces the effect of a major discrepancy due to individual verification and provides additional information for the final validation. This could be used to elaborate new strategies of decisions for in vivo dose verification by introducing decision-making based on statistical variation of the measured dose. Another advantage of the method would be to repeat the measurements periodically during the treatment to obtain a follow up and/or the trend of the delivered dose. For instance, in case of an increasing discrepancy, the medical staff could be alerted and consider to take into account and/or adapt the dosimetry regarding the observed evolution [35]. This study suggested us the following methodology for the validation of in vivo dose verification. The measurements are performed during the first five sessions. At the first session, if the deviation between the expected and the reconstructed dose is larger than 7% (2 SD of individual measurement (see Fig. 5)) the patient file must be reviewed to identify the discrepancies and continue the treatment. After the fifth day, if the average of the deviations is larger than 5% (2 SD of repeated measurements (see Fig. 6)) the discrepancy must be questioned and explained to continue the treatment.

The second part of this study (on prostate treatments only) allowed us to define procedures to include more people like physicists technician in routine tasks. Actions levels were defined for each treatment beams used in our prostate irradiation protocol. They may be directly compared to the results observed with others methods of EPID transit dosimetry [25,26]. These actions levels are determined for the first fraction checked and also for the mean observed over the first five fractions. Other fractions give information on the treatment variation. These tolerances are optimized and adapted to our process and we expect to separate deviations due to errors in treatment process from deviations due to approximations in the in vivo dose computation.

In our study, the CT scan performed during the treatment preparation was used. We expect to be more accurate by using the daily cone beam CT [36] and this study is under way.

The main remaining problem in the use of EPID for transit dosimetry is its over response regarding low energy

Table 2 Results and actions levels determined form the study on the 20 patients treated for prostate cancers on three linear accelerators.

	Plan (Fields sum)	Field 1 (180°)	Field 2 (270°)	Field 3 (310°)	Field 4 (50°)	Field 5 (90°)
Mean	0.984	1.000	0.961	0.986	0.991	0.971
Median	0.988	1.000	0.962	0.991	0.995	0.976
1SD	0.0137	0.0227	0.0147	0.0235	0.0183	0.0159
2SD	0.0274	0.0454	0.0293	0.0469	0.0366	0.0318
Mean − 2SD	0.957	0.954	0.932	0.939	0.955	0.940
Min	0.960	0.959	0.930	0.942	0.952	0.940
Mean + 2SD	1.012	1.045	0.991	1.033	1.028	1.003
Max	1.014	1.048	0.989	1.030	1.016	0.990

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photons. We observed this phenomenon for the low energy photon beams, when a large patient thickness is crossed and/or when the detector is placed closer to the patient [37]. In these cases, the EPID response is strongly dependant of the scatter component and the modification of the quality indexes due to patient transmission. In addition, EPIDs are not under electronic equilibrium. Some authors propose to add a cupper layer [10] or to work with direct-detection configuration [38]. This customization could not be done without manufacturer development to be safety implemented in every radiotherapy center. Portal dosimetry is used in our department since the aSi EPIDs were installed (more than 5 years ago) and their lifetime and performance do not seem to be shortened. However this point will have to be investigated in the future [39].

Conclusion

A formalism for patient in vivo dose verification on photon beam central axis is proposed. It is based on the back projection of transit dose measurements. This formalism was developed to use the aSi EPID as detector. With this type of detector, the method provides a very quick and simple way of patient dose verification at the point of prescription usable in clinics. However, the proposed formalism is general and can be used with any type of detector. Dose verifications on real patients have been performed and repeated for several consecutive sessions and the results are within the usual tolerance of "classical" in vivo dosimetry. To be able to repeat measurements at every session without diminishing the patient throughput opens new perspectives for the patient safety. We are working, in a partnership with the company DOSIsoft (Cachan, FR), on the validation of the formalism outside the central axis of the beam to be able to reconstruct the dose at off axis localization as for organ at risks verifications. A primary/scatter separation algorithm is now under investigation to improve the computation in irregular fields with modulated fluences.

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