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ABSTRACT

In radiation therapy using electron and photon beams the dosimetry chain consists of several sequential phases starting by the realization of the dose quantity in the Primary Standard Dosimetry Laboratory and ending to the calculation of the dose to a patient. A similar procedure can be described for the dosimetry of epithermal neutron beams in boron neutron capture therapy (BNCT). To achieve the required accuracy of the dose delivered to a patient the quality of all steps in the dosimetry procedure has to be considered. This work is focused on two items in the dosimetry chains: the determination of the dose in the reference conditions and the evaluation of the accuracy of dose calculation methods. The issues investigated and discussed in detail are: a) the calibration methods of plane parallel ionization chambers used in electron beam dosimetry, b) the specification of the critical dosimetric parameter i.e. the ratio of stopping powers for water to air, $(S/r)_{air}^{water}$, in photon beams, c) the feasibility of the twin ionization chamber technique for dosimetry in epithermal neutron beams applied to BNCT and (d) the determination accuracy of the calculated dose distributions in phantoms in electron, photon, and epithermal neutron beams.

The results demonstrate that up to a 3% improvement in the consistency of dose determinations in electron beams is achieved by the calibration of plane parallel ionization chambers in high energy electron beams instead of calibrations in ^{60}Co gamma beams. In photon beam dosimetry $(S/r)_{air}^{water}$ can be determined with an accuracy of 0.2% using the percentage dose at the 10 cm depth, *%dd(10),* as a beam specifier. The use of *%dd(10)* requires the elimination of the electron contamination in the photon beam. By a twin ionization chamber technique the gamma dose can be determined with

uncertainty of 6% (1 standard deviation) and the total neutron dose with an uncertainty of 15 to 20% (1 standard deviation). To improve the accuracy of the twin ionization chamber technique in epithermal neutron beams, improvements in chamber materials, and development of neutron beam calibration facilities for chambers are required. The general accuracy achieved by treatment planning systems is approximately 4% for photons and 5 to 7% for electrons. Large (> 10%) deviations in calculated doses are possible even when relatively modern calculation approaches are used. For treatment planning systems used in BNCT, intercomparisons and similar validation procedures as for treatment planning systems for electron and photon beams should be performed.

CONTENTS

LIST OF PUBLICATIONS

This thesis is based on the following original publications, which are referred to in the text by their Roman numerals, Publ.I-V.

I Kosunen A, Järvinen H. Application of $Li_2B_4O_7$ TL detectors for control of inhomogeneity calculations by radiotherapy treatment planning systems. Radiat. Prot. Dosim. 1990; 34: 257–260.

II Kosunen A, Järvinen H, Vatnitskij S, Ermakov I, Chervjakov A, Kulmala J, Pitkänen M, Väyrynen T, Väänänen A. Intercomparison of radiotherapy treatment planning systems for external photon and electron beam dose calculations, Radiother. Oncol. 1993; 29: 327–335.

III Kosunen A, Rogers DWO. Beam quality specification for photon beam dosimetry, Med. Phys. 1993; 20: 1181–1188.

IV Kosunen A, Järvinen H, Sipilä P. Optimum calibration of NACP type plane parallel ionization chambers for absorbed dose determination in low energy electron beams, IAES-SM-330/41, 1994: 505–513.

V Kosunen A, Kortesniemi M, Ylä-Mella H, Seppälä T, Lampinen J, Serén T, Auterinen I, Järvinen H, Savolainen S. Twin ionization chambers for dose determinations in phantom in an epithermal neutron beam, Radiat. Prot. Dosim. 1999; 81: 187–194.

NOMENCLATURE

AAPM	American Association of Physicists in Medicine
BNCT	Boron Neutron Capture Therapy
CoP	Code of Practice
CT	Computed Tomography
ESTRO	European Society for Therapeutic Radiology and Oncology
IAEA	International Atomic Energy Agency
IEC	International Electrotechnical Commission
IPEMP	Institution of Physics and Engineering in Medicine and Biology
ISO	International Organization for Standardization
J_{10}/J_{20}	Ratio of ionization at depth of 10 cm to that at depth of 20 cm
LET	Linear Energy Transfer
MC	Monte Carlo
MRI	Magnetic Resonance Imaging
NACP	Nordic Association of Clinical Physics
PP	Plane Parallel
PSDL	Primary Standard Dosimetry Laboratory
QA	Quality Assurance
QC	Quality Control
SSDL	Secondary Standard Dosimetry Laboratory
TE	Tissue Equivalent
TLD	Thermoluminescent dosimetry
TPR_{10}^{20}	Tissue-phantom-ratio. Ratio of doses at depths of 20 cm and 10 cm
TPS	Treatment Planning System
WHO	World Health Organization
2D	Two dimensional
3D	Three dimensional
A150	Plastic substitute material for muscle tissue
C	Factor for an ionization chamber to convert the calibration factor in laboratory conditions to that in a user beam
$C_{n,K}^{TE}$	Factor for a TE ionization chamber to convert the air kerma calibration factor in a ^{60}Co gamma beam to that for neutron radiation
$C_{g,K}^{TE}$	Factor for a TE ionization chamber to convert the air kerma calibration factor in a ^{60}Co gamma beam to that for photon radiation
D_n	Absorbed dose to a medium from neutron radiation
D_g	Absorbed dose to a medium from photon radiation

$d(80\%)$	Depth of the 80% dose level
$\overline{E}_0$	Mean energy at phantom surface
$\overline{E}_Z$	Mean energy in phantom at depth z
g	Fraction of energy of secondary charged particles that is lost to bremsstrahlung in air in a ^{60}Co gamma beam
$(K)_2^1$	Ratio of kermas for medium 1 to medium 2
k_m	Factor to take into account the non-air equivalence of the chamber wall in air in a ^{60}Co gamma beam
k_{att}	Factor to take into account the attenuation and scatter of photons in the wall of the chamber in a ^{60}Co gamma beam
N	Calibration factor for an ionization chamber
N_D	Absorbed dose -calibration factor for an ionization chamber
N_K	Air kerma -calibration factor for an ionization chamber
p	Factor to take into account the difference of the electron and photon/neutron fluence in an ionization chamber and in a phantom*
Q	Electric charge
R_{50}	Half value depth for electrons
R_p	Practical range for electrons
r_{gas}^{wall}	Gas-to-wall dose conversion factor
$(S/r)_2^1$	Ratio of mass collision stopping powers for medium 1 to medium 2
W	Energy required to produce an ion pair
$\%dd(10)$	Percentage depth dose at 10 cm
$(m/r)_2^1$	Ratio of mass energy absorption coefficients for medium 1 to medium 2
$1sd$	One standard deviation

* Perturbation factor p can have a subscripts c or u depending on the beam considered i.e. subscript c for calibration beam and subscript u for user beam. In Publ. IV symbol $p_{u,\,pp}$ is used for perturpation factor both in calibration and user beams.

1 AIM OF THE STUDY

The aim of this study is to analyse and improve the reliability of dosimetry of electron and photon beams in external radiation therapy and neutron beams in boron neutron capture therapy (BNCT). High energy electron and photon beams and epithermal neutron beams are considered. The work is focused on two items in the dosimetry chains: determination of the dose in the reference conditions and evaluation of the accuracy of the dose calculation methods.

The specific aims of the study are:

a) to investigate the effect of the calibration methods of the plane parallel ionization chambers on the accuracy of dose measurements in electron beams,

b) to establish a new radiation beam quality specifier for dosimetry of photon beams,

c) to evaluate the feasibility of the twin ionization chamber method for dose determinations in epithermal neutron beams applied to BNCT,

d) to study the accuracy of the calculated dose distributions in phantoms in electron, photon and epithermal neutron beams.

2 INTRODUCTION

Radiation therapy is a clinical modality dealing with the use of high doses of ionizing radiations in the treatment of patients mostly having malignant tumors. The aim of radiation therapy is to deliver a sufficiently high absorbed dose to a defined target volume resulting in the eradication of the tumor with as minimal a damage as possible to the surrounding healthy tissues.

Radiotherapy with external high energy photon beams is the most common radiotherapy modality [116]. Electron beams are used either as the primary mode of radiation therapy or combined with photon beams. High energy electron and photon beams are typically produced by linear accelerators and also by microtrons [59, 107]. The traditional ^{60}Co gamma units and betatron electron accelerators are still in use in some radiotherapy centres. For a typical multi-energy medical linear accelerator the accelerating potential spans between 4 and 25 MV. The mean energy of the electron beams is between 4 and 22 MeV and of photon beams between 1.5 and 7 MeV [59, 77]. In external beam therapy, to optimize the dose distribution in a patient the dose is typically delivered from different directions by beams of modified shape and intensity.

Boron Neutron Capture Therapy (BNCT) is a special radiation therapy modality where the dose is targeted to the tumor cells by a pharmaceutical containing the ^{10}B-compound [9]. In the modern BNCT neutron beams the mean energy of the spectrum is adjusted at the epithermal energy range (from 0.5 eV to 10 keV), so that the final thermalisation of neutrons takes place in the target tissue [32]. The potential sources for BNCT neutron beams are nuclear reactors, proton accelerators and neutron emitting isotopes. So far, only nuclear reactors have been used for neutron production in BNCT treatments [32]. In BNCT the therapeutic dose is produced by the high-LET particles released in a $^{10}B(n,\alpha)^{7}Li$ reaction, as the target area is exposed to thermal neutrons. Most of the unwanted radiation in tissue is gamma radiation from the neutron capture of hydrogen where 2.23 MeV photons are emitted. Other significant components inducing dose are the recoil protons of epithermal and fast neutrons from the neutron capture of nitrogen [31].

The concept of dosimetry covers the determination of absorbed dose to a medium both by measurement and by calculation. Dosimetry covers also the two different aspects of radiation: the description of the radiation quality

and the description of the energy deposited in a medium (the absorbed dose) [57]. Metrology is a field of knowledge of measuring a value of a physical quantity, including both the theoretical and the practical aspects of measurement [53]. Measurement of absorbed dose of ionizing radiation is an exceptional area in metrology as the accuracy achieved in measurements at laboratory conditions is close to the accuracy required in typical applications. The requirement of accuracy is most emphasized in radiotherapy where the reference dose in a clinic should be measured with the same level of accuracy as in a dosimetry laboratory [6, 16, 44, 74, 117].

To assure the consistency of dosimetry and the metrological traceability of calibrations, the dosimetry procedures in radiation therapy are guided internationally. For common radiotherapy modalities, such as external beam therapy by high energy electrons and photons or by fast neutrons, the guidance for dosimetry is established by organizations of metrology and medical physics [18, 39, 47, 52, 80]. In these radiotherapy modalities the improvements in the dosimetric methods and data come into effect only after the improvements are implemented in the widely used dosimetry reports and Codes of Practice (CoP) for dosimetry. For uncommon and specific type of radiotherapy modalities, such as BNCT, no uniform international guidance for dosimetry exists, so far. In BNCT the work towards coherent dosimetry procedures was started in 1998, when a European project for a CoP of epithermal neutron beams in BNCT was launched [111].

In dosimetry of radiation therapy the importance of quality assurance (QA) has been emphasized since the establishment of systematic dosimetry procedures [44, 79, 80, 81]. During the recent years interest in quality-oriented systems has increased in the medical field. The American Association of Physicists in Medicine (AAPM) has published recommendations for a comprehensive QA for radiation therapy in 1994 and the European Society for Therapeutic Radiology and Oncology (ESTRO) has presented a frame for a quality system based on the guidelines of the International Standardization Organization (ISO) in 1997 [1, 30, 55]. In the BNCT community the validity of quality-oriented treatment procedures has been recognized [103] and the baselines for a dosimetry quality program has been presented by *Rassow et al.* [95].

The experimental dosimetry of electron and photon beams is based on the use of ionization chambers. Ionization chambers are also recommended as reference dosemeters for mixed neutron and gamma beams of fast

neutrons. In dosimetry of epithermal neutron beams various types of dosemeters are required for dose characterization, because of the mixed neutron and gamma radiation field and the spread of the neutron energy and fluence. The twin ionization chamber technique has been applied for the determination of neutron and gamma doses in epithermal neutron beams for BNCT [93, 100].

The primary measurement standards at the Primary Standard Dosimetry Laboratories (PSDL) and the dissemination of absorbed dose to users have traditionally been based on physical quantities “in air”. The calibrations of radiotherapy dosemeters have been, and mainly still are, performed for air kerma in a ^{60}Co gamma beam [10]. During this decade calibrations for absorbed dose to water in PSDLs have become available and CoPs based on the absorbed dose to water are under development [12, 99]. In PSDLs calibrations for absorbed dose to water in other than ^{60}Co gamma beams are possible [10, 20]. According to the hierarchic system of metrological laboratories the calibrations of user dosemeters are commonly performed in Secondary Standard Dosimetry Laboratories (SSDL). To improve the availability of SSDL services world-wide the International Atomic Energy Agency (IAEA) and the World Health Organization (WHO) have established a network of SSDLs [41]. SSDLs provide a direct linkage of national dosimetry standards to the international measurement system and offer calibration, training and quality audit services for radiation therapy clinics [34, 39, 41]

The work is concentrated on the metrological aspects of experimental dosimetry and the evaluation of the accuracy of the calculated doses. The wide use of NACP-type plane parallel (PP) chambers in electron beam dosimetry and the reported individual variations of the type-related critical parameters of these chambers in a ^{60}Co gamma beam used for calibration have lead to investigations of different calibration methods of NACP PP chambers [Publ. IV]. In photon beam dosimetry the validity of the commonly used beam specifier, i.e. the tissue-phantom-ratio, TPR_{10}^{20}, as a general beam specifier is investigated and the accuracy of the alternative photon beam specifier, i.e. the percentage depth dose at 10 cm depth, *%dd(10)*, is studied through the Monte Carlo (MC) simulations of photon beams [Publ. III]. For epithermal neutron beams the twin ionization chamber technique has been applied, so far, to determine only the epithermal+fast neutron dose and the gamma dose. In this study the feasibility of ionization chambers for the measurement of the total neutron dose and the gamma dose is investigated [Publ. V]. To investigate the accuracy of calculated

doses by treatment planning systems (TPS) of electron and photon beams and to improve test methods for QA an intercomparison of TPSs was performed [Publ. I and II]. For epithermal neutron beams a comparison of calculated and measured dose distributions is also presented [Publ. V].
The dosimetry procedures of epithermal neutron beams for BNCT are described parallel to those of electron and photon beams. The methods and formalism for the measurement of dose in the reference conditions, the methods for dose calculation by TPSs, and the principles for QA in radiotherapy are described. Finally, the outcome of the study is discussed and the implementations of the results into the international CoPs for dosimetry are described.

3 QUALITY OF RADIOTHERAPY DOSIMETRY

3.1 Dosimetric accuracy requirements

In radiation therapy with external radiation beams the absorbed dose at the specification point in a patient should be known with an overall uncertainty of 3.5% (*1sd*)* [16, 44, 74, 117]. The main reason for the requirement of high accuracy in dose delivery is the typically narrow dose margin between the dose needed for tumor control and the dose causing complications for healthy tissues. The size of the dose margin depends on the biologic characteristics of tissues, the size and shape of the irradiated tissue volumes, the quality of radiation, as well as the dose fractionation regime. Due to individual nature of the dose margin, 3.5% (*1sd*) uncertainty can be regarded as a general level accuracy requirement for dose delivery [16, 74, 117]. Besides the individual treatments, the high accuracy in dose delivery is essential to enable a reliable analysis and a comparison of results of different radiation therapy techniques and modalities [22, 23, 44].

In BNCT the succcess of a treatment depends on a number of parameters including tumor location and depth, boron content in tumor and healthy tissues, and neutron fluence. The biologic effectiveness depends not only on the quality of ionizing particles present in the tissues, but most importantly on the intracellular boron concentration [9]. Furthermore, the intracellular boron concentration is critically dependent on the type of the boron pharmaceutical used. Mainly due to this complicated, microdosimetric and pharmacokinetic nature of producing the therapeutic effect in BNCT, generally accepted accuracy requirements for the delivered dose are difficult to define and so far they do not exist.

In typical BNCT treatments of brain tumors 30% to 50% of the dose in the brain is produced by the beam radiation components (non-boron dose) [9, 93]. On the other hand, the dose from the neutron capture by boron is directly proportional to the thermal neutron fluence, also responsible for production of the main fraction of the beam dose in tissues. The delivery of

* According to the recommendations of International Committee for Weights and Measures (CIPM) the estimated accuracy is expressed in uncertainties. The uncertainties are classified to those estimated by statistical means (Type-A) and to those estimated by other means (Type-B). Both types of uncertainties are expressed by standard deviation (*1sd*). The overall uncertainty of a process includes combination of both types of uncertainties from all steps in a process [53].

a relatively high beam dose to tissues requires therapeutic accuracy intrinsically and the requirement of accuracy is even more pronounced due to the linkage of boron and non-boron doses through the thermal neutron fluence. Based on the arguments above the same level of accuracy as in the electron and photon radiotherapy can be justified for the beam dose in BNCT.

In the quality standards of ISO the quality is defined as "*the totality of features and characteristics of a product or service that bear on its ability to satisfy stated or implied needs*" [55]. Based on the requirements for curative radiotherapy it can be stated that the "need" for radiotherapy dosimetry is the "need of accuracy". Furthermore, to achieve or even get close to the required accuracy the features of the whole treatment and dosimetry procedures have to be considered.

3.2 Treatment and dosimetry procedures

The treatment procedures for external electron and photon radiation therapy and for BNCT radiotherapy includes several sequential phases. The schematic presentation of the treatment procedures is presented in Table I.

Treatment planning includes all the steps prior to treatment delivery. Dose planning is a process used to prescribe the target volume, dose and the dose fractionation regime and also the number, orientation, type and characteristics of the radiation beams [16, 30]. In BNCT the planning for the concentration and distribution of boron in a patient is a part of dose planning [9, 27].

In the external electron and photon beam dosimetry the dosimetry chain starts by the realization of the dose quantity in the PSDL leading by an unbroken chain of calibrations of ionization chambers to the measurement of the absorbed dose to water in the reference conditions in a clinic and, finally, ending to the calculation of the dose to a patient [30, 39, 44, 46]. The beam specific reference dose of the treatment unit is used for the calibration of the beam monitoring detectors (or the timer) and for the normalization of the further relative dose measurements required [39, 44]. Dosimetry enters the treatment procedure at the phase of dose planning, where the measured characteristics of the beam and the calibration factors for beam monitor detectors are used as an input [2, 46, 110].

Table I*. The schematic presentation of the treatment procedure of external electron and photon radiation therapy and BNCT [9, 16, 27, 30, 46, 48, 65, 93].*

Treatment procedure				
Treatment planning			**Treatment delivery**	**Follow-up**
Tumor localization	**Dose planning**	**Plan verification**		
• CT, MRI, X-ray imaging • Patient positioning (fixation)	• Prescription of target volume, dose and fractionation • Dose calculations by TPS (optimization) • Plan of the boron concentration and distribution in the patient. [2)]	• Simulation of the patient set-up • Simulation of the radiation fields in the patient by X-rays [1)]	• Patient set-up • Dose delivery • Verification - recording and verifying systems - *in-vivo* dosimetry - portal imaging [1)] • Determination of boron concentration (blood samples)[2)]	• Medical examinations of the patient • Evaluation of results
[1)] Actions performed only in electron and photon beam radiotherapy. [2)] Actions performed only in BNCT.				

The dosimetry procedure of epithermal neutron beams in BNCT consists of the same elements as the dosimetry chain of electron and photon beams. The differences arise from the complexity of radiation components and from the variety of dosemeters. The physical quantity of the neutron beam related to the therapeutic dose is the thermal neutron fluence. The thermal neutron fluence rate is typically determined by a metal foil activation technique. The measured neutron fluence is traceable to the standards of activity [43, 45]. With a twin ionization chamber technique it is possible to determine both the neutron and gamma doses traceable to the standards of dose quantities (air kerma and absorbed dose).

Metrology in radiotherapy dosimetry deals with the dose measured in the reference conditions in a clinic and is mainly related to its traceability to the international measuring system. Besides the actual determination of the dose to a patient (the dosimetry chain) a lot of dosimetry actions are required in measurements of the input data for TPS, in QA of treatment units and TPSs, and *in-vivo* dose measurements.

3.3 Quality assurance of dosimetry

It can be summarized that the elements affecting the quality of dosimetry in radiation therapy are the means of dosimetry (methods and equipment) as well as the possibilities and practice to carry out the dosimetric actions (quality policy, organization, education). The means of QA for dosimetry are the establishment of performance criteria, quality control (QC) and quality audit measures [1, 2, 30].

Through the use of the methods, data and dosemeters of CoPs it is intended to unify both the accuracy of dose with respect to the absolute dose and the consistency of dosimetry between different radiotherapy clinics and modalilities. The dosimetry recommendations include also guidance for QC. The performance requirements for manufacturing of the medical electrical radiation therapy equipment and the recommendations for QC of this equipment are established by the International Electrotechnical Commission (IEC) [50, 51]. The basic structure of the organization of a radiotherapy clinic and the expertise of the staff of the clinic are outlined in the published comprehensive QA recommendations [1, 30].

Quality audit site visits and mailed dosimetry services are organized by SSDLs and by other national and international organizations [28, 34, 58, 114]. Dosimetric intercomparisons are a widely used method for external verification of the quality of dosimetry. The intercomparisons are performed either at some specific level in the dosimetry chain [10, 71, 102] or in order to evaluate some specific treatment technique [21].

4 MEASUREMENT OF DOSE IN THE REFERENCE CONDITIONS

By an ionization chamber the absorbed dose to a medium, D can be determined according to a relation:

$$D = QNC \tag{4.1}$$

where Q is the measured charge, N is the individual calibration factor of the chamber for the dose quantity and beam quality used in the calibration laboratory, and C is the conversion factor of the calibration from the laboratory irradiation conditions to the reference conditions in a clinic. For Q the corrections for the influence quantities of charge collection and for the ambient climatic conditions are required [51]. In a clinic the reference absorbed dose in external electron and photon beams is measured in defined reference conditions in a water phantom [39]. The reference conditions fix the beam geometry (attenuation and scattering conditions) and the position of the chamber in the beam so that the conversion factor can be specified.

In a mixed neutron and gamma beam the neutron dose can be determined by a tissue equivalent (TE) ionization chamber [47]:

$$\frac{D_n}{C_{n,K}^{TE}} + \frac{D_g}{C_{g,K}^{TE}} = NQ \tag{4.2}$$

In Equation 4.2 D_n and D_g are the neutron and gamma doses. $C_{n,K}^{TE}$ and $C_{g,K}^{TE}$ are the conversion factors relative to air kerma for neutron and gamma radiation qualitites, respectively. To determine both the gamma and neutron doses a twin ionization chamber technique can be applied and a similar equation as for a TE chamber (Equation 4.2.) can also be written also for a non-hydrogenous neutron insensitive gamma ionization chamber [47].

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4.1 Calibration of ionization chambers

In the calibration process the calibration factor N and the conversion factors C are determined (Equations 4.1 and 4.2). The accuracy of

N depends on the characteristics of the chamber and the calibration method used. N is determined by a direct comparison of the ionization chamber to a standard dosemeter. The dose quantity, in terms of which N is determined, is either the air kerma or the absorbed dose to water in a ^{60}Co gamma beam or in a beam with a quality closer to the user beam. In the SSDLs, typically, only ^{60}Co gamma sources are available for high energy calibrations. If other beam qualities are used, the calibrations have to be performed either in a PSDL or in a radiotherapy clinic.

The conversion factors C can be determined experimentally by using a reference dosemeter and specific beam qualities or theoretically by methods based on the cavity theories of dosimetry [8]. The accuracy of C depends on the dose quantity used to determine N (air kerma or absorbed dose), the reference dose quantity used in a clinic, the characteristics of the radiation beams and the characteristics of the chamber. In the dosimetry of electron and photon radiotherapy the reference dose quantity is the absorbed dose to water and the measurements are performed in a water phantom [39, 44]. For neutrons the fluence to dose conversion depends critically on the chemical composition of the material, and the reference dose is defined directly to the target tissue material e.g. “ICRU adult brain” for treatments of brain tumors in BNCT [47, 56, 93, 100]. In spite of the definition of the reference dose to the target tissue material for neutrons, the dose measurements and calculations can be performed in “simple” materials as water or PMMA [91, 104]. For reference dosimetry of photon beams small (< 1 cm^3), air filled, graphite wall thimble type ionization chambers are recommended. For dose measurements in electron beams, depending on the electron beam energy, both thimble chambers and flat, coin shaped PP chambers with graphite windows are suitable [51, 39]. For neutron dosimetry ionization chambers made of TE plastic and flushed with TE gas are recommended [47].

The formalism of the conversion factors C, based on cavity theories for dosimetry, for the ionization chambers typically used in the electron, photon and neutron dosimetry are summarized in Table II.

Table II. *Conversion factors C for typical ionization chambers in electron, photon, and neutron dosimetry. Abbreviations are explained in the text. [7, 39, 40, 47, 81]*

Type of the ionization chamber	**Dose quantity and radiation quality used for individual calibration of the chamber**	**Radiation quality to be measured**	**Conversion factor**
Thimble	Air kerma ^{60}Co gamma	Electrons and photons	$C_K = (1-g)k_{att}\dfrac{[(S/r)^{water}_{air}]_u}{[(S/r)^{wall}_{air}(m/r)^{air}_{wall}]_c}\dfrac{(W_{air})_u}{(W_{air})_c}p_u$
Plane parallel	Air kerma ^{60}Co gamma	Electrons	$C_K^{pp} = (1-g)k_m k_{att}[(S/r)^{water}_{air}]_u\dfrac{(W_{air})_u}{(W_{air})_c}p_u$
Thimble and plane parallel	Absorbed dose to water ^{60}Co gamma or electron/photon	Electrons and photons	$C_D = \dfrac{[(S/r)^{water}_{air}]_u}{[(S/r)^{water}_{air}]_c}\dfrac{(W_{air})_u}{(W_{air})_c}\dfrac{p_u}{p_c}$
TE-thimble	Air kerma ^{60}Co gamma	Neutrons	$C^{TE}_{n,K} = (1-g)k_{att}\dfrac{[r^{wall}_{gas}K^{tissue}_{wall}]_u}{[(S/r)^{wall}_{gas}(m/r)^{air}_{wall}]_c}\dfrac{(W_{gas})_u}{(W_{gas})_c}p_u$

In Table II. $(S/r)^1_2$, $(m/r)^1_2$ and $(K)^1_2$ are the ratios of mass collision stopping powers for electrons, ratios of mass absorption coefficients for photons, and ratios of kermas for neutrons, respectively. All ratios are expressed in medium 1 to medium 2. r^{wall}_{gas} is the gas-to-wall conversion factor for neutrons. W is the energy required to produce an ion pair in the gas of the ionization chamber.

k_m is the factor which takes account of the non-air equivalence of the chamber wall in a ^{60}Co gamma beam. k_{att} is the correction factor for attenuation and scatter of photons in the chamber relative to air in a ^{60}Co gamma beam. p is the product of several factors and takes account of the perturbation of the electron and photon (or neutron for TE chamber) fluences in the chamber compared to those in the phantom medium. g is the

fraction of energy of secondary charged particles that is lost to bremsstrahlung in air in a ^{60}Co gamma beam (value of g is 0.003) [39, 47]. Subscripts c and u refer to the calibration and user beam qualities, respectively.

In this work the calibration procedures of NACP PP chambers [72] are studied
and the individual variations of the values of factors k_m, $k_m k_{att}$ and p_c are determined experimentally for several PP chambers [Publ. IV].

The variation of C_K, C_K^{pp}, and C_D relative to the user beam quality is mainly defined by the variation of $(S/r)_{air}^{water}$ and a minor contribution is induced by p_u. For electron and photon beams the value of the ratio $(W_{air})_u/(W_{air})_c$ can be considered to be unity [7, 40]. In the reference conditions in water for typical accelerator spectra the variation of the values of $(S/r)_{air}^{water}$ is approximately 13% for electron beams and 6% for photon beams [7, 40]. In electron beams the change of electron spectra and $(S/r)_{air}^{water}$ relative to the depth in water have to be considered. For neutrons the values of $C_{n,K}^{TE}$ change approximately with a factor of three relative to neutron energy due to change of critical parameters K_{TE}^{brain} and W_{TE} [56, 100, Publ.V]. In this study the values of $C_{n,K}^{TE}$ were evaluated for the total epithermal neutron spectra (including thermal neutrons) and a change of 13% relative to the depth up to 150 mm was found [Publ.V]. According to the results of *Raaijmakers et al.* the variation of the value of $C_{n,K}^{TE}$ is approximately 5 to 6% for the epithermal and fast components of the spectra they used [92]. For a total neutron spectrum the value of $C_{n,K}^{TE}$ is close to 1.30, whereas for epithermal and fast neutrons it is close to 0.90. In all the studies mentioned for neutrons the changes of the values of $C_{n,K}^{TE}$ are referred to brain tissue and assuming the value of p_u to be equal to one.

4.2 Specification of beam quality

For the evaluation of the conversion factor C, the energy spectrum of the particle fluence of the beam has to be specified. In the dosimetry of electron and photon beams the spectrum of an individual beam is not explicitly determined. In these beams relatively simple, easily measurable beam quality specifiers are used [7, 12, 15, 39, 40]. These beam quality specifiers

are based on the indication of dose or ionization characteristics relative to the depth in water at a specific beam geometry. The relations between a beam specifier and $(S/r)_{air}^{water}$ and p are determined experimentally as well as by MC calculations. In CoPs these relations are given as tables or figures for typical accelerator spectra and common ionization chambers [7, 39].

In the CoPs for electron beams the selections of $(S/r)_{air}^{water}$ and p are based on the use of two measured beam energy specifiers, i.e. the depth of 50% of the maximum dose, R_{50}, and the practical range of electrons, R_p [39, 40, 52, 80]. First, R_{50} is used to determine the mean energy at the surface of the phantom, $\overline{E}_0$ [12]. $\overline{E}_0$, in turn, and R_p are used to determine the mean energy at the reference depth in a phantom, $\overline{E}_Z$. The realtions of $\overline{E}_0$, R_p, and $\overline{E}_Z$ are based on MC simulations of the beams. Finally, $\overline{E}_0$ is used for the selection of $(S/r)_{air}^{water}$ and $\overline{E}_Z$ for the selection of p.

The accuracy of R_{50} as an electron beam specifier depends on the characteristics of electron scattering and on the level of photon contamination in the beam in water [12, 39]. In the recent studies *Sorcini et al.* investigated the effects of the energy and angular spread on the characteristics of the depth dose curve in water produced by a high energy electron beam [108, 109]. They also introduced novel range parameters for evaluation of $\overline{E}_0$ [109]. The effect of photon contamination in electron beams has recently been studied by *Klevenhagen et al.* [62], *Ding et al.* [25], and *Sorcini et al.* [108, 109]. According to these studies the absorbed dose determined by ionization chambers in heavily photon contaminated electron beams ($\overline{E}_0 \geq 17$ MeV) may be too low, even as much as 1.0 to 1.2%. *Ding et al.* and *Sorcini et al.* have also developed methodologies to take account of the contribution of contaminant photons to the measured absorbed dose in electron beams. A simplified procedure where both the reference depth and $(S/r)_{air}^{water}$ are defined by $\overline{E}_0$ is presented by *Burns et al.* [19]. In the approach by *Rogers et.al.*, where the absorbed d ose to water is used as a calibration quantity, the conversion factor C_D is expressed directly by R_{50} [98, 99].

For photon beams a widely adopted beam quality specifier is the tissue-phantom-ratio, TPR_{10}^{20} [7, 15, 39]. TPR_{10}^{20} is defined as a ratio of the absorbed dose to water at the depth of 20 cm to that at the depth of 10 cm on the central beam axis in a water phantom, with a 100 cm source-to-detector-distance and a field size of 10 cm × 10 cm at the detector level. The

precursor of TPR_{10}^{20}, the ionization ratio, J_{10}/J_{20}, was used to specify the nominal accelerating potential, $(S/r)_{air}^{water}$, and p [14, 80]. Other types of photon beam quality specifiers, mainly used to specify the nominal accelerating potential, are the depth at which the dose fell to 80% of the maximum dose, *d(80%)*, and the percentage dose at the 10 cm depth, *%dd(10)* [17, 67]. In this study the validity of the *%dd(10)* as a beam specifier i.e. to determine $(S/r)_{air}^{water}$ is investigated [Publ. III]. *Karlsson and Nyström* have demonstrated the use of a device for measuring the half value thickness of a photon beam in a narrow water column. The half value thickness they used to determine $(S/r)_{air}^{water}$ especially in non-reference conditions [60, 89].

For accurate neutron dosimetry the characterization of the neutron spectrum is essential [47]. For epithermal neutron beams this requirement is emphasized due to the spread of the neutron spectrum at the thermal energy range and due to the energy dependence of dosemeters. For the determination of the neutron spectrum in a phantom a set of metal activation foils and wires with different cross-sections relative to neutron energy has been applied [43, 45, 91]. In this work the neutron spectrum was calculated by using deterministic methods and verified by the activation foil technique [Publ. V].

5 CALCULATION OF DOSE TO A PATIENT

5.1 Treatment planning systems

The core of a TPS is the algorithm responsible for the calculation of the dose distribution in a patient. The algorithm should take into account all beam geometries and beam modification devices as well as the full extent of the patient anatomy [46]. The dose calculation algorithms can be classified into "correction" based approaches (matrix methods, methods of beam generating functions, and methods of separation of the primary and scattered dose components) and into those using beam modelling [22, 46, 70, 83]. The "correction" algorithms reconstruct the beam in a homogenous medium and use separate corrections for beam shaping, patient contour, and heterogeneities. With the "correction" algorithms a separate calculation procedure is also required for the calculation of the output factors (monitor units) of the modified beams [70]. The "correction" based algorithms are typically 2D calculation approaches, although permitting the dose in 3D locations. By the model based algorithms the fluence transport in a medium is directly modelled by either deterministic (radiation transport equations and 3D convolution methods) or stochastic means (MC calculations). The model algorithms are suitable for efficient 3D calculations and capable of calculating the dose in "absolute units" i.e. dose per unit fluence.

In spite of the differences in the principles for dose calculation, the "correction" and model based approaches also have a lot of common features. All calculation algorithms require some measured data of the beam as an input. In the model based approaches the measured input data is mainly needed for "tuning" the beam specific parameters [70]. All dose calculation methods are also approximative: the "correction" based approaches due to their initial nature and the model algorithms due to the limited calculation time in clinical use and/or the inadequate input data.

Although the trend in TPSs for electron and photon beams is towards the 3D convolution model based dose calculation algorithms [37, 70], the "correction" type dose calculation approaches are still common [2, 4, 101]. Due to long calculation times, MC techniques are still mainly used as a complementary method for convolution techniques [62, 102, 122] and as a reference method in the verification of other calculation methods [24]. An extensive list of MC based treatment planning projects is included in the work of *DeMarco et al.* [24]. In this study the accuracies of the six TPSs,

using “correction” type dose calculation approaches for electron and photon beams, were compared [Publ. II].

For the treatment planning in BNCT the fluence and dose calculations are mainly performed by MC methods [83, 125]. Deterministic radiation transport calculation methods, common in reactor physics, are also used for dose calculations in epithermal neutron beams [83]. *Raaijmakers et al.* have modified algorithms designed for external electron and photon radiotherapy and used them for dose calculations of the mixed epithermal neutron and gamma beam in BNCT [94].

The implementation of a modern conformal treatment technique requires a TPS with totally 3D dose calculation algorithms. An efficient use of a 3D algorithm also requires accurate 3D patient data and a visualization for 3D images [82, 110]. These characteristics are standard features in the newest TPSs as well as software tools for image fusion, automatic contouring, beam's eye view display, digitally reconstructed radiographs, and dose volume histograms [2, 101]. All these features have to be included also in the QA procedure of a TPS [2].

According to ICRU the calculated relative doses should not differ by more than 2% (low dose gradient area) or 2 mm (high dose gradient area) from dose measurements [46]. A detailed summary on the performance requirements for external electron and photon beams has been published by *Van Dyk et al.* [120]. The tolerance values by *Van Dyk et al.* can be summarized as 3–4% (/ 4 mm) for photons and 5–7% (/ 5 mm) for electrons. The percentage tolerance values are expressed relative to the maximum dose higher values being typically adopted at high dose regions. Tolerances in the distance (mm) are for high dose gradient regions. Tolerance values are valid for tests of a complicated geometry using radiation fields of a modified shape and intensity together with inhomogeneities. *Van Dyk et al.* considered their tolerance values as practical achievable criteria and they also pointed out that the criteria are not limits beyond which no further improvements are necessary. For the accuracy of the calculation of the beam output (absolute dose) they recommended a tolerance of 1%.

5.2 Verification of calculated doses

In commissioning and in QA of a TPS, to verify the accuracy of the calculated doses, calculated and measured doses are compared. Typically the comparisons are performed in separate tests for different shapes of beam and phantom surface and inhomogeneities [105, 120]. The calculated doses are also verified by the *in-vivo* measurements of a patient and the measurements of the delivered dose in a phantom [2, 106, 120]

The accuracy of the calculated dose depends on the accuracies of measured data, data entry and output, calculation algorithm, and the physical data [46, 88, 120]. For the measurement of the comparative relative doses (non-reference geometries) a variety of dosemeters can be used. The selection of the most suitable dosemeter should be based on the optimization of the energy, angular, and dose rate dependences of the dosemeter. For electron and photon beams typical on-line dosemeters are small ionization chambers and diodes. Passive dosemeters, such as TLD and films, can be used, especially in solid phantoms [61, 71]. For dosimetry in phantoms, the novel application of fricke dosimetry is the use of fricke gels read out by nuclear magnetic resonance [90]. In this study lithium tetraborate ($Li_2B_4O_7$) TLD, ionization chambers, silicon diodes and diamond detectors are used for the measurements of relative dose distributions in electron and photon beams [Publ. I , Publ. II].

For the verification of calculated doses accessory softwares have been developed, although not widely implemented in the commercial TPSs [2, 24, 35, 69]. AAPM Task Group 23 provides a test package for the quantitative analysis of treatment planning systems for photon beam radiation therapy. The test package includes measured input data and comparative data for the verification [4]. The Collaborative Working Group of the National Cancer Institute in the USA also provides measured verification data for electron algorithms [105].

For *in-vivo* dosimetry in electron and photon beams semiconductor diodes and TLD are used and recommended [75, 119]. Methods to use electronic portal image devices for *in-vivo* dosimetry of the patient and for QA of accelerators have been developed. [11, 78, 119].

For the experimental verification of calculated doses of epithermal neutron beams the dosemeters used in fast neutron therapy can be used if special attention is paid to their response for thermal neutrons [45, 73]. In BNCT

dosimetry for the detection of thermal neutron fluence in phantoms, activation foils [91, 112], boron doped proportional counters [5, 112], and diodes [29, 36] have been used. For neutron and gamma dosimetry, TLD [115, 124], fricke gels [33] and microdosimetric techniques [66, 123] have successfully been applied. Especially for *in-vivo* dosimetry in BNCT prompt gamma detection from the neutron capture of boron has been developed [121].

6 DISCUSSION

In the estimation of the overall uncertainty of a process the uncorrelated sub-uncertainties are added in quadrature and the importance of the largest uncertainties is pronounced [54]. In the dosimetry chain for electron and photon beams the “weakest links” are the determination of the calibration factor of an ionization chamber for the user beam quality and the calculation of the dose to a patient [16]. For dosimetry of epithermal neutron beams in BNCT the same steps as for electron and photon beam dosimetry can be considered as most meaningful. This work is concentrated on studing these steps of the dosimetry procedures.

6.1 Calibration of ionization chambers

At the time of closing this study the general attempt to calibrate radiotherapy dosemeters in terms of absorbed dose to water instead of air kerma is going to be fullfilled. For electron and photon beams, AAPM, IAEA and ICRU are preparing dosimetry recommendations based on absorbed dose [42, 49, 99]. Due to the elimination of conversions of quantities the advantages of the absorbed dose to water calibrations are obvious for photon dosimetry [7, 96, 97], whereas for electrons these advantages are partly lost [98]. Primary standard equipment for absorbed dose are mainly developed for photon radiation [12] and the radiation quality dependent systematic components of the uncertainties in the dose conversion procedure for electrons are not cancelled out to the same extent as in case of photons [98]. However, regardless of the quantity used for calibration (air kerma or absorbed dose to water), $(S/r)_{air}^{water}$ and p of the user beam are directly related to the measured absorbed dose i.e the dose to a patient.

For electron dosimetry by PP chambers, the chamber to chamber variations of the fluence perturbation effects (k_m, $k_m k_{att}$ and p_c) in a ^{60}Co gamma beam increase the uncertainty of the measured dose. In CoPs these effects are assumed to depend only on the type of the chamber and the beam quality. If a PP chamber is calibrated in a ^{60}Co gamma beam (for air kerma or absorbed dose to water) the uncertainty of the absorbed dose measured in the reference conditions in an electron beam by this chamber is 3.4 to 3.5% ($1sd$) [40]. If a PP chamber is calibrated for absorbed dose to water in a high energy electron beam, the uncertainty of the measured dose is 2.9% ($1sd$), even if a thimble type chamber with an air kerma calibration is used as a secondary standard [40]. The described improvement of the accuracy is

mainly due to the elimination of the individual variations of the fluence perturbations of PP chambers in a ^{60}Co gamma beam. In this study 3% chamber to chamber variations of k_m, $k_m k_{att}$ and p_c factors are found*. Publ. IV is adopted as a reference study in the CoPs for electron dosimetry by IAEA and IPEMB [40, 52].

Since Publ. IV, *Nilsson et al.* have studied the effects of chamber and phantom materials on the perturbation factors p_u of PP chambers in electron beams [84].
The results of *Nilsson et al.* verify that for NACP chambers p_u in electron beams is close to unity in water, the deviation being less than 0.5%. However,
as related to p_u, even with well guarded chambers the contribution of lateral scatter might not be negligible [84]. *Ding and Cygler* have determined p_u relative to R_{50} in electron beams for other types of PP chambers using NACP PP chambers as a reference [26].

Burns et al. have demonstrated deflections from theoretical predictions in the ion collection efficiency (recombination effect) of NACP PP chambers in electron beams of high dose per pulse [20]. Using the commonly applied two voltage method for the determination of the recombination correction, *Burns et al.* obtained up to 1% over estimation for recombination correction. For the accurate calibration of NACP PP chambers in electron beams, the total saturation behaviour of the collection efficiency has to be determined. In the dose measurement the effect of the overestimated recombination is cancelled out if the same “dose rate per pulse” and collecting potentials are used both in the calibration and in the actual dose measurement [20, 40]. The possible overestimated recombination effects can partly explain the large variations of the perturbation effects of PP chambers obtained in this study. This item is under research at the moment. *Nisbet et al.* have studied the polarity effects of the collection potential in NACP chambers and emphasize the requirement of polarity correction i.e. the use of both polarities of the collection potential in accurate measurements [85]. The methods of accurate recombination and polarity corrections have been implemented in the recent dosimetry CoPs of IAEA and IPEMB [40, 52].

* The manufacturing of the NACP PP chambers by Dosetek company has been finished in 1997 (Varian-Dosetek Oy, Metsäneidonkuja 8 FIN-02130 Espoo, Finland). At the moment, NACP type chambers are manufactured only by Scanditronix Medical AB, Stålgatan 14, S-754 50 Uppsala, Sweden.

For the calibration of PP chambers in electron beams, the arrangement of laboratory conditions in a clinic can be difficult due to practical considerations. In Finland, the calibrations of PP chambers are performed by the staff of the SSDL-Helsinki, during their regular site visits to clinics. By this arrangement the same reference thimble chamber can be used as a standard and the consistency of calibrations can be increased.

In dosimetry of epithermal neutron beams, the main drawback of using a TE ionization chamber for neutron dose determinations in a phantom is the subtraction of the signal due to large gamma dose [Publ. V]. In this study, in a highly thermalised neutron beam the uncertainty of 6% (*1sd*) of the gamma dose is manifested as an uncertainty of 18% (*1sd*) in the neutron dose measured by a TE chamber. Another significant source of uncertainty is the uncertainty of the conversion factor $C_{n,K}^{TE}$. For an A150-plastic TE ionization chamber $C_{n,K}^{TE}$ to brain tissue can be determined with an uncertainty of 7 to 10% for the total epithermal neutron spectrum [Publ. V]. The energy response of a chamber and the uncertainty of $C_{n,K}^{TE}$ can be reduced if TE chambers made of brain tissue equivalent materials instead of muscle tissue equivalent A150-plastic can be used. So far, only liquid brain substitute materials are available [104]. The uncertainty of $C_{n,K}^{TE}$ can also be expected to be reduced through the calibrations of chambers in neutron beams. Unfortunately, the availability of these types of calibrations in neutron beams at PSDLs is limited. The total uncertainty of the determination of the neutron dose to brain tissue by an A150-plastic TE chamber in a phantom is 15 to 20% [Publ.V, 93, 100].

6.2 Specification of photon beam quality

For typical clinical photon beams $(S/r)_{air}^{water}$ can be expressed within a variation of 0.7% relative to TPR_{10}^{20} and with an uncertainty of 0.7% (*1sd*) [6, 7, 39, 86]. For TPR_{10}^{20} as a general photon beam specifier the following limitations can be stated [Publ. III, 68]: a) The relation between $(S/r)_{air}^{water}$ and TPR_{10}^{20} depends on the filtration of the beam; for the light filtered beams as used in some PSDLs as well as for the high energy racetrack microtron beams differences up to 1.0% in $(S/r)_{air}^{water}$ compared to the typical clinical beams are found for the same value of TPR_{10}^{20}. b) The relation between $(S/r)_{air}^{water}$ and TPR_{10}^{20} is not linear and the accuracy of

TPR_{10}^{20} is decreased at higher energies; 1% change in TPR_{10}^{20} leads to 0.4% change in $(S/r)_{air}^{water}$ at the TPR_{10}^{20} value of 0.8 and to 0.1% change at the TPR_{10}^{20} value of 0.7. c) As a ratio, TPR_{10}^{20} is not a very illustrative parameter for beam specification.

In this study the characteristics of TPR_{10}^{20} and an alternative photon beam specifier, *%dd(10)*, are investigated by simulating photon beams by an EGS4 MC code [Publ. III]. A linear relationship within 0.2% between $(S/r)_{air}^{water}$ and *%dd(10)* was found both for clinical and for lightly filtered photon beams. The main disadvantage of *%dd(10)* as a photon beam specifier is the effect of the electron contamination of the beam at the depth of the dose maximum. The contribution of electrons to the dose at the depth dose maximum in the phantom is between 2 to 5% and increases with the increasing beam energy [Publ. III, 68, 107]. Without corrections in clinical photon beams the effect of contaminant electrons introduces a maximum deviation of 0.7% to $(S/r)_{air}^{water}$ when determined by *%dd(10)*. As a continuation to Publ. III of this study *Li and Rogers* have studied methods to eliminate the contaminant electrons by metal filters [68]. They define a cubic relation within 0.2% between $(S/r)_{air}^{water}$ for a realistic, contaminated photon beam and *%dd(10)* when the contaminant electrons are excluded from the beam by a 1 mm thick lead filter.

In the measurement of *%dd(10)*, attention has to be paid to the type of the ionization chamber used. Due to the change of perturbation factors, p_u, of thimble and cylindrical chambers relative to the depths of the dose maximum and 10 cm in water [Publ. III], PP chambers are preferred in the measurements of *%dd(10)*. Mainly for practical considerations, especially for the use of a lead filter, the outcome of the use of *%dd(10)* as a photon beam specifier for dosimetry will be verified through the experiences in practical dosimetry work. Comparative dose measurements using different beam quality specifiers will give more information of practical aspects. This kind of a study is planned for the future. Publ. III of this study and the study of *Li and Rogers* [68] are used as a basis for adopting *%dd(10)* as a photon beam specifier in the forthcoming dosimetry recommendations of AAPM [99].

The exclusion of the contaminant electrons from a photon beam is desirable not only for dosimetry but also for reducing the skin dose of a patient. It can be expected that the use of the *%dd(10)* specifier in dosimetry will

encourage the manufacturers to develop medical accelerators with less contaminated photon beams.

6.3 Verification of calculated doses

For electron dose calculations only a few intercomparisons of TPSs have been published since Publ. II. Dose calculation algorithms for electrons have been tested by several groups, most recently by *Samuelsson et al.* for a 3D pencil beam algorithm [102]. They used basically the tests described by the Collaborative Working Group of the National Cancer Institute [105]. The general consistency between measured and calculated relative dose distributions found by *Samuelsson et al.* was within the 5 to 7% (/ 5 mm) limits stated by *Van Dyk et al.* [120]. The largest deviations detected by *Samuelsson et al.* were near the edges of the radiation fields (7%) and in and under the bone and lung inhomogeneities (2 to 4mm). The authors considered the 4 mm deviations related to the doses of the lung inhomogeneity too large to be fully accepted [102]. The calculated results for the beam output were within 1% of the measured values. Although the tests used by *Samuelsson et al.* are not totally comparable to the tests used in this study [Publ. II], the results of both studies illustrate the same behaviour of the limitations of the dose calculation algorithms. In both studies the largest inaccuracies in dose calculations were detected in the lung inhomogeneities where calculations overestimate the dose. In this study the maximum differences of measured and calculated doses were 10%.

Since Publ. II, intercomparisons for photon dose calculations by TPSs have been performed at least by *Van Bree et al.*, *Masterson et al.*, *Aird et al.*, *Cheng et al.* and by *Alam et al.* [3, 4, 21, 71, 92, 118]. *Van Bree et al.* and *Cheng et al.* tested the accuracy of the calculated tangential beams in breast treatment [21, 92, 118]. *Van Bree et al.* circulated a breast phantom in six clinics having different TPSs of 2D calculation algorithms. In the study of *Cheng et al.* TPSs also with 3D calculation features were included in the intercomparison and no measurements were performed. *Aird et al.* investigated the consistency of the measured and calculated doses of nine TPSs using a “head and neck” phantom and a “bronchus” phantom. *Mastersson et al.* tested six non-commercial 3D algorithms and *Alam et al.* compared the calculations of two TPSs. In all these tests the general consistency of the measured and calculated relative doses were 3 to 4% (/3mm). Similarly, as for electrons, the largest discrepancies were related to the dose calculations near or in the lung inhomogeneities. In a few studies also large individual (clinic related) discrepancies were found. *Van Bree et*

al. detected up to 10% and *Aird et al.* up to 7% differences between the calculated and measured doses. These were mainly explained by the inaccuracies in the input data [3, 92, 118]. *Alam et al.* detected up to 19% inaccuracies of the calculated doses for a pencil beam based algorithm. They thought the discrepancy to be related to the inaccuracies in the user dependent fitting parameters of the algorithm [4]. The calculated beam output was typically within the limits of the estimated uncertainty of the measured dose, i.e.1% (*1sd*), expressed relative to the dose in the reference conditions in a clinic.

The accuracy of the calculated photon doses was also verified in the large intercomparison of 62 radiotherapy clinics by *Thwaites et al.* and the similar intercomparison of 4 clinics by *Nisbet et al.* [87, 113]. In these studies the consistencies of the measured and calculated doses were compared in five points in a simple phantom with a lung substitute insert. As in the TPS tests the general consistency of the calculated and measured doses (including the beam output in these comparisons) was better than 3 to 4%. *Thwaites et al.* found also large individual variations of –8% to +16% for a few TPSs. The largest deviations were due the exclusion of the lung inhomogeneity correction and the software error of a TPS.

For photon beams, in addition to the intercomparisons mentioned, a large number of tests are performed for individual algorithms and treatment techniques. Intensive studies for a model based 3D pencil beam calculation algorithm have been performed by *Knöös et al.* and *Hurkmans et al.* [38, 63, 64]. The limitations of the algorithm were clearly illustrated as 3 to 14% deviations of relative doses with an increasing trend relative to the increasing beam energy being detected in and close to lung inhomogeneities [64]. In the lung inhomogeneity study MC calculated dose distributions were used as a reference. In phantom studies for small fields and curved surfaces (simulated head and neck area) the algorithm overestimates the dose from 2 to 5% [38]. The absolute output was calculated with an accuracy better than 3% [63].

In the studies reviewed above, ionization chambers, diodes, TLD, and film dosimetry were used for the measurement of doses. For the same type of $Li_2B_4O_7$ pellets as used in this study [Publ. I] the energy dependence has been studied by MC simulations by *Mobit et al.*. [76]. According to their results the energy response (relative to dose in water) of $Li_2B_4O_7$ pellets is constant within 1% in the energy range of 2 to 20 MeV of monoenergetic electrons. For photons this variation was 3% in the energy range of 1 to 10

MeV [76]. The findings of *Mobit et al.* support the results of this study [Publ. I], that considering the energy response, the $Li_2B_4O_7$ pellets are well suitable for the measurements of relative dose distributions in a tissue equivalent phantom in electron and photon beams. The tests used for intercomparison of TPSs in this study [Publ. II] are similar to those used more recently [3, 4, 21, 71, 92, 105, 118, 120]. For the validation of algorithms and TPSs these types of tests can be adopted from the existing test packages [4, 105]. For quality audit purposes unspecific but sensitive comprehensive tests with phantoms including curved surfaces and inhomogeneities could be more practical [87, 106, 113].

Based on the results of this study [Publ. II] and the results of the reviewed recent studies for TPSs for electron and photon beams the following conclusions can be made: a) The accuracy level stated by *Van Dyk et al.* [120] i.e. 3 to 4% (/ 4 mm) for photons and 5 to 7% (/ 5 mm) for electrons is generally achieved by TPSs. b) The accuracy level stated by ICRU (2% / 2 mm) [46] can be achieved in simple geometries and occasionally in geometries of curved surfaces and with inhomogeneities. The criteria of ICRU can be considered objective for the accuracy of a TPS. c) Large inaccuracies (> 10%) are possible even with a modern TPS.

For epithermal neutron beams used in BNCT intercomparisons of TPSs have not been performed, mainly because of a small number of systems available. For the validation of an MC based TPS *Zamenhof et al.* performed a test in a head phantom [125]. They detected an approximately 10 to 15% consistency between the measured and calculated relative neutron dose rates and a 5 to 6% consistency for the gamma dose rates [125]. Using a semi-empirical calculation algorithms, *Raaijmakers et al.* found a 2% consistency between the measured and calculated relative thermal neutron fluences and a 5% consistency for gamma ray doses in a solid head phantom [94]. Also up to 30% deviations for the calculated thermal neutron fluences were found close to the air boundaries [94]. In this study the consistency of the measured and calculated doses was 5 to 6% for gamma radiation and 18% for neutron radiation [Publ. V]. In all the studies mentioned for epithermal neutron beams the uncertainties of doses (*1sd*) are at the same level as the consistency of measurements and calculations.

In the verification of the performance of a modern TPS, either in electron and photon radiotherapy or in BNCT the statement of J.R. Cunningham [22] is still in a great validity: *"All calculation procedures must be tested by comparison with carefully carried out experiments. The importance of*

experimental testing, preferably by the creator of the calculation method, cannot be overestimated. Although several calculation procedures are available, being approximate, they must be expected to have limitations and it must be realized that there will be irradiation conditions that will fool them".

7 SUMMARY

In electron beam dosimetry, the accuracy of dose determinations can be improved remarkably through the direct calibrations of plane parallel ionization chambers in high energy electron beams. These calibrations can be performed fluently by an SSDL during periodic site visits to a clinic. In photon beam dosimetry, the accuracy of dose determinations can be improved by using the percentage dose at the 10 cm depth, *%dd(10),* as a beam quality specifier. To quantify the final improvement of the accuracy in photon dosimetry, the experiences of the users of *%dd(10)* and the results of the comparative dose measurements using different beam specifiers are to be waited. For electron and photon beam reference dosimetry, the results of this study have been adopted in international Codes of Practices for dosimetry.

In dosimetry of epithermal neutron beams a reasonable accuracy in the determination of gamma and neutron doses by ionization chambers was achieved. To raise the accuracy close to the level achieved in electron and photon beam dosimetry, improvements in tissue equivalent chamber materials and calibration methods in neutron beams should be seeked and used. For dosimetry of epithermal neutron beams in boron neutron capture therapy (BNCT) the work towards coherent dosimetric methods has been started under the European program. The results of this study are available for this program and for the forthcoming Code of Practice in BNCT dosimetry in Europe.

In dose calculations by treatment planning systems for electron and photon beams, the development of dose calculation algorithms has not diminished the value of experimental dose verification. Discrepancies of calculated doses can be expected also for modern treatment planning systems and comparative studies shall be performed. In BNCT the value of intercomparisons is emphasized because of the unique, individual beam facilities and a variety of dosimetry methods used. Furthermore, the relatively large uncertainties of the measured doses in epithermal neutron beams in BNCT broaden the limits of consistency between individual clinics. For the quality assurance of a treatment planning system in BNCT the similar validation procedures as used for electron and photon beams should be performed.

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