The Clinical Commissioning of Beams for Neutron Capture Therapy

14

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Contents

14.1 Introduction

The optimal energy for a neutron beam intended for BNCT for brain tumours is often referred to as epithermal, i.e. above thermal neutron energy (i.e. above 0.025 eV) [11, 46]. Depending on neutron production and on the design of the filter and the collimators, a neutron beam will exhibit different characteristics with respect to photon and fast neutron contamination [20]. Unique beam filter design has emerged through the computer optimisation process performed at each facility (e.g. [8, 16, 27]), which calls for individual characterisation of each neutron beam. Careful investigation and

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reporting of the properties of the radiation source, as well as the treatment details of clinical trials performed, are equally important in all radiotherapy modalities. The radiation absorbed dose delivered is traditionally one of the principal parameters in radiotherapy, as it is correlated to tissue response [18]. Thus, the uncertainty in the delivered absorbed dose is a treatment parameter that must be kept as low as possible. A few studies have discussed the acceptable level of total uncertainty in the dosimetry of a radiotherapy regime. The ICRU (Report 24 [21]) recommended that radiotherapy dosimetry should aim for an overall uncertainty of no more than 5 %, which has been interpreted as referring to an interval of 2 standard deviations (2 SD) [4]. Other authors suggested [12] that the uncertainty of the dose delivery in external photon and electron therapy should be no more than 3 % (1 SD) for curative treatments. Based on radiobiological considerations, Mijnheer et al. [41] found that the uncertainty of the dose delivery should be no more than 7.0 % (2 SD) in both photon and fast neutron therapy. Until information specifically relevant for NCT is made available, it is reasonable to assume the evaluation provided by Mijnheer et al. [41] applies also to NCT. The uncertainty associated with each individual step in the treatment procedure must thus introduce substantially lower dosimetric uncertainty in order to keep the overall uncertainty within these limits (Ahnesjö et al. [1]).

In the characterisation of mixed neutron-photon beams, it is necessary to quantify each dose component individually as the absorbed dose distributions and the relative radiobiological effect of the components in tissue are different (e.g. [14, 43–45]). We will now limit the discussion to the case of BNCT, but the formulation in this chapter can fairly easily be adapted to for instance gadolinium NCT. In BNCT, the irradiated tissue is subjected to (primarily) four biologically relevant absorbed dose components:

1. The photon absorbed dose
2. The fast neutron absorbed dose
3. The nitrogen absorbed dose
4. The boron absorbed dose

It is here suggested that the following definition is used: the photon absorbed dose is delivered by electrons produced in photon interactions. The boron absorbed doses and the nitrogen absorbed dose are delivered by the charged particles produced by neutron capture in boron and nitrogen, respectively. The fast neutron absorbed dose is the absorbed dose delivered by neutron scatter in hydrogen (producing recoiling protons). Please note that by this definition the "fast neutron absorbed dose" is delivered by neutrons with rather low kinetic energy down to some conveniently selected cut-off energy such as 0.5 eV. Other neutron interaction processes occur that give rise to absorbed dose in tissue, although these are of smaller relevance in BNCT and can often be considered negligible in comparison to the listed. A rigorous dose calculation in neutron beams would thus require a full simulation of neutron, photon and charged particle interactions, and a Monte Carlo-based approach is suitable and could be adopted for the purpose. In clinical BNCT, simplifications are often made in order to make the treatment planning process faster ([52, 53], see Chap. 16 for full details).

The topic of the following sections is the many measures needed to be taken before a radiotherapy treatment can be started.

14.2 Clinical Acceptance

The purpose of the clinical acceptance tests is to ensure that the equipment are safe to use in the clinic. The clinical acceptance procedure encompasses all tests needed to verify that the delivered equipment is meeting the specifications stipulated in the contract. The tests included are agreed upon as part of the purchase. The level of details of the specifications can vary and may even be non-existing such as in the case of a home-grown system. The acceptance procedure then involves the point-wise check of all delivered systems using previously agreed upon customer acceptance procedures. The tests are performed by representative/s of the manufacturer and the clinic. From the clinic's side, the person in charge of such a process is a certified medical physicist expert (EU Directive 97/43). In the case of a non-existing set of customer acceptance procedures, it is worthwhile to formulate a set of tests required to be fulfilled in order to ensure that the systems are safe for clinical use; there are many sources of information that could be drawn upon to this end (e.g. IAEA TRS-430, [5]).

All important parts of a facility must be subjected to an acceptance procedure, e.g. the beam and gantry (if applicable), the patient couch, the imaging system/s, the interlocks, radiation protection and safety, etc. In this chapter, however, we will limit the discussion to the beam. The acceptance and commissioning of the treatment planning system is detailed in Chap. 16. The main topics in interest related to the beam are the beam monitoring system and the beam properties (i.e. beam quality) and the reproducibility. Obviously, a beam intended for BNCT should not be heavily contaminated with photons and fast neutrons, or has a poor reproducibility in terms of beam quality or intensity, as that would compromise patient safety. Such serious problems would need to be corrected before a commissioning phase is started.

The performance of the beam monitors needs to be investigated during the acceptance procedure [9]. In particular, the accuracy, reproducibility and linearity of the beam monitors with neutron and photon fluence and fluence rate need to be carefully investigated as part of the acceptance. As an example, Fig. 14.1 shows one of the initial tests of the four beam monitors installed at the BNCT facility at Studsvik, Sweden. In the simple test shown in Fig. 14.1, the reactor power was stepwise increased, and the beam monitor count rate was recorded. As can be seen, the count rate was occasionally erroneously elevated in a few points. The problem was subsequently identified as a programming error in the control software and was corrected.

During the clinical acceptance phase, it is advisable to study parameters previously suggested by Zamenhof et al. [74]: advantage depth, advantage depth dose rate and advantage ratio. The parameters give an indication whether the neutron beam is well suited for BNCT, such an analysis was provided for instance by Kiger et al. [27] and Giusti et al. [16]. The advantage depth and advantage ratio parameters give an indication of the contamination of photons and fast neutrons. It must be pointed out that it is not sufficient to rely on a computer model for the generation of such data as impurities of the construction materials might significantly alter the beam properties; for instance, small impurities in materials at crucial positions in the beam line might impact the resultant photon component of the

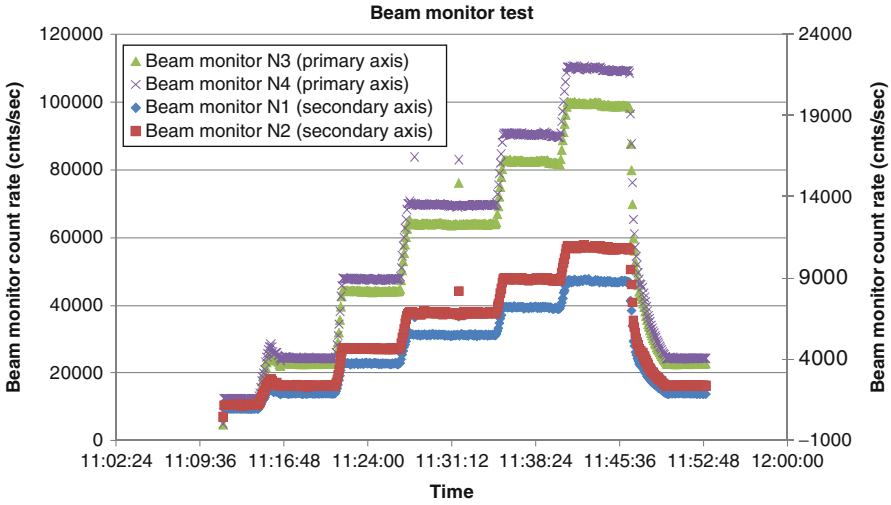

Fig. 14.1 Initial tests of the four beam monitors installed at the Studsvik facility, Sweden. The reactor power was stepwise increased, and the count rate of the beam monitors was recorded. As can be seen, the count rate was occasionally erroneously elevated in a few points. A programming error of the control system was detected and subsequently corrected.

beam. A computer model must be verified against measured data before it is used for calculation of advantage depth, advantage depth dose rate and advantage ratio.

Measurements free in air (or in a mini-phantom) (see for instance, [10]) have some interest for the validation of the computer model, but the clinical relevance of such data is less than in phantom data. In the end, the primary focus is to be able to get convergence of measured and calculated data using the treatment planning system in a tissue-equivalent phantom. The thermal neutron fluence and dose distribution in phantom are not overly sensitive to the photon and neutron spectra [17, 25]. For instance, for pure photon beams, it has been shown that it is adequate to know the average and spread of the energy in order to calculate a depth dose curve in water with high accuracy [25]. In-air measurements are therefore largely left out of the present treatise, and we limit the scope of the discussion to some brief comments. It should be noted that in-air data is useful for the purpose of comparison with existing and decommissioned neutron beams used for therapy such as that compiled by Harling et al. [20]. A beam with larger contamination of photons and fast neutrons than the previously used for BNCT is not advisable to accept for therapeutic purposes. One should be careful, however, to make sure that the same parameters are compared. Specifically, the kerma (kinetic energy released in matter) in air without the presence of the detector needs to be reported (refer to [50], for a full discussion). One should also be aware that significant deviations in calculated and measured in-air data have been observed, even for very elaborate computer models (for instance, [16]).

14.3 Commissioning

The steps that have been taken in the commissioning of facilities to date include characterisation of the beam in air, validation of the computer model usually constructed using a Monte Carlo-based model (e.g. MCNP, Briesmeister et al. [13]), dosimetry under reference conditions and finally clinical commissioning of the beam. These steps differ from those generally taken in commissioning of conventional radiotherapy equipment. However, dosimetric calculations in pure (or nearly so) photon beams are somewhat simpler and do not generally require a Monte Carlo-generated source file for dosimetric calculations of high accuracy, which is generally considered the case for BNCT – though some authors have experimented with a more simplistic approach for epithermal beams [58]. A comprehensive discussion regarding treatment planning systems in BNCT is provided in Chap. 16, and an overview of dosimetric detectors and methods for use in epithermal neutron beams is provided in Chap. 13.

As the beam has been deemed safe in the acceptance procedure, the aim of the commission process is thus to gather all data required for the clinical use. It is generally a quite extensive set of measurement data needed as input for or as verification data for the treatment planning system. The data set will then serve as reference for subsequent quality assurance procedures in which a subset of the commissioning parameters is checked. The procedure is commonly referred to as dosimetry under non-reference conditions. In addition to that, measurements need to be performed where the data is acquired in terms of absorbed dose per beam monitor unit (MU) for a reference point in a phantom, which is generally referred to as dosimetry under reference conditions [2, 4, 71].

14.3.1 Dosimetry Under Reference Conditions

Traditionally, BNCT dosimetry is loosely based on ICRU Report 26 [22] describing neutron dosimetry in biology and medicine and ICRU Report 45 [23] describing clinical dosimetry in fast neutron therapy. ICRU Report 45 was not intended for BNCT, which was also explicitly stated in the report, and it does not address a number of key issues for adequate BNCT dosimetry. The report serves however as a good source of information for neutron dosimetry in general. A report on BNCT dosimetry was recently stipulated by an international work group [71]; the following discussion follows largely the recommendations of that unique work.

14.3.1.1 Choice of Dosimeters

Two types of ionisation chambers are generally used for the determination of photon and fast neutron absorbed dose, while the thermal neutron fluence is best determined using activation detectors [34, 39, 49, 50, 54, 62, 63]. The detectors can be calibrated with a low uncertainty (about 1 %, 2 SD) at standards laboratories [50]. Two ionisation chambers of identical geometrical design but with different material choice is referentially used, which commonly referred to as the “twin” or

"paired ionisation chambers". A common choice is to make use of one chamber with a wall and central electrode of tissue-equivalent plastic (A-150 plastic) and flushed with tissue-equivalent gas ("TE/TE chamber") and one with a wall of magnesium and flushed with argon gas (a "Mg/Ar chamber"). The latter is often referred to as a "neutron-insensitive chamber", which is a fairly reasonable assumption looking at the neutron cross section of the materials, but in reality, oxidation of the magnesium causes a significant neutron response [50, 56]. An alternative ionisation chamber construction is to make use of graphite for wall and central electrode and flushing the chamber with carbon dioxide gas [63]. Graphite is a detector material in widespread use for photon dosimetry in the conventional field [4].

A significant amount of work has been performed using various kinds of dosimeters in epithermal neutron beams, including for instance thermoluminescence dosimeters [6, 7], dosimetry gels [67, 69], diodes [60], prompt gamma methods ([31]; Verbakel et al. [48, 70]), scintillator materials [33], proportional counters [42], activation detectors [10] and bubble detectors [15]. An overview of dosimeters in epithermal beams is presented in Chap. 13. For dosimetry under reference conditions in epithermal beams (often referred to as "absolute dosimetry"), it must be considered that ionisation chambers and activation (primarily) gold foils are currently the standard. Even so, somewhat surprisingly, the data appearing in open literature regarding correction factors to apply in clinical beams in the typical format found for other radiotherapy disciplines, i.e. reports AAPM TG51 [2] and IAEA TRS 398 [4], are scarce.

The use and corrections applicable to ionisation chamber measurements were provided in detail in the IAEA TRS-398 protocol [4], and the discussion provided in this reference is relevant also for measurements in neutron beams. Briefly, ionisation chamber signal collected needs to be corrected for temperature and pressure, polarisation, recombination effects and the electrometer charge collection correction. In addition, there is possibly also an unwanted signal arising from activation of parts of the detector materials when placing an ionisation chamber in the neutron beam. The signal caused by activation could be difficult to account for in practice considering that the irradiation history might not be known in sufficient detail. At the very least, the error introduced needs to be estimated and included in the uncertainty analysis.

14.3.1.2 Choice of Phantom

The ICRU Report 45 promotes the determination of absorbed dose to tissue inside a tissue-equivalent phantom, for instance in a water phantom. In fast neutron beams, the choice is reasonable given that the absorbed doses to water and tissue are comparable, and hence, the corrections required to account for the differences in the neutron/photon interaction properties are rather close to unity. This has then become the tradition within the neutron therapy community.

The situation is different in an epithermal neutron beam, however, where the total neutron absorbed dose during BNCT differs quite substantially in tissue and water. This dosimetric difference is due to the contributions of boron and nitrogen

neutron capture, causing the corrections related to the interaction properties to deviate quite strongly from unity. In addition, the geometry of the irradiated object affects the absorbed dose rate considerably in an epithermal neutron beam [51, 59]. Therefore, reporting absorbed doses resulting from boron and nitrogen capture, photons and fast neutrons to tissue inside a water phantom are not of high clinical relevance. By determining and reporting absorbed doses to the materials in which the measurements were performed, the problem is avoided in the sense that the clinical relevance of the values is not implied. This concept was introduced in Munck af Rosenschöld et al. [49]. More importantly, the suggested methodology allows adaptation of the mathematical formalism and dosimetric procedures concerning ionisation chambers to that formulated in the IAEA TRS-398 protocol [4]. The procedures in the IAEA report form the basis for radiotherapy dosimetry in general, and the exception is then only the fast neutron therapy field.

The effect of phantom material composition and size has been studied previously in epithermal neutron beams [32, 59, 65, 73]. In a previous work, an artificial "liquid brain" mixture was found to serve as an appropriate phantom material for dosimetry in epithermal neutron beams [65]. In other works, the authors used an ellipsoidal phantom for dosimetry under reference conditions to have a better representation of a human head [19, 63]. The material and geometric corrections from PMMA to brain tissue containing boron applicable in the Studsvik beam were presented in Munck af Rosenschöld [51], and strongly indicate their importance in BNCT dosimetry.

The international report on dosimetry of BNCT suggests the use of a water phantom for dosimetric measurements [71], the reference serves as an excellent and comprehensive guide for BNCT dosimetry, and this chapter largely adheres notation and methods described in that publication.

Though the geometry and material composition of a phantom have a large impact on the mixed neutron and photon radiation field of a beam optimised for BNCT, for dosimetry under reference conditions, a water phantom of a simple geometrical shape appears the best choice. Water is readily available, is cheap and practical to use and is also the choice for all other radiotherapy disciplines. Further, having a simple phantom geometry and composition to use for dosimetry under reference conditions simplifies future standardisation of measurement methodology and the collection and tabulation of correction factors for recommended dosimeters. However, in the subsequent step of commissioning the treatment planning system, it is useful to investigate the accuracy of the system to handle the effects of various geometrical shapes and composition in order to match the treatment situation more closely.

14.3.1.3 General Formalism

The commonly accepted formalism used in all disciplines of radiotherapy is here adopted and extended to cover neutron therapy [4], similar to what was previously suggested and presented [49], [50]. When a detector is calibrated in terms of absorbed dose to water is used at the reference depth in a water phantom for a

reference beam quality (Q_0) and in the absence of the detector, the absorbed dose is given by [2, 4]

$$D_{w,Q_0} = M_{Q_0} \cdot N_{D,w,Q_0} \tag{14.1}$$

In this work, it is assumed that the detector response could be separated into a signal arising from photons (index γ), fast neutrons (index *fn*) and thermal neutrons (index *tn*), giving

$$M_Q = M_Q^{\gamma} + M_Q^{fn} + M_Q^{tn} \tag{14.2}$$

Here, M_Q is the total detector response corrected for quantities affecting the measurement. The detector reading includes a response caused by interactions in the detector structures in the active medium in the detector. The corrected detector reading can be related to the absorbed dose to water (D_w) at the point of measurement in beam quality (Q) through the following equations:

$$D_{w,Q}^{\gamma} = M_Q^{\gamma} \cdot N_{D,w,Q}^{\gamma} \tag{14.3}$$

$$D_{w,Q}^{tn} = M_Q^{tn} \cdot N_{D,w,Q}^{tn} \tag{14.4}$$

$$D_{w,Q}^{fn} = M_Q^{fn} \cdot N_{D,w,Q}^{fn} \tag{14.5}$$

Three detectors with a different response to photons, thermal and fast neutrons are used in order to resolve the equation system arising from Eqs. 14.2, 14.3, 14.4, and 14.5. Instead of the dose to water from thermal neutrons (Eq. 14.4), it might be more convenient to refer to the thermal neutron fluence.

14.3.1.4 Photons

The calibration factor, $N_{D,w,Q}^{\gamma}$, needs to be known in order to derive the photon absorbed dose in the absence of the detector in the mixed beam. In a mixed radiation field, the calibration factor needs to be corrected by a beam quality correction factor that accounts for differences in perturbation effects and sensitivity (energy response) of the chamber compared to the calibration field. This yields a chamber calibration factor that can be used in the mixed radiation field Q, i.e.

$$N_{D,w,Q}^{\gamma} = N_{D,w,Q_0} \cdot k_Q^{\gamma} \tag{14.6}$$

where N_{D,w,Q_0} is the chamber calibration factor provided by a standards laboratory, herein assumed to be the quality of ^{60}Co gamma-rays, and k_Q^{γ} is the beam quality

correction factor applicable to the mixed radiation field for photons. This factor is therefore equivalent to the k_Q [2] and k_{Q,Q_0} factors [4] given by the recent dosimetry protocols based on absorbed dose to water standards. The k_Q factor is equal to unity for the reference beam quality by definition. The k_Q^γ factor for a mixed beam needs to be calculated. To my knowledge, presently, only data is available for a magnesium-walled and argon-flushed and an A-150-walled ionisation chamber for a decommissioned epithermal neutron beam in open literature [49]. In that reference, it was also shown that the beam quality of the epithermal neutron beam was similar to ^{60}Co gamma-rays which therefore is a reasonable reference beam quality. Equations 14.1 and 14.3 give

$$k_Q^\gamma = \frac{D_{w,Q}^\gamma / M_Q^\gamma}{D_{w,Q_0} / M_{Q_0}} \tag{14.7}$$

Preferably, the k_Q^γ factor is known through measurements in a number of beams. This is however not realistic in epithermal neutron beams given the dosimetric complexities involved of mixed beams and the lack of methods for absolute dosimetry methods. Instead, one has to rely on a calibration in a pure photon beam and calculations for the determination of a suitable correction of stopping power ratios and perturbation effects. Assuming that the detector signal per unit absorbed dose to the gas inside the ionisation chamber is the same regardless of the beam quality, one has

$$k_Q^\gamma = \frac{D_{w,Q}^\gamma / D_{\text{gas},Q}^\gamma}{D_{w,Q_0} / D_{\text{gas},Q_0}} \tag{14.8}$$

where D_{gas} is the absorbed dose to the detector gas originating from photons in the mixed beam (Q) and in the calibration (Q_0) beam. The assumption made in Eq. 14.8 is in fact the same as used in conventional photon and electron beam dosimetry, i.e. the average energy required for producing an ion pair in the detector gas is constant for the two beam qualities Q and Q_0 (c.f. IAEA TRS 277, Eqs. (5a) and (5b), [3]). All the factors in Eq. 14.8 can be calculated using a Monte Carlo computer program with a model of the detector and the two radiation beams and thus giving the k_Q^γ factor [49]. In lack of calculated data for the beam of interest, it might be necessary to assume that the k_Q^γ factor is equal to unity and assign an appropriate uncertainty.

14.3.1.5 Thermal Neutrons

The absorbed dose to water at point of interest can be derived as (assuming charge particle equilibrium)

$$D_{w,Q}^{tn} = f_{w,Q}^{tn} \cdot \Phi_{w,Q}^{tn} \tag{14.9}$$

where $f_{w,Q}^{tn}$ is the fluence-to-kerma conversion factor (i.e. "kerma factor") for water at the reference point in water in beam quality Q applicable for the thermal

neutron group fluence $\phi_{w,Q}^{tn}$. High-purity gold foils are recommendable for the determination of $\phi_{w,Q}^{tn}$, which is given by the following relation:

$$\dot{\phi}_{w,Q}^{tn} = A_{\mathrm{sat},Q} \cdot \left(\frac{\dot{\phi}_{w,Q}^{tn}}{A_{\mathrm{sat},Q}} \right)_{\mathrm{MC}} \tag{14.10}$$

Here, $\dot{\phi}_{w,Q}^{tn}$ is equal to the neutron fluence rate of the thermal group at the reference point without the presence of the foil, $\left(\dot{\phi}_{w,Q}^{tn}\right)_{\mathrm{MC}}$ is the corresponding thermal group fluence rate per source particle calculated by means of the Monte Carlo method for beam quality Q, $A_{\mathrm{sat},Q}$ is the measured saturated activity of the gold foil in Bq per gram of the sample and $\left(A_{\mathrm{sat},Q}\right)_{\mathrm{MC}}$ is the corresponding calculated saturated activity of the gold foil in Bq per gram of the sample and per source particle using the Monte Carlo method. The factors $\left(\dot{\phi}_{w,Q}^{tn}\right)_{\mathrm{MC}}$ are calculated in the position of the foil without the presence of the foil, and $\left(A_{\mathrm{sat},Q}\right)_{\mathrm{MC}}$ is calculated with the gold foil included in the computer model, preferably using Monte Carlo method. Thus, the ratio intrinsically includes the appropriate correction for the perturbation caused by the foil itself on the neutron field in the phantom at the reference position, within the limits of the accuracy of the Monte Carlo model.

The user could perform a comparative measurement using for instance a high-purity germanium crystal detector set-up with fixed settings of the analysis program (see for instance, Knoll [29] for information on such systems). Thus, allowing for a conversion between a signal measured (M_Q^{tn}) and the saturated activity reported by the standards laboratory for a fixed set of experimental conditions.

14.3.1.6 Fast Neutrons

The beam quality correction factor for fast neutrons is given by (cf Eq. 14.8)

$$k_Q^{fn} = \frac{D_{w,Q}^{fn} / M_Q^{fn}}{D_{w,Q_0} / M_{Q_0}} \tag{14.11}$$

where the factors were defined previously. Assuming that the detector reading can be written as the product of the absorbed dose delivered to the detector gas, $D_{\mathrm{gas},Q}$, the inverse of the average energy required to produce an ion pair in the detector gas for the actual charged particle spectra for beam quality Q, $\left(e/W\right)_Q^{\mathrm{eff}}$, and the mass of the detector gas, m_{gas}, gives

$$M_Q = D_{\mathrm{gas},Q} \cdot \left(e/W\right)_Q^{\mathrm{eff}} \cdot m_{\mathrm{gas}} \tag{14.12}$$

Inserting Eq. 14.11 in Eq. 14.12 gives

$$k_Q^{fn} = \frac{D_{w,Q}^{fn} \cdot W_Q^{fn,\mathrm{eff}} / D_{\mathrm{gas},Q}^{fn}}{D_{w,Q_0} \cdot W_{Q_0}^{\mathrm{eff}} / D_{\mathrm{gas},Q_0}} \tag{14.13}$$

Multiplication of dividend and divisor by $\left(f_m / f_t\right)_Q^{fn}$, i.e. the kerma factor ratio for the detector wall material, which is A-150 plastic (index = m), and muscle tissue (index = t) weighted by the actual neutron spectra at the point of interest gives

$$k_Q^{fn} = \frac{1}{k_t} \cdot \frac{D_{w,Q}^{fn} / D_{\mathrm{gas},Q}^{fn}}{D_{w,Q_0} / D_{\mathrm{gas},Q_0}} \cdot \left(f_m / f_t\right)_Q^{fn} \tag{14.14}$$

In Eq. 14.14, $k_t = \frac{\left(f_m / f_t\right)_Q^{fn} \cdot W_{Q_0}^{\mathrm{eff}}}{W_Q^{fn,\mathrm{eff}}}$, which is a simplified form of the neutron sensitivity factor for a tissue-equivalent detector for muscle tissue, k_t, that was calculated by Jansen et al. [24] as a function of neutron energy; it was given in its complete form in the ICRU Report No. 45. Calculation of the factors in Eq. 14.14 is possible by means of the Monte Carlo method.

14.3.2 Dosimetry Under Non-Reference Conditions

For clinical use, central-axis percentage depth dose (PDD) curves beam profiles (typically at several depths), and beam components as a function of distance from the aperture need to be measured. It might be practical and beneficial to use other dosimeters for the determination of the relative distributions as compared to the preceding chapter. For instance, using dosimeters with less need for MC derived corrections with high signal to noise appear attractive for use as long as the relative sensitivity to the beam dose components of the dosimeters can be established accurately.

The usefulness of varying the field size is probably less for NCT than for conventional photon therapy, so the number of useful field size combinations are likely to be less. The magnitude of the beam dose components and the relative distribution of the components does vary for as a function of aperture size for epithermal neutron beams (Raaijmakers et al. [57]). Therefore, if different field sizes or beam apertures are available, the dosimetry procedures need to be repeated for each beam.

14.4 Clinical Dosimetry

Once the dosimetric properties of the beam have been determined with sufficient accuracy and reproducibility, the following step involves the implementation of the accumulated data into the treatment planning system (TPS). If the implementation is done accurately, the TPS is then able to simulate a treatment set-up and derive the

resulting dose distribution, allowing for a certain amount of optimisation. Commissioning and use of a TPS are presented in Chap. 16. We here limited the discussion to a few comments regarding the actual implementation of beam data.

In the commissioning of the TPS, one needs to compare the calculated data in phantom vs. the measured data. At that point, it might be necessary to adjust the relative magnitude of the dose components in the computer source description in order to improve the agreement towards the measurements. In this comparison, it is of great importance to make sure that the same dosimetric data is used in all steps of the process, i.e. the same kerma factors and/or stopping power data are used in the TPS as are used in the derivation of absorbed dose in the preceding step. In the author's opinion, it is reasonable to normalise the TPS calculations towards the thermal neutron group fluence per beam monitor unit at the reference point in a water phantom. This might be advisable considering that the kerma factor for the thermal neutron absorbed dose in water is low (and the transition from fluence to absorbed dose does not improve the accuracy of the procedure). Then, adjust the photon intensity coming from the beam to match the measured photon absorbed dose per beam monitor unit at the reference point. The measurement of the fast neutron absorbed dose is generally very uncertain using the paired ionisation chamber technique (see [56], and others); therefore, in a similar fashion, adjusting the relative intensity of the fast neutron component of the beam based solely on ionisation chamber measurements for a well-optimised epithermal neutron beam is questionable.

The geometry and the material content of the irradiated volume in an epithermal neutron beam have great impact on the dose distribution ([19]; Wojnecki et al. [73]; [51]). The TPS ability to account for such effects correctly should be independently verified using calculations or phantom experiments (or both). The absorbed dose of a single treatment field to a patient (D_{pat}) of the dose component i to be delivered is given by the simple relation:

$$D_{\mathrm{pat},i} = \left(\frac{D_{\mathrm{pat},i}}{D_{\mathrm{ref},i}} \right)_{\mathrm{TPS}} \cdot \left(\frac{D_{\mathrm{ref},i}}{M} \right)_{\mathrm{Measured}} \cdot M \tag{14.15}$$

Here, M is the total number of beam monitor unit counts, $D_{\mathrm{ref},i}/M$ is the measured absorbed dose of component i per beam monitor count under reference conditions and the $D_{\mathrm{pat},i}/D_{\mathrm{ref},i}$ ratio is calculated using the TPS. Note that for $i = boron$ and $i = nitrogen$, $D_{\mathrm{ref},i}$ is replaced by $\phi_{\mathrm{ref},i}$ (i.e. the thermal neutron fluence determined under reference conditions).

14.5 Quality Assurance

In order to ensure safe radiotherapy, continuous quality assurance (QA) of equipment and procedures is of paramount importance. The subject of QA in radiotherapy has been discussed extensively in the literature (see, e.g. [35, 36]) and specifically

for a BNCT facility [9]. Rassow et al. [60] has compared the QA of medical accelerators and an epithermal neutron beam, which constitutes a good starting point for a QA programme. QA of beam output, photon contamination and neutron quality, as well as the stability of dosimeters, is of importance for a safe clinical practice. Raaijmakers et al. [56] investigated the long-term stability of these parameters for the epithermal neutron beam in Petten facility.

The same procedures as for conventional radiotherapy apply to the QA of epithermal neutron beams; therefore, the recent report by the American Association of Physicists in Medicine (AAPM) task group report 142 [28] provides a guideline and provides tolerances that could arguably be used also for epithermal neutron beams.

In addition to the standard tests of the neutron beam and the dosimeters, quality assurance procedures need to be established for the boron concentration measurement of tissue samples (Kobayashi et al. [26]; [30, 37, 38, 47, 55, 64, 68]), the measurement system for activation measurements [9] and in vivo dosimetry [51, 66, 72].

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