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APPLICABILITY OF THERMOLUMINESCENT DOSIMETERS IN X-RAY ORGAN DOSE DETERMINATION AND IN THE DOSIMETRY OF SYSTEMIC AND BORON NEUTRON CAPTURE RADIOTHERAPY

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ABSTRACT

The main detectors used for clinical dosimetry are ionisation chambers and semiconductors. Thermoluminescent (TL) dosimeters are also of interest because of their following advantages: i) wide useful dose range, ii) small physical size, iii) no need for high voltage or cables, i.e. stand alone character, and iv) tissue equivalence (LiF) for most radiation types. TL detectors can particularly be used for the absorbed dose measurements performed with the aim to investigate cases where dose prediction is difficult and not as part of a routine verification procedure. In this thesis, the applicability of TL detectors was studied in different clinical applications. Particularly, the major phenomena (e.g. energy dependence, sensitivity to high LET radiation, reproducibility) affecting on the precision and accuracy of TL detectors in the dose estimations were considered in this work.

In organ dose determinations of diagnostic X-ray examinations, the TL detectors were found to be accurate within 5% (1 S.D.). For *in vivo* studies using internal irradiation source, i.e. for systemic radiation therapy, a method for determining the absorbed doses to organs was introduced. The TL method developed was found to be able to estimate the absorbed doses to those critical organs near the body surface within 50%. In the mixed neutron-gamma field of boron neutron capture therapy (BNCT), TL detectors were used for gamma dose and neutron fluence measurements. They were found able to measure the neutron dose component with the accuracy of 16%, and therefore to be a useful addition to the activation foils in BNCT neutron dosimetry. The absorbed gamma doses can be measured with TL detectors within 20% in the mixed neutron-gamma field, which enables *in vivo* measurements at BNCT beams with approximately the same accuracy.

In this study, the uncertainties of TL dosimeters were found to be high but not essentially greater than those in other measurement techniques used for clinical dosimetry. Therefore, it is concluded, that the TL dosimeters are capable of determining the absorbed doses to tissue in different clinical exposure conditions.

CONTENTS

LIST OF PAPERS

This thesis is based on the following papers:

I M. Toivonen, C. Aschan, S. Rannikko, K. Karila and S. Savolainen, Organ dose determinations of X ray examinations using TL detectors for verification of computed doses. Radiat. Prot. Dosim. 66, 289-294 (1996).

II C. Aschan, M. Toivonen, J.S. Lampinen, M. Tenhunen, K. Kairemo, T. Korppi-Tommola, A. Jekunen, P. Sipilä and S. Savolainen, The use of TL detectors in dosimetry of systemic radiation therapy. Acta Oncologica, 38, 189-196 (1999).

III C. Aschan, M. Toivonen, S. Savolainen, T. Seppälä and I. Auterinen, Epithermal neutron beam dosimetry with TL dosemeters for boron neutron capture therapy. Radiat. Prot. Dosim. 81, 47-56 (1999).

IV C. Aschan, M. Toivonen, S. Savolainen and F. Stecher-Rasmussen, Experimental correction for thermal neutron sensitivity of gamma-ray TL dosemeters irradiated at BNCT beams. Radiat. Prot. Dosim. 82, 65-69 (1999).

V C. Aschan, J.S. Lampinen, S. Savolainen and M. Toivonen, Monte Carlo simulation of the influence of adjacent TL dosemeters on TL readings in simultaneous measurements in BNCT beams. Radiat. Prot. Dosim. 85, 349-352 (1999).

VI M. Toivonen, V. Chernov, H. Jungner, C. Aschan and A. Toivonen, The abilities of LiF thermoluminescence detectors for dosimetry at boron neutron capture therapy beams. Radiat. Meas. 29, 373-377 (1998).

The papers will be referred to in the text by their Roman numerals.

1 AIM OF THE PRESENT STUDY

TL dosimeters are widely used in clinical applications. The aim of this thesis is to determine the abilities of TL dosimeters in terms of precision and accuracy in measuring the absorbed doses in macroscopic level to patients, and to phantoms. For this work, the dose estimations have been done in the cases of

1) diagnostic X-ray studies (paper I)
2) gamma ray irradiations (paper II)
3) mixed neutron-gamma field irradiations (paper III)

The special technology required by the mixed neutron-gamma field is concerned in papers IV, V and VI.

2 INTRODUCTION

Absorbed dose is defined as an amount of energy absorbed per unit mass of an irradiated material [1]. It is a quantity of fundamental interest, and may be specified in any medium for any type of ionising radiation. Clinical dosimetry is performed in order to determine absorbed doses to patients who are exposed to ionising radiation, either for therapy or diagnosis.

In therapy, the absorbed dose distribution in a phantom must be known to ensure that the prescribed absorbed dose is delivered to the target volume in the patient. Furthermore, while the measurement of absorbed dose distribution in a phantom is essential in treatment planning, the ultimate check on the absorbed dose delivered to the patient can only be made by *in vivo* absorbed dose measurements. Therefore, the applications of clinical dosimetry in radiation therapy may be divided into groups according to the used method [2]: i) *in vivo* measurements in the region of interest or at some convenient region of the body, ii) *in vitro* measurements, and iii) measurements in anatomical phantoms. The most useful phantom for absorbed dose measurements both in radiation therapy and diagnostic radiology is one which is designed, as far as it is practicable, to simulate the structure of the human body.

Traditionally, dosimetry in diagnostic radiology [2] has been mainly restricted to personnel dosimetry and local investigations, but also it has a broader application for instance by checking the quality of the used irradiation beams. According to the national and international guidelines, e.g. EU directives, the absorbed doses to the patients, exposed to radiation from diagnostic radiology, have to be determined [3]. Therefore, *in vivo* measurements on patients undergoing radiological examinations or nuclear medicine studies are performed. Also, as in therapy, anatomical phantoms are useful for absorbed dose measurements in diagnostics.

In clinical practice, the main detectors used for dosimetry are ionisation chambers and semiconductors. Ionisation chambers are commonly used for phantom measurements because of their accuracy and practicality [4,5], but for *in vivo* measurements they are not frequently used because of the high voltage applied and the cables attached to the chamber. Semiconductor diodes are routinely used for absorbed dose measurements of clinical studies. These measurements make use of the advantages of semiconductors, such as ease of handling and the dose determination in real time [6,7,8,9]. Thermoluminescent (TL) dosimeters are widely used for radiation detection in the fields of environmental, industrial and personnel applications, just to mention a few. The theory of TL dosimetry, and the abilities of different TL materials for use in several applications, have been summarised in a variety of books [10,11,12,13,14,15,16]. The main advantages of TL dosimeters are [17]: i) wide useful dose range, ii) small physical size, iii) reusability and therefore, iv) economy, v) no need for high voltage or cables,

i.e. stand alone character, and vi) tissue equivalence (LiF) for most radiation types. These make TL detectors a useful tool for clinical dosimetry; since its first use for *in vivo* dosimetry during radiotherapy [18], the use of TL detectors has become an important technique for clinical dosimetry [12,13,14,19,20,21].

Two important characteristics of the detectors used for clinical dosimetry are precision and accuracy, defined as follows [13]:

> *Precision* is a term related to the random uncertainties associated with the measurement, i.e. the uncertainties that have been derived by statistical methods from a number of repeated readings. In order to define the precision of a set of measurements, the standard deviation may be used.
>
> *Accuracy* is a statement of the closeness with which a measurement is expected to approach the true value. Accuracy includes the effect of both systematic and random uncertainties. The value of a quantity is understood to be considered as ”true” either by theoretical considerations or by comparison with a fundamental measurement. The indicated value is the value of a quantity as indicated by the relevant measuring device, sometimes also called ”reading” or measured value.

The precision and accuracy of TL dosimeters in the dose estimations is composed of several parameters [13,22]. The sources of errors that affect the precision and accuracy in determining the absorbed dose have been studied and reviewed by Robertson [23], and more recently in the field of clinical applications by Toivonen [24,25]. According to Toivonen, the accuracy requirements [26,27] of the conventional radiotherapy, performed with external radiotherapy beams with photons and electrons, can be achieved using TL detectors.

TL detectors can be, and commonly are, used for the absorbed dose measurements performed with the aim to investigate cases where dose prediction is difficult and not as part of a routine verification procedure. Among these cases are, for example, new radiotherapies which have been developed for patient treatment during the past decades. Absorbed dose determination in these radiotherapies, e.g. radioimmunotherapy and boron neutron capture therapy, is more complicated (see e.g. refs. [28,29,30,31]) compared with the external radiotherapy. High uncertainties may be present in the dose determination due to patient anatomy, i.e. geometry, inaccurate irradiation source definition or the radiation quality, among other things.

In this thesis, the abilities of TL dosimeters in terms of precision and accuracy are studied in different clinical exposure conditions: i) phantom studies with external X-rays, ii) phantom and *in vivo* patient studies using internal irradiation source emitting

gamma radiation, and iii) phantom studies with an external epithermal neutron beam containing a gamma ray component.

3 ABSORBED DOSE MEASUREMENTS

3.1 Diagnostic X-rays

In X-ray diagnostics, TL dosimeters are commonly used for determining absorbed doses to patients [32,33,34,35], as well as in phantom studies [36,37]. For this study (paper I), TLD-100 (LiF:Mg,Ti) TL dosimeters from Harshaw[1] were used to measure the absorbed doses. The measurements were performed in a Alderson-Rando phantom [38], which is a commercially available phantom[2] with anatomical unhomogenities in respect to the lungs and skeleton. Four different X-ray examinations were performed: abdomen AP, chest PA, skull LAT and modified lumbar spine LAT (paper I). In the irradiations, the TL dosimeters were fixed into the holes of the Alderson-Rando phantom (see Fig. 2a in paper I) so that the flat surface of the detector always faced the beam. The absorbed doses to various organs of the phantom were derived from the doses of the reference points, i.e. those points measured directly with TL detectors or interpolated from them.

3.2 Internal gamma ray source

In systemic radiation therapy (SRT), e.g. in radioimmunotherapy (RIT) with tumour-associated monoclonal antibodies [39], internally administered radionuclides are used for the treatment of cancer. For therapy, it is essential to know the absorbed doses for besides the tumour also for normal tissues, especially to those critical organs limiting the treatment. The dose estimations for RIT are frequently performed using calculation methods: MIRD formalism [40,41,42,43] or the point dose kernel technique [44,45]. However, there is also a need to measure the absorbed doses in SRT [46,47].

Dose estimation methods in RIT have recently been summarised by Fisher [28], Buchsbaum and Wessels [48], and Strand *et al.* [29]. New methods to measure the absorbed doses with implanted TL dosimeters have been developed. However, so far the mini-TL dosimeters have been implanted only into animals and tumour model systems receiving RIT [49,50,51,52,53]; no results with patients have been reported. There are certain limitations in direct organ dose measurements with the implantation of dosimeters in the tumour or in the critical organs. These are mainly the decay of detector

[1] Bicron, Ohio, USA.
[2] The Phantom Laboratory, New York, NY, USA.

sensitivity with time [46,47,54,55] and the requirement for surgical procedures. Therefore, for routine clinical use the dosimeter must be placed on the skin of the patient or in accessible cavities when determining absorbed doses to a tumour or to critical organs. For example, TL detectors have been inserted in the peritoneal cavity for the measurement of the tumour dose arising from beta particles emitted by ^{131}I [56].

During 1996, six patients [57] underwent eight RIT trials with ^{131}I-labelled monoclonal antibody in Helsinki University Central Hospital. The aim of the dose estimations was to establish the magnitude of the absorbed doses to radiosensitive organs. As described in detail in paper II, an independent method was developed in order to determine the absorbed doses to the critical organs during SRT. The method is based on the TL dosimeters placed on the skin of the patient. Therefore, four reference points were marked on the skin of the patients for dose estimations. The first two (posterior) points represented the right and left kidneys, and two points were selected on the upper abdomen. Measurements with TL dosimeters were made by taping the dosimeters to each reference point for a short period of time. Also, a phantom study was performed to establish the method (paper II); the absorbed doses were measured with TL detectors on the surface of the elliptical water phantom. The absorbed doses to the points of interest, situated at various distances from the surface of the phantom, were derived using the developed method. For comparison, the absorbed doses were also measured with TL detectors situated inside the phantom at each point of interest.

3.3 Mixed neutron-gamma field

Boron neutron capture therapy (BNCT) [58,59,60] utilises epithermal neutrons for the treatment of malignant tumours. In BNCT ^{10}B is introduced into the tumour cells, and the selective therapeutic dose is delivered by the neutron capture reaction $^{10}B(n,\alpha)^{7}Li$ when exposed to a neutron fluence. In order to evaluate the quality of an epithermal beam for BNCT the desired epithermal neutron fluence and the undesired fast and thermal neutron fluences as well as gamma fluence have to be determined. The gamma dose to tissue when irradiated with an epithermal beam is composed by gamma-rays present in the beam and from the capture reaction $^{1}H(n,\gamma)^{2}H$.

The main dosimeters for characterising and controlling BNCT beams are activation foils and paired ionisation chambers [30]. TL dosimeters are also used because of the following advantages: i) with TL dosimeters it is possible to measure depth dose curves and profiles at the same time, with one irradiation, and ii) because of their small size TL dosimeters may be possible detectors for *in vivo* use. In previous studies, TLD-700 (^{7}LiF:Mg,Ti) TL dosimeters have been used in the gamma dosimetry of BNCT [30,61,62,63]. However, difficulties have been encountered arising from a

small ^{6}Li content in the enriched ^{7}Li. Therefore, Raaijmakers *et al*.[30,31] have applied a method in which the TL detectors are shielded from thermal neutrons using a ^{6}Li containing cap in the epithermal neutron beam of BNCT. Also, a theoretical method for determining correction factors for thermal neutron sensitivity of TL detectors has been developed [64].

As described in detail in paper III, two new TL dosimeter types[3], MTS-Ns and MCP-7s, were selected for use for neutron and gamma detection, respectively, in the mixed neutron-gamma field of BNCT. For comparison the conventional TLD-100 and TLD-700 detectors of Harshaw were also used. The measurements with TL dosimeters were performed in the cylindrical PMMA (polymethylmetacrylate, $(C_5H_8O_2)_n$) and water phantoms at the epithermal neutron beam of the TRIGA Mark II research reactor FiR 1 [65,66]. Reactor powers of 300 W for a PMMA phantom and 450 W for a water phantom were used. The measurement time was half an hour for both phantom types. TL detectors used for measurements in the water phantom were inserted in holes in polypropylene discs. These holes were water isolated with paper and self-adhesive tape of polypropylene on both sides of the discs. Since no thermal neutron shields were used in the measurements, the correction for the thermal neutron sensitivity of the gamma detectors were based on the theoretical method [64].

4 UNCERTAINTIES PRESENT IN THE MEASUREMENTS

The uncertainties of the absorbed dose measurements, performed with different radiation qualities with TL dosimeters, are presented in Table 1. The chosen measurement techniques as well as other major phenomena affecting on the total uncertainty, and therefore the use of TL dosimeters in clinical applications in general, are discussed in detail in this chapter.

4.1 TL characteristics

Thermoluminescent sensitivity may be defined [10] as the amount of light released by the phosphor per unit of radiation exposure, i.e. as the TL signal per kerma or mean absorbed dose. In the uncertainty evaluations of the papers I, II, III and IV, the uncertainty arising from the detector sensitivity is taken into account concerning the calibration of the detectors, since in practice the uncertainties caused by the calibration source and detector sensitivity are linked together. Because of the deviation in readout values of the individual detectors, illustrating the deviation of detector sensitivities

[3] TLD Niewiadomski & Co., Krakow, Poland.

specification of the uncertainty u_i	estimated u_i (1 S.D.) %						
	TLD-100			TLD-700	MCP-7s	MTS-Ns	$CaSO_4$:Dy
	X-ray	γ	n + γ	n + γ	n + γ	n + γ	γ
u_1 = u(TL characteristics)	4	3	15	17	20	16	8
u_2 = u(phantom measurements)	3	15	3	3	3	3	
u_3 = u(*in vivo* measurements)							45
$u_{total} = (u_1^2 + u_2^2 + u_3^2)^{1/2}$	5	15	16	17	20	16	46

Table 1. The uncertainties present in the absorbed dose determinations of the clinical applications of this thesis. The measurements with TL detectors were performed either in the phantoms or *in vivo* using different radiation fields composed of X-rays, gamma rays (γ), or neutrons and gamma rays (n + γ).

among a detector batch, individual calibration of the dosimeters was performed for all the used solid discs and pellets (papers I, II, III, IV).

In the radiation field composed of X-rays, the absorbed dose measurements were performed using TLD-100 detectors, which are sensitive enough (hundreds of counts per mGy) for the purpose. The uncertainty of the whole calibration chain was estimated to be 3% (1 S.D.) (Table 1 in paper I) including the uncertainty of the reference irradiation (~ 2% (1 S.D.)). Since the individual calibration, i.e. reference irradiation, was performed using a $^{90}Y/^{90}Sr$ source, the reference air kerma, used for the determination of the detector sensitivity, was calibrated for each X-ray quality of interest by using a diagnostic dosimeter. Therefore, energy dependence (i.e. detector sensitivity as a function of photon energy) of TLD-100 detectors [67,68] was not considered when evaluating the uncertainties of the measurements (paper I). Also, in order to avoid the uncertainty caused by the sensitivity change of the reader used over the irradiation cycles, the calibration detectors irradiated with the $^{90}Yr/^{90}Sr$ source were read frequently. The energy response of LiF TL detectors in a phantom differs from that in free air [69], due to spectral differences caused by attenuation and scatter of X-rays. A specific factor was used to correct for the isotropic angular distribution of photons scattered from the phantom. Also, a conversion factor was used to convert between the air kerma and tissue kerma. Both of these factors are slightly dependent on the X-ray quality [70,71], and their uncertainty was estimated to be 2.5% (1 S.D.).

When measuring the absorbed doses from gamma radiation of radioimmunotherapy, two types of TL detectors were used. $CaSO_4$:Dy powder was selected for use for patient studies because of its high sensitivity. However, the response of $CaSO_4$:Dy powder is highly energy dependent [67], and the relative sensitivity of the $CaSO_4$:Dy dosimeters varies as a function of distance from the irradiation source, since

the radiation arriving at the water-air interface has a large amount of low energy scattered photons as a result of Compton interactions of the 364 keV gamma radiation. Therefore, TLD-100 discs were used for determining the sensitivity of $CaSO_4$:Dy dosimeters as a function of distance from the source, as well as in the phantom measurements performed to establish the developed method. In the Monte Carlo simulation (paper II), the sensitivity of the used TLD-100 detectors was found to be independent within 1% on the source to detector distance, and furthermore, energy in the range of interest. The uncertainty caused by the energy dependence of $CaSO_4$:Dy detectors was estimated to be as high as 7% (1 S.D.) (Table 4 in paper II) even if calibrated with TLD-100 detectors at various distances from the ^{131}I source.

In BNCT, the absorbed dose to tissue is composed of dose components caused by thermal, epithermal and fast neutrons, and gamma rays. These reaction specific dose components must be known separately because of their different biological effectiveness. In our study (paper III), ^{7}LiF:Mg,Cu,P (MCP-7s) pellets with low sensitivity for neutron radiation [72,73,74,75] were chosen for gamma separation. However, they were found to be more sensitive to thermal neutrons than the literature indicated [75], and therefore a detailed study was performed of the thermal neutron sensitivities for both the MCP-7s and TLD-700 TL detectors (paper IV). According to this study, the uncertainty in deriving the introduced figures of merit, used for determining correction factors [64] for thermal neutron sensitivities, is approx. 13% for both the MCP-7s and TLD-700 TL detectors. Therefore, the uncertainty caused by the thermal neutron sensitivity is the dominant one compared with the other uncertainties caused by the characteristics of the used TL materials (i.e. photon sensitivity or low energy dependence [67,76]).

Experimentally observed thermal neutron sensitivities of the LiF TLD-100 detectors have been found to vary [77]. These variations in sensitivity are mainly due to the self-shielding of TL detectors which can vary from a few percent to 50% depending on the geometry, i.e. thickness, of the TL detector [14]. Therefore, besides the traditional TLD-100 detectors also two-layer detectors (MTS-Ns) with an ultra-thin active LiF:Mg,Ti layer on a passive base were selected for use as a neutron radiation sensitive dosimeters for BNCT (paper III). These new detectors have been developed and studied for personal neutron dosimetry [78], and found to provide essential improvements in their characteristics especially with respect to self-shielding.

The response of TL material to neutrons is mainly dependent on the neutron capture cross sections of its constituent elements [79]. The ^{6}Li abundance in natural Li, causing the thermal neutron sensitivity of LiF detectors, is about 7% [80]. Because of the high capture cross section and, therefore, the high sensitivity of LiF detectors to thermal neutrons, a lowered reaction power or shortened exposure time has to be used for irradiation with a high neutron fluence rate of BNCT beams. However, those may

cause additional measurement uncertainty. Therefore, it might be useful to reduce the neutron sensitivity of LiF detectors by adding ^{7}Li to natural Li. According to statistical analysis (paper VI), this reduction would not significantly increase the random uncertainties of neutron fluence measurements. Problems may arise, however, from the increased relative gamma sensitivity of these special LiF detectors with the reduced neutron sensitivity. The gamma component of the induced TL signal would not be insignificant compared to the neutron component, as it is when using TL detectors composed of natural Li [81], and it might be difficult to separate the gamma and neutron radiation induced signals. When using the commercial LiF:Mg,Ti detectors MTS-Ns and TLD-100, the gamma subtraction had no significant effect on neutron dose: the gamma ray induced TL signal had a magnitude of approx. 1% of the net TL signal (paper III).

The reproducibility of the used solid detectors, TLD-100 and TLD-700 from Harshaw and MTS-Ns and MCP-7s from TLD Niewiadomski & Co., were also studied for the applied readout and annealing procedures (paper III). The reproducibility of the MTS-Ns detectors was found to be 6% (1 S.D.) in the mixed neutron-gamma field. Even though the standard annealing procedure [82] of LiF:Mg,Ti detectors (400°C ± 5°C for one hour followed by 75°C ± 3°C for 24 h pre-irradiation and / or 110°C ± 3°C for 10 min post-irradiation) had been used, the reproducibility might not have improved since the 400^{o}C temperature annealing used regenerates these detectors completely, and residual background readout values do not explain any instability in the detector sensitivities. For this study, shortened annealing times were selected for use with the aim of avoiding possible damages to TL detectors resulting from a long-term treatment at a high temperature [24,25]. According to Carlsson [83] and Toivonen [25], the shortened annealing procedure with a lower annealing temperature (approx. 300°C) reproduces relative detector sensitivities better than the procedure where the TL detectors are heated up to 400°C in free air.

The Harshaw detectors reproduced their readings accurately both after successive gamma irradiations (1 S.D.=3% for TLD-100 and 1% for TLD-700) and irradiations in the mixed neutron-gamma field in which the standard deviations (1 S.D.) were 4% and 2% for TLD-100 and TLD-700, respectively. Therefore, the lower reproducibility in integrating the glow area of the neutron specific 270^{o}C glow peak compared to the 210^{o}C glow peak may be the source of uncertainty in the mixed field dosimetry with the LiF:Mg,Ti detectors compared to gamma dosimetry. In our recent study (paper VI) on the response characteristics of the LiF:Mg,Ti (MTS-Ns) detectors, the reproducibility of the readout values has been improved by using an advanced readout technique with linear heating and glow curve analysis. The reproducibility of the MTS-Ns detectors as well as TLD-100 detectors may also be worsened by radiation damages due to neutron radiation. Gambarini and Sinha Roy [81] have reported about irreversible radiation

damage in dosimeters having high sensitivity to thermal neutrons (TLD-600 with 96.5% ^{6}Li), showing a memory effect on the previous thermal neutron irradiation history which is not restored by anneal treatment. In the same study, ^{7}LiF dosimeters showed a response unaffected by the thermal neutron irradiation [81].

The MCP-7s detectors reproduced their readings accurately (1 S.D.=3%) after successive gamma irradiations. After a few irradiations in the mixed neutron-gamma field, the reproducibility of the TL readings was found to be only 7% (1 S.D.). The reason for the poorer reproducibility was assumed to be in the high temperature glow peaks of the ^{7}LiF:Mg,Cu,P TL material generated by neutron radiation. These glow peaks are not released during the annealing procedure used: it is recommended by the manufacturer that MCP-7s detectors be prepared by heating at 240°C ± 5°C for ten minutes, but since a sufficiently stable oven was unavailable to meet this temperature requirement, the preparation was made by heating the detectors in a Vinten Toledo 654 -reader[4] with a reading temperature of 240°C for 40 s followed by rapid cooling with the normal rate of the heater planchet (from 240°C to 80°C in about 20 seconds). By using the recommended or longer annealing time with an advanced readout technique and background subtraction method, the standard deviation, i.e. the random uncertainty, may be smaller than the obtained 7% which represents gross signals without any background subtraction. Furthermore, the use of an annealing temperature higher (e.g. 260°C) than the standard 240°C improves the reproducibility of ^{7}LiF:Mg,Cu,P detectors [84,85].

The accuracy of the measurements performed with TL detectors depends also on the dose level used: for the most TL detectors, the induced TL signal as a function of exposure is non-linear above a certain dose range. For example, the TL response is supralinear for LiF:Mg,Ti detectors at the doses exceeding approx. 1 Gy, and therefore, non-linearity corrections have to be introduced, especially if high-temperature peaks are included in the glow curve [86,87]. In this study, because of the dose levels used (varying from a few mGy (papers I, II, III) to hundreds of mGy (paper IV)), no uncertainty was assumed to be caused by the supra- or sublinearity of the detectors [88,89,90].

One of the important dosimetric characteristics of a TL material is fading, i.e. the loss of signal during storage. The phenomenon has been studied extensively, both experimentally and theoretically, for LiF:Mg,Ti detectors but the data reported in the literature show considerable variation [91]. According to the manufacturers of the TL detector types used (Harshaw, TLD Niewiadomski & Co.), fading varies from 5%/yr (LiF:Mg,Ti and LiF:Mg,Cu,P) to 2% in one month ($CaSO_4$:Dy) at room temperature. Therefore, fading, as well as the uncertainty caused by it, were assumed to be negligible for the measurements in which the solid detectors were read according to a fixed time schedule (paper I), or immediately after use, preheated at 135°C for 16 s (papers II,III).

[4] Vinten Instruments Ltd, Great Britain.

Also, the readout procedure and the reader itself may cause uncertainty to the detector readout values, i.e. to the sensitivity and reproducibility of the detectors. The readers used, Vinten Toledo 654 (papers II, III, IV) and Dosacus[5] (paper I), and the errors caused by them, have been thoroughly studied by Toivonen [24,25], and the manual planchet reader (Toledo 654) has been found more stable than an automatic reader based on gas heating (Dosacus). It has also been stated [24,25] that it is possible to improve the precision of TL measurements by improving readout techniques and readers; as mentioned previously, the reproducibility of MTS-Ns detectors was improved by using a Risø[6] TL/OSL system with a linear heating profile and a glow curve analysis (paper VI).

In the measurements with TL detectors, additional uncertainty may also be caused by detector handling. In this study, detectors were carefully handled either by mechanical or vacuum tweezers with the aim of avoiding uncertainty. As TL detectors lose sensitivity when frequently handled with mechanical tweezers [13], the detecting surface of the dosimeters was not touched while using mechanical tweezers. No cleaning of the used detectors was needed nor, therefore, performed. Besides, cleaning is good to be avoided [13] since it may change the crystalline surface structure of the detector and, therefore, its response characteristics.

4.2 Phantom measurements

Additional uncertainty arises, when deriving absorbed doses to individual organs of the Alderson-Rando phantom from the doses of the reference points, i.e. those points measured directly with TL detectors or interpolated from them (paper I). The maximum error, caused by the use of one point instead of the whole organ in determining the absorbed dose of the organ, was found to be 11% (paper II). The organ used for this approximation was the kidney, the volume of which was estimated from the literature [92] to be 150 cm^3. In the X-ray study, one or a few reference points represented the smallest organs, but dozens of reference points were needed for large organs (paper I). Therefore, an uncertainty caused by the use of points, representing the entire volume of the organ, is estimated to be 3% (1 S.D.). Also, an uncertainty of approx. 5% is caused to soft tissue in close contact with bone through interface effects.

In order to determine the uncertainties present in the phantom measurements, performed with large number of consecutive TL detectors, Monte Carlo simulations were used (paper V). In the simulations, detectors positions were selected to be the same as those in actual measurements (paper III) at the BNCT beam of FiR 1. Also, the perturbation of the neutron and gamma fluences caused by the polycarbonate

[5] Rados, Finland.
[6] Risø National Laboratory of Denmark, Denmark

($(C_{16}H_{14}O_3)_n$) frames, with the aid of which the TL detectors have been situated in the measurements in the phantom, were determined using Monte Carlo simulations. As a result, no significant fluence differences were found to occur in the cases of thermal and epithermal neutrons or gamma rays: the neutron and gamma fluence rates were similar within 3% for the simulations with and without neighbouring ^{nat}LiF or ^{7}LiF TL detectors and polycarbonate frames at various measurement points in the phantom. The spatial uncertainty of the measurements performed in the water phantom was ± 0.5 cm due to uncertain positioning of the TL detectors in their thin holder frames (paper III).

According to the Burlin cavity theory [93], a one sided cavity effect occurs when irradiating the phantom medium surrounded MCP-7s TL detectors (paper III). Because of the one sided effect and the isotropic angular distribution of the gamma rays and secondary electrons, the increase in the dose to the active LiF was estimated to be less than 10%. Therefore, a reduction of 5% was made for the kerma, measured with MCP-7s detectors, to correct for the error caused by the cavity effect.

Often, a considerable disturbance of the dose distribution near the interface between the detector and medium may result due to the different scattering properties of the two materials. Therefore, a study (paper II) was performed to determine whether the phantom-TL material interface effects on the surface of the phantom, caused by the nonequilibrium of secondary electrons, have to be taken into consideration in determining absorbed doses from the readout values of the powdered $CaSO_4$:Dy TL detectors. The phantom study showed no thickness effect in the response of $CaSO_4$:Dy powder on the surface of the phantom. Therefore, no additional uncertainty was due to measurements performed on the surface of the phantom where the requirement for the secondary electron equilibrium is not completely met.

4.3 *in vivo* measurements

The uncertainty, present in the absorbed dose measurements performed *in vivo* (paper II), was evaluated with the help of phantom measurements and calculations based on the use of a point dose kernel technique. A bolus factor was determined with an accuracy of 4% from the phantom study to correct for the lack of back-scattering from the bolus tissue in the reference point compared to that in the organ of interest. Also, the attenuation coefficient for the broad beam geometry (including scattering) was determined, and found to be 0.05 ± 0.01 cm^{-1} (paper II).

CT images of the patients were used in estimating the distances; the uncertainty in measuring the individual distances from the (point) irradiation source to the reference point or to the organ of interest was estimated to be 20% (1 S.D.). The uncertainty caused by the use of a point source instead of a volume distribution of the activity

incorporated in the cancer tissue was calculated to be 10% (1 S.D.). As already mentioned in chapter 4.2., the maximum error caused by the use of a point of interest instead of the whole organ in determining the absorbed dose was found to be 11%. The total uncertainty of measuring absorbed doses of the organs with the TL dosimeters placed on the skin of the patient was found (paper II) to be below 50% (1 S.D.). The total uncertainty is high compared to that of the phantom study (15%), and is mainly explained by the uncertainty in the absorbed dose in the reference point, which arises from the uncertain radioactivity distribution in the body (25%, 1 S.D.), and by the characteristics of the disease: no exact tumour area can be delineated, and therefore the measuring distances from the (point) irradiation source to the reference point or to the organ of interest is uncertain.

5 DISCUSSION

In this thesis, the abilities of TL dosimeters for clinical dosimetry were studied in terms of the precision and accuracy. The measurements with TL detectors, both in the phantom and *in vivo*, were performed for clinical applications in which the dose prediction is difficult and not as part of a routine verification procedure, such as RIT and BNCT. Also, a phantom study with external X-rays was performed.

According to the ICRU [26], the accuracy in the dose determination should be within ± 5%, or even lower, in the conventional radiotherapy. In diagnostics, there is not so far these kinds of requirements for accuracy. Furthermore, since the radiation damages in diagnostics are non-deterministic, the needed accuracy is not so high as in therapy. Therefore, it is found (Table 1) that the uncertainty, caused by TL characteristics and the measurement arrangements, is at a tolerable level in the case of TLD-100 detectors exposed with diagnostic X-rays. As well as being the only reasonable measurement technique, the TL detectors are also accurate enough for the verification of the diagnostic studies in the anatomical phantoms.

$CaSO_4$:Dy is a sufficiently sensitive TL material for the *in vivo* dosimetry of the internal radionuclide source. However, $CaSO_4$:Dy detectors are not tissue-equivalent and their uncertainty, caused by the TL characteristics of the $CaSO_4$:Dy dosimeters, is high (Table 1). This inaccuracy is dominated by the energy dependence of the $CaSO_4$:Dy powder. Therefore, it was found to be essential that the most appropriate detector types, i.e. independent of the energy, are selected for use in clinical applications in which the irradiation of a wide energy range is applied.

When using an internal irradiation source, both in the phantom and *in vivo*, the measurement uncertainties were dominant (Table 1) compared with that caused by the TL characteristics. Therefore, special care has to be taken while positioning the TL

detectors for the (*in vivo*) measurements. Also, the patient anatomy and geometry as well as the radionuclide distribution have to be studied thoroughly. If possible, the placement of the detectors should be marked into the patient while studying his / her anatomy e.g. with CT.

The low precision of the gamma dosimeter MCP-7s, irradiated in the mixed neutron-gamma field of BNCT, is mainly due to the poor reproducibility of the detector. Also, the inaccuracy of the gamma dose measurements, performed both with TLD-700 and MCP-7s detectors, is caused by the high uncertainty present in deriving the correction factors for the thermal neutron sensitivity of the TL detectors used. The reproducibility of MCP-7s detectors might be reduced considerably (e.g. to be < 1% as in [89,94,95]) from the obtained 7% (paper III) by using a proper oven for annealing, and the recommended annealing procedure. However, there are still questions about the optimum thermal treatment and readout procedures for LiF:Mg,Cu,P TL detectors (e.g. [84,96,97,98,99,100]). The use of thin layer detectors, such as MCP-7s, for the gamma dosimetry should also be re-considered because of their uncertainty due to the cavity effect.

The uncertainty of the thermal neutron correction factor mainly arises (paper IV) from the uncertainty of the spectrum averaged fluence-to-kerma conversion factor, used in theoretical derivation of the correction factor [64]. The uncertainty of the used fluence-to-kerma conversion factor is 10% [101], and it also causes uncertainty to (relative) neutron fluence measurements (paper III) performed with TLD-100 and MTS-Ns detectors. As well as in the case of the external X-ray irradiation beam, the measurement arrangements in the phantom did not increase significantly the total uncertainty (Table 1) when irradiating at the mixed neutron-gamma field of BNCT. It was found that even if lower precision, the accuracy of the dose estimations, performed with TL detectors in the phantoms, is not essentially worse when compared with that performed with ionisation chambers [102].

The use of TL detectors for clinical dosimetry may be laborious. In many applications a large number of TL detectors have to be used for the measurement of the absorbed dose which requires care and accuracy in order to keep the detectors in order. Also, the detectors should be handled with care in order to avoid additional uncertainty and loss of sensitivity (see e.g. ref. [13]). The recommended readout and annealing procedures should also be followed even though they are cumbersome in many cases. In spite of these drawbacks, TL detectors may be for many clinical applications the only available, and reasonable, measurement technique. If handled with care, their uncertainties can easily be reduced, if not to a tolerable level, but to the same level achieved with other methods.

6 SUMMARY

In many clinical applications, the use of TL dosimeters is the only available tool for dosimetry. TL detectors are especially useful in difficult geometries where the best use can be made of their advantages such as their stand alone character and small physical size. In this thesis, the abilities of TL detectors were studied in different clinical applications, in which TL dosimeters were found to be capable of making absorbed dose measurements.

Paper I contains organ dose determinations of X-ray examinations using TL detectors. The measured absorbed doses are compared with the computed doses. TL detectors are found to be accurate within 5% (1 S.D.), and are therefore capable of verifying the calculated absorbed doses.

In **Paper II**, a method for determining the absorbed doses to organs in systemic radiation therapy is introduced. The developed method is based on TL detectors placed on the patient's skin. A phantom study is carried out to establish the method. It was found that in SRT the TL method can be used to estimate the absorbed doses to those critical organs near the body surface within 50%.

The characteristics of TL detectors, irradiated at the mixed neutron-gamma field of BNCT, is studied in **Paper III**. A code of practice for relative neutron fluence measurements, performed with TL detectors in the mixed neutron-gamma field of BNCT, is presented. TL detectors (both the thin-layer MTS-Ns and TLD-100) were found to be able to measure the neutron dose component with an accuracy of 16%. Also, a new gamma dosimeter, the MCP-7s, was found to be less sensitive to thermal neutrons and therefore more suitable for BNCT dosimetry than the conventional TLD-700. An experimental method for determining correction factors for thermal neutron sensitivity of TL detectors used in gamma dosimetry of BNCT is presented in **Paper IV**.

A Monte Carlo study was performed on the influence of adjacent TL detectors to TL readings in simultaneous measurements at BNCT beams (**Paper V**). In the study, it is concluded that several TL dosimeters can be used in simultaneous measurements in the phantoms without the need for the correction factors arising from the shielding of the other TL dosimeters present in the measurement, or the polycarbonate frame used in the measurements.

Paper VI is a statistical analysis of the effects of sensitivity reduction on the random uncertainty of the measured gamma and neutron doses at the BNCT beam. It was found that the reduction does not significantly increase the random uncertainties.

The uncertainties of TL dosimeters were found to be high but not essentially more significant than those in other measurement techniques used for clinical dosimetry. Also, the precision and accuracy of the absorbed dose measurements performed with TL detectors may be improved by: i) selecting the appropriate detector type (e.g. energy

dependence) for the measurement purpose, ii) using the recommended thermal treatment procedure, and iii) careful handling of the detectors. It is shown in this thesis that the absorbed gamma doses can be measured with TL detectors within 20% in a mixed neutron-gamma field, which enables *in vivo* measurements at BNCT beams with approximately the same accuracy. However, as reported in paper II, a major uncertainty in the *in vivo* measurements may arise from the placement of the TL detectors. Therefore, the positioning of the detectors, as well as the other arrangements of the measurement, have to be planned carefully beforehand, and if possible, also verified while studying the anatomy of the patient.

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REFERENCES

[1] International Commission on Radiation Units and Measurements, Radiation quantities and units, ICRU Rep. 33, ICRU Publications, Bethesda, Maryland (1980).

[2] Lindskoug, B.A. Clinical applications of thermoluminescent dosimetry. Strahlenterapie 161, 91-95 (1985).

[3] Council Directive 97/43/Euratom of 30 June 1997 on health protection of individuals against the dangers of ionizing radiation in relation to medical exposure.

[4] International Atomic Energy Agency. Absorbed dose determination in photon and electron beams. An international code of practice. IAEA Technical Reports Series No. 277, Vienna, Austria (1987).

[5] Svensson, H. and Brahme, A. Recent advances in electron and photon dosimetry. In: Radiation Dosimetry. Physical and Biological Aspects. Orton, C.G. (ed.). New York: Plenum Press (1986).

[6] Rikner, G. and Grusell, E. General specifications for silicon semiconductors for use in radiation dosimetry. Phys. Med. Biol. 32, 1109-1117 (1987).

[7] Rikner, G. and Grusell, E. Patient dose measurements in photon fields by means of silicon semiconductor detectors. Med. Phys. 17, 870-873 (1987).

[8] Nilsson, B., Ruden, B. and Sorcini, B. Characteristics of silicon diodes as patient dosimeters in external radiation therapy. Radiother. Oncol. 11, 279-288 (1988).

[9] Li, C., Lamel, L.S. and Tom, D. A patient dose verification program using diode detectors. Med. Dosim. 20, 209-214 (1995).

[10] Cameron, J.R., Suntharalingam, N. and Kenney, G.N. Thermoluminescent Dosimetry. Madison: University of Wisconsin Press (1968).

[11] Becker, K. Solid State Dosimetry. Cleveland, Ohio: CRC Press (1973).

[12] McKinlay, A.F. Thermoluminescence Dosimetry. Bristol: Adam Hilger (1981).

[13] Oberhofer, M. and Scharmann, A. (eds.). Applied Thermoluminescence Dosimetry. Bristol: Adam Hilger (1981).

[14] Horowitz, Y.S. (ed.). Thermoluminescence and Thermoluminescent Dosimetry. Vol. I - III. Boca Raton, Florida: CRC Press (1984).

[15] McKeever, S.W.S. Thermoluminescence of Solids. Cambridge: Cambridge University Press (1985).

[16] Mahesh, K., Weng, P.S. and Furetta, C. Thermoluminescence in Solids and Its Applications. Ashford, Kent: Nuclear Technology Publishing (1989).

[17] Attix, F.H. Introduction to Radiological Physics and Radiation Dosimetry. New York: John Wiley & Sons (1986).

[18] Daniels, F., Boyd, C.A. and Saunders, D.F. Thermoluminescence as a research tool. Science 117, 343-349 (1953).

[19] Kron, T. Thermoluminescence dosimetry and its applications in medicine--Part 1: Physics, materials and equipment. Australas. Phys. Eng. Sci. Med. 17, 175-199 (1994).

[20] Kron, T. Thermoluminescence dosimetry and its applications in medicine--Part 2: History and applications. Australas. Phys. Eng. Sci. Med. 18, 1-25 (1995).

[21] Kron, T., Schneider, M., Murray, A. and Mameghan, H. Clinical thermoluminescence dosimetry: how do expectations and results compare? Radiother. Oncol. 26, 151-161 (1993).

[22] Kirby, T.H., Hanson, W.F. and Johnston, D.A. Uncertainty analysis of absorbed dose calculations from thermoluminescence dosimeters. Med. Phys. 19, 1427-1433 (1992).

[23] Robertson, M.E.A. Identification and reduction of errors in thermoluminescence dosimetry systems. Thesis. University of Surrey. Weybridge, Surrey: D.A. Pitman (1981).

[24] Toivonen, M. Improved processes in therapy dosimetry with solid LiF thermoluminescent detectors. Thesis. Report STUK-A101, Helsinki: Finnish Centre for Radiation and Nuclear Safety (1991).

[25] Toivonen, M. Improving the accuracy of TLD systems in clinical applications. Radiat. Prot. Dosim. 47, 497-503 (1993).

[26] International Commission on Radiation Units and Measurements, Determination of absorbed dose in a patient irradiated by beams of X or gamma rays in radiotherapy procedures, ICRU Rep. 24, ICRU Publications, Bethesda, Maryland (1976).

[27] Johansson, K.-A., Hanson, W.F. and Horiot, J.C. Workshop of the EORTC Radiotherapy Group on quality assurance in cooperative trials of radiotherapy: a recommendation for EORTC Cooperative Groups. Radiother. Oncol. 11, 201-203 (1988).

[28] Fisher, D.R. Radiation dosimetry for radioimmunotherapy: an overview of current capabilities and limitations. Cancer (Suppl.) 73, 905-911 (1994).

[29] Strand, S.-E., Jönsson, B.-A., Ljungberg, M. and Tennvall, J. Radioimmunotherapy dosimetry - a review. Acta Oncol. 32, 807-817 (1993).

[30] Raaijmakers, C.P.J., Konijnenberg, M.W., Verhagen, H.W. and Mijnheer, B.J. Determination of dose components in phantoms irradiated with an epithermal neutron beam for boron neutron capture therapy. Med. Phys. 22, 321-329 (1995).

[31] Raaijmakers, C.P.J., Watkins, P.R.D., Nottelman, E.L., Verhagen, H.W., Jansen, J.T.M., Zoetelief, J. and Mijnheer, B.J. The neutron sensitivity of dosimeters applied to boron neutron capture therapy. Med. Phys. 23, 1581-1589 (1996).

[32] Banu, H., Alam, M.N., Chowdhury, M.I., Kamal, M., Bardhan, D.K. and Chakraborty, D. Assessment of occupational and patient dose from diagnostic and therapeutic radiation exposure using thermoluminescent dosimetry. Health. Phys. 74, 478-480 (1998).

[33] Lavoie, C. and Don, C. In vivo measurement method of ovarian dose during barium enema examinations. Br. J. Radiol. 70, 291-295 (1997).

[34] Gentry, J.R. and DeWerd, L.A. TLD measurements of in vivo mammographic exposures and the calculated mean glandular dose across the United States. Med. Phys. 23, 899-903 (1996).

[35] Moulin, G., Chagnaud, C., Waultier, S., Le Brigand, B., Espagnan, M., Dessi, P., Bartoli, J.M., Chatenet, P. and Botti, G. Radiation dose to the lenses in CT of the paranasal sinuses. Neuroradiology 38, Suppl. 1, S127-129 (1996).

[36] Goldstein, A. Exposure and dose in panoramic radiology. Med. Phys. 25, 1033-1040 (1998).

[37] Damilakis, J., Prassopoulos, P., Perisinakis, K., Faflia, C. and Gourtsoyiannis, N. CT of the sacroiliac joints. Dosimetry and optimal settings for a high-resolution technique. Acta Radiol. 38, 870-875 (1997).

[38] Alderson, S.W., Lanzl, L.H., Rollins, M. and Spira, I. An instrumented phantom system for analog computation of treatment plans. Am. J. Roentgenology, 87, 185-195 (1962).

[39] Kairemo, K.J.A. Radioimmunotherapy of solid cancers. A review. Acta Oncol. 35, 343-355 (1996).

[40] International Commission on Radiation Units and Measurements. Methods of assessment of absorbed dose in clinical use of radionuclides. ICRU Report No. 32. Washington, DC (1979).

[41] Loevinger, R. and Berman, M. A revised schema for calculating the absorbed dose from biologically distributed radionuclides. MIRD Pamphlet No. 1, Revised, New York: The Society of Nuclear Medicine (1976).

[42] Snyder, W.S, Ford, M.R, Warner, G.G. and Watson, S.B. ‘’S’’, Absorbed dose per unit cumulative activity for selected radionuclides and organs. MIRD Pamphlet No. 11, New York: The Society of Nuclear Medicine (1975).

[43] Watson, E.E., Stabin, M.G. and Siegel, J.A. MIRD formulation. Med. Phys. 20, 511-514 (1993).

[44] Giap, H.B., Macey, D.J., Bayouth, J.E. and Boyer, A.L. Validation of a dose-point kernel convolution technique for internal dosimetry. Phys. Med. Biol. 40, 365-381 (1995).

[45] Giap, H.B., Macey, D.J. and Podoloff, D.A. Development of a SPECT-based three-dimensional treatment planning system for radioimmunotherapy. J. Nucl. Med. 36, 1885-1894 (1995).

[46] Strandh, M. and Strand, S.-E. In vivo absorbed dose measurements with mini-TLDs. Parameters effecting the reliability. Acta Oncol. 35, 713-719 (1996).

[47] Strand, S.-E., Strandh, M. and Spanne, P. Electron microscopy and computed microtomography studies of in vivo implanted mini-TL dosimeters. Acta Oncol. 32, 787-791 (1993).

[48] Buchsbaum, D.J. and Wessels, B.W. Introduction: Radiolabeled antibody tumor dosimetry. Med. Phys. 20, 499-501 (1993).

[49] Griffith, M.H., Yorke, E.D., Wessels, B.W., DeNardo, G.L. and Neacy, W.P. Direct dose confirmation of quantitative autoradiography with micro-TLD measurements for radioimmunotherapy. J. Nucl. Med. 29, 1795-1809 (1988).

[50] Langmuir, V.K., Wessels, B.W., Mendonca, H., Yorke, E.D. and Montilla, L. Comparisons of sectioned micro-TLD dose measurements with predicted dose from 131-I-labeled antibody. Med. Phys. 19, 1213-1218 (1992).

[51] Wessels, B.W. and Griffith, M.H. Miniature thermoluminescent dosimeter absorbed dose measurements in tumor phantom models. J. Nucl. Med. 27, 1308-1314 (1986).

[52] Wessels, B.W., Vessella, R.L., Palme, D.F., Berkopec, J.M., Smith, C.K. and Bradley, E.W. Radiobiological comparison of external beam irradiation and radioimmunotherapy in renal cell carcinoma xenografts. Int. J. Radiat. Oncol. Biol. Phys. 17, 1257-1263 (1989).

[53] Williams, J.A., Edwards, J.A., Wessels, B.W., Dillehay, L.E., Wanek, P.M., Poggenburg, J.K., Wharam, M.D., Order, S.E. and Klein, J.L. Targeting and therapy of human glioma xenografts in vivo using radiolabeled antibodies. Int. J. Radiat. Oncol. Biol. Phys. 19, 633-642 (1990).

[54] Strandh, M., Strand, S.-E. and Norrgren, K. The accuracy of in vivo absorbed dose measurements using mini-TLDs. Antibody Immunoconj. Radiopharm. 5, 335 (1992).

[55] Demidecki, A.J., Williams, L.E., Wong J.Y., Wessels, B.W., Yorke, E.D., Strandh, M. and Strand, S.-E. Considerations on the calibration of small thermoluminescent dosimeters used for measurement of beta particle absorbed doses in liquid environments. Med. Phys. 20, 1079-1087 (1993).

[56] Stewart, J.S.W., Hird, V., Snook, D., Sullivan, M., Hooker, G., Courtenay-Luck, N., Sivolapenko, G., Griffiths, M., Myers, M.J., Lambert, H.E., Munro, A.J. and Epenetos, A.A. Intraperitoneal radioimmunotherapy for ovarian cancer: pharmacokinetics, toxicity, and efficacy of I-131 labeled monoclonal antibodies. Int. J. Radiat. Oncol. Biol. Phys. 16, 405-413 (1989).

[57] Kairemo, K.J.A., Jekunen, A.P., Bondestam, S., Korppi-Tommola, E.T., Savolainen, S. and Paavonen, T.K. Detection of pseudomyxoma peritonei with radioimmunohistochemistry and radioimmunoscintigraphy. Cancer Biotherapy Radiopharm. 11, 325-334 (1996).

[58] Barth, R.F., Soloway, A.H., Fairchild, R.G. and Brugger, R.M. Boron neutron capture therapy for cancer. Cancer 70, 2995-3007 (1992).

[59] Slatkin, D.N.A. A history of boron neutron capture therapy of brain tumors. Brain 114, 1609-1629 (1991).

[60] Gahbauer, R., Gupta, N., Blue, T., Goodman, J., Grecula, J., Soloway, A.H. and Wambersie, A. BNCT: status and dosimetry requirements. Radiat. Prot. Dosim. 70, 547-554 (1997).

[61] Perks, C.A., Harrison, K.G., Birch, R. and Delafield, H.J. The characteristics of a high intensity 24 keV iron-filtered neutron beam. Radiat. Prot. Dosim. 15, 31-40 (1986).

[62] Perks, C.A., Mill, A.J., Constanine, G., Harrison, K.G. and Gibson, J.A.B. A review of boron neutron capture therapy (BNCT) and the design and dosimetry of a high-intensity, 24 keV, neutron beam for BNCT research. Br. J. Radiol. 61, 1115-1126 (1988).

[63] Perks, C.A., Constantine, G. and Birch, R. The design and dosimetry of an Al/S/Ar filtered neutron beam. Radiat. Prot. Dosim. 23, 329-332 (1988).

[64] Croft, S. and Perks, C.A. Corrections to gamma ray dosimetry measurements made in Harwell's two high intensity filtered neutron beams using ^{7}LiF thermoluminescent dosemeters owing to their neutron sensitivity. Radiat. Prot. Dosim. 33, 351-354 (1990).

[65] Savolainen, S., Auterinen, I., Kallio, M., Kärkkäinen, M., Kosunen, A., Aschan, C., Benczik, J., Färkkilä, M., Hiismäki, P., Kaita, K., Kulvik, M., Ryynänen, P., Seppälä, T. Serén, T., Tanner, V., Toivonen, M., Vähätalo J. and Ylä-Mella, H. The Finnish boron neutron capture therapy program - an overview on scientific projects. In: Advances in Neutron Capture Therapy, Vol. I, Medicine and Physics. Larsson, B., Crawford, J. and Weinreich, R. (eds.), pp. 342-347. Amsterdam: Elsevier (1997).

[66] Auterinen, I. and Hiismäki, P. The epithermal neutron irradiation station for boron neutron capture therapy (BNCT) at the FiR 1 in Otaniemi. Med. Biol. Eng. Comput. 34, Suppl. 1, Part 1, 299-300 (1996).

[67] Bassi, P., Busuoli, G. and Rimondi, O. Calculated energy dependence of some RTL and RPL detectors. Int. J. of Appl. Radiat. Isot. 27, 291-305 (1976).

[68] Gamboa-deBuen, I., Buenfil, A.E., Ruiz, C.G., Rodriguez-Villafuerte, M., Flores, A. and Brandan, M.E. Thermoluminescent response and relative efficiency of TLD-100 exposed to low-energy X-rays. Phys. Med. Biol. 43, 2073-2083 (1998).

[69] Zoetelief, J. and Jansen, J.Th.M. Calculated energy response correction factors for LiF thermoluminescent dosemeters employed in the seventh EULEP dosimetry intercomparison. Phys. Med. Biol. 42, 1491-1504 (1997).

[70] Seuntjens, J., Thierens, H., Van der Plaetsen, A. and Segaert, O. Conversion factor f for X-ray beam qualities, specified by peak tube potential and HVL value. Phys. Med. Biol. 32, 595-603 (1987).

[71] Morgan, T.J. and Brateman, L. The Energy and directional response of Harshaw TLD-100 thermoluminescent dosimeters in the diagnostic X-ray energy range. Health Phys. 33, 339-341 (1977).

[72] Horowitz, Y.S. and Shachar, B.B. Thermoluminescent LiF (Mg,Cu,P) for gamma ray dosimetry in mixed fast neutron-gamma ray radiation fields. Radiat. Prot. Dosim. 23, 401-404 (1988).

[73] Pradhan, A.S. and Bhatt, R.C. Thermoluminescence response of LiF (Mg,Cu,P) and TLD-100 to thermal neutrons, ^{241}Am alphas and gamma rays. Radiat. Prot. Dosim. 33, 247-250 (1990).

[74] Bilski, P., Olko, P., Burgkhardt, B., Piesch, E. and Waligorski, M.P.R. Thermoluminescence efficiency of LiF:Mg,Cu,P (MCP-N) detectors to photons, beta electrons, alpha particles and themal neutrons. Radiat. Prot. Dosim. 55, 31-38 (1994).

[75] Wang, S.S., Cai, G.G., Zhou, K.Q. and Zhou, R.X. Thermoluminescent response of ^{6}LiF(Mg,Cu,P) and ^{7}LiF(Mg,Cu,P) TL chips in neutron and gamma ray mixed radiation fields. Radiat. Prot. Dosim. 33, 247-250 (1990).

[76] Piesch, E. and Burgkhardt, B. Thermoluminescence and personal dosimetry: potentials and limits. In: Techniques and Management of Personnel Thermoluminescence Dosimetry Services. Oberhofer, M. and Scharmann, A. (eds.), pp. 125-129. Dordrecht, the Netherlands: Kluwer Academic Publisher (1993).

[77] Gibson, J.A.B. The relative tissue kerma sensitivity of thermoluminescent materials to neutrons. Radiat. Prot. Dosim. 15, 253-266 (1986).

[78] Budzanowski, M., Bilski, P., Olko, P., Niewiadomski, T., Burgkhardt, B. and Piesch, E. New TL detectors for personal neutron dosimetry. Radiat. Prot. Dosim. 47, 419-423 (1993).

[79] Ayyangar, K., Lakshmanan, A.R., Chandra, B. and Ramadas, K. A comparison of thermal neutron and gamma ray sensitivities of common TLD materials. Phys. Med. Biol. 19, 665-676 (1974).

[80] Ayyangar, K., Reddy, A.R. and Brownell, G.L. Some studies in thermoluminescence from lithium fluoride and other materials exposed to neutrons and other radiations. In: 2nd Int. Conf. on Luminescence Dosimetry, Gatlinberg, Tennessee, September 1968, USAEC, CONF-680920, pp. 525-542 (1968).

[81] Gambarini, G. and Sinha Roy, M. Dependence of TLD thermoluminescence yield on absorbed dose in a thermal neutron field. Appl. Radiat. Isot. 48, 1467-1475 (1997).

[82] Horowitz, Y.S. The annealing characteristics of LiF:Mg,Ti. Radiat. Prot. Dosim. 30, 219-230 (1990).

[83] Carlsson, C.A. Thermoluminescence of LiF : dependence of thermal history. Phys. Med. Biol. 14, 107-118 (1969).

[84] Bilski, P., Budzanowski, M. and Olko, P. Dependence of LiF:Mg,Cu,P (MCP-N) glow curve structure on dopant composition and thermal treatment. Radiat. Prot. Dosim. 69, 187-198 (1997).

[85] Toivonen, M., Chernov, V., Jungner, H., Auterinen, I. and Toivonen, A. Response characteristics of LiF:Mg,Cu,P TL detectors in boron neutron capture therapy dosimetry. Radiat. Prot. Dosim. 85, 373-376 (1999).

[86] Attix, F.H. Further consideration of the track interaction model for thermoluminescence in LiF (TLD-100). J. Appl. Phys. 46, 81-88 (1975).

[87] Perks, C.A. and Marshall, M. The dose response of individual glow peaks in ^{7}LiF (TLD-700) read out normally and after UV phototransfer. Radiat. Prot. Doism. 33, 271-274 (1990).

[88] Harris, C.K., Elson, H.R., Lamba, M.A.S. and Foster, A.E. A comparison of the effectiveness of thermoluminescent crystals LiF:Mg,Ti and LiF:Mg,Cu,P for clinical dosimetry. Med. Phys. 24, 1527-1529 (1997).

[89] Fill, U.A. and Regulla, D.F. Use of Chinese LiF:Mg,Cu,P (GR-200) TL detectors at therapy-level absorbed doses. Appl. Radiat. Isot. 49, 791-793 (1998).

[90] Horowitz, Y.S. LiF:Mg,Ti versus LiF:Mg,Cu,P: the competition heats up. Radiat. Prot. Dosim. 47, 135-141 (1993).

[91] Horowitz, Y. Fading in LiF:Mg,Ti. Radiat. Prot. Dosim. 32, 147-148 (1990).

[92] Servomaa, A., Rannikko, S., Nikitin, V., Golikov, V., Ermakov, I., Masarskyi, L. and Saltukova, L. A topographically and anatomically unified phantom model for organ dose determinations in radiation hygiene. Report STUK-A87, Helsinki: Finnish Centre for Radiation and Nuclear Safety (1989).

[93] Burlin, T.E. A general theory of cavity ionizations. Br. J. Radiol. 39, 727-734 (1966).

[94] Fill, U.A. and Regulla, D.F. Measurements with Chinese LiF:Mg,Cu,P (GR-200) detectors using a volume-type oven. Appl. Radiat. Isot. 48, 133-134 (1996).

[95] Muñiz, J.L. and Delgado, A. A study of LiF GR-200 for radiotherapy mailed dosimetry. Phys. Med. Biol. 42, 2569-2576 (1997).

[96] Zha, Z., Wang, S., Shen, W., Zhu, J. and Cai, G. Preparation and characteristics of LiF:Mg,Cu,P thermoluminescent material. Radiat. Prot. Dosim. 47, 111-118 (1993).

[97] Bos, A.J.J. and Meijvogel, K. Influence of thermal treatment on glow curve and thermoluminescence emission spectra of LiF:Mg,Cu,P. Radiat. Meas. 24, 239-247 (1995).

[98] Piters, T.M. and Bos, A.J.J. Influence of the cooling rate on repeatability of LiF:Mg,Cu,P thermoluminescent chips. Radiat. Prot. Dosim. 33, 91-94 (1990).

[99] Saez-Vergara, J.C. and Romero, A.M. The influence of the heating system on the hypersensitive thermoluminescence material LiF:Mg,Cu,P (GR-200). Radiat. Prot. Dosim. 66, 431-436 (1996).

[100] Oster, L., Horowitz, Y.S. and Horowitz, A. Glow curve readout of LiF:Mg,Cu,P (GR-200) chips at maximum temperatures between 240°C and 280°C: elimination of the residual signal. Radiat. Prot. Dosim. 49, 407-411 (1993).

[101] Caswell, R.S. and Coyne, J.J. Kerma factors for neutron energies below 30 MeV. Radiat. Res. 83, 217-254 (1980).

[102] Kosunen, A., Kortesniemi, M., Ylä-Mella, H., Seppälä, T., Lampinen, J., Serén, T., Auterinen, I., Järvinen, H. and Savolainen, S. Twin ionisation chambers for dose determinations in phantom in an epithermal neutron beam. Radiat. Prot. Dosim. 81, 187-194 (1999).