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Study of dosimetric properties of a thermoluminescent system for dose audit in radiation therapy

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Abstract

For regular monitoring of gamma therapy units against the backdrop of insufficient provision of modern dosimetric equipment and, in some extreme cases, the lack of qualified personnel, external audits across all radiological departments are practically the only way to independently verify the accuracy of calculations of therapeutic dose delivered to patients.

The relevance of this study is justified by the need to comply with the provisions of national legislative documents in the field of the nuclear energy application: The Law of Ukraine "On Human Protection against Impact of Ionizing Radiation", The Norms of Radiation Safety of Ukraine (NRBU-97), and The State Standard of Ukraine "Measurement of Ionizing Radiation. Metrological Support. General Provisions" (DSTU 3240:2015).

The requirements of these documents regarding the accuracy and reliability of dose measurements necessitate the use of high-quality detectors. Hence, among all thermoluminescent materials used for thermoluminescent dosimetric (TLD) audits of therapeutic photon beams, preference is given to a LiF, Mg, Ti phosphor (TLD-100) due to its dosimetric properties: tissue equivalence, long-term retention of dosimetric information (low fading), high sensitivity, a wide range of measurable doses, and excellent reproducibility of results in repeated measurements.

The use of such detectors makes it possible to conduct regular dosimetric audits of operational gamma therapy units using the "postal dosimetry" method. This approach is of paramount importance as it allows for clinical dosimetry errors to be timely detected, enhances the quality of radiation therapy, and consequently, the standards of treatment.

When conducting TLD audits, regular monitoring of dosimetric properties and parameters of both the phosphor itself and the TL measuring instrument shall be ensured, as well as a set of correction factors and the value of the expanded uncertainty shall be determined.

Keywords: radiation safety; radiation therapy; thermoluminescent dosimeters; thermoluminescent powder; uncertainty; γ-dosimetry; ionizing radiation.

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Determination of correction factors and adjustment coefficients

Thermoluminescent dosimetric audit (TLD-audit) is a key tool for external quality control in radiation therapy. This "postal dosimetry" method involves the distribution of standardized TL detectors among radiology departments. A medical institution irradiates the detectors under specified geometry (e.g., in a water phantom at a depth of 5 or 10 cm) with a therapeutic dose (e.g., 2 Gy). The capsules are then returned to the audit centre for readout and analysis. Comparing the absorbed dose measured by the centre with the dose delivered by the clinic allows for systematic dosimetry errors to be verified, the performance of gamma therapy units verified, and the compliance with national safety standards assessed. The relevance of TLD audits in Ukraine is driven by insufficient provision of modern dosimetry equipment and the lack of qualified personnel in many medical institutions, making regular external control practically the only way of independent verification.

The absorbed dose determined using TL detectors irradiated on teletherapy units is influenced by factors such as fading*, non-linearity of thermoluminescent

* During the period between the irradiation of a TL detector and its readout, the magnitude of the TL signal decreases. This partial loss of the energy deposited by ionizing radiation in the phosphor during irradiation is known as the fading.

detector (TLD) response to the dose, energy dependence, and the presence of a holder.

To accurately identify the factors influencing the calibration of a dosimetric system (comprising the PCL-3 thermoluminescent reader and TLD-100 powder), correction coefficients were studied and determined [1]. Specifically, when determining the correction factors, the daily drift of the PCL-3 reader was accounted for based on the readings of control powder. The control powder used was TLD-100, irradiated with an absorbed dose in water of 2 Gy under standard conditions and aged for 5 months to achieve a stable TL signal [2].

The absorbed dose in water was calculated using the formula [2]:

$$D_w = M \cdot N \cdot f_{lin} \cdot f_{en} \cdot f_{hol} \cdot f_{fad}, \quad (1)$$

where D_w is the absorbed dose in water, Gy;

M is the TL-signal value, corrected for daily fluctuations of a TL reader using a correction factor accounting for instrument sensitivity drift, rel. units;

N is a calibration coefficient of a TL-system, Gy/rel.unit;

f_{lin} is a correction factor accounting for the non-linearity of the TL-signal dependence on the irradiation dose, dimensionless;

f_{en} is a correction factor accounting for the dependence of a TL-signal on the energy of ionizing radiation, dimensionless;

f_{hol} is a correction factor accounting for the influence of a standard irradiation holder on the TL-signal value, dimensionless;

f_{fad} is a fading correction factor, dimensionless.

The system calibration coefficient (N) is defined as the inverse of the TL-signal per unit dose and is calculated by the formula [2]:

$$N = \frac{1}{n} \cdot \sum_{i=1}^{n} \frac{D_i}{M_i - M_{background}}, \quad (2)$$

where N *is a* calibration coefficient of a TL-system, Gy/rel.unit;

D_i is the absorbed dose in water (2 Gy) used to irradiate TL detectors, Gy;

$M_{background}$ is a mean value of a background signal from non-irradiated TL detectors, rel.unit;

M_i is the TL signal value of the i-th detector irradiated with a dose of 2 Gy, rel.unit;

n is the number of detectors irradiated with the specified dose.

When determining a calibration coefficient (N) of a TL system, the irradiation of TL capsules with an absorbed dose in water of 2 Gy in a standard water phantom was performed. A standard holder (which is used by the IAEA laboratories to conduct TLD audits) was used for irradiation in the water phantom, allowing only for one capsule to be irradiated per procedure. When irradiating subsequent capsules in this phantom, an uncertainty in the capsule positioning arises.

An alternative solution is to use a PMMA phantom, which does not only allow for 3 TL-capsules to be simultaneously irradiated, reducing the total irradiation time, but also ensures constant geometry conditions for the capsule placement. However, considering that medical institutions exclusively irradiate TL-detectors (capsules) in a water phantom, and that to calibrate a TLD system on the National Primary Measurement Standard of absorbed dose and absorbed dose rate units for X-ray and gamma radiation (the NSC "Institute of Metrology") a PMMA phantom is used, the difference in scattering properties between water and PMMA when calculating doses based on calibration in PMMA shall be accounted for. This prevents discrepancies in determining the absorbed dose which can be avoided by establishing the relation between the TL signal of capsules irradiated in water (M_{water}) and in the PMMA phantom (M_{PMMA}), i.e., the correction factor $F_{PMMA \to water}$, determined by the formula:

$$F_{PMMA \to water} = M_{PMMA} / M_{water}. \quad (3)$$

The study of dosimetric properties of a TLD system was divided into four sequential stages, aiming at the gradual refinement of the methodology and the consideration of all negative critical influencing factors [2]. Stage I involved performing a baseline calibration in a water phantom to determine the initial metrological characteristics of the system. Stage II included re-calibration to assess the system stability and identify the need for regular monitoring. Stage III was a transition to calibration in a PMMA phantom against the State Primary Measurement Standard (the NSC "Institute of Metrology"), thereby reducing the positioning uncertainty and increasing the resolution. Finally, Stage IV implied a comprehensive approach: parallel irradiation in water and PMMA (see Table 1) to directly determine the environmental conversion coefficient $F_{PMMA \to water}$, equal to 1.0042. This approach made it possible to systematically account for technological, methodological, and metrological aspects of preparing the system for audits in medical institutions across Ukraine.

Table 1

Calibration coefficients of the TL-system according to the study stages

Stage	Phantom material	TL-system calibration coefficient ($\times 10^{-5}$)
I	Water	3.5200
II	Water	3.8253
III	PMMA	3.5682
IV	Water	3.5770
	PMMA	3.5619

When calibrating a TL system using a PMMA phantom, the readings of TL detectors shall be cor-

rected using the heterogeneous media correction factor $F_{PMMA\to water}$.

To determine the calibration coefficient of a TL system, TL capsules were irradiated with an absorbed dose of 2 Gy under standard conditions in phantoms made of different materials – PMMA and water [1]. Based on the results of these measurements, the calibration coefficients of the system were determined, their values presented in Table 1.

According to Table 1, the calibration coefficients of a TL system obtained using both the water phantom and the PMMA phantom are similar in value and bear no statistical difference ($p<0.05$), except for the Stage II value. The lowest uncertainty was achieved during calibration in the PMMA phantom against the National Primary Measurement Standard (NSC "Institute of Metrology"). The deviation of the TL system obtained at Stage II is slightly higher than the other values, which highlights the necessity of calibrating TLD systems before each stage of a TLD audit.

Determination of correction factors

The correction factor accounting for the influence of a standard holder for TL-detectors (f_{hol}) compensates for the reduction in the TL signal due to partial attenuation of a gamma radiation beam by the standard plexiglas holder during irradiation of TL detectors. At the IAEA Dosimetry Laboratory, which conducts TLD audits, it has been established that for a depth of 5 cm and ^{60}Co energy, this factor is f_{hol} = 1.008, and at 10 cm depth – f_{hol} = 1.018 [3, 4].

The correction factor for the energy dependence of the TL-signal (f_{en}) is applied when measuring absorbed doses in radiation beams with energy different from ^{60}Co energy since TL-detectors are calibrated at ^{60}Co energy, f_{en}=1.

The correction coefficient for the non-linearity of the TL-signal versus irradiation dose, is calculated using the formula [2]:

$$f_{lin_i} = \frac{M_{Do} D_o}{M_{Di} D_i}, \qquad (4)$$

where M_{Do} is the TL signal value when the TL detector is irradiated in water with dose D_o=2.0, Gy (rel. units);

M_{Di} is the TL signal value when the TL detector is irradiated with absorbed dose in water D_i , Gy (rel. units).

The TL signal non-linearity function was studied against the dose across all stages (I–IV). To determine the non-linearity correction factor, TL detectors were irradiated with 6 doses (0.5, 1.5, 1.75, 2.0, 2.25, 2.5 Gy), using 5 capsules per dose. The parameters of the non-linearity correction factor function were determined by linear approximation of experimental data according to the equation: $f_{lin} = a_0 + a_1 \cdot D$ using the least squares method.

The values of the function parameters for f_{lin} are given in Table 2.

Table 2

Parameters of f_{lin} (D) according to the study stages

Stage	Regression Equation
I	f_{lin} = 1.0781 – 0.0391×D
II	f_{lin} = 1.0628 – 0.0314×D
III	f_{lin} = 1.0379 – 0.0189×D
IV	f_{lin} = 1.0159 – 0.0080×D

As it can be seen from Table 2, the value of the regression coefficient a_1 in the non-linearity function f_{lin} approaches zero, while the constant of the regression equation a_0 approaches unit one. This indicates that the dependence of f_{lin} on the dose is negligent yet shall be accounted for in calculations, especially if the dose significantly deviates from 2 Gy.

Fig. 1 shows the graphs of the f_{lin} non-linearity correction factor functions.

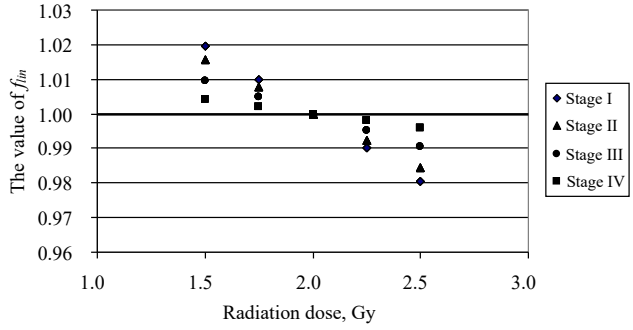

Fig. 1. Dependence of the non-linearity correction factor on the irradiation dose for Stages I–IV

At Stage I, it was revealed that the mean value of the non-linearity correction factor in the operating dose range (1.5–2.5 Gy) varied within ±2% (from 0.98 to 1.02), whereas at Stage IV, it varied within ±0.4% (from 0.996 to 1.004). Thus, with repeated use of a single batch of TL-powder across the stages "thermal treatment – irradiation – readout", a reduction in the degree of the TL-signal non-linearity against the dose is observed. This is associated with changes in dosimetric properties of the powder. Therefore, when calibrating TL-systems, the non-linearity of the TL-signal against the dose shall be studied regularly.

Compensation for the partial loss of thermoluminescence signal magnitude is achieved using the fading correction factor, calculated by the formula [2]:

$$f_{fad} = \frac{F_{fad}(\Delta t_{ref})}{F_{fad}(\Delta t_{user})}, \qquad (5)$$

where $F_{fad}(\Delta t_{ref})$ is the value of the fading function for the reference TL detector, dimensionless;

$F_{fad}(\Delta t_{user})$ is the value of a fading function for a clinical TL detector, dimensionless;

Δt_{ref} is the time between irradiation and readout of a reference TL detector, days;

Δt_{user} is the time between irradiation and readout of a clinical TL detector, days.

During Stages II–IV, the fading of the irradiated TL powder (TLD-100) was studied for a single powder batch. The fading function was determined according to the formula: $F(\Delta t) = b + c \times e^{-\left(\frac{\Delta t - 7}{d}\right)}$. The function parameters were calculated using the ORIGIN software [5].

Values of the TL powder fading function parameters for Stages II–IV are given in Table 3.

Table 3

Fading function parameters according to the study stages

Stage	TL Powder Fading Function
II	$y = 0.9701 + 0.02899 \times e^{-\left(\frac{\Delta t - 7}{22.38864}\right)}$
III	$y = 0.95309 + 0.0415 \times e^{-\left(\frac{\Delta t - 7}{44.42362}\right)}$
IV	$y = 0.95928 + 0.04263 \times e^{-\left(\frac{\Delta t - 7}{9.16225}\right)}$

Fig. 2 shows the results of TL signal measurements based on the study of Stage IV fading depending on the time interval (Δt) between the irradiation and readout of TL powder. The graph presents the TL signal normalized to its value at Δt=7 days.

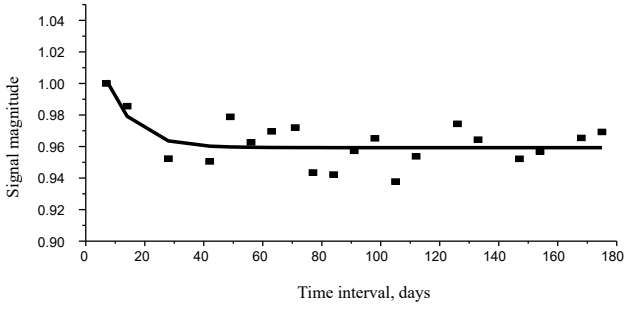

Fig. 2. Dependence of a normalized TL signal on the time interval between the irradiation and readout for Stage IV

For comparison, Fig. 3 shows the fading functions determined in Stages II and IV.

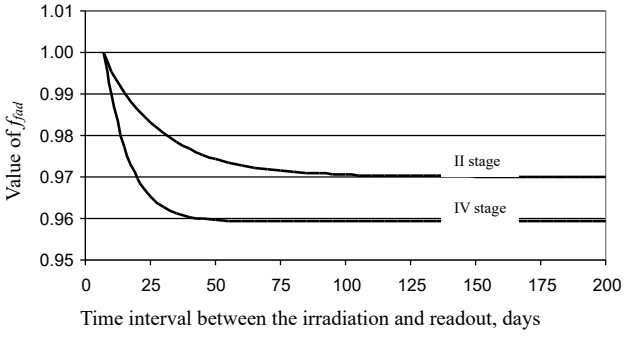

Fig. 3. The fading function curve determined at Stages II and IV

As it can be seen in Figs. 2 and 3, a fairly sharp decrease in the TL signal magnitude is observed between days 7 and 30. Beginning from day 40, the thermoluminescence signal becomes more stable, and from days 40–50, it remains practically unchanged with the increasing time interval Δt, indicating that both functions reach a plateau.

However, the functions exhibit different rates of dosimetric information loss during the first 30–40 days:

1. During the first use of the TL powder (Stage II), the fading study showed a lower information loss within 30 post-irradiation days (up to 3% of the initial value).

2. For subsequent applications of the same powder batch (Stage IV), the rate of information loss was higher, reaching 4% of the initial value before stabilizing.

When determining the relation of the two fading functions for Stages II and IV, a stable ratio of 1.010 ± 0.005 is observed between them beginning from days 30–40 (Fig. 4).

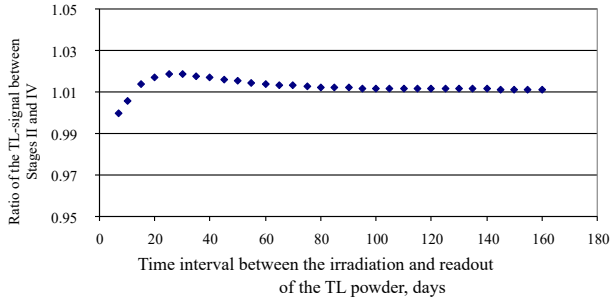

Fig. 4. Comparison of fading functions for Stages II and IV

When calculating the absorbed dose using the ratio of values from two fading functions – one for the time interval between the irradiation and readout of a reference TL detector Δt_{ref}, and another for clinical TL detectors Δt_{user} (see Formulas 1 and 5) – it is highly probable that a single fading function determined for a given powder batch can be used in calculations. This is particularly relevant since studying the fading function takes approximately 5 months (150–160 days per year), significantly reducing the time required for TLD audits.

The dependence of the fading coefficient f_{fad} on the time interval ($\Delta T_{user/ref} = \Delta T_{user} - \Delta T_{ref}$), between the irradiation of reference TL detectors at the centre and at medical institutions, determined according to the Stage IV fading function (Fig. 5), was studied.

As it can be seen in Fig. 5, even when TL detectors are irradiated at the centre and medical institutions within the same ±10-day "window" of the TLD audit stage, the fading coefficient f_{fad} significantly varies if the readout occurs 15 days after the irradiation day: f_{fad} ranges from 0.987 to 1.036. Therefore, the f_{fad} correction factor shall be included in the absorbed dose calculations for medical institutions.

At the same time, when the readout of TL detectors is performed 30 days after the irradiation, the correction factor f_{fad} changes but negligibly: in the

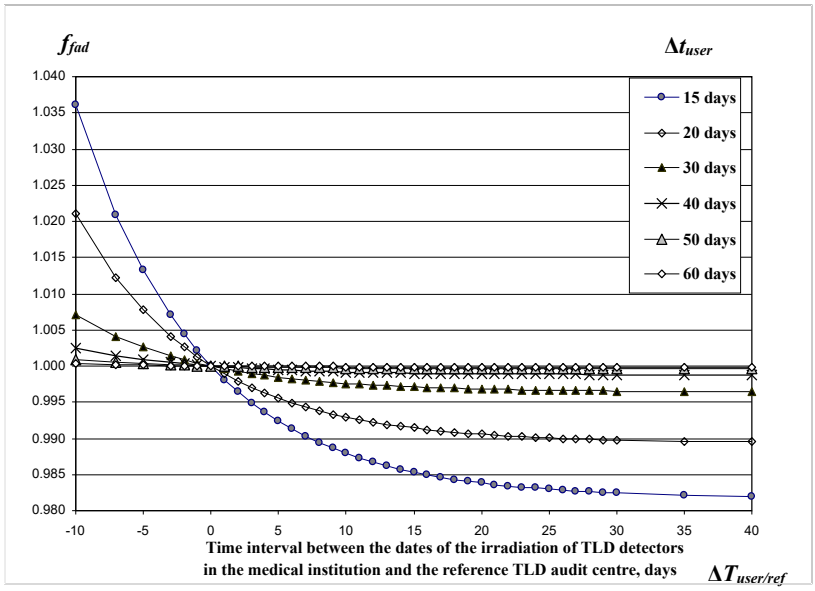

Fig. 5. Dependence of the fading function coefficient on $\Delta T_{user/ref}$ for different time intervals Δt_{user} between the irradiation and readout of TL detectors of a medical institution

range from 0.997 to 1.007, regardless of the time interval between the day of irradiation at the centre and in the medical institution, i.e. the indicated values of the coefficient change within the permissible error of its determination ±0.7% [6]. When the time interval between the irradiation and readout of TL detectors is increased to 50–60 days, the correction factor f_{fad} is practically equal to one (Fig. 5).

If subsequent studies confirm the stability of parameters b, c, d in the fading function $F(\Delta t) = b + c \times e^{-\left(\frac{\Delta t - 7}{d}\right)}$, the fading correction could be omitted for TL detector readouts performed ⩾30 days after the irradiation.

Overall, the fading studies represent the most time-consuming phase of TLD-audit organization and performance. This study shall be extended to different TL powder batches to explore minimizing this procedure without compromising the dose calculation accuracy.

Conclusions and discussion of results

In this study, the key correction factors required for the accurate operation of a TLD system were systematically studied and determined. The factors ensure the calculation of the absorbed dose in water during TLD audits and include:

• The non-linearity correction factor, which accounts for the non-linear dependence of the TL-signal on the irradiation dose. Its dependence on the dose was found to be minor yet shall be considered, especially for doses significantly deviating from the 2 Gy calibration point. Repeated use of the powder reduces this non-linearity.

• The energy dependence factor, which is equal to one since the photon energy of ^{60}Co therapy units in clinics is identical to the gamma energy of the reference source used for the system calibration.

• The holder correction factor, which compensates for the attenuation of the radiation beam by the standard plexiglass holder.

• The fading correction factor, which compensates for the signal loss over time between the irradiation and readout. For time intervals exceeding 30 days, the fading effect stabilizes, and the factor approaches unit one. This allows for the use of a single fading function determined for a specific powder batch, significantly simplifying the determination of the absorbed dose.

The environmental conversion factor $F_{PMMA \to water} = 1.0042$ was determined. This factor enables the use of a PMMA phantom for multi-capsule system calibration against the state primary measurement standard, while ensuring correct dose calculations in the water phantoms used in clinical facilities.

The practical application of these coefficients in TLD audits shows that under typical conditions where the time between the irradiation and readout exceeds 30 days, the impact of fading is minimal. In cases where reference and clinical detectors from the same powder batch are irradiated and read out simultaneously, the correction of fading is not required.

For the obtained data and coefficients to be properly applied during TLD audits, the next crucial step is to perform a detailed calculation of the combined standard uncertainty of the entire measurement process, incorporating the uncertainties of each individual correction coefficient used in the absorbed dose determination formula.

Вивчення дозиметричних властивостей термолюмінесцентної системи для аудиту доз у променевій терапії

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Анотація

Для здійснення постійного контролю апаратів гамма-терапії в умовах недостатнього забезпечення сучасним дозиметричним обладнанням, а в деяких випадках і відсутністю кваліфікованого персоналу, охоплення усіх радіологічних відділень щорічним зовнішнім контролем є практично єдиною можливістю незалежної перевірки точності розрахунків лікувальної дози, що відпускається пацієнтові.

Актуальність цієї роботи визначається необхідністю виконання положень національних законодавчих документів у галузі використання ядерної енергії: Закону України "Про захист людини від впливу іонізуючого випромінення", Норм радіаційної безпеки України (НРБУ-97), Державного стандарту України "Вимірювання іонізуючих випромінень. Метрологічне забезпечення. Основні положення" (ДСТУ 3240-2015).

Вимоги цих документів щодо точності та надійності дозовимірювань обумовлюють необхідність використання високоякісних детекторів. Саме тому з усіх термолюмінесцентних матеріалів, які використовують у проведенні ТЛД-аудиту терапевтичних струменів фотонного випромінення, перевагу віддають люмінофору LiF, Mg, Ti типу TLD-100, завдяки його дозиметричним властивостям: тканиноеквівалентності, тривалості збереження дозиметричної інформації (низькому федингу), високій чутливості, великому діапазону вимірюваних доз, чіткому відтворенню результатів при багаторазових вимірюваннях.

Саме використання таких детекторів робить можливим проведення регулярного дозиметричного аудиту діючих гамма-апаратів методом "доза-поштою". Цей підхід є важливим, оскільки дозволяє вчасно виявляти помилки клінічної дозиметрії, підвищити якість променевої терапії та, як наслідок, підвищувати якість лікування.

При проведенні ТЛД-аудиту необхідно забезпечити регулярність контролю дозиметричних властивостей та параметрів як самого люмінофора, так і вимірювального ТЛ-приладу, а також визначити низку факторів корегувальних коефіцієнтів та величину розширеної невизначеності.

Ключові слова: радіаційна безпека; променева терапія; термолюмінесцентні дозиметри; термолюмінесцентний порошок, невизначеність, ɣ-дозиметрія, іонізуюче випромінювання.

References

1. Ozerskyi K. Determination of correction factors and correction coefficients for calculations of the absorbed dose. *Ukrainian Metrological Journal*, 2024, no. 2, pp. 51–57. doi: https://doi.org/10.24027/2306-7039.2.2024.307280
2. Technical Reports Series No. 398. Absorbed Dose Determination in External Beam Radiotherapy: An International Code of Practice for Dosimetry Based on Standards of Absorbed Dose to Water. Vienna, IAEA, 2000. 242 p.
3. Council Directive 2013/59/Euratom of 5 December 2013 laying down basic safety standards for protection against the dangers arising from exposure to ionising radiation, and repealing Directives 89/618/Euratom, 90/641/Euratom, 96/29/Euratom, 97/43/Euratom and 2003/122/Euratom. Available at: https://eur-lex.europa.eu/eli/dir/2013/59/oj
4. Izewska J., Georg D., Bera P. et al. A methodology for TLD postal dosimetry audit of high-energy radiotherapy photon beams in non-reference conditions. *Radiother. Oncol.*, 2007, vol. 84, pp. 67–74. doi: https://doi.org/10.1016/j.radonc.2007.06.006
5. Izewska J., Gajewski R., Gwiazdowska B. et al. TLD postal dose intercomparison for megavoltage units in Poland. *Radiother. Oncol.*, 1995, vol. 36, no. 2, pp. 143–152. doi: https://doi.org/10.1016/0167-8140(95)01604-f
6. Kirby T.H., Hanson W.F., Johnston D.A. Uncertainty analysis of absorbed dose calculations from thermoluminescence dosimeters. *Med. Phys.*, 1992, vol. 19, no. 6, pp. 1427–1433. doi: https://doi.org/10.1118/1.596797