KAUNAS UNIVERSITY OF TECHNOLOGY
FACULTY OF MATEMATICS AND NATURAL SCIENCES

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THE UNCERTAINTY ASSESSMENT METHODOLOGY IN CLINICAL THERMOLUMINESCENCE DOSIMETRY APPLICATIONS

Master's Final Degree Project

Supervisor
Lect. dr. Jurgita Laurikaitienė

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(Title and code of study programme)

'The uncertainty assessment methodology in clinical thermoluminescence dosimetry applications'

DECLARATION OF ACADEMIC INTEGRITY

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Kaunas

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SUMMARY

In developed countries more than 20 % [1] of all radiation exposure can be attributed to the medical field. All medical applications require some type procedures that guarantee their quality control. This is very important because any errors or miscalculations can lead to negative effect to patient health. Dosimetric measurements are a part of quality control measures. In-vivo dosimetry using thermoluminescent dosimeters is preferred in this practice, due to the broad availability of detector materials and variety of detector size and shape. High measurement uncertainties is the main disadvantage of this method. This problem in personal dosimetry is being solved applying different calibration and uncertainty evaluation methodologies, but radiotherapy field still lacks this kind of practice.

In this work uncertainty assessment methodologies for both systematic and random uncertainty values evaluation were applied for medical practise and used for low energy radiotherapy.

It was found that after exposure to 2 Gy of dose TLDs measured doses from 1,6 GY to 2,26 Gy, variations in results being from -14% to +20,7%. Using of correct uncertainty estimation methodology expanded uncertainty of TLD measurements can be reduced by ~20-21% with the result reproducibility of ~ 88% for dose range of 2-4 Gy. After comparison of doses measured by TLDs and by GafChromic films it was found, that in the case of low scattering, dose measurement errors were lower than 10 %.

SANTRAUKA

Išsivysčiusiose šalyse daugiau nei 20% [1] visos apšvitos gali būti priskiriamos medicinos sričiai. Visos medicininės apšvitos procedūros turi būti aprašytos kokybės užtikrinimo programoje, taip yra užtikrinama jų kokybės kontrolė Tai ypač svarbu, kadangi bet kokios apšvitos klaidos ar klaidingi skaičiavimai gali turėti įtakos paciento sveikatai. Dozimetriniai matavimai yra kokybės kontrolės dalis. In-vivo dozimetrijoje termoluminescenciniai dozimetrai (TLD) dėl plačios detektorių medžiagų ir dydžio ir formos įvairovės, dozimetriniams matavimams yra plačiai naudojami. Tačiau didelis matavimo neapibrėžtumas yra pagrindinis šio metodo trūkumas. Ši problema profesinėje dozimetrijoje yra išspręsta taikant skirtingas kalibravimo ir neapibrėžtumo vertinimo metodikas, tačiau tokios praktikos vis dar trūksta radioterapijoje.

Šiame darbe panaudota neapibrėžčių nustatymo metodika, įvertinant sistemines ir atsitiktines neapibrėžties vertes buvo pritaikyta klinikinėje praktikoje ir panaudota mažų energijų radioterapijoje.

Darbo metu buvo nustatyta, kad po apšvitinimo 2 Gy doze TLD išmatuotos dozių vertės kito nuo 1,6 Gy iki 2,26 Gy, kas sąlygojo gautų rezultatų kitimą nuo -14% iki + 20,7%. Taigi naudojant teisingą neapibrėžčių nustatymo metodiką galima išplėstinę neapibrėžtį sumažinti iki 20-21% su rezultatams atsikartojant iki 88%, kai doze yra 2-4 Gy ribose. Palyginus dozes, išmatuotas TLD ir GafChromic filmais buvo nustatyta, kad atveju kai sklaida nedidelė išmatuotos dozės paklaidos buvo mažesnės nei 10%.

Abbreviations

TLD – thermoluminescent dosimeters;

IAEA – International Atomic Energy Agency;

ALARA – as low as reasonably possible;

CT – computed tomography;

SPECT – single photon emission computed tomography;

PET- positron emission therapy;

SRT – superficial radiotherapy;

LINAC- linear accelerator;

3D-CRT - three-dimensional conformal radiation therapy;

IMRT -intensity modulated radiation therapy;

MLC -multileaf collimator;

IGRT- image guided radiation therapy;

CBCT - cone-beam computed tomography;

MRI – magnetic resonance imaging;

WHO – World Health Organization;

ICRU – International Commission Radiation Unit;

EPID – electronic portal imaging detector;

CCD – charge-coupled device:

MOSFET – metal-oxide semiconductor field effect transistor;

PSD – plastic scintillation detector;

OSLD – optically stimulated dosimeters;

RPLD - radiophotoluminescent dosimeters;

TL – thermoluminescence;

PMT- photomultiplier;

UV – ultraviolet;

ICRP -International Commission on Radiological Protection;

PMMA -Poly (methyl methacrylate);

FSD – field-surface distance;

STD – source – target distance;

RCF – reader calibration factor;

ECC – element correction coefficient.

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INTRODUCTION

Thermoluminescent dosimeters (TLDs) are one of cheapest and most accessible types of dosimeters. Beside of this they also can come in various forms and materials, resulting large range of sensitivity. TLDs have very wide range of applications in personal monitoring, industry and medical field.

In medical applications, where any type of error or mistake can result negative effect to patient's health dosimetric practices are important issues. In this filed TLDs can be used for equipment and treatment planning quality assessment. Even though a lot of TLD's features, such as small size and possibility to use them for dose measurements on both surface of phantom or patient and in volume are very useful in medical practice, TLDs also have cons. Mainly due to relative high errors, uncertainties and time-consuming reading process a lot of time TLDs are not used for measurements and other type of dosimeters, if at all, are selected.

In personal dosimetry this problem is solved with the extensive methodologies that were developed by many different controlling institutions such as IAEA. Medical field due to it's specifics and variations of parameters used still lacks unified methodology. Unified methodology of error and uncertainty not only would let us to improve quality of the treatment, but also would help to reduce cost of clinical dosimetry, since other dosimetric methods are much expensive. During this work methodology of TLD calibration and uncertainty assessment used for personal dosimetry will be applied to the radiotherapic use.

Main task includes:

1. Calibration of TLD dosimeters, using other dosimetric method in order to obtain reference dose and evaluation of TLD reader effect on results.
2. Calculation of absorbed doses, errors and combined system uncertainty, using methodology developed by IAEA for personal dosimetry and evaluation of these results. Evaluation of result repeatability, when measuring dose multiple times
3. Application of mentioned TLD methodology for dose measurements in phantom volume
4. Application of mentioned TLD methodology for evaluation of shielding effect on dose.

The irradiation of TLDs procedure was done performed in Oncology Hospital of the Hospital of Lithuanian University of Health Sciences Kaunas Clinics

LITERATURE REVIEW

In developed countries, such as Lithuania more than 20% of radiation exposure can be attributed to medical sources [1]. This radiation exposure in medical practice is usually due in three main procedure types. These procedures are radiation diagnostic or radiation diagnostic, radiation therapy or radiotherapy and nuclear medicine [2]. Due to negative effects of radiation to the patient's body, these procedures require throughout analysis of patient's situation for confirmation, that procedure is necessary and will not harm patient more then it's positive affect on either treatment or diagnosis of the disease. Because of this in medical radiology principle known as ALARA (As low as reasonably achievable) is widely applied [3]. This principle guaranties than both positive and negative effects of procedure is evaluated and compared. This means that procedure is optimized for best results with the lowest negative effects for patient (dose that is used during procedure should not negate usefulness the procedure itself).

1.1.1 Radiation diagnostics and nuclear medicine

Largest part of exposure can be attributed to radiodiagnostic, mainly due frequent use of standard diagnostic procedures such as x-ray, mammography and CT scans [3]. All of these procedures are based on x rays passing through the body to form pictures on film or on a computer or television monitor, which are viewed by a radiologist.

During nuclear medicine's diagnostic procedures, such as single photon emission computed tomography (SPECT) and positron emission tomography (PET) scans, a very small amount of radioactive material (radiopharmaceuticals) is inhaled, injected, or swallowed by the patient [2]. Then during a nuclear medicine exam, a special camera is used to detect energy given off by the radioactive material in patient's body and form a picture of organs and their function on a computer monitor. A nuclear medicine physician views these pictures. The radioactive material typically disappears from patient's body within a few hours or days [3].

1.1.2 Radiation therapy

Patients who have been diagnosed with the cancer (malignant tumour) usually have few different treatment options. These include surgery, radiation therapy, chemotherapy or mixture of these methods. The choice of method depends on many different factors, which include [3][4]:

- Location of tumour (in which organ it is located);
- Size of the tumour;

- Type of tumour;
- Possibility of tumour spread to other organs.

Even though, as mentioned before, type of treatment depends entirely on the individual situation, in many cases, radiotherapy (radiation therapy) is preferred. This happens because radiotherapy is painless and produces much less negative effects to patient's health when compared to other treatment methods, in addition to that, with advancements in treatment methods and equipment, treatment can be applied very precisely to the region of tumour, thus protecting other organs from unnecessary exposure [4].

When speaking about radiation therapy, two main types that are used for patient treatment must be mentioned [3]: brachytherapy, when radiation source, such as cesium (^{131}Cs, ^{137}Cs), cobalt (^{60}Co), iodide (^{125}I), radium (^{226}Ra) or other [1][4], is inserted inside a patient's body and then is moved near the region of tumour, and external beam radiation therapy. First method is used for some cancer types such as prostate cancer, but second method is much more common and is safer for both the patient and the personnel who are present during the treatment [1].

1.1.3 External beam radiotherapy

During external radiotherapy two different types of radiation can be used [1]: electromagnetic radiation, e.g., x-rays and particles, e.g., electrons. Usually, therapeutic and diagnostic x-rays and gamma is in range are produce, when the tube voltage is in range of kilovolts (kV) and megavolts (MV) [4], and energy of therapeutic electrons is in range of megaelectronvolts (MeV) [5]. X-rays that are useful in medicine are produced when electrons are accelerated to high energies and then hits the target. In diagnostics and therapy x-rays are separated in such energy intervals [3][5]:

- Superficial x-rays (35-60 kV) – usually penetrate up to 5 mm and are mostly used for skin treatment [5];
- Diagnostic x-rays (20 -150 kV)- used in radiodiagnostic procedures [5];
- Orthovoltage x-rays (200-500 kV) and supervoltage x-rays (0,5 -1 MV)– have penetration dept of 4-6 cm, used for treatment of skin, superficial structures, ribs [3];
- Megavoltage x-rays (1-25 MV) – are used to treat tumours that are located deep in patient's body. Megavoltage x-rays are preferred for therapy, because they are attenuated less then lower energy photons and penetrate much further, thus resulting lower does to skin [6]. Also, megavoltage x-rays have much higher relative biological effectiveness.

Fig 1. Example of linear accelerator [7].

Currently external treatment is delivered by using following equipment:

- Superficial radiotherapy (SRT) machines – produces low energy (20-150 kV) x-rays, that are used for treatment of skin conditions [5];
- Orthovoltage x-ray tubes – produces orthovoltage x-rays (200-500 kV) and are used for treatment of skin cancer [5];
- Linear accelerators or "LINACs" - produce megavoltage x-rays as bremsstrahlung spectrum by rapidly decelerating electrons in target material (usually tungsten) [1]. Then intensity and shape of beam is modified with the help of collimators. Example of LINAC is seen in Fig.1
- Cobalt units – radiation is produces with the help of radioisotope cobalt-60, which produces stable dual energy beams of 1.17 and 1.33 MV [4]. Nowadays cobalt-60 is mostly replaced by linear accelerators because they can generate much higher energies, but still can be used in specific applications, such as gamma knife [6].

In radiotherapy, which uses electrons instead of photons, radiation is produced in similar way, as it is in case of x-ray [3][6]. When target is removed from standard x-ray unit we can get high energy electrons, which are then directed to treated part of patient's body. These electron beams usually have energies of 4-20 MeV, with the energies above 18 MeV being used very rarely [4]. Produced dose rapidly decreases in depth, resulting low penetration depth of 1-5 cm and highest distribution of dose near the surface. Because of this, electron beams are used for treatment superficial lesions [4].

One of the most rarely used types of external beam radiotherapy is hadron therapy. It includes both proton and neutron therapy. Proton therapy is used because characteristics of proton beam allow effectively reduce dose to nearby healthy organs and tissue [4][5]. Neutron therapy is used for radioresistant types of tumour, that very difficult to remove using conventional type of x-ray radiation therapy [4]. Hadron therapy is used rarely and is available in only small amount of treatment centres in the world.

In order to deliver correct treatment, dose which is delivered to volume of tumour must be sufficiently high and patient's healthy tissue also must be protected from unnecessary exposure. This requires throughout planning which is performed by medical physicist with the help of specialized software.

External radiotherapy can be used in either the higher doses, to try to cure the cancer [10] (curative radiotherapy) or in the lower doses, to relieve pain and other symptoms and slow down cancer progress (palliative radiotherapy) [3].

Usually, during the treatment patients receive the radiation dose via multiple smaller exposure sessions. This approach is known as a dose fractionation [10]. Healthy tissue regenerates more quickly than tumour tissue do after exposure to radiation. Fractionated radiotherapy, therefore, gives the healthy tissue a chance to recover between sessions. As healthy tissue and tumour tissue react to radiation differently, method is used to both optimize treatment process and to reduce negative effects of treatment. Fractionated radiotherapy is typically spread out over a time period of several weeks (usually between five and eight weeks) [1] [3].

1.1.4 Advanced external beam radiation methods

As mentioned before, external beam radiation therapy is delivered through two main methods as electromagnetic radiation in form of x-ray and gamma or as particles, mainly electrons. In recent years, standard radiation treatment is increasingly used alongside with some type of imaging. These

advanced techniques that can help to improve both effectiveness and quality of treatment. Main techniques include [4] [6]:

- Three-dimensional conformal radiation therapy (3D-CRT) typically uses computed tomography (CT), magnetic resonance imaging (MRI) or positron emission tomography (PET) to define location, shape of tumour and volume which surrounds it. Based on this, complex plans are developed to deliver a radiation dose distribution within the patient, where the regions of high dose are concentrated within the tumours. Then with the help of collimators radiation beams are adjusted accordingly. Higher doses of radiation can be delivered to cancer cells while significantly reducing the amount of radiation received by surrounding healthy tissues [6] [8].

  3D-CRT is used to treat tumours that in the past might have been considered too close to vital organs and structures for radiation therapy. For example, 3D-CRT allows radiation to be delivered to head and neck tumours in a way that minimizes exposure of the spinal cord, optic nerve, salivary glands and other important structures [8].
- Intensity modulated radiation therapy (IMRT) is the form of 3D-CRT; it allows radiation to be modified to the shape of tumour. IMRT splits beam in many smaller beams, with intensity that can be adjusted individually for each one of them by the multi-leaf collimator (MLC), that are attached to the linear accelerator. It decreases dose received by healthy tissue even more than standard 3D-CRT [6].

  Due to its complexity, IMRT needs slightly longer treatment times than conventional radiotherapy. Longer planning time and additional safety checks are also required before the start of the treatment [8].
- Image guided radiation therapy (IGRT) includes CT, ultrasound or x-ray imaging as part of treatment process. Special on-board imaging system for IGRT, most commonly cone-beam CT (CBCT), is attached to a linear accelerator [8]. With the help of imaging techniques position of tumour and surrounding anatomies are taken and compared with the simulation scans. This helps identify possible movement of tumour and tissue and then adjust treatment plan according to it [8].

1.2 Importance of dosimetry in medical applications

Dosimetry in Medical Physics involves the patients and phantom dosimetry as well as that for the occupationally exposed personnel and the environmental monitoring in hospitals.

1.2.1 In Radiotherapy

In external beam radiation therapy, where radiation exposures are intentional in order to obtain a direct benefit to the patient's health, it is very important to ensure that delivery of radiation doses to tumour tissue is accurate and of radiation administered to healthy tissue is minimal. Even though external beam radiation therapy is relatively save method with the low probability of accidents, sometimes accidental exposure is unavoidable.

Correctly delivered treatment plan has many potential benefits, including better tumour control, lower toxicities to healthy tissues, better quality of life for patient and possibility for improved survivability rate [9]. A treatment plan which is delivered poorly may have opposite results. Side effects of these accidental exposure can be classified as either acute side-effects and injuries (deterministic effects) or long-term complications (stochastic effects) [10]. An acute side-effect results from radiation damage to rapidly dividing tissues while a late complication is the result of irreversible damage to cells that sustain long-term tissue integrity. At some cases, late complication from radiation therapy can lead to a radiation-induced malignancy [9].

Radiation incidents can be either caused by human errors or by system [9]. One or more of the parameters involved in a patient irradiation may have a systematic error, which can lead to suboptimal patient treatment [11]. Probability of this is increasing, because newly developed radiation treatment techniques and their implementation into the clinic require an increasing level of alertness to verify the safe and accurate delivery of the prescribed treatment. In order to reduce probability of accidental exposure or to identify, as soon as possible, when the plan is delivered poorly, additional safeguard system is needed.

Dose is main parameter which helps to in identify if the treatment was delivered correctly. In order to limit the errors arising during the treatment course, some international organizations such as World Health Organization (WHO), International commission Radiation Unit (ICRU) and International atomic energy agency (IAEA) have recommended the implementation of quality assurance programmes, which consist of making use of all the necessary steps to ensure achievement of the intended dose delivery [10] [11]. These programs verify the correct functioning of all components in the radiation therapy chain including the treatment planning and delivery systems. Usually there are many processes that are involved in the dose delivery in radiation therapy and analysis of all them is quite difficult. Thus, an overall check of the procedure is therefore recommended and can be performed only by means of dosimetry. While dosimetry is mainly based

on the quality of the radiation beam as the long-term goal, in practice dosimetry helps to analyse the quantity of the dose absorbed by the patient. Dose quantities, measured during the procedure represent convenient indicators in the assessment of diagnostic practice and population exposure, estimating the health risk owing to stochastic effects of radiation and assessing the potential of deterministic effects and helps to prevent them. These quantities also allow comparing the risk from different types of procedures, irradiation geometries and radiation type [11].

The dosimetric goal of patient treatment has 2 components [10]: verification of the delivered dose and verification of the patient's positioning. Patient's positioning verification has been significantly aided by the availability of on-board–imaging systems such as electronic portal imaging detector (EPID is the process of using digital imaging, such as a CCD video camera, liquid ion chamber and amorphous silicon flat panel detectors to create a digital image with improved quality and contrast over traditional portal imaging), megavoltage CT and cone beam CT [12]. The verification of the delivered dose requires comparisons of measured and calculated dose distributions. Isodose lines, coloured two-dimensional maps are used to visualize measured and calculated dose distributions or three-dimensional surface plots. The assessment of the final uncertainty between the prescribed dose and dose that it delivered to the patient is an effective way of checking the entire dosimetric procedure.

1.2.2 In Radiodiagnostic

During radiodiagnostic the doses applied to patients does not require such level of control as in radiotherapy because the result of a radiological examination does not depend on the dose as much as a therapeutic exposition [13]. In radiology, the pertinence of an exposure is determined by the quality of the image and rarely a strict control of the exposition is required because it is more important the benefit obtained by improving the diagnosis that the radiation risks. However, clear evidence exists, in practice that the doses received by patients submitted to the same type of radiological examination vary very much from one patient to another [14]. In addition, medical radiology contributes very much to the collective dose of the population [10]. Optimization of the radiodiagnosis requires the evaluation of the effectiveness of the diagnosis as well as the measurements of the absorbed dose of the patients. So, it is necessary to have criteria for establishing the image quality required to make the diagnosis sure and to determine the dose to the patient. Dosimetry must be directed to [13]:

a) Establish that the doses received by the patients are in accordance with the optimal performance of the equipment (as a part of the quality control program);

b) Compare the doses among different equipment and techniques for optimizing the design and performance of new equipment.

c) Estimate the risk to the patient.

1.3 Dosimetric methods

There are several different ways of measuring absorbed doses from ionizing radiation. Two main types are external, also known as in vivo and internal, in vitro dosimetry [11].

Internal dosimetry is the measurement of doses due to nuclear substances that have entered the body by way of ingestion, inhalation or other means. In vitro dosimetry refers to most of other physics measurements in phantoms [11]. Doses at depth are difficult, if not possible to obtain without invasive procedures. Internal dosimetry (in-vitro) involves two steps:

1. The level of radiation inside a person's body is estimated using one of three methods [14] [15]:

- In-vivo bioassay (direct measurement of radioactivity in the body);
- In-vitro bioassay (measurement of radioactivity in a person's urine or feaces);
- Measurement of radioactivity in air.

2. The resulting internal radiation dose is calculated.

Due to the complexity of the measurements in-vitro dosimetry is rarely used in medical applications. It is much more commonly used in dose monitoring for workers [14].

In vivo dosimetry refers to measuring dose received by the patient during the treatment. During external dosimetry the dose is measured when the radiation source is outside of the patient's body. In terms of dose, external dosimetry is concerned with radiation that can penetrate the skin: beta, photon, and neutron radiation [15]. Since photons and beta interact through electronic forces (interactions between charged particles) and neutrons interact through nuclear forces, their detection methods and dosimetry are substantially different [15][16]. The fundamental basis of external dosimetry is the determination of the absorbed energy in matter and, more specifically, human tissue.

Even though, dose distribution, during the treatment in patient's body is simulated by planning software, it may contain errors, because algorithms used for calculation of tissue inhomogeneity and interfaces are not completely accurate. In vivo dosimetry provides an additional safety step which helps to ensure that systematic errors have not been acquired during the treatment planning process.

If an inconsistency is found, the plan can be corrected before the patient begins to suffer irreversible problems due to the error [16].

Two major technologies are used today to monitor radiation exposure from ionizing radiation. The technologies are generally categorized as passive, or delayed readout, dosimetry and active, or real-time, dosimetry [15].

An active dosimeter produces a radiation-induced signal and displays a direct reading of the detected dose or dose rate in real time. Active dosimeters include diodes, metal-oxide semiconductor field effect transistors (MOSFETs), plastic scintillation detectors (PSDs) and electronic portal imaging devices (EPIDs) [15]. These detectors, containing integral build up can be placed both superficially and intracavitary and results are available immediately. Ionization chambers are rarely used of in vivo dosimetry but in some applications, such for monitoring a high-dose-rate brachytherapy procedure, the can be used (with the protective sleeve) for superficial or in intracavitary locations [11]. Active detectors can measure total dose during the treatment delivery and are capable to measure time-related dose fraction also known as dose rate, which can give additional useful in formation in some situations [16].

A passive dosimeter produces a radiation-induced signal, which is stored in the device. Passive dosimeters include radiographic and radiochromic films, implantable MOSFET detectors, thermoluminescent dosimeters (TLDs), optically stimulated dosimeters (OSLDs) and radiophotoluminescent dosimeters RPLDs [17]. Passive dosimeters do not require any additional power source and are integrating - they show the complete dose which is absorbed during the treatment. These detectors do not provide immediate results but require additional time which ranges from minutes to hours for their read out [17]. This happens when dosimeters are processed whit special reader, and the output is analysed. Films offer two-dimensional dose measurements, while TLD's, OSLD's, RPLD's and MOSFET's provide point dose information [18]. Because of this TLD's, such as LiF dosimeters (LiF-TLD) are always chosen as the gamma dosimeter for human radiation therapy [19].

1.3.1 Thermoluminescence dosimetry

Thermoluminescence dosimeters (TLDs) is one of most commonly used types of dosimeters. They are used in different fields of both the medicine and the industry. One of most important of these TLD characteristics is their size – they are small, which allows them to be adhered to the patient without causing it any discomfort or interfering with his movement [19]. As mentioned before, they can be used for point dose measurements. Because of this it is possible to use them for surface and

volume dose measurements when using the as system of multiple dosimeters, as they were used during this project. Other advantages of TLD should be also mentioned [20][21][22]:

- The existence of thermoluminescent materials that are nearly tissue equivalent;
- TLD's are sufficiently accurate and sensitive and for both personal, environmental monitoring;
- Small size solid dosimeters can be adapted for both manual and automatic processing;
- TLD's processing is simple;
- TLD's are suitability for extremity and skin beta dosimetry;
- Thermoluminescent materials usually are stable under varying environmental conditions;
- TLD's can be reused multiple times;
- TLD's response to the dose and dose rate is linear over a large range.

Mechanism of TLD is based on luminescence principle [21]. For this process three elements are essential: recombination centres, mobile (or charge) carriers, and traps. Also, for explanation we need to look at the electronic energy band model and assume that energy states can exist in the forbidden band. These energy states are known as metastable states [22] and have a relatively long lifetime. Mobile carriers freely travel in the crystal lattice until they are trapped in metastable states or recombine with other charges [22][23].

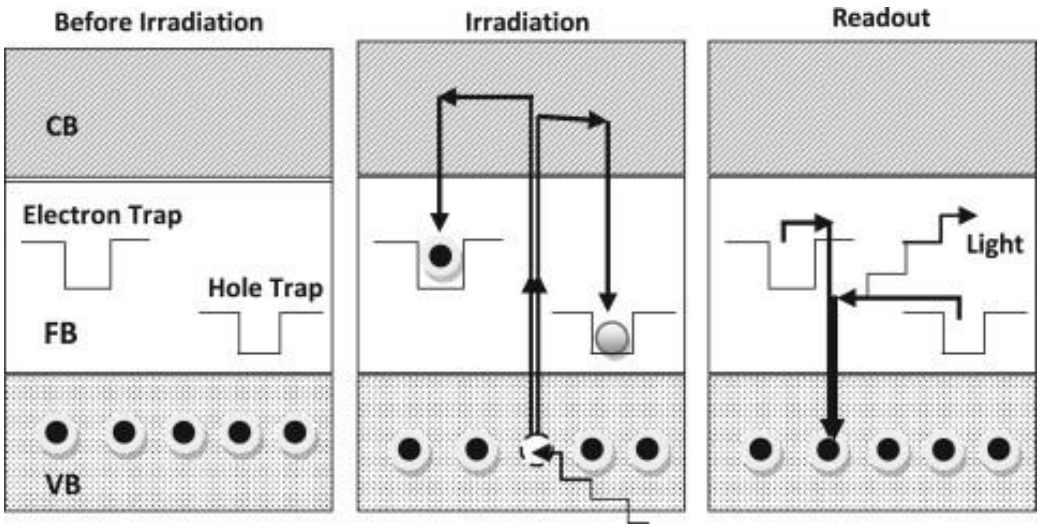

Fig 2. Simplified scheme of thermoluminescence process [20]

When a crystalline material is irradiated its structure is alternated due to ionization. A certain amount of energy is absorbed by insulating material, in this case, a thermoluminescent material, and this creates mobile carriers (both electrons and holes). The holes remain in valence band and the electrons travel from valence band to conduction band. They freely travel in the crystal lattice until they are trapped in metastable states (caught by crystalline imperfections, traps)) or recombine with

other charges. Usually the electrons will not have enough energy to escape, until energy threshold is reached [20][21][23].

This usually happens when external energy source in introduced. Then charge carriers leave the traps and starts to travel along the crystalline lattice until they recombine at luminescence centres and charge carriers return to their original states before the irradiation and excess energy is emitted in a form of visible light photons [21][24]. In case of thermoluminescence this external energy is provided by external the heat source. Threshold energy, which is needed for the process to start is called activation energy or trap depth [24][25].

In thermoluminescent detectors number of trapped electrons and holes is the same as the number of the electron- hole pairs that are produced during the exposure. Ideally, every trapped electron and hole produces one energy photon [26].

1.4 TLD materials

In nature there are more than 1000 different TL materials [25], but only few of them can be used in medical applications. These include dosimeters that are approximately equivalent to the human tissue, such as lithium fluoride (LiF), lithium borate ($Li_2B_4O_7$) and beryllium oxide (BeO) [26][28] and materials that have much higher atomic number, but better sensitivity such as calcium sulphate ($CaSO_4$), calcium fluoride (CaF_2) and aluminium oxide (AlO_3) [25][27][29].

1.4.1 LiF based materials

Most commonly used TL material in clinical dosimetry is lithium fluoride. At first this material produced with impurities of titanium (Ti 15 ppm (particles per million)) and magnesium (Mg 300 ppm) [28]. This type of material is called LiF; Mg,Ti. Magnesium is main source of electron traps and titanium helps to create luminescence centres. LiF based materials are available in many different physical forms and dimensions [31]. Even materials that have identical form factor may vary because of different production process, which leads to different crystal structure and variations in doping. The range of applications for LiF can be increased by using different isotopes of Lithium, which leads to differences in cross sections and sensitivity of mentioned material [32]. This type of materials includes well known TLD600 (enriched with ^{5}Li) and TLD700(enriched with ^{7}Li) [33].

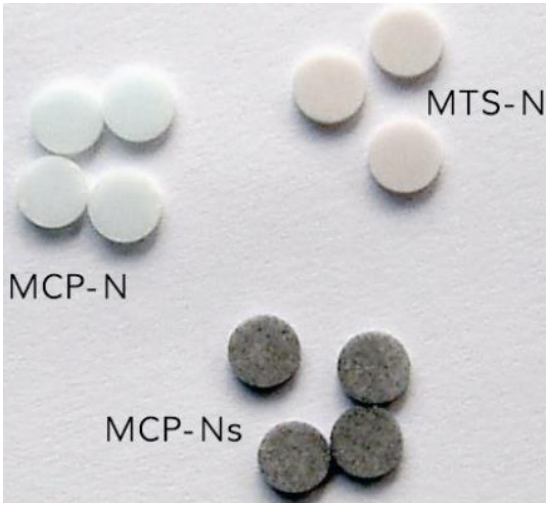

Fig 3. Different pellets of LiF based TLDs: MTS-N is LiF: Mg,Ti (TLD) MCP-N and MCP-Ns is LiF: Mg, Cu, P [30]

Newer variation LiF based thermoluminescent material is lithium fluoride which is doped with 100 ppm phosphorus, 50 ppm copper and 2000 ppm of magnesium (LiF: Mg,Cu,P) [31][33]. This material is tissue equivalent andis about 30 times more sensitive than standard LiF:Mg,Ti. Also, this material requires much shorter annealing cycle. Because of all these factors LiF:Mg,Cu,P is preferable alternative to other TL materials [34].

1.4.2 $Li_2B_4O_7$ based materials

Lithium borate is material which is completely tissue equivalent. It is typically doped with about 0,1% Mn. This concentration can is adjusted in order to match different types of tissue. Additionally, lithium borate has simple, uncomplicated glow curve and simple annealing procedure [35]. This material is more sensitive them LiF:Mg,Ti, but glow peak maximum is emitted in wavelength of 600 μm [35][36], which are near detection limits of most photomultipliers. Compared with LiF, lithium borate has a wider, more uniform energy response to photons but is more sensitive to thermal neutrons. Lithium borate's sensitivity to thermal neutrons is due to the ^{6} Li and ^{10}B content [36]. The ^{6}Li can be reduced (7.4% down to 0.01%) for LiF dosimeters suitable for beta and photon radiations. Pure ^{7}Li and natural lithium or lithium enriched in ^{6} Li are used to measure mixed gamma and thermal-neutron radiations [35][36].

This material starts to fade much more easily the standard lithium fluoride and at high dose lithium borate is prone to discolouring, which then leads to reduction of thermoluminescence signal. Even though this material has many disadvantages it still extensively used in medical applications because of its tissue equivalence [31].

Table 1. Properties of different TL materials [27]

TLD material	TLD100 (LiF: Mg,Ti)	LiF: Mg,Cu,P	TLD800 ($Li_2B_4O_7$)	$CaSO_4$: Mn	TLD900 ($CaSO_4$: Dy)
Physical density (g/cm^2)	2,64	2,64	2,3	2,61	2,61
Effective atomic number	8,2	8,2	7,4	15,3	15,3
Energy response 30 keV/1,25 MeV	1	about 30	0,3	70	about 15
Temperature of main glow peak (^0C)	195	210	200	110	220, 250
Maximum wavelength of emitted light (nm)	400	380	600	500	480, 570
Fading of main glow peak at 20 ^0C	<10% per year	2% in 3months	10% per year	50% per day	6% in 6 months
Typical annealing procedure	1h: 400^0C 20h: 80^0C	¼ h: 250^0C 2h: 100^0C	½ h: 300^0C	½ h: 400^0C	½ h: 400^0C (up to 700^0C may be used)
Useful dose range (Gy)	$5*10^{-3}$ to10^3	10^{-3} to 100	10^{-2} to 10^4	$<10^{-4}$ to 100	10^{-4} to 100
Principle dosimetric applications	Personal, radiotherapy, TLD600: neutrons		Diagnostic, radiotherapy	Environmental, low dose, high fading	Environmental, personal
Additional information	Complex glow curve, most common TLD, available as ^5Li and ^7Li	Good S/N ratio, No suppralinea-rity, low temperature annealing	Tissue equivalent	Low dose TLD, high sensitivity	Complex glow curve, high sensitivity

1.4.3 $CaSO_4$ based materials

$CaSO_4$ is not tissue equivalent and high atomic number material with the high sensitivity. This material is one of most sensitive TL materials, with the very high fading ratio (up to 50% per day) [37]. $CaSO_4$ can be doped with many different materials, such as thulium (Tm), dysprosium (Dy) and samarium (Sm) [31]. The most stable of these detectors is $CaSO_4$:Dy (TLD 900). TLD 900 is available in many different forms and has low fading, which not exceed 5% [38]. But glove curve is very

complex and has 10 different peaks. $CaSO_4$:Dy has similar dose response to that of bone and sometimes can be used for dose prediction in superficial radiotherapy [37][38].

Other TLD materials include CaF_2, calcium fluoride (CaF2), beryllium oxide (BeO), aluminium oxide(Al2O3) and magnesium orthosilicate (Mg2SiO4) [29][31]. Not like the others, BeO is tissue equivalent, but toxic and required high temperature of 600^0C for complete annealing [31]. Other materials, like film, they require additional filters in order to match their energy response to that of tissue. Because of this they are mainly used for only environmental monitoring. These materials are relatively cheap. CaF2 is very sensitive to radiation, but it over- responds to low energy x-rays. Al_2O_3 is also referred as sapphire detector. It has complex glow curve with fast fading in the low energy peaks (detection limit is about 0,3 μGy) [30]. Slow heating rates helps to improve detection limit but results slow processing of detector. Both BeO and Al_2O_3 are sensitive to light [27][31].

1.5 Thermal background

Thermoluminescence measurements usually are affected by the presence of a thermal background. This signal is induced by irradiation, but by high-temperature signal, which is present during the processing of the dosimeter [39]. There are at least three different sources of this signal, that can be easily identified and distinguished. These include [39][40]:

- Instrumental background (also known as PMT dark current, constant which is not related to the temperature);
- Emission of thermal radiation from heated dosimeter and heating element (also known as black body radiation, which is directly related to the temperature);
- Emission from chemiluminescence effects, which are occurring on the surface (e.g. surface oxidation).

High-temperature part of the measured glow-curve is usually a superposition of TL signal related to the previous exposition of the sample to ionizing radiation of the mentioned above signals, that are non-radiation induced [39]. Because of this, it is very important to find out a method of separation of the radiation and non-radiation induced signals. The effect of instrumental background can be eliminated simply by checking PMT dark current value before the reading process [39]. This is usually done by the reader. Effects of chemiluminescence can be also suppressed significantly by introducing the inert gas, such as Ar or N_2 [31] into the heating chamber during the reading process. Thermal radiation effects may be reduced by appropriate optical filters that is applied to the reader. Additionally, other methods, such as, chemical cleaning of dosimeters surface and postprocessing and fitting of received data helps to eliminate these high temperature signals [39][40].

1.6 Glow curves

Escape from traps can happen at different energies, meaning that charge carriers are released at different temperatures. This can be seen in the material's characteristic glow curve, plot of light intensity against temperature. When TLD material is heated after the irradiation, light is emitted with the higher intensity at peaks that are situated at specific temperatures. Glow curve usually has one or more of these peaks. Amplitude or the area under one peak, at a constant temperature, is proportional to the total number of charge carriers that are captured in the traps. Most of thermoluminescence readers are using this property. Based on the total emission of one or more glow curve peaks measurements can be made. In the range, in which their thermoluminescent response as a function of dose is linear, readout of material is very simple and direct [30][40][41].

Changing concentration of impurities leads to changes in thermal equilibrium of crystal defects and other lattice disorders, which then could significantly affect the structure of both glow peak and the properties of TLD material [27][40]. Glow peaks of the same material can be significantly different for different types of radiation. The intensity of these peaks is mainly dependent on the amount of radiation dose, ideally a single, stable glow peak should be situated around 200 ^{O}C [40].

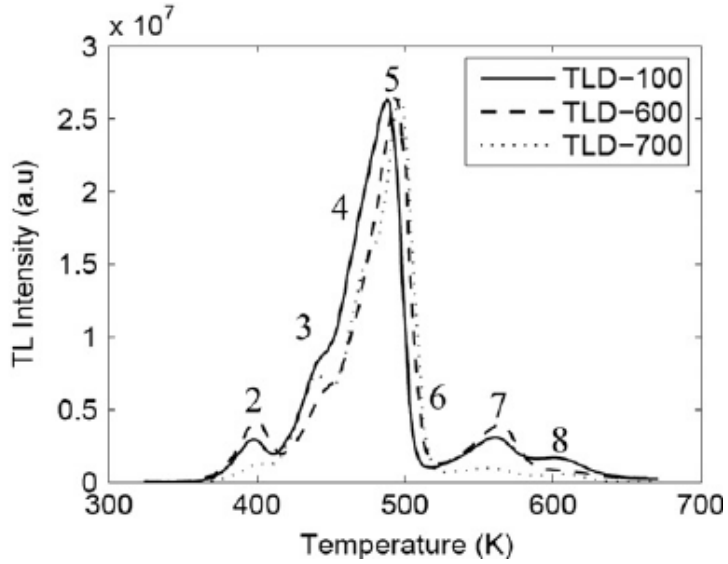

Fig 4. Glow curves of TLD-100, TLD-600 and TLD-700 [42]

Easiest and most accurate thermoluminescence readout from glow curves can be obtained when the peak is in a narrow temperature (around 50-60 degrees) range [40]. Curve with multiple glow peaks could also be accepted if the peaks are well separated or if we have satellite peaks, the difference between the amplitudes of main and satellite peaks is about 10 times [31]. Peak position of the glow curve changes significantly at the temperatures that are higher than 300 ^{O}C, and most of thermoluminescent materials become unsuitable for standard dosimetric applications [27].

Other factors that may influence glow curve [43][47]:

- Annealing at a high temperatures and other thermal treatments can may influence TL in a way that removes defects that may appear inside TLD structure. This action with effect of irradiation greatly enhances TL [43][44]. Several processes are responsible for that phenomenon, which is called ‘sensitization’ or ‘the pre-dose effect’. Annealing that is performed at insufficient temperatures can have negative results in obtained glow curve, such as having additional peaks that appear because of impurities [47].
- Mobile carriers can also be released from traps when they absorb optical, not thermal energy. Because of this, unwanted exposure of light (infra-red, visible, or UV), can reduce number of the trapped charge carriers, thus reducing TL intensity [45]. This causes so called “bleaching spectrum” and the interpretation of glow curve can much more be complicated. Energy that is needed for this process is usually greater than the thermal activation energy and is not directly related to the kinetic parameters of the TL peak [43].
- Some of released charge carriers can be recaptured again in other traps [44]. Phototransfer technique is based on this phenomenon. In case of technique material at first is irradiated via x-ray or gamma and then additionally exposed to monochromatic UV light, at selected wavelength [46]. This causes mobile careers release from deep traps and their partial re-trapping at smaller energy traps. Additional peaks then appear that cannot be seen without additional UV irradiation. By comparing results optical absorption spectrum specific wave length can be selected. [46] This greatly enhances resolution of TL curve and helps see and analyse peaks that cannot be seen due to masking by the black body radiation or for study of peaks that cannot be observed directly. This also can cause negative effect when because this phenomenon additional the light emission is register thus increasing intensity values which are register by measurement system, i.e. TLD reader [31].

1.7 Uncertainties and errors

In radiodiagnostic and radiotherapy the uncertainty is usually associated with the measurement and is often expressed in the terms of both accuracy and precision. The precision of measurements represents the confidence or probability that the measured value is within a certain defined range around the true value, or rather that the true value is within a certain range of the observed value [31][48][49]. While TLD as dosimeters are cheap and use to use they also have number of uncertainties. These uncertainties arise from number of sources. The overall uncertainty of a dosimetric system can be evaluated from the combined effects of the two main types of uncertainty

[50] (Type A, also known as random, and Type B, also known as systematic). This methodology was first recommended in 1981 by the Comité International des Poids et Mesures (CIPM) [53] and included methods for combining the various components of uncertainty [50]. The combined standard uncertainty u_C can be calculated using equation [48]:

$$u_C = \sqrt{u_A^2 + u_B^2} \tag{1}$$

Standard random, type A uncertainty usually can be calculated by statistical methods [50]. This type of uncertainty can be reduced by increasing number of times the dose was measured. If a measurement of a dosimetric quantity x is repeated N times, then the best estimate for x is the arithmetic mean value of all measurements x, i [50]:

$$\bar{x} = \frac{1}{N}\sum_{i=1}^{N} x_i \tag{2}$$

From this standard deviation σ_x, characterizing the average uncertainty of individual result then can be calculated [48]:

$$\sigma_x = \sqrt{\frac{1}{N-1}\sum_{i=1}^{N}(x_i - \bar{x})^2} \tag{3}$$

Type A uncertainty is expressed as $u_A = \sigma_{\bar{x}}$. Standard deviation of mean $\sigma_{\bar{x}}$ then can be calculated [48]:

$$\sigma_{\bar{x}} = \frac{1}{\sqrt{N}}\sigma_x = \sqrt{\frac{1}{N(N-1)}\sum_{i-1}^{N}(x_i - \bar{x})^2} \tag{4}$$

Typical sources of random uncertainties are [51]:

- In homogeneity of detector sensitivity, which depends on the dosimeters production quality.
- Variability of detector readings due to limited sensitivity and background;
- Variability of detector readings at zero dose.

Systematic, type B uncertainties cannot be evaluated be evaluated by statistical means [48]. Evaluation of these uncertainties is based on other knowledge of the measurement system than statistical analysis of measurement data like that from specifications and certificates or experience. There are few different sources that are usually considered to cause systematic uncertainties. B type uncertainty u_B is obtained by evaluating of separate uncertainties $u_{B,i}$ for each individual uncertainty *i* [50]:

$$u_B = \sqrt{\sum u_{B,j}^2} \tag{5}$$

In IAEA's safety standard RS-G-1.3 it is assumed that type B uncertainties can be represented by rectangular probability density distribution [52]:

$$u_{B,i=\frac{a_i}{\sqrt{3}}} \tag{6}$$

where a_i is half- range value (*HRV*) of that parameter i. It is calculated by equation [52]:

$$HRV = \frac{\max(i) - \min(\mathrm{i})}{2} \tag{7}$$

TLD dependence on radiation beam energy (photon energy dependence) is most important factor influencing measurement results. TLDs are highly sensitive to the level of energy to which they are exposed. This means that if, for example, dosimeters were calibrated at one energy these dosimeters should be used at the same energy. In practice this adherence to single energy is difficult, since in practical applications both patient sizes and conditions may vary, thus requiring different energies [52].

Directional dependence with the angle of incidence. Dosimeters usually exhibit this type of dependence because of their physical size and the energy of incident radiation [51].

Non- linear dose response in TLD. Ideally, the dosimeter reading should be linearly proportional to the dosimetric quantity. [52] However, at a certain dose range a non-linearity can sets in. This particularly a problem at low and high dose readings. A dosimeter and its reader may both exhibit non-linear characteristics, which then need to be corrected. In modern TL dosimetry non-linearity uncertainty value is within 5% for doses that are between 10 μGy to 1 Gy [51].

TLD Signal fading. This is where dose information that is stored on the TLD in the form of electrons trapped at higher energy levels starts to decrease even before reading of TLD. Level of fading depends of type of TLD, the type of exposure and type of annealing process. Trap dept and temperature of glow peak is main factors that govern this process. Glow peaks at temperature range from 170 to 200 ^{o}C exhibit lowest fading [49] and detectors with the glow curves at higher temperatures are observed to fade faster. Some materials can also exhibit anomalous fading, during which high temperature peaks can start to fade at much lower temperatures. [51] Additional, unwanted fading can occur due to exposure to sunlight, room light or the UV. Usually, level of fading can vary in between 1% year to 7% in first week after exposure [49][51].

Validity of calibration. Even if dosimeters are irradiated with the same uniform dose at the same conditions, their sensitivity (efficiency) may vary [52]. The thermoluminescence efficiency can be expressed as emitted light per unit of absorbed dose. The variance in the sensitivity of a

thermoluminescence dosimeters is unavoidable but can be reduced from 10-15% when dosimeters are not calibrated to 1-2% when they are calibrated [49]. Because of this, calibration is critical. Even then, this is far from perfect, since for calibration known dose is needed. This dose is usually measured by measurement unit, which is also prone to error and need to be calibrated to some primary or secondary source [52].

Variations in TLD and reader performance. As mentioned before individual TLD efficiency vary one from another [48] [52]. Calibration factor is used to eliminate this uncertainty, but it is calculated by using mean values, which is not perfect solution [38]. Also, if we use other TLD reader on the same dosimeter we can get different results. Reader performance is dependent on its age, the rigor, recency of calibration, model type and technical support [51].

Other factors that affect type B uncertainty include effects due to exposure to types of ionizing radiation that are not intended to be measured by the dosimeter, i.e. electrons or neutrons when measuring x-ray, effects from mechanical shock and variation in local natural background [51] [52]. When dosimetric measurements are done in controlled environment, such as during medical exposures, these uncertainties can be considered to be equal to zero and does not have significant impact to the final uncertainty. [52]

Even though every factor, that was mentioned before, affects combined uncertainty, in practice, the uncertainties caused by the energy and angular dependence of the response of the dosimeter receive more attention than any other source of error [38]. This happens because the effects from all other uncertainty components are assumed to be much smaller. Therefore, type B uncertainty due to the energy and angular dependence, characterized by the resultant standard deviation $u_{B(E,a)}$, and the uncertainties due to all other Type B uncertainties, characterized by the resultant standard deviation $u_{B(0)}$, are separated in two different values. This can be expressed by equation [52]:

$$u_B = \sqrt{u_{B(E,a)}^2 + u_{B(0)}^2} \tag{8}$$

Then expanded uncertainty is expressed [52]:

$$1{,}96U_c \leq 0{,}5 \times (0{,}33 + 0{,}50) \tag{9}$$

According to the ICRP recommendations overall uncertainty of dosimetric system should not differ by more than: - 33% or + 50% (at the 95% confidence level) from the dose equivalents that would be indicated by other type, calibrated dosimeter [52]. This means that uncertainty is significant

and measured changes in dose should significantly higher than possible errors if they are not to be obscured by uncertainties.

1.8 Summary of theoretical overview

In the medical practice, especially in external beam therapy evaluation of patient's radiation exposure has very important role. Dosimetric measurements not only helps to confirm that dose is delivered correctly to the critical volume but helps to prevent accidents and optimize procedures for use in future applications.

Due specifics of both exposed object (patient body) and patient's response to the dose, from many different dosimetric methods passive dosimetry, such as film or luminescent dosimetry is preferred. While films usually have higher sensitivity, they can be used only for 2D dose measurements. TLDs and other types of luminescent dosimeters can be used for point dose measurements, who then can pe used for volume dose evaluation.

One of negative points of TL materials is high error and uncertainty values. Errors appear mainly due to incorrect calibration and incorrect use of materials. On the other hand, uncertainties are influenced by many different factors and can be evaluated only by combination of both statistical and experimental means. For personal dosimetry, where use of TLDs is much more common these problems are eliminated by detailed error and uncertainty evaluation methodologies.

MATERIAL AND METHODS

2.1 Equipment.

2.1.1 Gulmay medical D3225 orthovoltage x-ray treatment unit

Samples were exposed with the help of Gulmay medical D3225 orthovoltage x-ray treatment unit (Gulmay Medical Ltd., UK). This unit consists of a tube, the high voltage generator, the control console, and the cooling system. Energy conversion takes place within the x-ray tube, which is a main component of the system. X-ray tube is capable of delivering x-rays at voltages from 20 to 225 kV with an inherent filtration of 0.8 mm 17 beryllium [40]. The x-ray tube contains two principal elements: cathode, which provides a source of electron and wolfram anode, which is angled 20º relative to the beam axis and acts as the target for electrons and releases x-rays. Additional components include [55] [56]:

- Expansion bellows (provide space for oil to expand);
- Tube envelope (evacuated);
- Tube housing;
- Cooling dielectric oil;
- Rotor;
- Induction stator;
- Tube window.

The cathode and anode are contained in the envelope, which provides vacuum, support and electrical insulation. In this x-ray treatment unit, the envelope is formed from ceramic.

Fig 5. Gulmay medical D3225 control console

The energy used for this process is provided from the generator, connected by an electrical circuit connected to the system. Quantity (exposure) and quality (spectrum) of the x-radiation

produced can be controlled by adjusting three principle electrical quantities, that are applied to the tube [56][57]:

- The voltage or electrical potential applied to the tube, kVp (peak voltage). This factor controls quality of x-ray beam. If the kVp is increased, then the kinetic energy of the electron at the point when it starts to interact with the target will be increased. By increasing the maximum photon energy will also increase the average photon energy. High-energy x-ray photons have a greater probability of penetrating matter.
- The electrical current that flows through the tube or mA controls quantity of beam. As the mA setting is increased, more power is applied to the filament, which heats up and releases more electrons. Increasing mAs results more photons.
- Duration of the exposure or exposure time or s also affect quantity of generated photons. So, both mA and s are included in one parameter mAs.

By adjusted these parameters and by adding specific filtration and collimation x-ray beam can beam adjusted to the needs of patient.

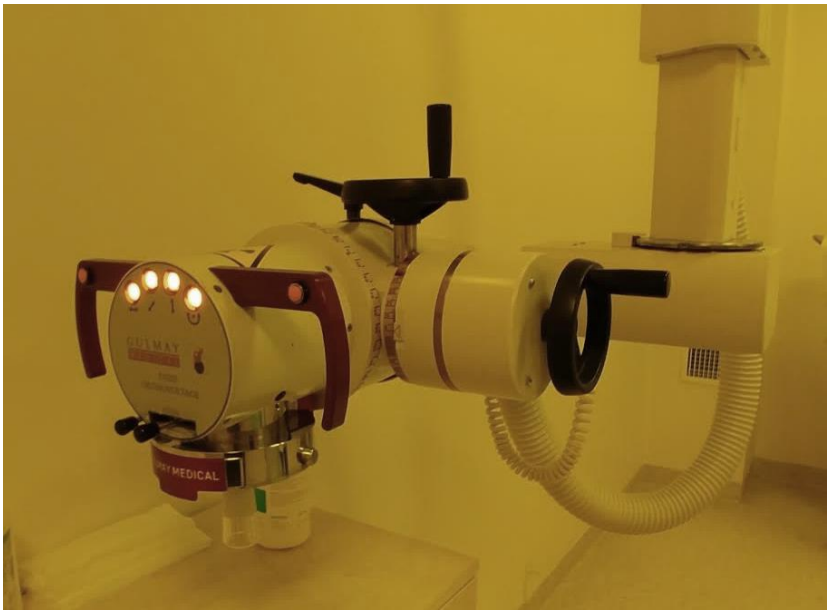

Fig 6. Gulmay medical D3225 orthovoltage x-ray treatment unit

The unit has several applicator cones (with different length and openings) that define treatment distances and field sizes. These open applicators constructed of steel and copper with an end frame of clear PMMA defining the treatment aperture. The distances from the focal spot to the centre of the surface defined by the end of the applicators were within the manufacturer's specification, i.e. within 0.5 mm in all cases. The machine is equipped with a single transmission chamber controlling the

beam output. The kV/filter combinations and the focus surface distance (FSD) of the applicators used in the present study are presented in Table 2.

Table 2. Filter/Tube potential combinations of the Gulmay D3225 unit [56].

Tube potential (kV)	**Filter**	**Reference field size**	**Focus-surface distance**
30	0,8 mm Al	3 cm diameter	20 cm
80	2 mm Al	3cm diameter	20 cm
120	2 mm Al	10 x 10 cm^2 square	50 cm
200	0,5 mm Cu	10 x 10 cm^2 square	50 cm

2.1.2 Ionization chamber PTW 23342-1720

Calibrated ionization chamber PTW 23342-1720 was used as reference system for both calibration of TLD and result's comparison [58]. This soft x-ray ionization chamber is standard detector for measurements in skin therapy. During the measurements chamber was placed in the depth of solid, PMMA phantom (T2962) [45]. This type of chamber has very high energy dependence in the voltage range from 10 kV to 100 kV [59].

Fig 7. Ionization chamber PTW 23342 and phantom T2962 system used as reference system for measurements.

When measuring, PTW 23342 shows charge produced in ionization chamber due to irradiation in the form of charge (nC). Due to this, mathematical operation needs to be performed for correct charge to dose conversion [58]. Since values of external factor such as temperature and atmospheric pressure are not constant and can change during the chamber's calibration process and during the measurements correction coefficient needs to be calculated. This coefficient is known as temperature-pressure correction coefficient and can be calculated by equation [59]:

$$k_{tp} = \left(\frac{273,2+t_m}{T_c}\right) \times \left(\frac{P_c}{P_m}\right) \quad (9)$$

Where t_m is measured temperature (in ^{o}C), T_c is temperature measured during chambers calibration (in K), P_c is atmospheric pressure measured during calibration and P_m is measured atmospheric pressure. Both T_c and P_c values can be found in chamber's calibration certificate (in our case T_c =293,2 K and P_c=1013,2 hPa).

When coefficient is known corrected value of measured charge is calculated [59]:

$$q_{tp} = q_m \times k_{tp} \quad (10)$$

Then corrected charge to dose conversion operation is performed by following equation [59]:

$$D = q_{tp} \times N_K \quad (11)$$

In this equation N_K represents detector calibration factor which is found chambers calibration certificate (N_K=1,111*10^9 Gy/C)

2.1.3 Equipment. Rialto TLD reader

For dosimetric measurements Rialto TLD reader was used. TLD reading process is based on heating of material from ambient temperature up to 300-400 ^{O}C [31]. Then emitted light is collected and measured quantitively. Rialto system consists of reader unit, monitor and keyboard. For dosimeter cooling process nitrogen is used. Usage of nitrogen also to helps to reduce the signal interference produced from impurities in the air and reduce oxidation processes.

Fig 8. TLD reader Rialto system

During the measurement dosimeter needs to be placed on a tray, which goes inside the chamber. Inside the reader unit two measurement chambers are located. Inside each of them separate heating (heating coil is usually used as heating element, because it can assure good contact with the tray and the dosimeter and measurement systems (measurement systems consist of light collection and detection system) are used [60][61]. Due to this two different dosimeters can be read at same time. In order to assure reduced errors and uncertainties inside the chamber temperature is kept constant. Thermocouple is used for temperature measurements during heating cycle. [60]

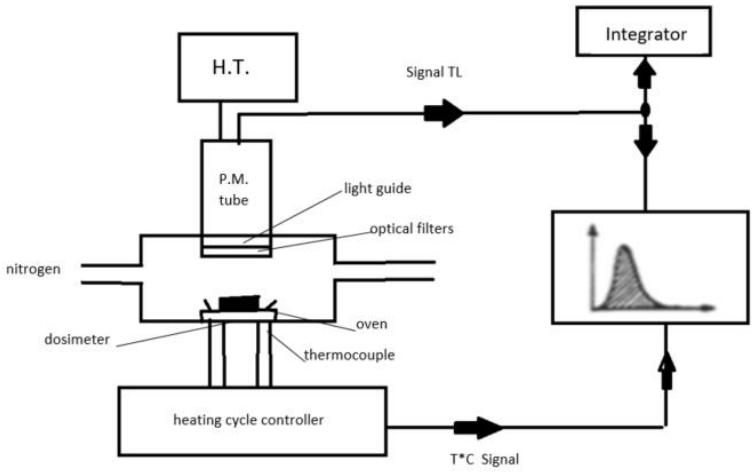

Fig 9. Structure of TLD reader [62].

There are two different heating cycles that can be used during the reading: multiple stage and continues heating cycle. As mentioned before different materials have different glow curves with specific glow peaks and can sometimes more than of them. In order to identify these different peaks multiple stage heating cycle is used more commonly. Also, usage of multiple stage heating helps to

eliminate less stable low temperature peaks and obtain results with the lower probability of error. Multiple stage heating has 4 main stages:

1. *Pre-heating stage*. During this stage dosimeters are heated to low temperatures of about 100 -160 ^{o}C [60].

   It removes low temperature, unstable peaks, that do not give us any useful dosimetric information.
2. *Reading stage.* During this stage dosimeters are heated to temperature of 300 ^{o}C [60] (exact temperature can differ and is dependent on specific TLD which is used during the measurements.) When mentioned temperature is reached trapped electrons are freed and are measured by measurement system.
3. *Annealing stage.* During this stage dosimeter's temperature is increased to 300-400 ^{o}C [60]. When this temperature is reached last trapped electrons are released and dosimeters are restored to the state before irradiation.
4. *Cooling.* Nitrogen is used to cool heated dosimeters to the room temperatures.

After cooling dosimeters can be irradiated and measured again.

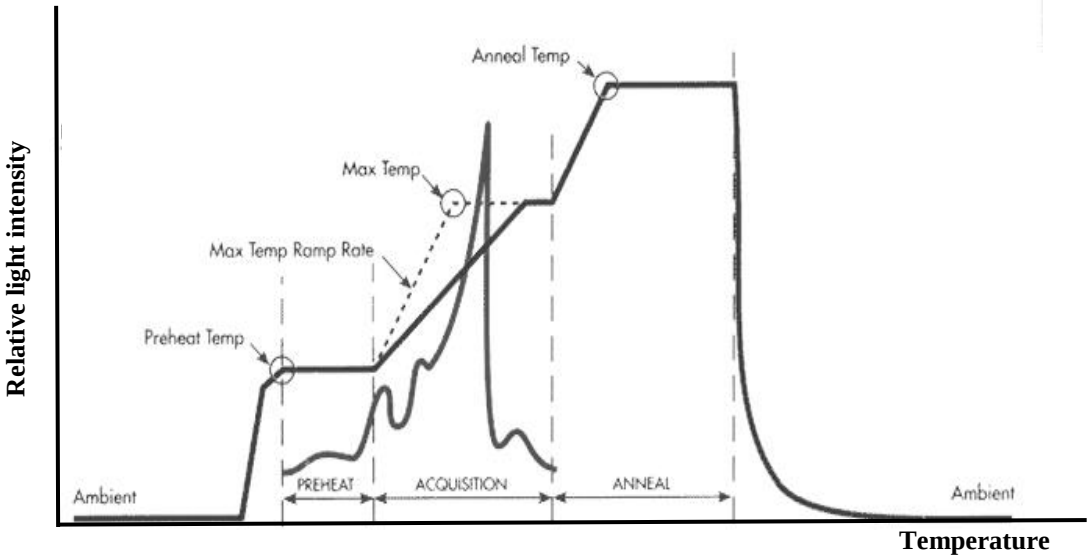

Fig 10. Multi-stage TLD heating cycle [61].

Light, produced due to thermoluminescence, passes through optical filters, then it enters the PMT through the light guide. The incident light is converted into electrical current and via current - frequency converter is output is converted into pulses and counted. Obtained number of counts is proportional to the dose, which was absorbed by dosimeter After the reading obtained signal is then processed and data is recorded by system. Later it can be accessed via user interface [60] [61].

2.2 Measurements

During this work two packets of TLDs were irradiated by X-ray therapy unit Gulmay medical D3225. Each one of the packets contained 40 TLD dosimeters.

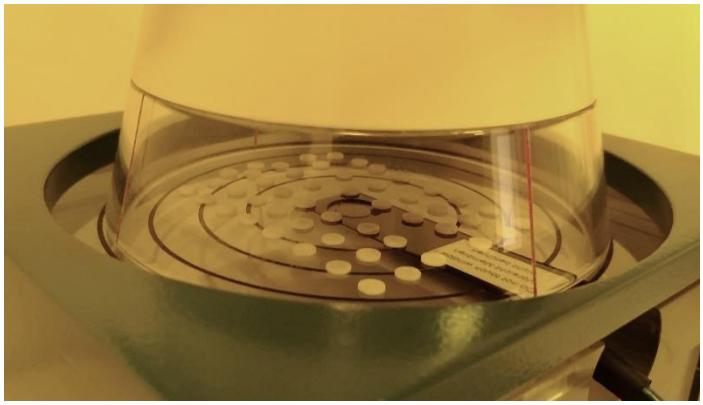

Fig 11. TLD placement during calibration.

Measurements were performed in this order:

1. During calibration dosimeters were placed on top of T2962 phantom in order to reduce differences between STD (source – target distance) of TLD's and ionization chamber, with the difference being only the width of TLDs (about 1-1,5 mm). These dosimeters were exposed to one exposition 2 Gy.
2. The same measurement was performed for both packets of dosimeters, without changing any of parameters. For the first packet measurements were repeated 2 more time and for second packet one more time. Parameters that were used for both measurements calculation of coefficients are given in the table below (table 3, No. 1).

Table 3. Parameters used for measurements

No.	Dose, MU	Dose, Gy	Voltage. kV	Electric current, mA	Filter type	Applicator type and diameter
1	165	2	120	20	Filter 5, 5,39 mm Al (0,22 mm Cu)	Applicator D, Diameter 10 cm
2	461	4	80	20	Filter 3, 2,44 mm Al	Applicator A, Diameter 2 cm

3. Then both correction and calculated and dose values were compared to the obtained reference values and uncertainties were calculated.
4. For RCF calculation measurement was performed for 5 additional dosimeters. They were exposed to 3 exposures whit each being 1 Gy (83 MU (Monito units)) per fraction.

5. After calculations were done additional measurements were performed, with other parameters (table 3, No. 2) in order to evaluate correctness of calculation:

a. 9 dosimeters were placed inside the PMMA phantom, simulating human head volume (fig.12). Dosimeters were place in three different volume point (three TLDs at each point.)

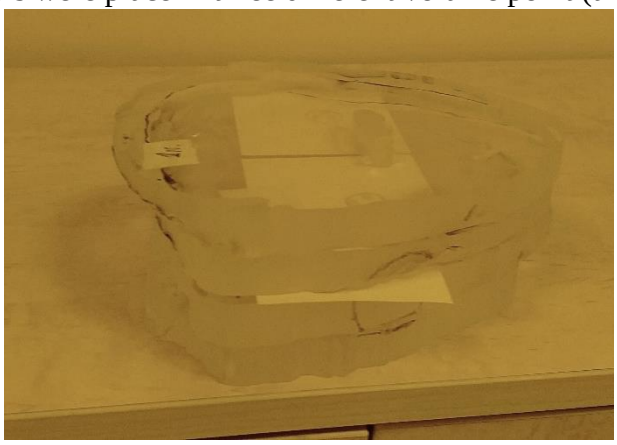

Fig 12. Human head volume PMMA phantom used for measurement 5a. Applicator cone was place on circle, which is seen in the picture.

b. 6 dosimeters were placed in pairs of two on top of phantom, then lead shielding was placed on top of one of pairs and additional pair was placed on top of shielding. Finally, applicator was placed in a such way that one of pair would be outside off field (as seen fig 13.)

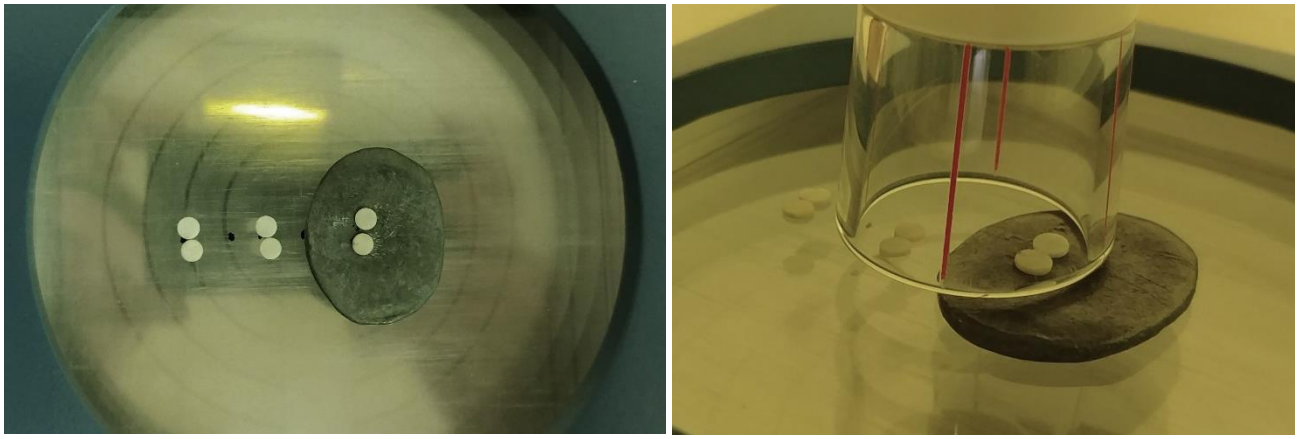

Fig 13. TLD placement during measurement 5b. a) from top; b) from side.

2.3 Calculations

As mentioned before, pulses that are registered and counted by reader are linearly proportional to dose, which was absorbed by TLD. In order to convert these obtained pulses (counts) to measurement dose additional calculation is needed. During this process few correction factors needs to be calculated. Dose value can be expressed by following algorithm [48]:

$$D = \frac{(N - Bgr) \times K_{con}}{RCF \times ECC_j} \quad (13)$$

Where D is dose (in Gy), N is number of counted pulses (counts), *Bgr* is background of individual dosimeter; ECC_j is individual correction factor for dosimeter j, K_{con} is pulse to dose conversion coefficient (counts/Gy) and *RCF* is reader calibration factor [49].

Following coefficient then must be calculated in following order:

1. Background of individual dosimeter (*Bgr*) is calculated;
2. *RFC* (reader calibration coefficient) is calculated;
3. *ECC* for all dosimeters are calculated;
4. Pulse to dose conversion coefficient is calculates

2.3.1 Background of individual dosimeter (Bgr)

This coefficient is calculated every time when TDL are read and annealed according to reading procedure. If a dosimeter is read after exposure additional heating is not needed. If number of counted pulses exceed 10000 dosimeters cannot be used for measurements [49]. When dose exceed 0,02 Gy and counts value does not exceed 500 this coefficient can be ignored due it's miniscule effect (less the 0,01%) to the final dose value. If this is the case *Bgr* is equal to 1. Since dose that we were using was equal or greater to 1 Gy and time between exposures was less than a day this coefficient was not calculated.

2.3.2 Reader calibration factor (RCF)

After calculation of we still need to express reader effect on the system, this value is expressed as Reader Calibration factor (RCF) [63]:

$$RCF = \frac{<Q>}{L} \tag{14}$$

In this formula L is a radiation quantity expressed in generic units (gU). Calibration dosimeters are selected automatically by the software and is calculated automatically during calibration procedure and requires knowing internal parameters of the reader [63].

If these parameters are not known, dose can be calculated from dose response curve. For this three different point of dose are selected [62]. For every point at least 5 different dosimeters are selected and irradiated. Then average values are obtained, and background values are removed. By dividing each value from each dose coefficients then are obtained [62]:

$$RCF_1 = \frac{<Q1>}{L_1}, \qquad RCF_2 = \frac{<Q2>}{L_2}, \qquad RCF_3 = \frac{<Q3>}{L_3} \tag{15}$$

This can be true only when following condition is met [62][64]:

$$RCF_1 \cong RCF_2 \cong RCF_3 \tag{16}$$

2.3.3 Individual element correction coefficient (ECC_j)

The Element correction coefficient (ECC) is factor which relates the specific TLD efficiency and the average efficiency of all TLDs (TLE). It is given by formula [62]:

$$ECC_j = \frac{<TLE>}{TLE_j} \tag{17}$$

In this formula:

ECCj – the element correction coefficient (ECC) of dosimeter j, <TLE> -average TLE of all TLDs and TLEj – TLE of dosimeter j. For <TLE> calculation a small subset of all the dosimeters is used. These dosimeters are called Calibration dosimeters (CD). The average value of all the CDs is compared with the efficiency of each of field dosimeters (FD), to calculate the ECC for each one individually.

In practice, quantity that is measured by the reader is the charge which is produced during the TL process. Since dose response is related to produced charges, the ECC can be expressed as [65]:

$$ECC_j = \frac{<Q>}{Q_j} \tag{18}$$

Where <Q> is the average measured charge of all dosimeters and Q_j - the charge measured from dosimeter j. If new dosimeters are added, the ECCs needs to be re-evaluated with the aim of having similar efficiency with the existing ones. In order to achieve this, the sensitivity of the Calibration dosimeters must remain constant.

2.3.4 Pulse to dose conversion coefficient

This coefficient helps to convert pulse value to the absorbed dose by comparing it to the reference value which was measured by calibrated dosimeter (in our case PTW ionization chamber). It calculated by two separated calculated by two separated methods [64]:

1. Five different dosimeters with ECC_j values ≈1 are selected. Then average value of counts is divided by dose measured with reference dosimeter:

$$K_{con} = \frac{\left(\frac{\sum_{i=1}^{5} Q_i}{5}\right)}{D_{ref}} \tag{19}$$

2. Average count value of all dosimeters is divided by reference dose:

$$K_{con} = \frac{<Q>}{D_{ref}} \tag{20}$$

From these two methods, the first one is preferred because it guaranties lower uncertainty and eliminates TLD with lower or higher sensitivity.

2.4 Calculation of uncertainty

If we assume that all factors who affects uncertainty do not corelate to each other combined type B uncertainty of system can be calculated by following equation [66]:

$$u_B(D) = \sqrt{u_B(E,a)^2 + \sqrt{{u_{Rialto}}^2 + {u_{hol}}^2 + {u_{lin}}^2 + {u_f}^2}^2} \tag{21}$$

In this equation following coefficients are described in literature [51]:

Reader uncertainty u_{rialto}:

Reader uncertainty is given by the manufacturer and it cannot exceed 5 % [42]. Maximum value was chosen for calculation.

Uncertainty in the holder correction u_{hol}

This value has been earlier estimated and evaluated (uncertainty is equal 0.30 %) by scientists [54].

The uncertainty in the dose response non-linearity correction u_{lin}

The uncertainty in the dose response non-linearity was estimated during IAEA study and for doses in range of 2-2.5 Gy is about 5 % [67].

The uncertainty due to fading u_f

TLD's are known to lose part some of absorbed dose in long periods of time. For TLD100 this loss is <10% per year. Measurements were completed during the same day. Uncertainty due to fading was estimated to be >1%. For calculations 1% was selected [27].

Uncertainty due to angular dependence $u_B(E,a)$

Dosimeter's angular dependence can vary from 5 to 10 %, as it is found in literature. During calculations largest given value was of 10 % selected [27].

Since energy dependence of TLD100 is~1 uncertainty should be lower than 10% In the literature uncertainty value ~8 % was suggested [32]. From all of this $U_{B(E,a)}$ was calculated and is equal to 12,81 %

Type A uncertainty u_A was calculated (eq.4) for every measurement separately and then combined uncertainty was calculated (eq.1)

RESULTS

3.1 Ionization chamber readings

In order to callibrate TLD's correctly (during measurement 1) both dosimeters and ionization chamber were exposed to the same X-ray beam. Since we were using 2 diferent packets of TLD's measurement was repeated for the second packet of TLD's as well. Corection coeficient k_{tp} was calculated and dose calculated (see table 4.) using formula 8.

Table 4. Dose measured with ionization chamber.

	Measured charge, C	**Calibration factor, Gy/C**	**Pressure, kPa**	**Temperature, ºC**	**k_{tp}**	**Dose, Gy**
T_1,P_1	$1{,}66\times10^{9}$	$1{,}11\times10^{9}$	100,55	21,8	1,0138	1,870
T_2,P_2			100,54	21,2	1,0118	1,866
Avg			100,55	21,5	1,0128	1,867

Both these values were compared beetween each other and average dose was calculated. Since the difference of average value and both measured doses was smaller then 0,1 %, average dose was selected for calibration of both TLD packets.

3.2 Dose response curve linearity

Even though reader's calibration factor (RCF) is calculated by the reader automaticaly additional measurement were performed in order to obtain TLD response curve in 2-4 Gy dose range. For this 5 different dosimeters were exposed to the doses of 2 Gy,3 Gy and 4 Gy and raw intensity of these dosimeters were measured (as seen in table 5.)

Table 5. TLD dose response values.

Applied dose	**2 Gy**	**3 Gy**	**4 Gy**
Measured intensity values (Raw counts)	3564001	5207895	7221370
	3432412	5918970	8579477
	3672631	5624266	6622571
	3559081	5824764	7071847
	3461923	5400283	7673816
Average value	3538010	5595236	7433816
Average value adjusted per dose	1842713	1858454	1865079

From these 3 average intensity values dose response curve was ploted. When measuring background values count were always lower than 400. Since background was less then 0,03 % of average values no substraction of background was performed.

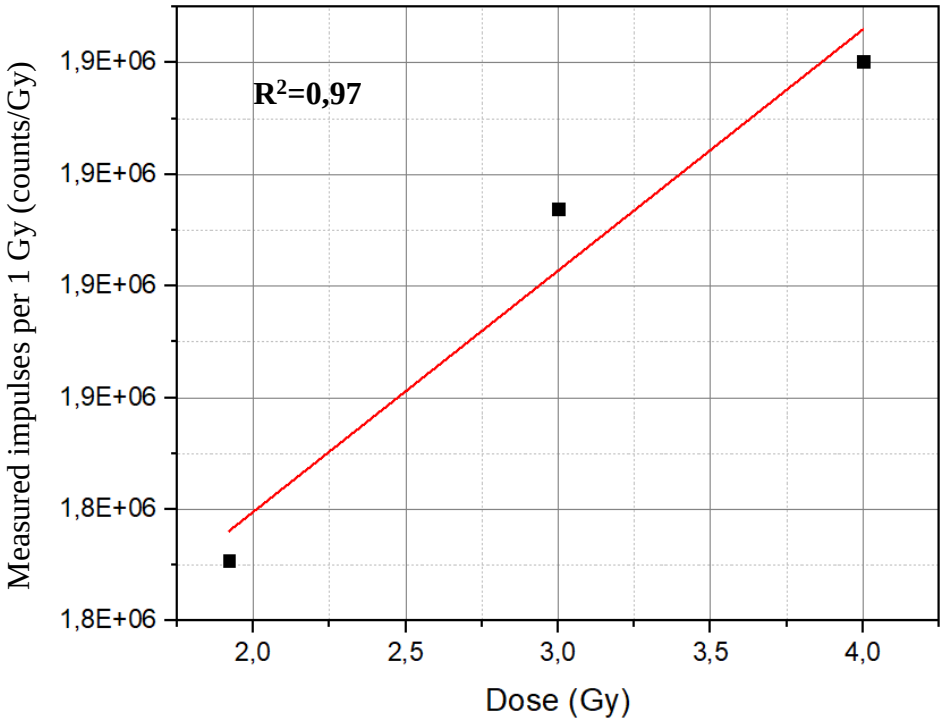

Fig 14. Obtained dose response values and fitted response.

From dose response intensity graph we can see that dose response is linear, as it should be accordind to the litherature. Obtained dose response curve was used instead and element correction coefficients were calculated using raw count values instead of corrected ones. Obtained values ranged from 0,75 to 1,27 for both packets of TLD's.

3.3 Calculation of ECC

Since dose to which TLDs were exposed is in non-linear respone intensity values corrected by the TLD reader were used. TLDs were exposed to 1,87 Gy dose with the parameters seen in table x. ECC. Values were calculated and the 5 dosimeters with highest and lowest ECC values were removed, this helped eliminate that can appear due to flawed dosimeters. Then again ECC for dosimeter were calculated. Results can be seen in table 65.

Table 6. Packet 1 of TLD's calibration coficient values

Dosimeter	Raw	Corrected	ECC	Dosimeter	Raw	Corrected	ECC
1	4680130	23690	1,0675	**29**	3730337	23680	1,0295
3	3183071	27190	0,7812	**30**	2767913	17320	1,1003
8	3545410	22030	0,9836	**31**	4616557	18930	1,1885
10	3123713	20410	0,9602	**32**	3184953	19930	1,0630
11	2878303	23290	0,8844	**33**	2568687	16300	1,2861
12	3097701	24250	0,7922	**34**	3236966	20250	0,9945
14	4584989	22640	1,0146	**35**	2864665	28180	0,7485
15	1747016	21050	1,0976	**37**	2699012	17130	1,1133
16	4068587	19410	1,2180	**38**	4662062	29170	0,7989
17	4207100	26600	0,7540	**39**	2841522	18040	1,0585
19	4205064	26590	0,7820	**40**	2579083	17280	1,1621
20	2637522	23470	0,8255	**41**	2305688	19630	1,0346
21	4022446	25530	1,0002	**42**	4316430	23250	1,0394
22	1310474	18200	1,0385	**43**	2680090	17010	1,2659
23	2556550	18230	1,0777	**45**	3024434	19200	1,3009
25	2539897	26120	0,7490	**46**	2609502	19470	0,9897
26	3040599	19030	1,0725	**47**	2640857	16760	1,2050
27	3102227	21690	0,9014	**48**	2915945	18410	1,1935
28	2844003	17800	1,1645	**50**	3430886	20660	1,2056

Calibration coeficients then were ploted as indexed intensity graph. From this plot we can see that calibration coeficients range from 0,7 to 1,3. These coeficients then were compared to the ones that were calculated using raw values. Difference beetwen these two calculated values ranges from -1,07% to 0,95 %, thus resulting uncertainty in range from 0,26% to 0,62%.

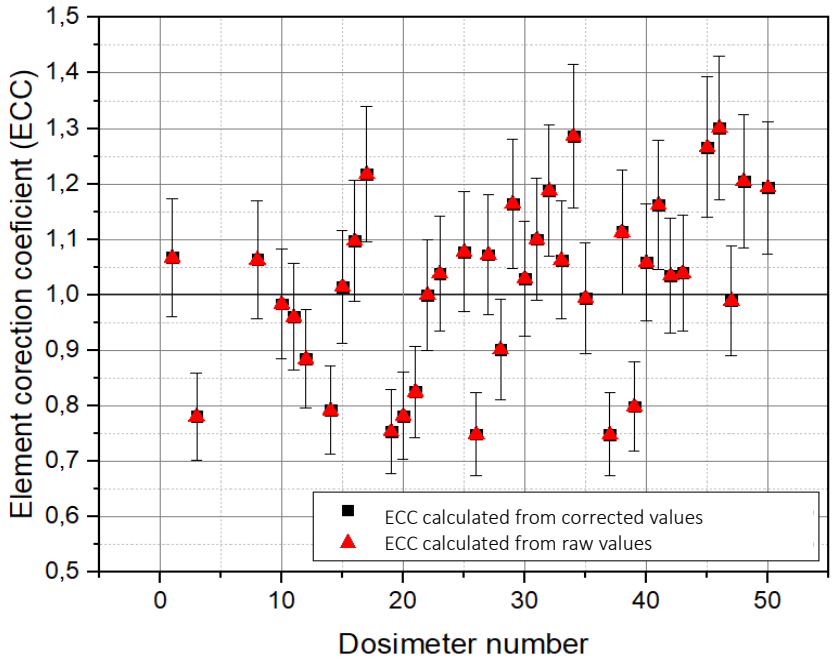

Fig 15. Calibration coeficients calculated for 1st set of TLDs.

Completely indentical callibration procedure was for TLD packet 2. From this ECC values were calculated as it is given in table 7.

Table 7. Packet 2 of TLD's calibration coficient values

Dosimeter	Raw	Corrected	ECC	Dosimeter	Raw	Corrected	ECC
1	2646681	16770	1,1973	**25**	4281073	27150	0,7396
2	3642743	22880	0,8776	**26**	4173383	26080	0,7699
4	2427913	15250	1,3167	**27**	2903343	18410	1,0907
5	2999207	19000	1,0568	**29**	4118170	26130	0,7684
6	2811452	17660	1,1370	**30**	3440925	21500	0,9339
7	2351945	14900	1,3476	**31**	2452847	15550	1,2913
8	3624787	22770	0,8818	**32**	4135156	25840	0,7771
9	3083241	19550	1,0271	**33**	4299374	27250	0,7368
10	2467972	15420	1,3021	**34**	2371000	14820	1,3549
11	4234858	26860	0,7475	**36**	2396279	15070	1,3324
12	2604787	16280	1,2333	**37**	2523606	15960	1,2581
14	2753364	17200	1,1674	**38**	2656661	16710	1,2016
16	4335391	27090	0,7412	**40**	3018709	18980	1,0579
17	2980658	18900	1,0624	**41**	4104669	25970	0,7732
19	2856135	18110	1,1087	**42**	4278742	26910	0,7462
20	3074431	19210	1,0452	**44**	2613823	16440	1,2214
21	2326590	14760	1,3604	**45**	3128694	19790	1,0146
22	2785761	17410	1,1533	**47**	4214525	26660	0,7532
23	2854733	18110	1,1087	**48**	4290962	26980	0,7442
24	2684482	16770	1,1973	**49**	2517176	16060	1,2502

Also calibration coeficient using Raw values were calculated. Then calibration coefficient – dosimeter index graph was ploted:

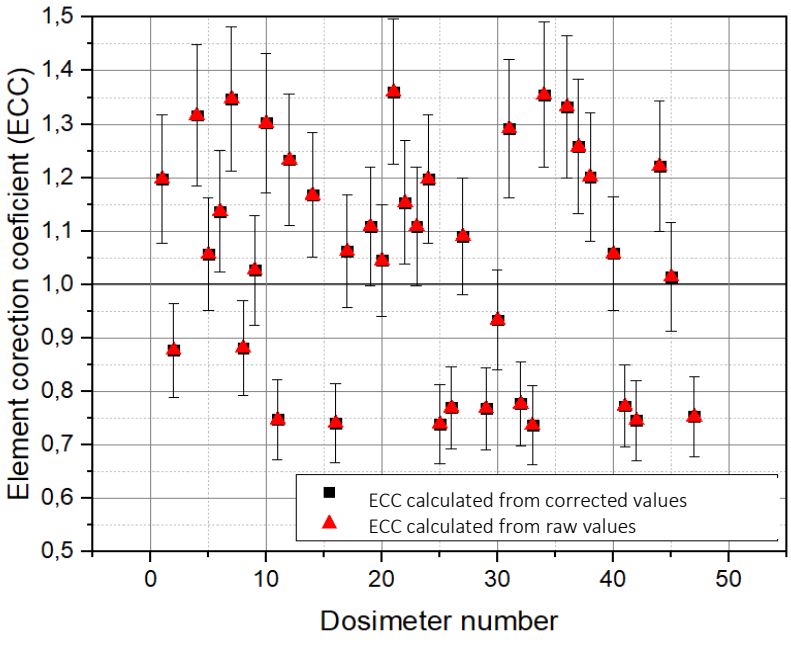

Fig 16. Calibration coeficients calculated for 2nd set of TLDs.

From plot we can see that callibration coeficient values ranges from 0,7 to 1,4. Values were compare beetween each other. Difference was in range from -0,85% to 1,25%, thus resultin uncertainties in range from 0,11% to 0,72%.

From calculation of ECC's and corection coeficients for both packets of TLD it was found that there is no significant difference with average difference being ~1%, uncertainties of thes values were also calculated for the use in calculation of combined uncertainty.

3.4 Calculation of dose and uncertainty

Doses of each TLD were calculated, from known reader measurements and TLD's individual calibration coeficients. Then using formulas no. 1, no.7 and known U_B value combined uncertainty was calculated. Results are given in table 8.

Table 8. Doses measured by TLD and calculated uncertainties for set 1.

No	Calculated dose, Gy	Uncertainty	No	Calculated dose, Gy	Uncertainty,	No.	Calculated dose, Gy	Uncertainty,
1	2,167	0,2111	**23**	1,683	0,2056	**39**	1,636	0,2065
3	1,820	0,2044	**25**	1,676	0,2057	**40**	1,720	0,2051
8	1,856	0,2045	**26**	1,749	0,2048	**41**	1,740	0,2049
10	1,679	0,2057	**27**	1,675	0,2058	**42**	2,071	0,2079
11	1,765	0,2047	**28**	1,776	0,2046	**43**	1,845	0,2045
12	1,646	0,2063	**29**	2,089	0,2084	**45**	2,140	0,2101
14	1,968	0,2056	**30**	1,633	0,2066	**46**	1,651	0,2062
15	1,980	0,2058	**31**	1,928	0,2050	**47**	1,730	0,2050
16	2,026	0,2067	**32**	1,815	0,2044	**48**	1,883	0,2047
17	1,718	0,2051	**33**	1,796	0,2045	**50**	2,134	0,2099
19	1,782	0,2046	**34**	1,725	0,2050	**—**	—	—
20	1,660	0,2060	**35**	1,807	0,2045	**—**	—	—
21	2,188	0,2119	**37**	1,634	0,2066	**—**	—	—
22	1,619	0,2069	**38**	1,997	0,2061	**—**	**—**	**—**

From the table we can see that measured dose values range from 1,63 to 2,18 Gy, with the average value being 1,82 Gy. Uncertainty ranges from 20,44 to 21,19 % with average value being 20,61 %. Both fose values an combined uncertainties were ploted in scatter plot.

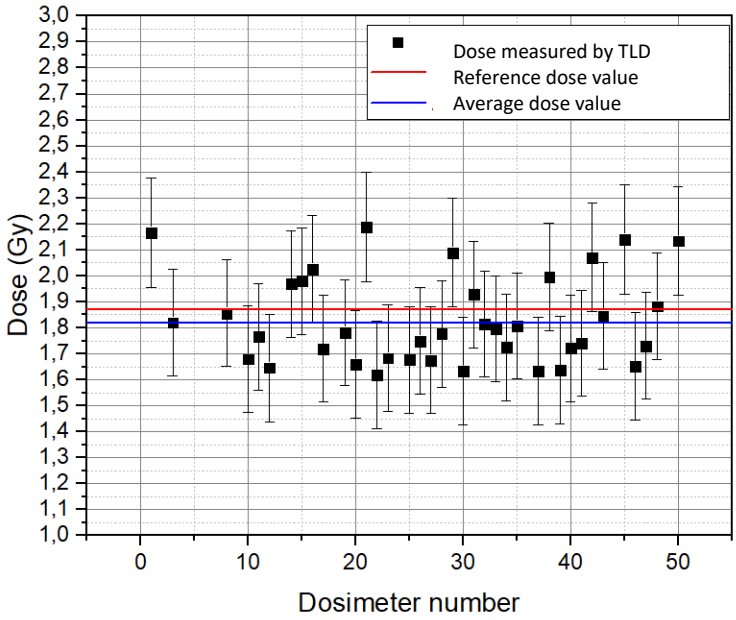

Fig 17. Doses measured during first measurement for first set of TLDs.

Also inside plot lines representing average measured dose value (blue line) and dose value measured with ionization chamber were ploted in order to show measured values are distrinuted around it.

Measurements was performrd for set 2 of TLDs. This measurement was performed in order to evalute both dose and uncertainty.

Table 9. Doses measured by TLD and calculated uncertainties for set 2.

No	Calculated dose, Gy	Uncertainty	No	Calculated dose, Gy	Uncertainty	No.	Calculated dose, Gy	Uncertainty
1	1,916	0,2365	**17**	1,881	0,2364	**33**	2,155	0,2399
2	1,692	0,2382	**19**	1,690	0,2382	**34**	1,731	0,2375
4	1,780	0,2369	**20**	2,081	0,2382	**36**	1,612	0,2399
5	1,757	0,2372	**21**	1,755	0,2372	**37**	1,672	0,2386
6	1,705	0,2379	**22**	1,815	0,2367	**38**	1,715	0,2378
7	2,173	0,2404	**23**	1,767	0,2371	**40**	1,715	0,2378
8	2,152	0,2398	**24**	1,672	0,2386	**41**	1,772	0,2370
9	2,294	0,2444	**25**	1,798	0,2368	**42**	2,203	0,2412
10	2,260	0,2431	**26**	2,108	0,2388	**44**	1,694	0,2381
11	1,833	0,2366	**27**	1,706	0,2379	**45**	1,677	0,2385
12	1,823	0,2366	**29**	1,980	0,2369	**47**	1,856	0,2365
14	1,764	0,2371	**30**	2,244	0,2426	**48**	2,199	0,2411
15	1,745	0,2374	**31**	1,704	0,2380	**49**	1,766	0,2371
16	2,292	0,2442	**32**	1,962	0,2367	—	—	—

From the table we can see that mesured dose values range from 1,67 to 2,29 Gy, with the average value being 1,88 Gy. Uncertainty ranges from 23,64 to 24,42% with average value being 23,84 %. Both fose values an combined uncertainties were ploted in scatter plot.

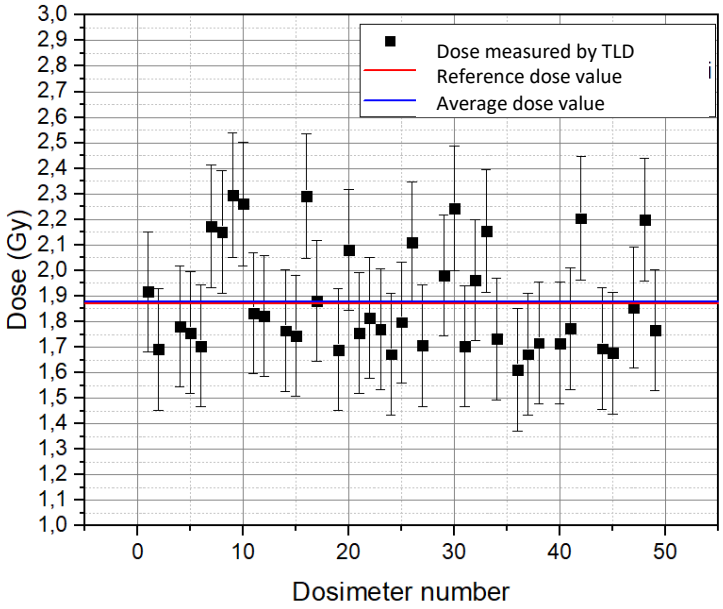

Fig 18. Doses measured durind first measurement for second set of TLDs.

Also inside plot lines representing average measured dose value (blue line) and dose value measured with ionization chamber were ploted in order to show measured values are distrinuted around it.

Second measuremet was performed again for set 1 of TLDs in order to evalute reapeatability of both dose and uncertainty.

Table 10. Doses measured by TLD and calculated uncertainties for set 1.

No	Calculated dose, Gy	Uncertainty	No	Calculated dose, Gy	Uncertainty	No.	Calculated dose, Gy	Uncertainty
1	2,020	0,2408	**21**	2,091	0,2418	**36**	1,638	0,2414
3	1,644	0,2409	**22**	2,107	0,2422	**37**	1,629	0,2416
6	2,116	0,2430	**23**	1,606	0,2421	**38**	2,123	0,2426
8	1,754	0,2394	**25**	1,721	0,2399	**39**	2,032	0,2406
9	2,101	0,2426	**26**	1,637	0,2414	**40**	1,641	0,2413
10	1,636	0,2411	**27**	1,752	0,2396	**41**	1,618	0,2418
11	1,858	0,2391	**28**	1,838	0,2391	**42**	2,260	0,2471
12	1,630	0,2412	**29**	1,794	0,2392	**43**	1,683	0,2405
14	2,184	0,2451	**30**	1,973	0,2397	**45**	1,952	0,2395
15	2,012	0,2406	**31**	2,217	0,2455	**46**	1,707	0,2401
16	1,612	0,2416	**32**	2,127	0,2427	**47**	1,764	0,2395
17	1,885	0,2392	**33**	2,026	0,2405	**48**	1,907	0,2392
19	1,712	0,2398	**34**	1,626	0,2416	**50**	1,825	0,2391
20	1,658	0,2406	**35**	1,645	0,2412	—	—	—

From the table we can see that mesured dose values range from 1,60 to 2,26 Gy, with the average value being 1,85 Gy. Uncertainty ranges from 23,91 to 24,55% with average value being 24,11 %. Both fose values an combined uncertainties were ploted in scatter plot.

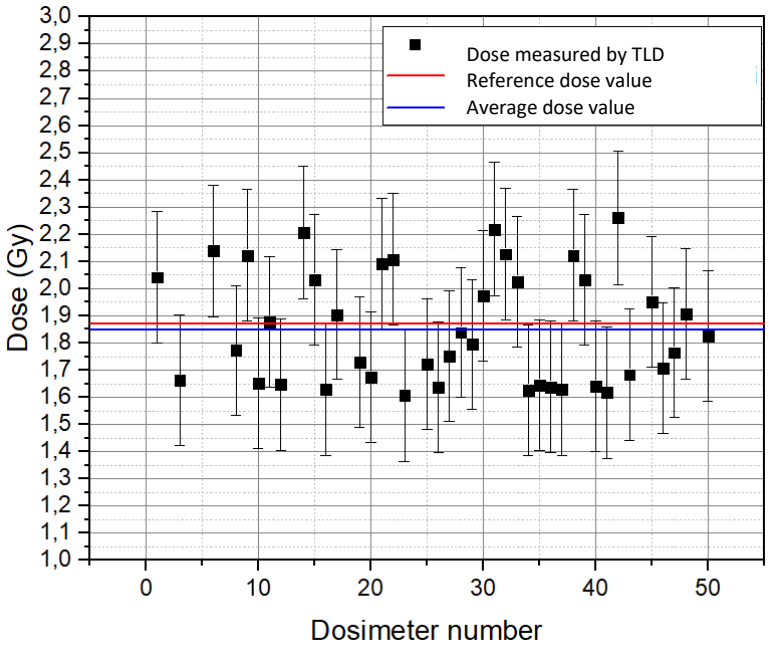

Fig 19. Doses measured durind second measurement for first set of TLDs.

Then results of both first and third measurements were plotted in the same graph in order to compare them and evaluate repeatability. From plotted graph it can be seen that most dose values from the first measurement are in range of uncertainty and the other way around. 5 of 40 measured dose values are outside of calculated uncertainty range of the ones that were calculated during the first and the second measurements.

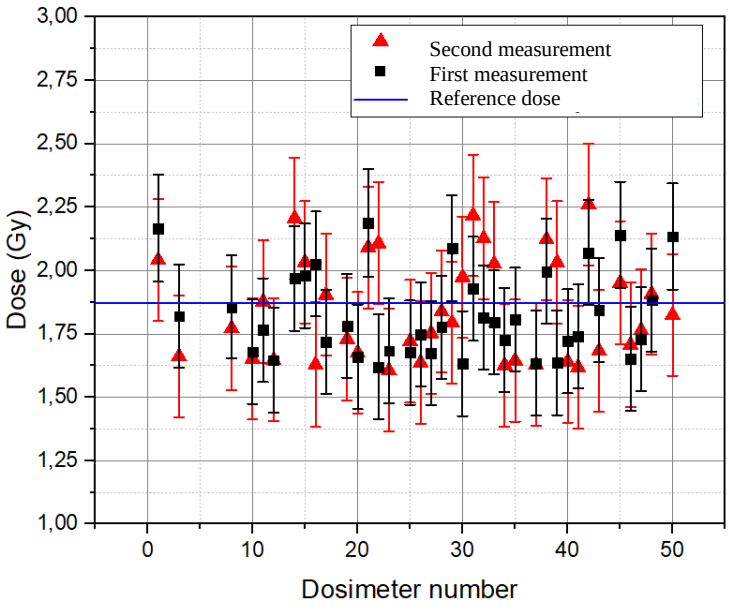

Fig 20. Comparison of doses 1st measured during 1st and 2nd measurement

3.5 Measurement of dose in dept

For this measurement dosimeters were placed inside PMMA phantom of human thyroid volume. Specifically, 9 dosimeters with the calibration coefficient ≈1 were selected. 3 TLDs were placed inside each of phantoms 3 pocket. Doses and uncertainties were calculated. Results are given in table 11.

Table 11. Dose measurements in phantom

Dosimeter position		**Dose, Gy**	**Uncertainty**	**Dose measured with film, Gy**	**Difference between doses, Gy**
Larynx		0,057	0,1229	0,107	0,5
		0,054	0,1229		0,53
		0,052	0,1229		0,55
Parotid glands	Left	2,524	0,1630	0,558	—
		2,555	0,1665		—
		2,322	0,1761		—
	Right	0,087	0,1630	0,84	0,03
		0,076	0,1665		0,08
		0,091	0,091		0,07

From results we can see that dose was highest in the pocket which is nearest to the applicator (average dose 2,46 Gy) and lowest in the pocket which was furthest from it (average dose 0,054 Gy). Also, from calculations we can see that uncertainties were highest at high dose ranging from 16,3 to 17,61% and lowest at low dose ranging from 12,29 to 12,31 %.

After that doses measured with TLDs were compared to the dose values who were measured by gafchromic films. Two points out three were compared: doses in right parotid volume and in larynx volume. Dose difference in right parotid volume were lower than 10%, doses in larynx volume were ~50%. Dose in left parotid volume were not compared due to very big difference in values.

3.6 Measurement of shielding effect on dose

For this measurement 8 TLD's with calibration coefficient ranging from 0,94 to 1,05 were placed on PMMA phantom in 4 different positions: 2 dosimeters were placed bellow piece of 1,5mm Pb, 2 dosimeters were placed on top of Pb, 2 dosimeters were placed on PMMA and aligned to central point of applicator and 2 were placed on PMMA, outside of applicator. After exposure to 4 Gy dose

and with the parameters found in table 3 absorbed doses were measured. Results are given in table 12 below.

Table 12. Absorbed doses measured during measurement 5b.

Dosimeter position	**Dose measured with TLD, Gy**	**Average dose value, Gy**	**Dose measured with gafchromic film, Gy**	**Difference between doses, Gy**	**Difference %**
1	0,117	0.13	0,17	0,04	33
	0,126				
2	3,657	3.803	3,8	0,003	<1
	3,950				
3	0,575	0,55	0,32	0,23	42
	0,527				
4	4,182	4,20	3,8	0,4	10
	4,210				

From results we can see that TLDs that were not shielded and under applicator received doses from 3,65- 4,20 Gy, thus meaning that doses ranged from -8% to 5% of set dose. Shielded TLD received ~0,55 Gy, TLDs area outside applicator received ~ 0,12 Gy due photon scatter. Dose intensity reduced due to shielding was represented as darker shade of colour. All obtained values were compared to the doses measured with gafchromic films. After comparison difference was calculated. It ranged from 1% to 42% of measured doses. Highest difference was calculated when doses were lower than >1 Gy (doses due to scatter). When doses were higher than 1 Gy calculated difference was below 10%.

CONCLUSIONS

1. Results of full calibration using 2 different TLD sets, containing 40 dosimeters in each, were compared between each other and was found that it differs from 0.25% to 0.95%, it means that RCF correction performed by the system was correct and only additional correction was needed for the uncertainty value evaluation of calculation algorithms. It was found that maximum uncertainty value 0.95%. Uncertainty values can be further reduced if proper calibration procedure would be performed for the reader or with the use of modern day equipment, because is RIALTO outdated system.

2. Irradiating TLD the dose simultaneously was measured with ionization chamber. For the first set absorbed doses (using TLD) were measured from 1.63 Gy to 2.18 Gy (average absorbed dose 1.82 Gy) and for second set: from 1.67 Gy to 2.29 Gy (average absorbed dose 1.88 Gy). Evaluating differences between the measured results (measured and average dose: 2.6% (1^{st} set) and 0.5% (2^{nd} set); minimal and measured doses: 12.8 % (1^{st} set) and 10.6% (2^{nd} set); maximal dose and average value: 16.6 % (1^{st} set) and 22.5% (2^{nd} set)); the results did not show difference higher than 25%, it means that measurements were performed correctly.

Also, uncertainties evaluation showed (20.4% ÷ 21.8% (1^{st} set) and 23.4 % ÷ 24.8% (2^{nd} set)), that uncertainty values due to IAEA recommendation (33% ÷ 50%) were not exceeded and average uncertainty of the system was reduced by 10 % and errors – 5%. Due to evaluated uncertainties of the dose measurements repeatability was registered of 88%.

3. Absorbed dose using TLD were measured using a prototype of PMMA slab head phantom. Evaluating absorbed doses for the parotid glands (0.08 Gy for the left and 2.5Gy for the right) and larynx (0.05 Gy) in compare to gafchromic films dosimetry, it was found that absorbed doses in right parotid gland were calculated with an error lower than 10%, for left side parotdi gland absorbed dose difference was registered >50%. This difference could be caused by experimental setup mistakes or inaccurate placement of dosimeters or even due to possible damage of dosimeter.

4. Evaluating effect of lead shielding to absorbed dose using TLDs and gafchromic films, it was observed that the measured dose differed from 1%-42% (directly in radiation field measured doses did not exceed 10% of measured value). Dosimeters placed outside field and under the shielding showed much higher (>10%) dose difference, it could be caused by the effect of backscattered or scattered irradiation from the lead shielding and phantom volume.

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