Accuracy of treatment planning calculations for conformal radiotherapy
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Chapter II

Verification of an energy transport model for dose calculation in a treatment planning system

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Verification of an energy transport model for dose calculation in a treatment planning system

Abstract
A procedure has been developed for verification of the radiation beam model in a modern treatment planning system, Helax-TMS. Aiming at general applicability, the beam model for photon dose calculations is based on a first principle description of the energy fluence components that exit the head of a linear accelerator. The dose to a patient is calculated by an energy deposition kernel that converts the incident energy fluence to deposited dose. The beam model is constructed on the basis of measurements supplemented with Monte Carlo data and a geometric description of the head of a linear accelerator. Because modelling steps are involved between measured basic beam data and the model parameters, the model and its implementation require specific attention during verification. In this paper the verification procedure is described with emphasis on verifications that are tailored to the model characteristics. A ‘concentric wall’ structure is set up which is shown to serve well the purpose of verification, because very few errors are found at the inner wall. Over a period of 5 years in 1% of all treatment plans a dose difference larger than ±2%, as compared to an independent check, was found.

1. Introduction

The treatment planning system (TPS) at a Radiation Oncology Department plays a key-role in the preparation of a radiotherapy treatment. The system provides the means to design treatment plans for delivery of the prescribed dose to the target volume (tumour with appropriate margin) while sparing the surrounding normal tissue as much as possible. To meet this requirement, the TPS should be able to exploit all degrees of freedom of the treatment machine, usually a linear accelerator (linac). A minimal margin usually exists between an eradicating dose to the tumour and a dose with unacceptable risks for complications. This implies that the TPS must be able to calculate dose distributions with sufficient accuracy, for all applied treatment techniques in any patient anatomy. As an indication, an accuracy in the dose distribution of 2-3% (1 S.D.) is considered as presently acceptable, and 1% (1 S.D.) as a possible future aim level.¹²

Many workers in the field agree that only a fundamental physical approach, based on first principles, will allow high accuracy of calculated doses to be obtained for all present (and future!) degrees of freedom in treatment delivery. The dose calculation model of the TPS studied in this paper and installed at our hospital (Helax-TMS 5.1.1, MDS Nordion, Kanata, Canada) is directly based on a description of the basic quantity in radiation: radiation energy that is transported from the treatment machine to the patient, where it is deposited as dose (energy per mass) in patient tissues. The transfer of this concept into a clinical TPS has been documented in a large number of publications, as discussed in a recent review article.¹

The aim of this paper is to present the verification procedure that we have developed to verify the beam model of our TPS, section 3, taking into account its physical background to describe energy transport, section 2. The scope is limited to megavoltage photons as generated by a linac, the most common application of radiotherapy.
2. The physics of a beam model

When irradiated materials and tissues and their cross sections are known in sufficient detail, Monte Carlo calculation provides the tools to calculate radiation interaction effects with any required accuracy. However, on present hardware dose calculation times are prohibitively long, thus urging simplified models for clinical purposes. The beam model in our TPS therefore applies various pre-calculated Monte Carlo results that are matched to the properties of the actual linac beam as defined by measurements and by the linac geometry.

Dose calculation in the TPS is performed in two distinct stages. In the first stage (section 2.1) the energy fluence is determined as it exits the linac in a particular field configuration. Energy fluence is defined as the number of particles passing a sphere of unit cross-sectional area, multiplied by the energy of the particles. Furthermore, in the first stage the relation is determined between energy fluence and the number of monitor units, and thus absolute dose. This is, however, outside the scope of this paper. In the second stage (section 2.2) the dose distribution is calculated that is deposited in a patient by this energy fluence. The determination of beam parameters to construct a beam model for an individual linac beam, the ‘beam-fit’, is subsequently described (section 2.3).

2.1 Description of the energy fluence of a linac

Megavoltage photons are generated as bremsstrahlung in the transmission target of a linac, Fig. 1. The fraction of these photons that exit the linac without further interaction constitutes the major contribution to the energy fluence. They represent the primary energy fluence, $\psi_{\text{prim}}$. This fluence is defined at the level of the isocentre in the maximum field size, $A_{\text{max}}$, as a variation $f(x,y|A_{\text{max}})$ upon a constant reference fluence $\psi_0$, thus: $\psi_{\text{prim}}(x,y|A_{\text{max}}) = \psi_0 f(x,y|A_{\text{max}})$. The function $f(x,y|A_{\text{max}})$ is being determined as a part of the beam-fit. When the field size is limited by collimators and shielding blocks, $A_{\text{max}}$ has to be replaced by the actual

![Fig. 1 Schematic transection of the head of a linear accelerator](image-url)
field size $A$. The fluence at other distances (the $z$-direction) can be derived by correction for the distance ($\sim 1/r^2$). For the purpose of clarity the present description is limited to isocentric distances only.

Another fraction of the bremsstrahlung photons will be scattered by the flatness filter, by the primary collimator, by the wedge, by the edges of the collimator, and, if present, by blocks and trays. This contribution to the fluence is called head scatter fluence, $\psi_{\text{hsc}}(x,y|A)$. The head scatter contributes 4% to 15% of the total dose, dependent on the linac construction, the field shape and accessories like compensators. The flatness filter and the primary collimator dominate the head scatter fluence. The head scatter is a function of geometry and position of all scattering parts of the linac head, as seen from the point of dose calculation $(x,y,z)$, Fig. 2.

The algorithm that performs the projection calculations has to take care of numerous interfaces, especially in the case of a multileaf collimator (MLC). Thus a robust implementation is required with a careful balance between efficiency and accuracy.\(^3\) The total energy fluence $\Psi_{\text{tot}}$ is defined by:

$$\Psi_{\text{tot}}(x,y|d) = \psi_0 \cdot \left( f(x,y|d) + \frac{\psi_{\text{hsc}}(x,y|A)}{\psi_0} \right)$$

A graphical representation of the dose distribution corresponding to a 6 MV-X MLC-shaped field at the level of the isocentre at 10 cm depth is shown in Fig. 3. For matter of clarity of display the dose distribution rather than the fluence distribution itself is shown, i.e. the fluence convolved with a pencil beam kernel, section 2.2. Also included is the phantom scatter dose, that is the dose resulting from photons that are scattered in the patient.

Inspection of Fig. 3 shows that the head scatter distribution is distinctly different from the primary fluence, both in homogeneity and in steepness of the penumbra. The most
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Fig. 3 Dose distributions of a 6 MV-X 10x10 cm$^2$ MLC-shaped field at 10 cm depth in water. In the total dose distribution (a) the area up to 20% is enhanced in the display. Inside the field boundary the dose decreases with distance to the centre. This is caused by the phantom scatter (c) and the head scatter (d), whereas the primary dose distribution (b) within the field is almost flat. The relative contributions of the dose components is indicated by a dose profile through the centre (e).

Important causes of these differences are the location, near the field defining collimators, and the extension of the flatness filter.

In Eq. (1) it can be seen that the energy spectrum of the photon beam is not explicitly included. An integration over an effective energy spectrum has already been performed, as described in section 2.3. This yields an increase in calculation speed of approximately a factor of 10. As a consequence, however, effects of spectral changes as a function of depth in a patient or as function of the off-axis position in the beam, can not be accounted for on the most appropriate place, Eq. (1), but have to be included as correction terms in the dose deposition stage.
2.2 Description of the dose deposition in a patient

The dose that is delivered to a patient upon irradiation with megavoltage photons is deposited by electrons (at higher energies also positrons) which induce tracks of ionization in tissue. The two quantities that determine the conversion between energy fluence and dose deposition are the total energy released to mass, terma, by the photon beam and the energy deposition pattern (kernel) around a photon interaction point. Given a homogeneous mass density $\rho$ and an exponential decrease as a function of the trajectory length (depth) with attenuation coefficient $\mu(E)$ at photon energy $E$, the terma differential in energy $E$, $T_E$, is defined as:

$$ T_E(\mathbf{r}) = -\frac{1}{\rho} \nabla \cdot \bar{\Psi}(\mathbf{r}) = \frac{\mu(E(\mathbf{r}))}{\rho} \bar{\Psi}(\mathbf{r}) $$

(2)

Kernels can be determined by Monte Carlo calculation, and in good approximation also by analytical calculations. In the calculations photons are forced to interact at a fixed location, yielding a point spread kernel. A graphic representation of a point spread kernel due to 6 MeV mono-energetic photon interactions is given in Fig. 4. A poly-energetic kernel can be composed by weighting of a set of mono-energetic kernels over the effective spectrum, refer to section 2.3. Such a poly-energetic point spread kernel can be described accurately in a homogeneous medium (water) by an analytical expression:

$$ h_{e}(r,\theta) = (A_{e}e^{-a_{e}r} + B_{e}e^{-b_{e}r})/r^2 $$

(3)

in which $r$ indicates the distance to the point of interaction and $\theta$ represents the azimuthal angle relative to the direction of the incoming photons. The parameters $A$, $a$, $B$ and $b$ are a function of this angle. By good approximation the A-term corresponds to the dose deposition of electrons recoiled at initial photon interaction (primary component), while the B-term represents the dose deposition by scattered photons (phantom scatter component). A correction on the B-term for off-axis spectral changes has recently been described. Such an analytic kernel description is a very effective pre-calculated summary of the probability distribution of ionization tracks. This yields a considerable reduction in calculation time in comparison to a full calculation of transport. The disadvantage is a decreased accuracy in cases where the kernel can not appropriately be scaled, in particular in electronic disequilibrium near interfaces of different tissue densities.

The calculation of the dose distribution in a patient can now be given by a single equation, in which $T_E$ represents the terma differential in energy, thus prior to spectral weighting:

$$ D(\mathbf{r}) = \int \int \int T_E(\mathbf{s}) h(\mathbf{r} - \mathbf{s}) d^3s dE = \int \int \int P(\mathbf{s}) h_{p}(\mathbf{r} - \mathbf{s}) d^3s + \int \int \int S(\mathbf{s}) h_{h}(\mathbf{r} - \mathbf{s}) d^3s $$

(4)

At the right side of this equation a separation has been made in the primary and phantom scatter component of both the terma (in P and S) and the kernel (in $h_p$ and $h_h$), in analogy to Eq. (3). This separation allows by good approximation the described spectral weighting. The head scatter component is then distributed over both components in the phantom.
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Eq. (4) describes in the case of an infinite homogeneous medium a convolution that could efficiently be calculated by a Fourier transform. Daily practice, however, presents finite media (patients) and density heterogeneities (lungs, air cavities, bone). In principle the kernel of Eq. (3) can be scaled for the density variations, followed by a dose calculation with Eq. (4). In case of a patient that is described by N points in each of three dimensions this requires a number of calculations in the order of \( N^7 \) (\( N^3 \) photon interaction points, scaling over in the order of N points, and \( N^3 \) dose deposition points). Calculation times would be prohibitively long. Therefore further approximations are needed. In our TPS two further approximations are implemented, a pencil beam kernel and a ‘Collapsed Cone’ kernel.

In the approximation based on a pencil beam kernel, also called line spread kernel, the point spread kernel of Eq. (4) is integrated along a ray-line of a beam. The results is stored as a pre-calculated analytical expression in the beam model of the particular linac/energy combination. With a line spread kernel only density variations along the ray-lines can be accounted for in the dose calculation. Also corrections for beam hardening can be included.

The Collapsed Cone approximation is based on a simplified form of a point spread kernel. The kernel is then subdivided into a number (M) cones that together constitute a \( 4\pi \) solid angle, Fig. 5. All the transport of recoil electrons as well as scattered photons that occurs inside a cone is taken together, ‘collapsed’, on the central, transport axis, of that cone. The orientation of the transport axes coincides as good as possible with the grid of point spread kernels, making axes of different kernels coincide as much as possible. Transport is then calculated only along the transport axes. By use of a recursive relation\(^5\) each transport axis need to be considered only once. On the way along an axis the energy of recoiled and scattered particles is ‘picked up’ and later on ‘delivered’ in dose deposition points. An instructive metaphor for this process is a regular bus service in which passengers are picked up and others delivered at every bus stop. In this way the number of dose calculations is reduced to an order of \( M \cdot N^3 \). Beam hardening and off-axis spectral changes are corrected for in our TPS by adaptation of the effective attenuation coefficients that are applied to calculate the primary and scatter component of the term, Eq. (4).

2.3 Data-acquisition for the beam model

Two fundamental quantities in the beam model are the energy fluence and the energy spectrum to perform the spectral weighting of the kernels. These two quantities can not be determined directly, and must thus be derived from other measurements, Fig. 6. The set of required measurements for an open beam (without a wedge) consists of:
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- Depth dose curves of four field sizes, 5x5, 10x10, 15x15 and 20x20 cm². Furthermore the output in a 10x10 cm² field in water and in air to exclude the head scatter component and to create phantom scatter normalized kernels. By comparison of a weighted set of depth dose curves that have been determined by Monte Carlo simulation of mono-energetic beams, the effective energy spectrum is determined. The remaining difference with measured depth dose curves in the build-up regions is used to determine a kernel for charged particle contamination, describing the dose of electrons and positrons that emerge from irradiated parts of the linac head.
- Dose profiles in two directions in a 10x10 cm² field at 10 cm depth. These are used to determine the effective source size of the tungsten target, which is included by convolution in the line spread and point spread kernels, Eq. (3). When an ionization chamber is applied, it is recommended to correct for its non-ideal behaviour. Recently we have derived line spread functions for this purpose, see also Chapter V and VI of this thesis.
- Line dose profiles in a star-shaped pattern at isocentric distance in the maximum field. With these profiles the function \( f(x,y|A_{\text{max}}) \) of Eq. (1) can be determined.
- Output factors in air for several field sizes. These are used to define the relative contribution of the head scatter component.
- Output calibration (dose per monitor unit) in the reference geometry to provide the link between fluence and dose.

Additional measurements are used by the vendor for quality assurance purposes, to check the consistency of the measurements, the quality of the beam fit and the implementation. A
similar set of measurements has to be done for all wedges. Furthermore a geometrical description of the linac head is required to model the head scatter calculation.

An example of the effective energy spectrum that has been determined for 6, 10 and 15 MV-X beams (SLi-15, Elekta, Crawley, UK) is shown in Fig. 7. The energy spectrum contains 28 bins spread over the range from 0.1 MeV to 50 MeV. This spread has been shown to be sufficient because attenuation coefficients and kernels parameters vary slowly with energy.\footnote{This reference is not visible in the image.}

![Effective energy spectra](image_url)

*Fig. 7 Effective energy spectra of 6 MV-X (circles), 10 MV-X (squares) en 15 MV-X (diamonds) photon beams (SLi 15, Elekta, Crawley). The fractional energy fluence per energy is plotted as a function of energy.*

### 3. Verification of the beam model

Prior to clinical application of a TPS, a verification programme, in this paper restricted to dosimetric verification, must assure the correct implementation of the beam model and define the accuracy of the dose calculations. This verification programme must be well-structured to meet all clinical requirements and it must be tailored to the characteristics of the beam model. In section 3.1 the structure of our verification programme is described. The characteristics of the TPS and/or the extent of the verification work may require specific tools and methods to be developed. The methods and tools that we have developed to advance the verification process to a level that meets the needs of conformal radiotherapy are presented in section 3.2. In section 3.3 the tailoring of the verification programme to the characteristics of the beam model is depicted.
3.1 The structure of the verification programme

Safe application of a TPS requires a multi-layer system of quality assurance (QA), as no single check procedure can be guaranteed to be completely fail-safe. The structure of such a QA system can be understood by the analogon of the defences of a medieval city based on several concentric walls, Fig. 8. The outer wall, the verification work (release test) by the vendor is outside the scope of this paper. The first inner wall represents the in-hospital verification (acceptance test) of a TPS in which new releases of software and beam data are verified prior to clinical use. Errors that might pass these walls should be caught by the most inner walls of system-oriented and patient-oriented quality control. In this section emphasis is placed on the tailoring of the in-hospital verification process to the characteristics of the TPS.

During the acceptance stage a set of tests is constructed that should represent all combinations of beam parameters that are clinically used. Each independent parameter can be interpreted as a new dimension in ‘verification space’. As is shown in section 3.1 of Chapter I, this implies an immense number of combinations to be verified. The concept of exploring these combinations is called ‘combinatorial verification’. Although not all ‘grid points’ in verification space need to be explored, it is shown that a restriction to the main axes of this space will miss unacceptable inaccuracies. The implementation of combinatorial verification thus becomes an essential consideration that determines its feasibility. This has been described in section 3.2 of Chapter I. The tools developed for this purpose are briefly reviewed in the next section.

An example of system-oriented QA is an automated checksum procedure that is applied on all patient-independent data in the TPS on a daily basis. Another example is a check of the Hounsfield values of the CT scanner and its representation in the TPS. For this purpose two reference bars, made of perspex and cork, have been mounted beneath the table insert on the CT-scanner. A regular check on the consistency of the corresponding values in our TPS has revealed an error of up to 15% in obese patients, which has been corrected for in a subsequent release of the CT software.

The inner defence wall of the QA system consists of a check on the monitor units for each patient and each field segment. An independent calculation based on an $S_c$, $S_p$ model\(^9\) using independent measurements is implemented in a spreadsheet. All differences larger than ±2% in comparison with the TPS are analysed. Over the period 1995-1999 discrepancies of more than ±2% were found in approximately 1% of all treatment plans, which gives an indication of the solidity of the outer walls. Examples of errors that were traced are: inadvertent inclusion in the patient contour of marker catheters, incorrect use of block and tray material, and an error in the head scatter calculations, that is illustrated in section 3.3.

This independent check is thus a valuable addition to the acceptance procedure, which can realistically not be expected to be fail-safe in all cases.

![Fig. 8 Schematic representation of the concentric multi-layer structure of the verification program of a treatment planning system.](image-url)
3.2 Methods and tools for verification

Faced with the multitude of measurements and comparisons for combinatorial verification of a TPS, we have developed dedicated tools and methods for this purpose.

Measurements are preferable done with a detector that is suited for routine use, without the need for extensive handling and for calibration procedures. A thimble ionization chambers meets these requirements well, except for its limited spatial resolution. This is caused by its size and the replacement of phantom medium, usually water, by detector air. To overcome this disadvantage, research has been done to define the Line Spread Function (LSF) of this type of detector, Chapters V and VI of this thesis. It was concluded that with a correction for the LSF a thimble ionization chamber is well suited for routine high-resolution dosimetry, which is required for verification of conformal radiotherapy treatment planning.

All data selection, profile handling, comparison and documentation that has to be done for combinatorial verification of a beam model has been implemented in a comparison program based on a line dose profile database, as described in section 3.2 of Chapter I. A macro-defined input prevents possible user interaction errors. The LSF-based correction is routinely performed with a maximum likelihood reconstruction technique that has been implemented in this profile comparison program.

A useful tool for the purpose of accuracy assessment is the field accuracy concept which has also been described in more detail in section 2.3 of the first chapter of this thesis. The field accuracy concept presents a practical method to handle the transition between dose difference criteria in low dose gradient regions and distance to corresponding isodoses in regions of high dose gradients. Field accuracy is defined as the smallest numerical value of dose difference in percent or isodose distance in millimetres, with pseudo-unit [%;mm], thus taking advantage of the close numerical correspondence of current dose and distance accuracy requirements. In this concept the dose is always related to the dose in a reference, isocentric geometry (SSD=90 cm, 10x10 cm² field at 10 cm depth in water), thus presenting absolute dose differences, including, in classical terms, discrepancies in output, depth dose curves and off-axis ratios. By application of this concept it can directly be assessed whether or not a criterion of 3 [%;mm], a common acceptance criterion, is met. As an example, Fig. 9 shows calculated as well as measured dose distributions of a 10x10 cm² MLC-shaped field. By inspection of the corresponding field accuracy plot it is directly seen that only in a small spot in the corner of the field the 3 [%;mm] criterion is slightly exceeded.
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Fig. 10 Head scatter dose distribution in an arrow shaped test field (1% dose per grey level). The distribution on the left shows irregularities up to 4% in dose, relative to 100% total dose centrally in the field. To the right the same situation in which the implementation error has been repaired.

Note that the mentioned criterion is used equally for all fields during the acceptance test. The field configurations that were used for the beam-fit have no distinctive status, for instance a requirement of exact reproduction of such fields. There is no argument for such a status in a planning system that is based on a beam model rather than on reproduction of beam data. Furthermore patients will seldom be exactly irradiated with any basic beam configuration.

3.3 Verification tailored to the characteristics of the beam model

The beam model in our TPS has two distinct stages: the calculation of energy fluence exiting the linac and the dose deposition in the patient.

Three quantities contribute to the calculated energy fluence: primary and head scatter photon contributions, Eq. (1), and charged particle contamination. Although a quantitative verification of the individual contributions is difficult to perform within a verification programme, checks for consistency of data and qualitatively correct correlation of these contributions with, for example, field size and source-surface distance, are part of our routine verification procedures. All line dose comparisons therefore not only include total dose but all dose contributions, i.e. due to primary and head scatter fluence, phantom scatter and charged particle contamination.

The complexity of the projection algorithm in the head scatter calculation necessitates specific attention during acceptance, in addition to profile based verification, for example, by inspection of two-dimensional distribution to trace possible irregularities. In Fig. 10 two 2D dose distributions are shown that are solely due to the head scatter component in the energy fluence, the last term in Eq. (1). In this arrow shaped field head scatter dose discrepancies of up to 4% were found. These appeared to be caused by an implementation error that has been corrected in a subsequent release. Note that verification solely along the main scan axes might have missed this dose calculation error.

Other elements of the acceptance procedure that are specifically tailored to this TPS are: a check of intermediate results of the beam fit, such as the effective spectrum, Fig. 7. Furthermore it is checked whether the beam parameters fall within reasonable physical limits and correspond to parameters in similar beams of other linacs.
The second stage in the dose calculation model is the calculation of dose deposition. The most challenging situation for this part of the model is a geometry with tissue heterogeneities. For this purpose the Collapsed Cone algorithm is specifically designed, Eq. (4) and Fig. 5. An example of the improvement in accuracy that is achieved by the Collapsed Cone representation of points spread kernels in comparison to pencil beam kernels is shown in Fig. 11. This is an experiment in which a phantom that simulates a lung tumour geometry is irradiated by a single beam, 6 MV-X or 15 MV-X. A multileaf collimator was used to make the dose distribution in the central plane conformal to the planning target volume (\(\varnothing 80 \text{ mm}\)), a spherical solid tumour of 50 mm diameter with 15 mm margin. This beam definition was done on the TPS with a pencil beam algorithm to assess the (lack of) accuracy of this calculation method. A detailed description of this study is given in Chapter IV of this thesis. A large difference between pencil beam calculations and measurements was found, especially at the edge of the target volume, 13% in a 6 MV-X beam and 21% in a 15 MV-X beam. This is due to the limitations in the pencil beam model which can not properly account for the extended transport range of recoil electrons and scattered photons in low density regions (lung/cork). Calculations done with the Collapsed Cone representation of a point spread kernel show close agreement with measurements, within \(\pm 2\%\) for both beam energies. This illustrates the need for a proper selection per treatment type of the beam model to be applied. In cases with large heterogeneities a pencil beam kernel results in unacceptable inaccuracy whereas it shows adequate accuracy, combined with practicable calculation times, in tissues with little variation in density.

Dose deposition calculated by Monte Carlo methods will become available in our TPS in the near future, but is expected to remain slower then Collapsed Cone calculations. The user
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will than have a choice of three algorithms for which the following applications can be thought of:

- pencil beam for straight-forward applications in mostly homogeneous tissues like the pelvic regions and the brain
- Collapsed Cone for heterogeneous environments like lungs, using multiple CT slices
- Monte Carlo for verification purposes and tumours near tissue interfaces such as occur in the head and neck region

New applications as IMRT/optimization might take advantage of this choice in speed and accuracy. For example, in optimization incidental Collapsed Cone iterations could be alternated with many pencil beam iterations, providing an acceptable result in a reasonable time span, also in heterogeneous environments.

Due to the separation in energy transport and dose deposition the increase in verification work will be less than proportional to the increase in the number of dose deposition algorithms, because they all have the first stage, energy transport, in common. Nevertheless, further streamlining of the verification process is inevitable.

4. Conclusions

In this paper the verification procedure is presented that has been developed to verify the beam model of our TPS, taking into account its physical background to describe energy transport. The specific, and partly complex, characteristics of this model require a tailored quality assurance programme. Methods and tools developed for this purpose include the concept of combinatorial verification implemented in a comparison program and a line dose profile database, application of a routine dosimeter with correction for its Line Spread Function, and the field accuracy concept to allow a direct comparison to current accuracy criteria. The presented ‘concentric wall’ structure serves well this purpose of a structured verification programme, because very few errors are found at the inner wall.

References

2. A.A. van ‘t Veld, this thesis, Chapter I.
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Acknowledgements

Fig. 2 has been drawn in analogy to ref [12]. Figure 5 and 6 are copied from an internal publication of MDS Nordion.