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THE SOFTWARE PROGRAM *PERIDOSE* TO CALCULATE THE FETAL DOSE OR DOSE TO OTHER CRITICAL STRUCTURES OUTSIDE THE TARGET AREA IN RADIATION THERAPY

P.H. van der Giessen
Dr. Bernard Verbeeten Institute, Tilburg, Netherlands

Abstract

An accurate estimate of the dose outside the target area is of utmost importance when pregnant patients have to undergo radiotherapy, something that occurs in every radiotherapy department once in a while. Such peripheral doses (PD) are also of interest for late effects risk estimations for doses to specific organs as well as estimations of dose to pacemakers. A software program *Peridose* is described to allow easy calculation of this peripheral dose. The calculation is based on data from many publications on peripheral dose measurements, including those by the author. Clinical measurements have shown that by using data averaged over many measurements and different machine types PDs can be estimated with an accuracy of ± 60% (2 standard deviatons). The program allows easy and fairly accurate estimates of peripheral doses in patients. Further development to overcome some of the constraints and limitations is desirable. The use of average data is to be preferred if general applicability is to be maintained.

1. Background and purpose

The incidence of cancer increases with age and as a consequence most patients entering a radiotherapy department are elderly. Nevertheless, this does not exclude the possibility of cancer occurring in younger people, at an age where they still have the prospect of establishing a family and having children. If young patients are treated with radiation it is essential that the dose to the gonads is kept as low as possible to keep the risks to the offspring at an acceptable level. Should pregnant patients be presented for radiation therapy and this therapy can not be postponed, keeping the dose to the fetus as low as possible is of utmost importance. Furthermore, there are times that conception occurs just prior to or during treatment. Knowledge of this dose at distances larger than a few centimeters outside the primary beam, which is called the peripheral dose (PD), is therefore essential in those cases. Computerised planning systems can accurately calculate the dose inside and at the edges of the primary beams; however, accurate dose calculations are usually limited to a few centimeters outside of the beam edges. Determination of the peripheral dose has been the subject of extensive investigation, the results of which we have published previously [1-3]. In these papers data were presented for photon energies from cobalt-60 gamma radiation to 6, 10, and 23 MV x-rays. These values were derived from measurements of the contributions to the PD from radiation scattered in the patient, leakage radiation, and radiation scattered from the collimator. Our own data were combined with other published data [4] and were used to generate a generalized method to estimate the peripheral dose for any arbitrary field size or shape at different depths. In patients an accuracy of ± 60% (2 standard deviations) could be obtained [5]. In view of the uncertainty of known risk factors, we consider this accuracy acceptable.

On the basis of this generalized method we decided to develop a software program to perform these calculations automatically and to make this program available to the radiotherapy community.

2. Structure of the program

The software is written in Delphi. Minimum system requirements are 4 MB RAM, and requires a 4 MB hard disk. It runs under Windows, version 3.11 and higher.

The data of our paper on a general applicable calculation method [4] form the basis for the calculation algorithm.

All graphical data from that paper are transformed into tabular data and intermediate values are determined by linear interpolation.

In figure 1 the input screen for one beam is presented showing also which input data are required. The maximum number of beams that can be calculated in one run is eight.

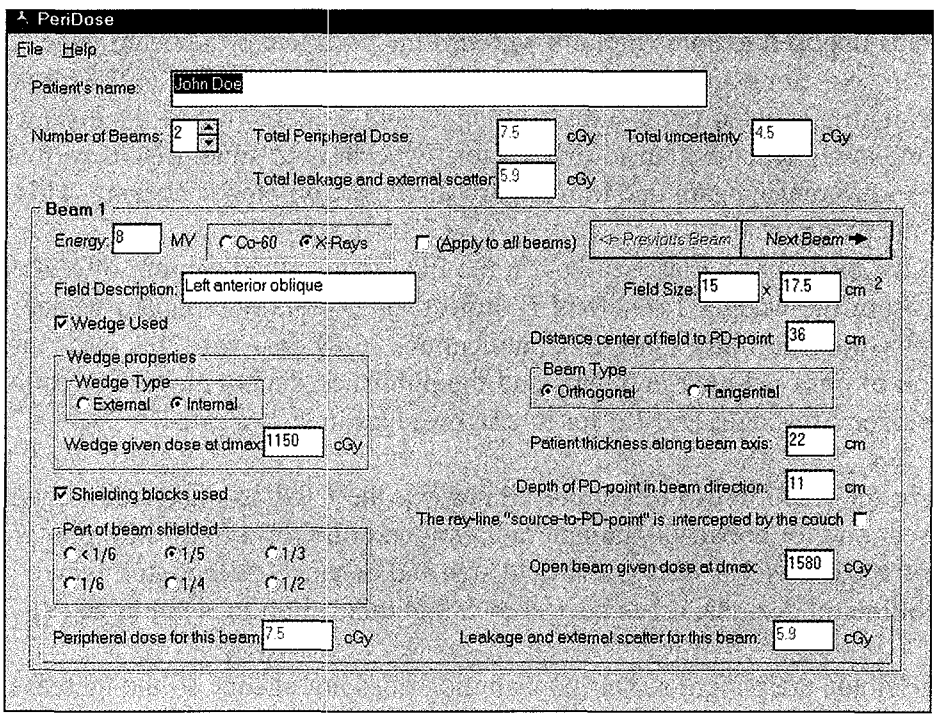

Figure 1. Input screen for beam 1 of two-beam calculation. At the top the total results are shown, at the bottom the results for beam 1

2.1. Orthogonal beams

In *the first step* the peripheral dose is calculated per beam as a percentage of the dose at depth of maximum dose (d_{max}) at a reference depth of 10 cm for a reference thickness of the patient of 20 cm. The equivalent square field size is used. A distinction is made between cobalt-60 gamma radiation and 4 to 25 MV photons.

The small variation of the PD for photon energies between 4 and 25 MV is accounted for by applying a correction factor in *the second step.*

Patient thickness is corrected for in *the third step.* When the primary beam travels through more tissue the contribution from patient scatter to the PD increases. The effect is greatest for

small distances. Variation of the PD with depth is accounted for in *the fourth calculation step.* There are two opposite effects involved. Close to the beam the patient scatter contribution increases with depth as a result of the forward directed Compton scattering. On the other hand the contribution of leakage radiation and scattered radiation from the collimator, referred to as collimator related radiation (CRR), decreases with depth because of attenuation. This decrease roughly follows the percentage depth dose distribution of the primary photon energy. Far away from the beam the CRR is the sole source of radiation so the correction factor follows the primary beam attenuation.

In *step five* a correction is made to the PD if the CRR is intercepted by the couch. This might be the case for posterior-anterior beams for target volumes further away from the PD point, for instance when treating targets in the thorax or head and neck, with the PD point in the pelvic area. The CRR will then be attenuated by the couch.

In our calculation model, distance is defined as the distance of the PD point to the beam central ray, as opposed to some authors who use the distance to the beam edge. Consequently in our model field elongation can have a considerable influence. The PD point is much closer to the edge of an elongated fields with the long axis in the direction of the distance vector than with the long axis perpendicular to that vector. Especially at small distances this can make a considerable difference, again due to foreward directed Compton scattering. This correction is *step six* of the program.
Wedges in the beam have a large effect on the PD by the added amount of scattered radiation emanating from the wedge. This effect is largest for externally mounted wedges and smaller for internally mounted ones. Only few publications [6-8] deal with this issue and based on a combination of our own measurements and the published data, a global correction factor of 4 is used for external wedges and 1.5 for internal wedges in *step seven.*

In step eight the fraction of the PD contributed by the CRR is calculated. Again two sets of data are used, one for cobalt-60 gamma radiation and one for 4 to 25 MV photons, giving the fraction of the CRR as function of the field size and distance. Although this will vary between different collimator designs, it has been shown that this variation is not large [9].

For wedged fields the patient scatter contribution does not change so the increase of the PD is caused entirely by the increase of the scattered radiation from the wedge. This is also accounted for in this calculation step by including this scatter in the CRR fraction.

In *step nine* the influence of blocks is addressed. Published data [2-3,8] have shown that the PD does not change significantly when shielding is introduced in the beam. This can be explained by assuming that the reduction of the patient scatter contribution due to partly shielding the incident beam is counterbalanced by the increased scattered radiation from the shielding blocks and tray.

In the *tenth step* the CRR is corrected for attenuation at other depths, as described in the explanation of step four.

2.2. Tangential beams

The program also offers the option to calculate the PD for tangential (breast) treatment techniques.

In this case the breast is the scattering volume and measurements were made for three breast sizes, which are called small, medium and large with field sizes to match. Interpolation by the program is based on the actual field size as stated by the user.

The program follows the same steps as for orthogonal fields with one exception. Since the patient scatter is determined by the breast size, there is no need for a correction for patient thickness. Furthermore, the depth of the PD point is defined differently. Since PD calculations in patients treated for breast cancer will often concern determination of the fetal dose, depth is now defined as the depth of the PD point (i.e., the fetus) in anterior-posterior direction

3. Results

The results of the calculations are presented in a simple way (Fig. 1). At the bottom of the screen the results per beam are shown, subdivided in the PD and the CRR contribution both in cGy. At the top the combined results for all beams are shown.

The data and results can be saved as a file with default extension *.pdd* and a hard copy of the results can be printed.

3.1. Constraints and limitations

Certain constraints have to be considered. An assumption is that the PD point is located more or less centrally and symmetrically in the body. Differences in the PD for deviations of the central position perpendicular to the plane through the beam axis and the distance vector of up to 5 cm are negligible; variations in distance and depth are accounted for.

The program cannot be used for other treatment modalities than photon beams. For electrons the scarce published data [10] and our own measurements indicate that the PD is roughly a factor of 4 lower, because there is hardly any scatter inside the patient and the CRR is much lower than for photons.

The program was not developed for use in intensity modulated radiation therapy (IMRT). During IMRT the number of monitor units delivered for a given target dose is much greater than in standard techniques. Consequently the contribution of CRR will be much greater but we are not aware of measurements on the exact magnitude of this contribution.

The program does not account for neutron production at higher photon energies. For 25 MV photons this can increase the PD by a factor 2.

3.2. Accuracy

We compared the calculations with clinical measurements and found a mean ratio of measured versus calculated PD of 0.92 with a standard deviation (SD) of 35% for all treatment techiques [5]. For tangential techniques only this was 1.12 and 26% respectively. We find it plausible that the program will be used most frequently for calculations in pregnant patients so the starting point of the program is an SD of 30%. The accuracy of the calculation is given as two SDs. The accuracy of the calculation is largest for open beams with limited shielding. In case of the use of wedges the program uses some average correction factors for internal and external wedges. The accuracy of these factors, however, is estimated to be of the order of ± 30%. When the PD-point is located further away from the central axis of the beam,

it is possible that the collimator-related radiation is intercepted by the treatment couch. In that case an attenuation factor is applied, based on our own measurements for our treatment couch. Data on the attenuation by couches from other manufacturers are not available.

The contribution of collimator-related radiation of linear accelerators to the PD is based on average data. However, some accelerators show higher collimator-related radiation values than others and there is also some dependence on collimator angle. The maximum difference is by a factor 2 [9]. For PD calculations at large distances, where the contribution is predominantly from collimator-related radiation, this can make some difference.

4. Discussion

A software program has been developed which allows the easy calculation of the peripheral dose in patients who are treated with megavoltage photon radiation. Within its constraints and limitations it allows a fairly accurate estimate of the dose at any point in the body outside the treatment area. Knowledge of the peripheral dose can help radiation oncologists in making important decisions in the treatment of cancer patients. Sometimes radiation therapy is the only viable treatment option when pregnant patients have to be treated and then it is of utmost importance to be able to estimate the risk to the fetus and compare this with the risk to the mother of postponing the treatment. Decisions on whether or not abortion should be considered may also depend on this information.

Another area where an estimate of the peripheral dose is of importance is in patients with pacemakers. Damage to pacemakers has been observed above 500 cGy [11] which is only a few percent of common clinical tumor doses. Assessment of doses to specific organs such as the thyroid may also be of interest to determine the possible risk of late effects such as carcinogenesis.

We feel that our program can be of great value for the professionals working in radiotherapy. We also feel that general applicability is desirable and therefore prefer the use of average data to the use of machine specific data, even at the cost of a small loss of accuracy. Situations where the PD has to be estimated are rare and usually occur unexpectedly. A calculation model should then be readily available since there is no time to perform extensive measurements on leakage radiation and collimator scatter.

Note: The program can be obtained from the author, preferably by e-mail request.

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