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Implementation of the International Code of Practice on Dosimetry in Diagnostic Radiology (TRS 457): Review of Test Results

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IMPLEMENTATION OF THE INTERNATIONAL CODE OF PRACTICE ON DOSIMETRY IN DIAGNOSTIC RADIOLOGY

(Technical Reports Series No. 457):
REVIEW OF TESTING RESULTS

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REVIEW OF TESTING RESULTS

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FOREWORD

In 2007, the IAEA published Dosimetry in Diagnostic Radiology: An International Code of Practice (IAEA Technical Reports Series No. 457). This publication recommends procedures for calibration and dosimetric measurement for the attainment of standardized dosimetry. It also addresses requirements both in standards dosimetry laboratories, especially Secondary Standards Dosimetry Laboratories (SSDLs), and in clinical centres for radiology, as found in most hospitals. The implementation of TRS No. 457 decreases the uncertainty in the dosimetry of diagnostic radiology beams and provides Member States with a unified and consistent framework for dosimetry in diagnostic radiology, which previously did not exist. A coordinated research project (CRP E2.10.06) was established in order to provide practical guidance to professionals at SSDLs and to clinical medical physicists on the implementation of TRS No. 457. This includes the calibration of radiological dosimetry instrumentation, the dissemination of calibration coefficients to clinical centres and the establishment of dosimetric measurement processes in clinical settings. The main goals of the CRP were to:

— Test the procedures recommended in TRS No. 457 for calibration of radiation detectors in different types of diagnostic beams and measuring instruments for varying diagnostic X ray modalities;
— Test the clinical dosimetry procedures, including the use of phantoms and patient dose surveys;
— Report on the practical implementation of TRS No. 457 at both SSDLs and hospital sites.

Testing of TRS No. 457 was performed by a group of medical physicists from hospitals and SSDLs from various institutions worldwide.

The present publication is a compilation of the results, findings and recommendations of the participants of the CRP and seeks to illuminate and highlight any issues that have arisen during the CRP period and thus supplement the work of TRS No. 457.

The IAEA wishes to express its gratitude to all authors and reviewers of this publication as listed at the end of the report.

The IAEA officers responsible for this publication were I.D. McLean, A. Meghzifene and F. Pernicka of the Division of Human Health.

EDITORIAL NOTE

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CONTENTS

1. INTRODUCTION

Supported by the recommendations of the IAEA Standing Advisory Group, the Scientific Committee of the IAEA/World Health Organization (WHO) Secondary Standards Dosimetry Laboratories (SSDL) Network in 1996, a series of consultants meetings were initiated in 1999 and supported by the coordinated research project (CRP E2.10.03, 2001–2003) with the task of producing an International Code of Practice for Dosimetry in Diagnostic Radiology.

In 2007 the task was completed with the publication entitled Dosimetry in Diagnostic Radiology: An International Code of Practice (IAEA Technical Reports Series No. 457) [1]. This publication is to advise users on the practice of dosimetry in diagnostic radiology for instrument calibration at standards dosimetry laboratories, especially at Secondary Standard Dosimetry Laboratories (SSDLs), and for application to patient dosimetry at radiological facilities.

It is important that the SSDLs that have been calibrating radiation detectors in terms of air kerma in X ray beam qualities suitable for radiotherapy [2] and/or radiation protection [3] dosimetry, be aware of the specific X ray beam qualities applicable in diagnostic radiology [4]. At present, most of the SSDLs in the network have not yet implemented the newly recommended beam qualities. This situation is expected to change significantly in the next few years.

Until recently there has been no standardization in the units or methodology for dosimetry for diagnostic radiology, as applied to the clinical work place. In 2005 the ICRU published a report entitled Patient Dosimetry for X rays used in Medical Imaging [5] which defined quantities and units for diagnostic radiology dosimetry along with some methodology. TRS No. 457 was developed in parallel with this ICRU publication and is complementary in its treatment of clinical dosimetry. It aims to give guidance to the Member States in essential work in diagnostic radiology dosimetry for 5 general modalities:

— General radiography;
— Fluoroscopy;
— Mammography;
— Computed tomography;
— Dental radiography.

The current publication reports on the substantive results and conclusions drawn from the coordinated research project (CRP) instituted to review the effectiveness of the implementation of TRS No. 457 both in the SSDL and clinical environment (Appendices I and II). The CRP ran 2005–2008, with 11 participants; 7 being primarily from SSDLs and 4 from clinical centres. At the first research coordination meeting, a set of 7 activities was agreed upon as shown in Table 1.

This report is made up of 5 sections in addition to this introduction. Section 2 takes a general look at the common areas of dosimetry shared by SSDLs and clinical centres including quantities, formalism and instrumentation as well as general observations relating to the implementation of TRS No. 457. Section 3 follows with work specific to SSDLs including work from activities 1–3 with some contributions from activities 5–7 as appropriate. Section 4 covers work specific to clinical centres and deals with activity 4, while Section 5 reports predominantly on the work of activities 5 and 6 with a concluding section containing recommendations arising from the CRP.

The object of this report is to discuss the work activities of the CRP and highlight the following issues

— General affirmation of the ability to implement TRS No. 457 as appropriate;
— Highlighting of areas of difficulty in implementation and suggestion of strategies for TRS No. 457, to assist in implementation;
— Identification of text that needs to be corrected, revised or added to in future revisions of TRS No. 457;
— Recommendations of further work needed in the field.

TABLE 1. ACTIVITY LIST

Activity 1	Setting-up calibration beam qualities at SSDLs
Activity 2	Development of calibration procedures including the uncertainty budget at SSDLs
Activity 3	Comparison of calibration of a selected instrument in selected beam qualities at SSDLs
Activity 4	Evaluation of measurement procedures in hospitals, including: 1. Research the feasibility of implementing the procedures described in TRS No. 457 for making dosimetric measurements using phantoms and for patient data collection. 2. Report on the availability of dosimetric instrumentation and recommended phantoms, and the possibility of phantom fabrication if needed. 3. Create uncertainty budget for each type of dosimetric estimation, including the dose estimation from patient data. 4. Compare phantom measurements with patient dose data for each modality.
Activity 5	Calibration of KAP meters at the SSDLs and at the clinical centres
Activity 6	TLD dosimetry audit for SSDLs and clinical centres
Activity 7	The implementation of practical peak voltage (PPV)

2. IMPLEMENTATION OF CODE OF PRACTICE

2.1. IMPLEMENTATION

A recent survey of SSDLs (Appendix III, Annex 1) shows that there are currently at least 15 laboratories giving services in the calibration of diagnostic radiology dosimeters. Another 13 sites indicate they plan to have a facility in the next three years. Of the 15 operating sites, 11 indicated they followed TRS No. 457 while 13 indicated they used IEC diagnostic quality beams. The survey shows there is a large range in diagnostic radiology calibration activity, with a range of 5 to 60 detectors a year being calibrated at a (non-commercial) SSDL facility, and a total of 335 detectors calibrated in 2007 at all active centres. The one commercial facility registered as an SSDL calibrated 4,881 detectors in the same period. At some facilities there is some confusion between ISO 4037 beam qualities and the use of IEC diagnostic radiology calibration beam qualities.

2.2. QUANTITIES AND UNITS

It was noted that the notation used by TRS No. 457 is similar to that used by ICRU 74, however in some cases there has been some simplification (Table 2). It is further noted for computed tomography (CT) that the manufacturers of CT scanners have adopted the units defined by the IEC [6]. This standard predates the ICRU standard. It is understood that currently both the ICRU and the IEC have committees reviewing aspects of CT dosimetry including the quantities and units.

2.3. DOSIMETRY FORMALISM

TRS No. 457 presents the dosimetric formalism that provides a calibration coefficient and beam quality corrections factors, k_Q for the dosimeter. The formalism is consistent with that used in the Code of Practice of Radiotherapy Dosimetry [7]. Calibration laboratories can provide the calibration coefficients in the calibration certificate either by stating the calibration coefficients for each radiation quality used in the calibration or by stating

TABLE 2. A COMPARISON OF QUANTITIES USED FOR ICRU 74 AND TRS No. 457

Quantity	ICRU	IAEA
Incident air kerma	$K_{a,i}$	K_i
Entrance-surface air kerma	$K_{a,e}$	K_e
Air kerma-area product	P_{KA}	P_{KA}
Air kerma-length product	P_{KL}	P_{KL}
CT air-kerma index free in-air	C_K (integration $-\infty$ to $+\infty$)	$C_{a,100}$ (integration –50 to +50 mm)
CT air-kerma index in the standard phantom	$C_{K,PMMA}$ ($C_{K,PMMA,100}$)	$C_{PMMA,100}$

the calibration coefficient for the reference quality along with the k_Q factors for each calibrated radiation quality. Clear examples of the expression of calibration coefficients described for both ways are included in TRS No. 457.

For uncertainty estimations, three scenarios of evaluation of uncertainties were introduced in TRS No. 457. Scenario 1 describes the case of a dosimeter that complies with IEC 61647 requirements; scenario 2 describes the case of a calibrated dosimeter with pressure and temperature corrections applied; and scenario 3 is the most comprehensive approach using a more accurate dosimeter with calibration and interpolation of calibration coefficients for user beam qualities. For scenario 3, the determination of calibration coefficients for the user radiation beam quality may require interpolation of the calibration coefficient from the calibrations coefficients stated in the calibration certificate (laboratory radiation beam qualities). Examples for interpolation of the calibration coefficients are presented in the Appendix IV.

2.4. DOSIMETERS

2.4.1. Stabilization

The ionization chamber and the measuring assembly require some time after switching on to stabilize. According to the specifications of most dosimeters used in clinical diagnostic radiology, only a few seconds are required for stabilization. Often, an initialization is automatically performed when the instrument is turned on, however this is not the case for reference ionization chambers and electrometers used in SSDLs, which should be left for ambient and electronic stabilization. A check should be made on the stabilization time for each detector and measuring assembly. In practice, this could be performed by measuring any leakage current, or by evaluating any continuous increase or decrease of the readings during unchanging exposure conditions.

2.4.2. Temperature and pressure correction

Ionization chambers and monitor chambers used in diagnostic radiology are vented and therefore correction for temperature and pressure is needed. This also includes KAP meters; however the temperature of the KAP meter may not always be the same as the ambient air temperature due to its location in the tube housing.

It should be noted that some electronic barometers make automatic corrections for altitude (that is they give the pressure corrected to sea level). Care should be taken when using these instruments. In addition, the response of a KAP meter to large pressure changes may be different to that expected from other chambers.

2.4.2.1. Dosimetry systems with automatic correction for temperature and pressure

There are some dosimeters that automatically perform temperature and pressure correction through either manual entry of temperature and pressure by the user, or automatic determination from internal sensors within the device. As an example, the RADCAL 2026 series dosimeter, for one of its commercial configurations, only has a sensor for temperature. The reading of the dosimeter is automatically corrected to a reference temperature of 22ºC

(where the calibration coefficient pertains) and ambient (room) pressure. In this case, correction only for pressure is needed, assuming that the measurement of temperature is correct. Additionally, clinical users and SSDLs should always pay attention to the reference conditions that are applied (most European SSDLs have a reference temperature at 20°C) either by instrument automatic corrections or to the calibration coefficient of the dosimeter.

For these reasons, the SSDL (during calibration procedures) and the user (during clinical measurement) should investigate the operation of an instrument with respect to the ambient conditions, in order to be aware of temperature and pressure corrections required, to avoid errors. However, such errors are not expected to be more than 3%. Temperature and pressure corrections should also note where they are measured, for example at the chamber position, so the correct temperature is indicated.

2.4.2.2. Solid state detectors

It is well known that solid state detectors do not need correction for temperature and pressure to be applied. Therefore the k_{TP} correction factor equals to 1.

2.4.3. Effects of radiographic versus continuous irradiation of detectors

Dosimeters that comply with the IEC 61674 standard are not significantly affected by different irradiation conditions (radiography or fluoroscopy) or dosimeter operation modes. A study conducted within the CRP showed that such equipment exhibited limits of variation within the suggested ±5% of the IEC standards with respect to the above mentioned conditions [8]. However, even if the dosimeter complies with the IEC requirements, the user should investigate its response and performance at various exposure conditions at a calibration laboratory or with clinical beams as appropriate. Moreover, the user should be aware of the technical specifications and limitations to use the instrument safely and avoid unacceptable errors in dose determination.

The user should also be aware that the different modes of operation cannot always be tested during calibration at SSDLs, as high dose rate X ray equipment is not available at most SSDLs (see Section 3.1.2). The user should investigate the validity of the calibration coefficients in cases where the use of the instrument is in modes other than those under which it has been calibrated.

As IEC 61674 does not contain separate requirements for the reference and field class dosimeters, TRS No. 457 includes recommended specifications for the reference class dosimeters for different applications in radiology (Table 5.2 in TRS No. 457).

2.5. USE OF NON-INVASIVE X RAY TUBE VOLTAGE MEASURING INSTRUMENTS

The quantity practical peak voltage (PPV) is used for measurements of the X ray tube voltage [9, 10]. Measurement of PPV is described in Appendix IV in TRS No. 457. The IEC Standard 61676 is used as a reference for the requirements of non-invasive measuring devices and the requirements in TRS No. 457 comply with those in IEC 61676. However, in contrast to IEC 61676, TRS No. 457 requires that tube voltage measurements be made exclusively using PPV, without the option of using other tube voltage quantities. The CRP participants found two published articles containing discussion of the application of PPV to mammography [11, 12] since the publication of IEC 61676 which support the application of PPV. However, systematic errors have been reported with a PPV measurement device that was reportedly associated with current variation during X ray production [13]. Furthermore, the need for mandating the exclusive use of PPV was questioned by the majority of the CRP participants. For practical purposes, modern X ray machines, which operate at high frequencies, are essentially constant potential, which means that kVp would be the same value regardless of the quantity being used.

3. IMPLEMENTATION AT SSDLs

3.1. ESTABLISHMENT OF A DIAGNOSTIC SSDL CALIBRATION FACILITY

3.1.1. Introduction

Activity 1 of the CRP was to test the establishment of facilities for diagnostic X rays and the calibration of selected radiation detectors at SSDLs.

Eight SSDLs from various countries (Brazil, Cuba, Czech Republic, Finland, Greece, Thailand, Vietnam and the IAEA) participated in the relevant CRP activities.

3.1.1.1. Environmental conditions

Calibration facilities at SSDLs should be maintained at a constant temperature within the irradiation room. The temperature should be monitored and checked continually, especially during calibration procedures.

It is good practice for temperature measurements to use two thermometers (mercury or thermocouples) which have a reading resolution better than 0.5ºC. The reference thermometer should have been checked at an appropriate primary laboratory and preferably also by calibration to the quantities of temperature at a laboratory traceable to the international measurement system. The reference thermometer can then be used to calibrate the working thermometer. This calibration could be carried out by placing the working thermometer close to the reference thermometer and comparing their readings.

During calibration measurements, the thermometer should be placed close to the ionization chamber. If this is not feasible, the difference of temperature between the two places should be known.

The pressure inside the irradiation room should be measured by a barometer. It is a good practice to use two barometers for pressure measurements; a reference barometer and a working barometer. Preferably, the barometers should also have calibration to the quantities of pressure at a laboratory traceable to the international measurement system.

3.1.1.2. Apparatus

TRS No. 457 describes in detail the apparatus required by an SSDL in order to establish diagnostic radiology dosimetry standards and to provide calibrations in these fields.

X ray systems for conventional radiology applications (other than mammographic)

All participating SSDLs and most SSDLs worldwide are equipped with X ray systems which operate in fluoroscopic mode. The tube current is in the range 0.5–30 mA, while the tube anode (target) is stationary with a larger target angle than that typically used in clinical diagnostic units. Therefore, these systems are more similar to therapy clinical X ray systems rather than those used for clinical radiology.

Table 3 shows the X ray system apparatus set in place at participating SSDLs with beam qualities RQR, RQA and RQT for activity 1.

Table 4 shows the conditions and dosimetry equipment used for HVL determination. For Greece, an additional 5.2 mm PMMA is used as permanent filtration for practical reasons, as the same system is used for other applications.

Monitor chambers

Five of the participating SSDLs used a monitor chamber in their X ray systems. Two SSDLs did not. TRS No. 457 (Section 6.4.4) describes the use and the suitability of a monitor chamber. The monitor chamber could be used for deducing the K_{air} or K_{air} rate reference value at the point of measurement, to check the stability of the X ray tube and apply any corrections; or to provide information about the beam on/off status. The way of using such a

TABLE 3. EQUIPMENT USED AT PARTICIPATING SSDL SITES FOR RQR, RQA AND RQT BEAM QUALITIES

	Brazil	Cuba	Czech Republic	Finland	Greece	Thailand	Vietnam
Unit	Pantak HF160	Pantak HF160	Seifert Isovolt US2	Seifert Isovolt160HS	Pantak 225HF	GE Isovolt Titan E	Pantak HF160
Target	Tungsten	Tungsten	Tungsten	Tungsten	Tungsten	Tungsten	Tungsten
Window	Be	Be	Be	Be	Be	Be	Be
Rectification	Constant potential	Constant potential	Full wave	Constant potential	Constant potential	3 Phase	Constant potential
Ripple	<0.15%	<0.13%	1%	<0.5%	<0.5%	<0.3%	<0.15%
Electrical power (kW)	3.2	3	3	2.5	3.2	4.5	3.2
Tube voltage range (kV)	5–160	5–160	0–150	160 max	5–220	5–320	5–160
Max tube current (mA)	50	50	20	30 & 19	30 & 20	45 mA	50
Operation. Mode	Fluoro	Fluoro	Fluoro	Fluoro	Fluoro	Fluoro	Fluoro
Shutter	Yes	Yes	Yes	Yes	Yes	Yes	Yes
Monitor Chamber	No	Yes	Yes	Yes	No	Yes	Yes
Use of the monitor chamber	Not used	Yes	Yes	QC check of output stability	Not used	Yes	Yes
Permanent filtration	1 mm Be	1 mm Be	3 mm Be	1 mm Be	1 mm Be + 5.2 mm PMMA	3 mm Be	1mm Be

TABLE 4. SET UP CONDITIONS USED AT PARTICIPATING SSDL SITES FOR THE MEASUREMENT OF HVL

	Brazil	Cuba	Czech Republic	Finland	Greece	Thailand	Vietnam
Focus to chamber distance (cm)	100	100	100	100	100	100	100
Focus to HVL filter holder distance (cm)	44	~ 35	63	59	42	50	65
Diameter of field at measurement point (cm)	10	6.1	1.9	3	5	12	7.6
Al attenuator purity	99.999%	99.9%	99.99%	99.999%	99.999%	99.99%	99.99%
Chamber	Radcal 10 × 5–6[a]	PTW 23361[a]	Exradin A3[b]	PTW 23344[c] NE 2561[a]	PTW 30001[a]	PTW[a]	Exradin A4[b]
Energy response	<1% *	0.73% **	1.2%**	<1% **	<0.5% *	—	1.5%**

[a] cylindrical; [b] spherical; [c] plane parallel

* as stated by the manufacturer at the diagnostic radiology X ray energy range

** as measured in the RQR series energy range

chamber depends on the procedures followed at each SSDL, in establishing and maintaining the standards and in calibrations.

Shutter

All participating SSDLs used a shutter in their X ray systems to control the irradiation time. The shutter transit time Δt should be known in order to perform adequate corrections if needed.

X ray system apparatus for mammography

While the use of continuous X ray systems is most common for mammography calibration at SSDLs, a number of centres use clinical mammography units with certain modifications. An example is given in Table 5.

The calibration of a mammography detector is performed free in air. In cases where the SSDL uses a clinical X ray mammography system, the cassette table (holder) and the compression paddle should be removed and the X ray beam positioned horizontally.

It should be considered that in clinical measurements of incident air kerma, K_i, the measurement conditions differ from those used in the dosimeter calibration, because the detector is positioned 45 mm above the cassette table with the compression paddle in the beam. Therefore, the beam quality will differ when compared to calibration conditions, and there may also be scattered radiation from the breast support table.

The magnitude of this scatter, measured as the backscatter factors[1] (BF), has been determined for various detectors and beam qualities (Mo/Mo, Mo/Rh and Rh/Rh beams has been investigated) as shown in Figs 1 and 2.

TABLE 5. SPECIFICATION OF THE X RAY MAMMOGRAPHY SYSTEM FROM GREECE WITH HVL MEASUREMENT PARAMETERS

X ray system		HVL determination	
Unit	GE SENOGRAPH 600 HF Senix	FCD (cm)	60
Target	Mo	Focus to HVL filter holder distance (cm)	25.8
Window	Be	Field (cm^2)	18 × 24 (rectangular)
Rectification	High frequency	Al attenuator purity	99.999%
Ripple	< 0,1%	Chamber	PTW TW 77337 plane parallel
Electrical power (kW)	15 kW	Energy response	1.7%, as measured at RQR-M series energy range
Tube voltage range	22–40 kV		
Max. tube current (mA)	600		
Operation mode	Graph		
Shutter	No, but tube current-time product set		
Monitor chamber	No		
Permanent filtration	30 μm Mo		

FCD: Focus to chamber distance

[1] Ratio of air kerma measured with the support to air kerma without the table support. The source to detector distance (SID) used was 65 cm.

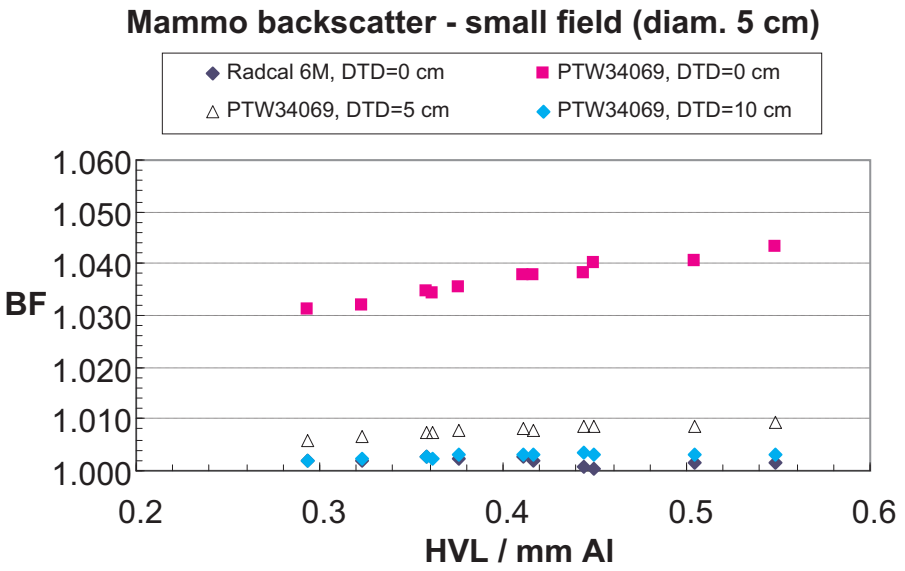

FIG. 1. Backscatter factor (BF) measurements for small field size. DTD stands for detector-to-table distance.

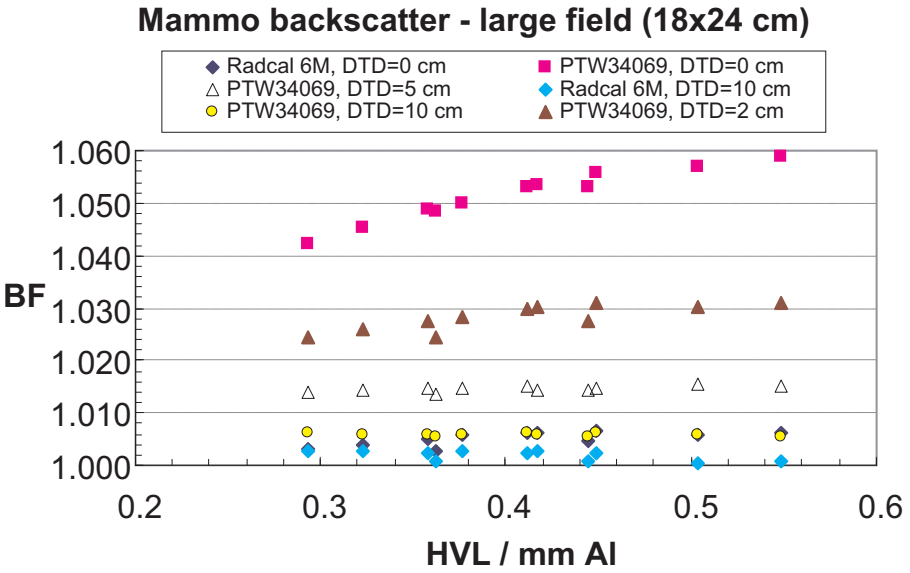

FIG. 2. Backscatter factor (BF) measurements for large field size. DTD stands for detector-to- table distance.

For some dosimeters, one can expect up to a 6% higher reading when backscatter is included. This should be taken into account when measuring the entrance incident air kerma or HVL.

The breast compression paddle hardens the X ray beam, resulting in higher HVL values. This may affect the response of high energy dependence dosimeters, such as those using semiconductor elements. In Section 3.1.2.3, the influence of the compression paddle on X ray beam quality is given. While this influence might be negligible for ionization chambers or solid state detectors that perform energy compensations[2], it could result in more than a 5% error in the case of solid state detectors without automatic energy compensation. The energy response of chambers in the mammography range is given in TRS No. 457, Table 5.2.

[2] Such solid state detectors should be subjected to QC checks to verify their energy compensation.

TABLE 6. ERRORS IN MEASUREMENT OF K_I FOR 6 COMMERCIAL MAMMOGRAPHY DOSIMETERS DETAILS IN STUDY [14].

Scenario	PTW	Inovision	Radcal	Piranha	Unfors	Dosimax	Solidose
Sc1 : Net dosimeter reading	—[a]	2.8%	0.7%	–5.8%	–1.1%	11.6%[(b)]	7.4%
Sc2 : N_K @ RQR-M2	1.4%	1.0%	–0.1%	0.6%	0.7%	11.7%	11.6%
Sc3 : N_K @ RQR-M4	0.1%	0.3%	0.2%	0.9%	0.9%	2.9%	2.4%
Sc4 : k_Q @ actual quality	–0.2%	0.0%	0.3%	1.0%	0.7%	–0.6%	–1.1%

[a] PTW net reading was in charge units (nC).
[b] A nominal correction factor of 1.8 was applied to net dosimeter reading (refer to the text [14]).

Table 6 presents the results of a study [14] where the errors from six commercial mammography dosimeters in air kerma K_i measurement, at 30 kV with 3 mm paddle thickness, arising from irradiation from the X ray beam (HVL of 0.39 mm Al), were measured under four scenarios.

In scenario 1, the K_i was the net dosimeter reading, without application of any correction factor or calibration coefficient. In scenario 2, which might be the most commonly applied scenario in clinics, the K_i was determined by applying the calibration coefficient N_K at RQR-M2 (28 kV) and air density corrections. In scenario 3, the N_K at RQR-M4, the hardest RQR-M series quality, and air density corrections were applied. In scenario 4, the k_Q (factor for energy corrections) at the actual X ray beam quality (at 30 kV with 3 mm of PMMA) was applied together with N_K at RQR-M2 and air density corrections.

The uncertainties for these error values were estimated to be 2% at 1 SD.

Apertures and usable beam sizes

TRS No. 457 does not define the size of reference chambers. However, it is stated that the secondary standard or detector to be calibrated should be totally within the beam and that the calibration field should be at least 1.5 times larger than the corresponding linear dimension of the detector. Furthermore, the variation in the air kerma rate over 80% of this field should not vary by more than 2% from the maximum value[3] .

One practical difficulty encountered at the IAEA Seibersdorf SSDL was in establishing a useable beam area that complied with the above mentioned criteria for all beam types. Beam profile measurements were made using an ionization chamber array with a resolution of 1 cm and a relative measurement accuracy of 1%. The greatest difficulties were with a large focal spot mammographic tube, using a 60 cm focus to detector distance, where a maximum variation in air kerma rate from the maximum of 5% was only achieved for a 40 mm diameter detector. Considerably better results were achieved at a measurement distance of 100 cm for a ceramic tungsten tube with a considerably reduced focal spot size.

The reasons for these difficulties were traced to the large focal spot size of the mammography X ray tube, in combination with the excessively tight beam collimation being applied in close proximity to the focal spot. This underlines the importance of using an X ray tube with as small a focal spot as the beam kerma rate will allow[4] and of careful design of the collimation system.

For the calibration of a specialized diagnostic radiology detector used for P_{KL} and P_{KA} measurements, small apertures are needed to create small X ray field sizes. The SSDL should investigate the accuracy of the determination of K_{air} at the point of interest. In calibrations of CT ionization chambers used for P_{KL} measurement, K_{air} is measured for the reference dosimeter in the plane of measurement for RQT qualities, without the lead rectangular aperture. A rectangular lead aperture with a width of between 20 mm and 50 mm and known to within 0.01 mm is then positioned in front of the user chamber (page 67, TRS No. 457).

[3] This is further described on page 88 of TRS No. 457.

[4] The focal spot size of a ceramic X ray tube proved to be about half the linear dimension of that of a comparable glass X ray tube, both tubes being able to operate with a similar tube current under similar tube voltage conditions.

In the case of KAP meter calibration measurements, a circular[5] or square lead aperture should be used with a diameter or width between about 40 mm to 60 mm. In this case, the measurement of K_{air} with the reference chamber is performed behind a lead aperture (page 68, TRS No. 457). Therefore the size of the beam behind the aperture with respect to reference chamber size should be considered.

When the HVL is to be determined, TRS No. 457 states that the field size should be small enough to just cover the detector (Appendix V, p. 262).

TRS No. 457 states that the lead aperture thickness should be 2 mm. Although this is adequate for RQR beams, 2.5 mm is recommended for beam qualities with heavier filtration (e.g. RQT). The criterion of 0.1% transmission is then met.

3.1.2. Establishing radiation qualities

TRS No. 457 gives the necessary data to establish radiation beam qualities and describes the procedures for the establishment of the diagnostic beam qualities, using IEC Standard 61267, 2nd ed: 2005 as a reference for the beam qualities RQR, RQA, RQT, RQR-M and RQA-M. It also provides guidance for the determination of HVL and the required additional filtration to establish a beam quality.

3.1.2.1. RQR beam qualities

Table 7 shows the RQR beam qualities, that have been established according to TRS No. 457 and the IEC 61267, at the participating SSDLs (Annex II).

TABLE 7. TRACEABILITY OF RQR QUALITIES

Country	Status as of July 2007	Traceability of standard/time
Brazil	RQR	PTB
Cuba	RQR	PTB
Czech Republic	RQR to be re-established*	PTB through SSDL Greece
Finland	RQR	PTB and BIPM
Greece	RQR	PTB
IAEA	RQR	PTB To be re-established
Thailand	RQR	PTB through SSDL Greece
Vietnam	RQR	PTB through SSDL Greece

* Czech SSDL has changed its site during 2007 and the new X ray unit has been installed. RQR spectra re-established in July 2009.

All participating SSDLs have succeeded in establishing RQR beam qualities according to TRS No. 457 and the IEC 61267 standard.

The amount of additional filtration used by participating SSDLs, in order to achieve RQR series beam qualities, is shown in Fig. 3. Note that when setting up a specified beam quality, Figure 6.9 from TRS No. 457 is a useful guide. Figure 4 shows an expanded version of that figure.

According to TRS No. 457, the acceptability criterion of the HVL value for each beam quality is that the K_{HVL}/K_0 ratio should be between 0.485 and 0.515, where K_{HVL} is the air kerma for the specified beam with an added attenuator equal to the HVL specified for the beam qualities (specified by TRS No. 457). An alternative method involves the determination of the beam HVL[6] from the measurement of K_0 and the K_i measured for two attenuator

[5] Precise circular apertures are generally easier to fabricate than rectangular ones.

[6] A log linear interpolation is required.

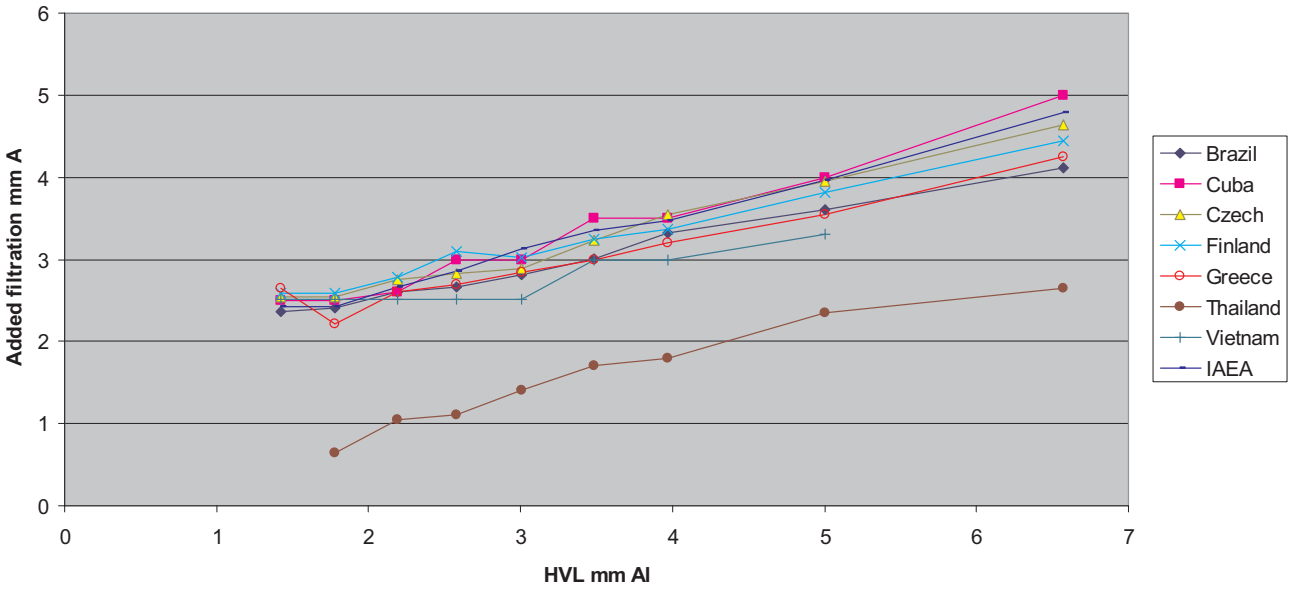

FIG. 3. The additional filtration used for the establishment of the RQR series qualities by participating SSDLs.

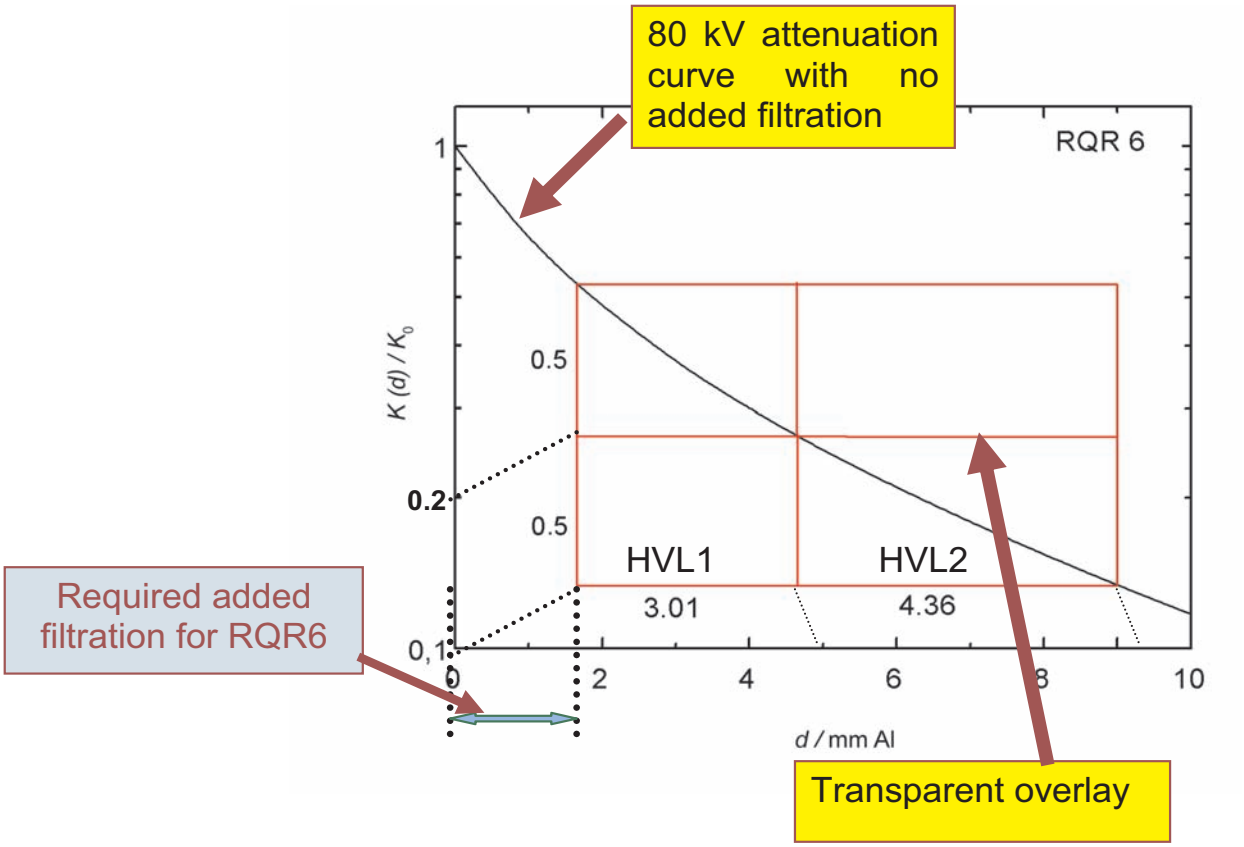

FIG. 4. An extension of Fig 6.9 from TRS No. 457 with additional explanation on the positioning of the overlay so that edges are parallel with the curve axis with the corners and centre of the overlay square on the attenuation curve.

thicknesses that bracket the expected HVL value. In such cases, the HVL acceptability criterion should be taken as the ratio of measured to specified HVL values, which should be between 0.957–1.044.

The homogeneity index (h) is a useful quantity to further specify the overall quality of the beam. It should be ±0.03 of the suggested IEC values (see the last columns of tables in Annexes II–V). This criterion is not met in only a few cases.

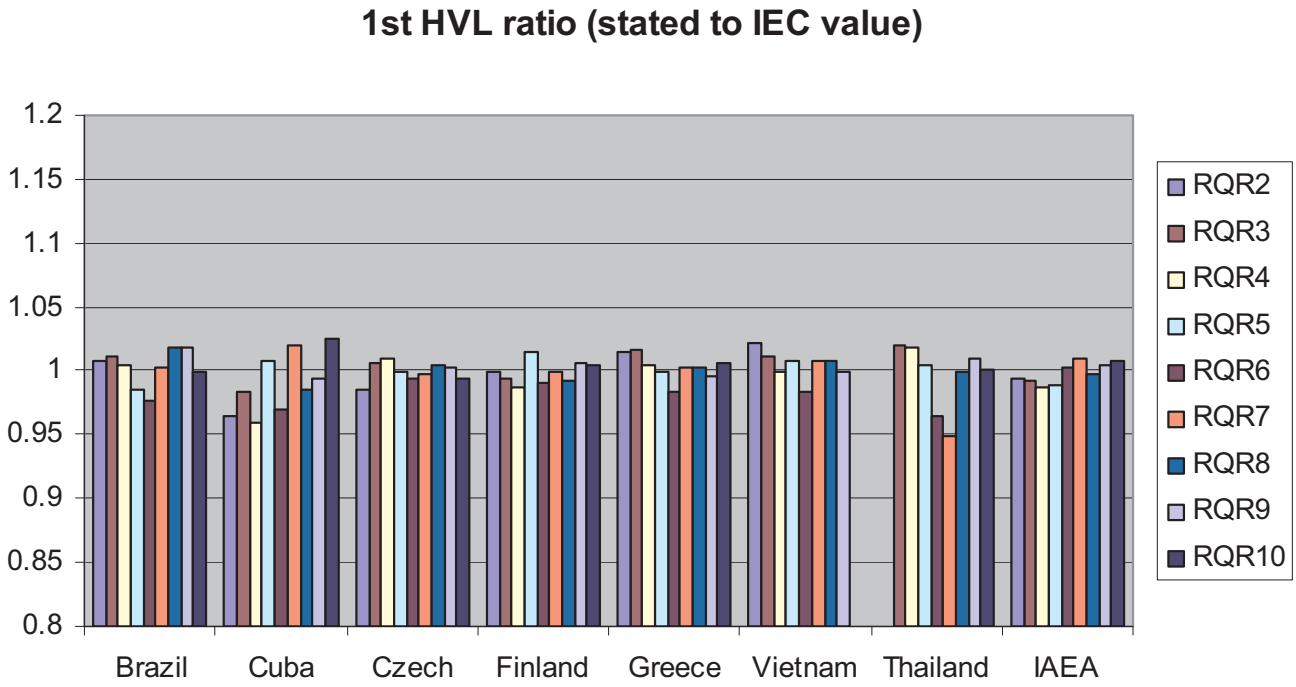

Fig. 5. The ratios of the measured HVL to the standard IEC HVL value for the RQR series, for the participating SSDLs (from Annex II).

3.1.2.2. Other TRS No. 457 beam qualities

Table 8 shows the RQT, RQA and RQR-M beam qualities according to TRS No. 457 and the IEC 61267 that have been established at participating SSDLs (Annex III–V).

TABLE 8. TRACEABILITY OF OTHER QUALITIES

Country	Status as of July 2007	Traceability of standard/time
Brazil	RQA being established	PTB
Cuba	RQT	PTB
Czech Republic	RQA, RQT, RQR-M to be established	PTB through Greece, (not RQA), RQR-M PTB through IAEA
Finland	RQT, W-Mammgraphyo	BIPM
Greece	RQT, RQR-M	PTB
IAEA	RQR-M, RQA Mammography — Rh	PTB RQA To be re-established NIST
Thailand		
Vietnam	RQT, RQA	PTB through Greece for RQT

Few SSDLs have established the non RQR beam quality series for several reasons. In the case of mammography, the RQR-M series require a Mo anode X ray tube and Mo filters, which are not available in some laboratories. To overcome this difficulty, one participating SSDL (Finland) established other mammography qualities based on a tungsten X ray tube target (Cuba is also planning to use a tungsten target based system). These W anode based qualities may be applied in mammography dosimetry, [15] however, they are not included in the IEC 61267 standard at the moment. Tungsten anode based mammography beam qualities might become necessary since some modern clinical mammography digital X ray systems use W anodes.

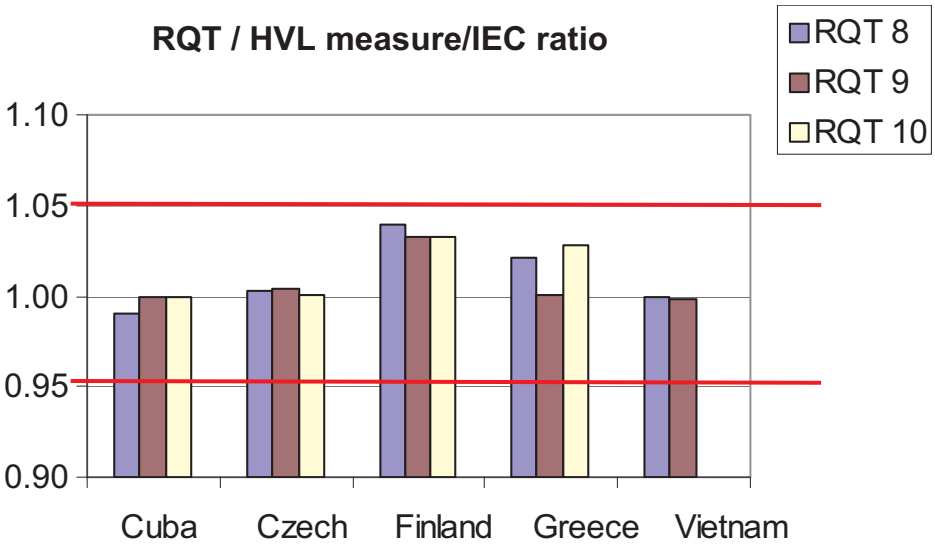

FIG. 6. The ratios of the measured HVL to the standard HVL IEC value ($HVL_{meas.}/HVL_{IEC}$ ratio) for the RQT series, for participating SSDLs. The acceptance limits are also shown.

The RQA qualities may also be difficult to establish due to the low air kerma rate exiting the high thickness of the required aluminium filters. In these cases, air kerma rate should be measured with chambers of appropriate sensitivity (e.g. 30cc volume). These also imply that RQA series establishment may need more effort and cost. If only one RQA quality is to be established, it should be RQA5.

The standard radiation qualities used at calibration laboratories as suggested in TRS No. 457 and IEC 61267 standards do not cover all the qualities encountered in clinical applications. Interpolations and extrapolations between the standard and clinical radiation qualities may be inaccurate if the response of the meter has strong energy dependence. This is seen particularly for KAP meters, mammography calibration and image intensifiers measurements.

3.1.2.3. Beam qualities not defined in TRS No. 457: Clinically useful qualities for mammography calibrations

Clinical X ray mammography systems may exhibit higher HVL values than indicated by the RQR-M series. Typically, HVL values are higher in clinical practice for similar target/filter combinations mainly because of the presence of the compression paddle in the beam. This is illustrated by comparing the typical clinical HVL values shown in Table 8.7 of TRS No. 457 with HVLs specified by the RQR-M series for equivalent tube potentials. Figure 7 further demonstrates this where the paddle is simulated by PMMA plates of 1, 2 and 3 mm thickness, which are equivalent to 0.05, 0.10 and 0.15 mm Al respectively.

It is clear that the use of some tube voltage and paddle thickness combinations result in HVL values that exceed the highest RQR-M series value (marked as dotted horizontal line in Fig. 7. Furthermore, other anode/filter combinations, increasingly used in digital mammography, might result even higher HVL values [16–18]. Therefore, it is concluded that IEC standard radiation qualities do not cover a sufficient range of HVL values for clinically used beam qualities and either extrapolation is needed or new beam qualities need to be applied[7].

Additionally, HVL values may vary considerably between systems, even of the same type and model. That is why tabulated HVL can only be recommended for general use if a greater uncertainty in the dose value is acceptable. The HVL should always be measured if possible. This is also recommended in TRS No. 457.

As suggested in TRS No. 457 and deduced from several studies [14, 15], reference class instruments that are designed for mammographic applications yield good response in different beams within 2–3%. Dosimeters having a flat energy response and with a calibration coefficient N_K for RQR-M2 could be safely used at all beam qualities

[7] Currently PTB offers mammographic calibrations for a broader range of target and filter materials at energies from 20 to 50 kV.

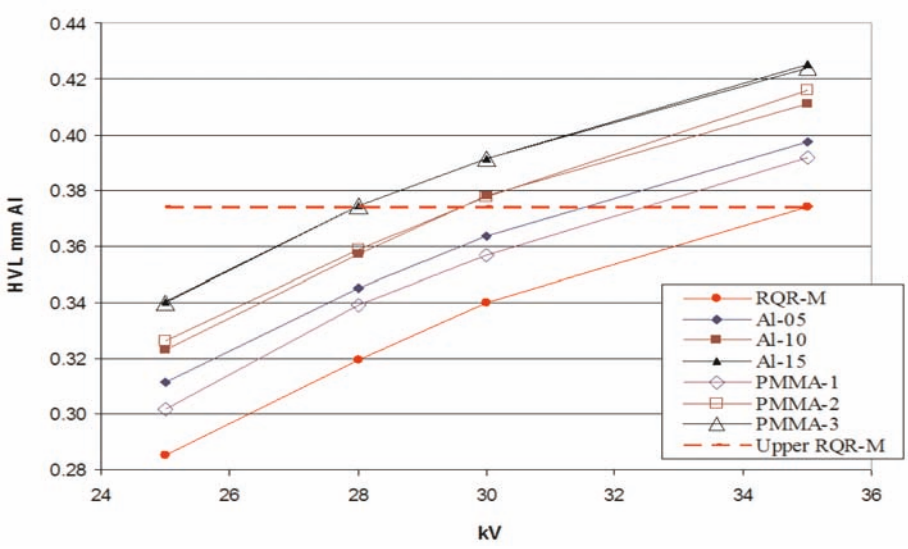

FIG. 7. Example of HVL values (mm Al) of the RQR-M qualities series and of beams with PMMA and Al additional filtration. The total filtration is for PMMA-1 : 30 μm Mo+1mm Al, PMMA-2 : 30 μm Mo + 2mm Al, PMMA-3 : 30 μm Mo + 3mm Al, Al-05 : 30 μm Mo + 0.05 mm Al, Al-10 : 30 μm Mo + 0.10 mm Al and Al-15 : 30 μm Mo+0.15 mm Al. The dotted line represents the highest HVL value of the RQR-M series beams. Data also shows the equivalence of PMMA and aluminium additional filtrations at all tube voltage (kV) range [14].

and paddle thicknesses. The introduced error is expected to be less than 5%. While some solid state detectors also appear to have a flat energy response due to real time corrections, less complex solid state dosimeters require manual correction through the use of an appropriate k_Q value.

3.1.2.4. Beam qualities not defined in TRS No. 457: Clinically useful qualities for KAP calibrations

The response of a typical KAP meter depends significantly on the spectra of the X ray beam [19, 20]. Typically RQR radiation qualities are used for diagnostic detector calibration at the SSDL; however, for calibration of KAP meters, they are not always sufficient. In order to achieve the accuracy level specified in Table 5.2 of TRS No. 457, a more comprehensive calibration may be necessary especially if the KAP meter will be used in X ray beams with added filters (e.g. for copper filtrations commonly used in fluoroscopy systems). The accuracy of KAP calibration at the laboratory can be improved if radiation qualities that are close to those used clinically are used by the SSDL. This is especially emphasized when portable KAP meters are calibrated at the SSDL. The portable KAP meters can be used either as a field instrument in the X ray units or as a reference instrument for calibration of the field KAP meters on-site.

For KAP meters, interpolation of calibration coefficients based on HVL alone as a radiation quality specifier may be insufficient and at least two radiation quality specifiers should be used [19] for this purpose, however this is non-trivial. Interpolation would be easier if calibrations were performed with a fixed filtration and a varying tube voltage [19]. The most suitable total filtration to be used for this purpose depends on the application and the required accuracy level. One possible option is to expand radiation qualities available at SSDLs by use of the filtrations of RQR and RQT qualities with adjusted high voltages. The usefulness of these qualities depends on the qualities used at clinics.

Typical relevant clinical radiation qualities are found in Appendix V. Typical clinically used total filtrations could be covered by using aluminium filtrations 2.5 mm–5 mm and aluminium filtration 3.5 mm– 4.5 mm Al together with copper filtration 0.1 mm–0.9 mm. Similarly, clinically relevant tube voltages varied from 50 kV to 150 kV. Based on the results in appendix V, aluminium filtrations close to 3 mm Al and 5 mm Al can be recommended. Additionally, copper filtration of at least 0.1 mm and 0.2 mm should be used together with ~4 mm Al filtration. With the additional copper filtration, the exact thickness of aluminium filtration is not important. Tube voltages of 50, 70, 90 and 120 kV are recommended. A table summarizing the recommended radiation qualities for calibration of KAP meters can be found in Appendix V.

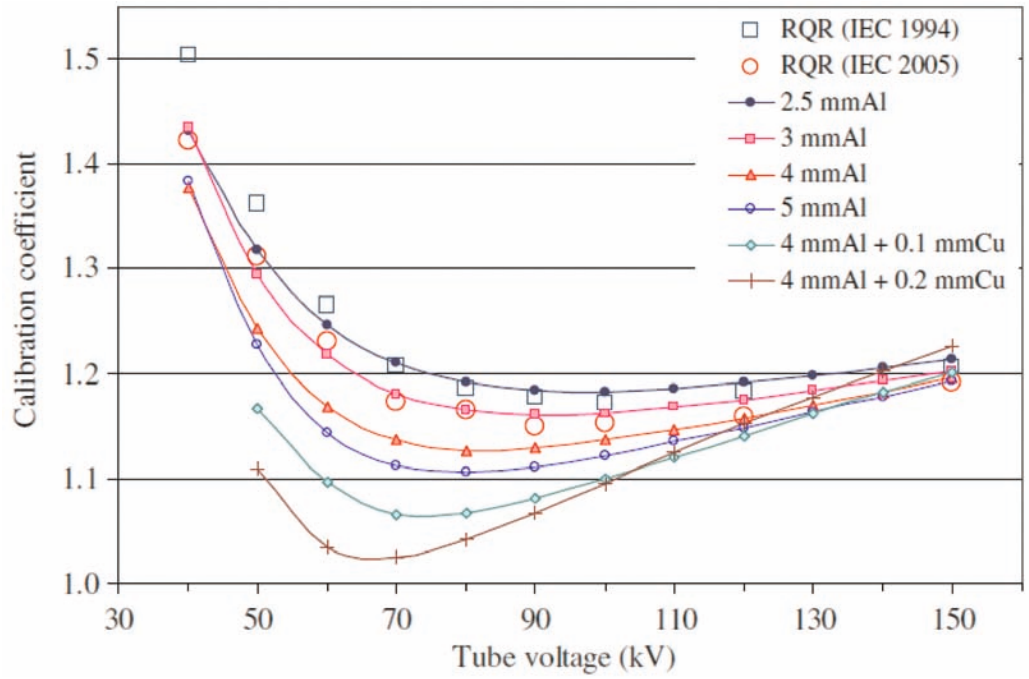

FIG. 8. Calibration coefficients for KAP meter with different tube voltages and filtrations. [19]

During the CRP, a novel KAP meter to be used as a reference instrument has been developed (patient dose calibrator, PDC, Radcal). The energy response of this instrument is markedly lower than that found with conventional KAP meters. This instrument is a promising tool for improving the accuracy of the calibration of a field KAP meter with a reference KAP meter [21].

3.1.2.5. RQC qualities

RQC qualities (3, 5 and 8) utilize copper filtration and have been developed for personnel who use additional copper filtration in the beam of fluoroscopic equipment to simulate the attenuation of a patient. This is commonly done by service engineers when adjusting the automatic brightness control system, and may also be used by physicists performing quality assurance measurements. It should be noted that these conditions bear little resemblance to clinical beam conditions experienced by image intensifiers and are not directly relevant to patient dosimetry. Therefore they were not referred to in TRS No. 457. However, calibration at these qualities may be asked from end users especially from companies which provide installation, maintenance, or service of X ray systems.

3.1.3. Uncertainties in the measurement of HVLs in establishing the diagnostic radiology qualities at the SSDLs

This section describes the uncertainties for the established diagnostic radiology qualities at SSDLs. Such standards concern, basically, the beam quality and, consequently, the X ray tube voltage (kV), the added filtration and the HVL measurements.

TRS No. 457 provides background and practical information about the calculation of uncertainties. In the IAEA CRP the participating SSDLs reported their uncertainty budget for the HVL measurements, as shown in Table 9.

The uncertainties at the different laboratories are very similar. However, there is no 'universal' uncertainty budget and different contributions have to be taken into account depending on the nature of the facility and on the type of instrument(s) involved in the procedures. SSDLs should be in a position to identify the sources of errors (influence quantities), evaluate them and quantify their influence to the result of a measurement/procedure.

TABLE 9. UNCERTAINTY IN HVL MEASUREMENTS

	Czech		Finland		Greece		Vietnam	
	A%	B%	A%	B%	A%	B%	A%	B%
X ray system								
kV stability of X ray system		0.3		0.5	0.01		0.5	
Accuracy of kV		0.15		0.8		0.29		
Stability of tube output — K_{air}	0.4			—	0.01		0.3	
Field homogeneity		0.3				0.50		0.2
Dosimeter								
Energy dependence of chamber		0.55		0.5		0.50		1.0
Chamber stability	0.3				0.02		0.3	
Electrometer accuracy		0.002				0.50		0.2
Electrometer stability				0.06				
Scale reading resolution		0.01		0.01		0.00		0.1
Temperature and pressure	0.5		0.08		0.20	0.40	0.2	0.1
HVL measurements								
Thickness of Al filters	0.2			0.6		0.50		0.2
Scatter radiation		0.2		0.8	0.10			0.2
Reproducibility of measurements	0.3				0.05		0.3	
Calculations of HVL		0.3		0.5		0.50		0.5
Quadratic sum	0.85	0.74	0.08	1.66	0.55	1.11	0.56	1.43
Combined uncertainty	1.13	%	1.32	%	1.24	%	1.41	%
Expanded uncertainty	2.3	%	2.64	%	2.49	%	2.82	%

3.1.4. Quality management system

The objectives of a SSDL should be:

— The establishment and maintenance of radiation dosimetry standards at a national level at an appropriate professional quality;
— The calibration and testing of dosimetry equipment;
— Irradiation services to users/customers.

A quality management system (QMS) is a essential in achieving these objectives and should:

— Ensure that all procedures and services are of high quality and consistent to scientific and internationally recognized standards;
— Ensure that all laboratory activities meet reasonable requirements of the client;
— Address management issues involved in day to day activities;

TABLE 10. STRUCTURE OF A QUALITY MANAGEMANT SYSTEM

Level	Subjects — information
LEVEL 1: Quality manual	• Quality system brief description • Policy of the laboratory • Commitment of the manager, staff for QA • Organization chart • Reference to written procedures, Job description
LEVEL 2: Job descriptions	• Who does this • Profession, Training and Experience requirements
LEVEL 3: Procedures	• Written procedures in detail • What, how, who and when • Reference to laboratory policy and ISO Std • Reference to working Guidelines & Instructions and ISO Standards
LEVEL 4: Working instructions	• Written description in detail of all quality control & test methods, calibration and procedures
LEVEL 5: Records and Files	• All necessary records, files and completed forms that support a procedure are kept and stored. • Evidence that QS is set and operates satisfactory
LEVEL 6: Documentation standard	• Any standard document, directive, scientific paper and guideline used in QS (e.g. legislation, ISO, IEC, IAEA, etc)

— Comply with legislative and regulatory requirements;
— Ensure that scientific standards have a central role at all levels of management and in various activities.

SSDLs are strongly recommended to adopt and set a QMS in accordance to ISO/IEC 17025[8].

3.1.4.1. Structure of QMS documentation

The documents of the QMS may be structured in six levels as shown in Table 10.

The documents of each level should provide all necessary information in order that the QMS is understandable, applicable, realistic and effective.

The **quality manual** gives a general overview of the QMS. The quality policy of a SSDL is a statement and a commitment, which determines the objectives and the goals of the laboratory to become and to be maintained as a laboratory of a high quality at an international level. Through the quality policy, the SSDL should declare and commit to:

— The high quality of activities and services;
— The high quality of equipment, including its calibration, maintenance, service etc;
— The traceability to scientifically recognized international primary standards;
— The provision of steering mechanisms, internal and external audits, quality control as well as the participation in intercomparisons;
— The scientific, reliable and high quality work of laboratory staff as well as its independency of any kind of financial, commercial and other kind of pressure and influence;
— The continuous education and training of laboratory staff;
— All possible efforts to meet the client’s requirements and to examine and resolve their complaints;
— All necessary provisions, including human and financial resources, in order to continually apply and improve the QMS in accordance to ISO standards.

[8] The ISO 9000:2000-series standards (alone) focus more on service procedures rather than the quality of services (outcome) and therefore, may be insufficient for SSDLs.

Job descriptions should provide details of the responsibilities and the authority of the staff. All jobs are distributed between staff members .

Jobs include any activity of the SSDL (scientific, technical, administrative, etc.). For example:

— Policy planning;
— Determination and approval of the annual budget;
— Approval of actions and activities;
— Participation in international organizations;
— Establishment;
— Development and maintenance of dosimetry standards;
— Issue of calibration certificates;
— Handling client's complaints;
— Record keeping;
— Participation in training of staff;
— Housekeeping, etc.

The assignment of a job to a certain staff member should be done according to predefined requirements and criteria. Also, certain predefined criteria should be applied for the recruitment of personnel for a given position.

The **procedures and working instructions** describe in detail the what, who, how and when should be done in order for the QS to operate satisfactory. They should refer to management and administrative issues; services; customer and third party issues; and technical support of the QMS. The idea should be to 'write what you do' but also 'do as you write'.

The staff would have a description of activities, and instructions to ensure that their work is carried out as required. A continuous improvement and homogenization of the job may be performed. Also, there would be evidence of the work performed for each client or activity.

Each **procedure or working instruction** may include the following:

— Introduction, scope and other issues that the procedure refers to;
— Method description in detail, basic scientific principles, tools and actions that should be followed as well as any precautions, and safety issues;
— Equipment to be used;
— Staff and their responsibilities;
— Record keeping details of all the records, files, electronic files, and working forms that are used;
— Working forms;
— Working instructions: detailed description of all steps to carry out a certain task (measurement, calibration, test, etc);
— Calculations, presentation of the results, certificates;
— References.

3.1.4.2. QMS application

The application of the QMS and its performance should be monitored through the use of qualitative and quantitative measures. These may include:

— Results from international intercomparisons and scientific projects. An intercomparison is the most reliable, independent and effective way to check the overall performance of an SSDL activity. SSDLs are strongly recommended to participate in intercomparisons run by international bodies (IAEA, regional metrology organizations, etc). Also, SSDLs may organize scientific projects (bilateral or multilateral) in order to verify their performance;
— Results from internal and external audits and inspection;
— Predefined indicators. For example, the number of identified non-conformities, the degree of the implementation (completion) of corrective/preventive actions, the number of client complaints, the client's satisfaction (from questionnaires).

A QMS needs human resources and staff availability in order to be effective and accurate. It is estimated that more than 30% of the time may be spent running a QMS, including all QC and QA of the SSDL. The SSDL should organize the QMS according to its policies, objectives, scopes and needs, to be a realistic and useful tool for improvement of the SSDLs activities and services.

3.2. CODE OF PRACTICE FOR DIAGNOSTIC CALIBRATIONS AT SSDLS

3.2.1. Introduction

Under Activity 2 of the CRP, the participating SSDLs should develop laboratory procedures and establish the uncertainty budget for the calibration of user dosimeters as described in Appendix I and II of TRS No. 457.

The intention of the CRP for this activity was to assess the degree of implementation of the quality system for diagnostic radiology dosimeter calibrations within the SSDLs. While this could have been done through an audit visit at each participating laboratory, considering the number of participants in the project, such a visit would have been unpractical due to time and cost constraints. Therefore, participants were requested to present a list of technical procedures developed for the implementation of the diagnostic radiology beam energy calibration.

The results of these activities are presented in the next sections.

3.2.2. Calibration procedures

The implementation of a calibration laboratory for diagnostic radiology is well detailed in Chapters 6 and 7 of TRS No. 457. A laboratory that follows these recommendations is in a good position to perform calibrations accordingly.

Calibration procedures were developed and published by five SSDLs. Table 11 lists the procedures developed by the participants.

Comments on calibration procedures:

- While it is sometimes convenient for intercomparisons to calibrate a diagnostic radiology detector alone or a read out device alone, it is not always feasible due to large connector variability.
- The calibration interval of the secondary standard could be extended beyond the suggested time of 2 years when its stability is controlled properly; including provision of documented procedures, provided this is consistent with national legislation.
- Some old CT dosimetry equipment has a readout in mR or mGy. This equipment should not be withdrawn from use if it is functioning satisfactorily apart from their inappropriate units. The SSDL may, upon the user's request, provide a calibration factor in terms of P_{KL} per reading (e.g. mGy mm/digit).

Ideally the calibration coefficient should be determined at the PSDL for the actual radiation qualities used in the SSDL. The SSDL should use exactly the same radiation qualities used in the calibration of their reference at the PSDL in order to avoid interpolations of the calibration coefficient of the SSDL reference instrument. However this is not always possible or practical as some PSDL do not provide calibrations for the standard IEC radiation qualities with exact matching of HVL and filtration. Some differences in the measured HVLs exist even with a similar set-up of the radiation beams, and the reliability of the calibration factor of the reference chamber of an SSDL can be improved through interpolation by HVL.

While in TRS No. 457 guidance is given to SSDLs on how to match radiation qualities, currently there is no guidance for SSDLs on the matching of the radiation qualities between the PSDL and the SSDL, and on the acceptable interpolations of the calibration coefficient of the reference instrument. by The CRP participants concluded that the interpolation, but not the extrapolation, of the calibration coefficient of the reference instrument of an SSDL relative to radiation beam quality is acceptable without losing the traceability. The uncertainties of these interpolations shall be included in the overall uncertainties of the measurements.

TABLE 11. PROCEDURES DEVELOPED BY THE SUPPORT FROM ACTIVITY 2

Code	Title	Release date	Revision
	Brazil		
PE2B019	Diagnostic radiology ionization chamber calibration	30/May/2008	0
PE2B020	Diagnostic radiology ionization chamber calibration — Certificate generation	30/May/2008	0
PE2B021	Diagnostic radiology laboratory staff training	30/May/2008	0
	Cuba		
PR/LSCD/035 version 3.0	Calibration of diagnostic dosimeters in conventional X ray beams	November 2007	In revision
	Note: The procedure is in revision for improvement and to introduce the data from the new calibration of the reference standard.		
	Finland		
DOS4.1.7a	Calibration of a diagnostic air kerma meter	9/July/2008	2
	Greece		
	Calibration of dosimeters used in diagnostic radiology	15/November/2001	2.3
	Calibration of kV meters used for non invasive measurements of high voltage in diagnostic radiology	30/November./2001	2.2
	IAEA		
DOLP.003	Maintenance of the secondary standard dosimetry system for mammography	1/January/1999	2005-02-04
DOLP.013	Calibration service for mammography ionization chambers	1/January/1999	2005-04-11
DOLI.0301	Half-value layer (HVL) determination for mammography beams	1/January/1999	2005-02-23
DOLI.3104	Operation and maintenance of the mammography X ray unit	14/October/2003	2007-09-19
DOLF.1301	A typical calibration certificate for mammography calibrations		
DOLF.1302	Summary of measurements for mammography calibrations		

3.2.2.1. Calibration of KAP meters

In this CRP project, calibrations of KAP meters were performed in SSDLs of Finland and Greece. Calibrations were performed for both incident and transmitted beams. The description of the calibration procedure in TRS No. 457 is short but clear and is easy to apply. The calibrations were performed with RQR radiation qualities according to TRS No. 457. In some cases, an extended range of radiation qualities were also used. Results of calibrations for RQR radiation qualities are summarized in Fig. 9. Typically, the KAP chamber attenuated the beam 10–20% but one chamber had much higher attenuation effect of 20–30%. The attenuation effect of KAP chambers is presented in Fig. 10.

Figure 9 shows that the KAP meter calibration factor varies rapidly with HVL for values of HVL of 2 mm and less. This means that any uncertainty in measured HVL could give rise to a large uncertainty in the calibration coefficient, and this should be accounted for in the resultant uncertainty budget for the calibration.

The effect of field size is integral for KAP measurements, however its effect on calibration is complex, as calibrating beams are generally not of uniform intensity particularly over larger areas. TRS No. 457 does not

recommend the use of different field sizes in KAP calibration. Field size dependence was tested at two sites. At one site, a Diamentor M4 KAP meter was used with three circular apertures (diameter ranging from 40 mm to 80 mm) and the difference in calibration coefficient was under 0.3%. In some routine calibrations, a 1.5% difference was measured for different field sizes. A second site used aperture diameters of 38 mm, 76 mm and 98 mm resulting in a calibration coefficient range of about 3%.

In TRS No. 457, a calibration method for a transmitted beam is described. This calibration method is needed for calibration of a field KAP at a SSDL, however it is not recommended as field KAP meters should ideally be calibrated in situ with either a calibrated dosimeter or a reference KAP meter. This is discussed in more detail in Section 5.1.4 of this report.

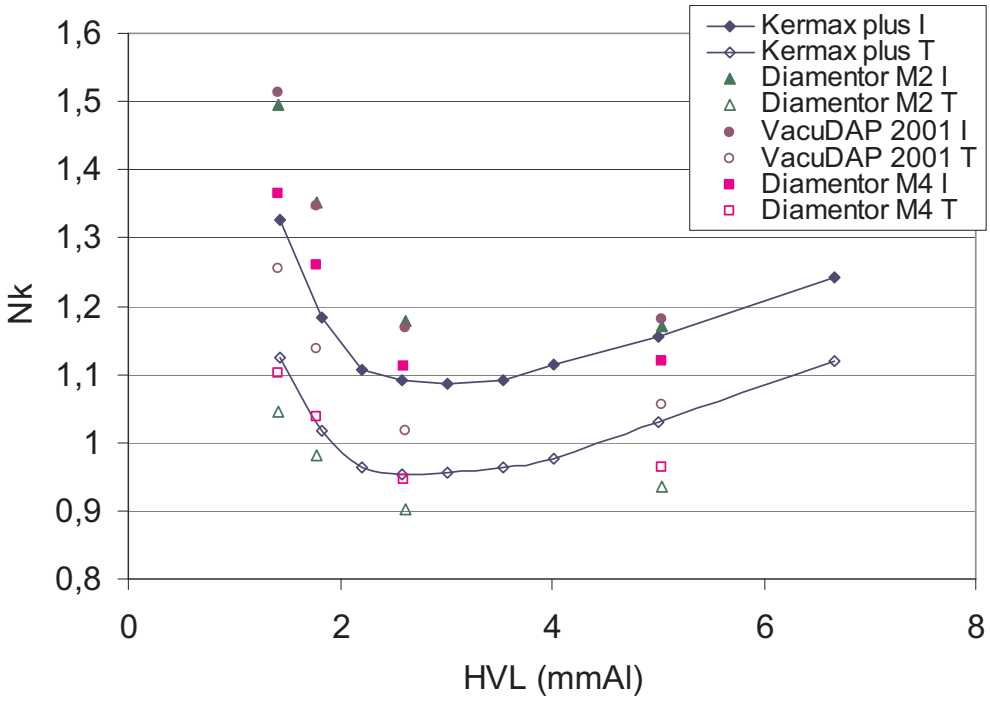

FIG. 9. Calibration coefficients (Nk) for KAP meters calibrated in SSDLs for transmitted (T) and incident (I) beams.

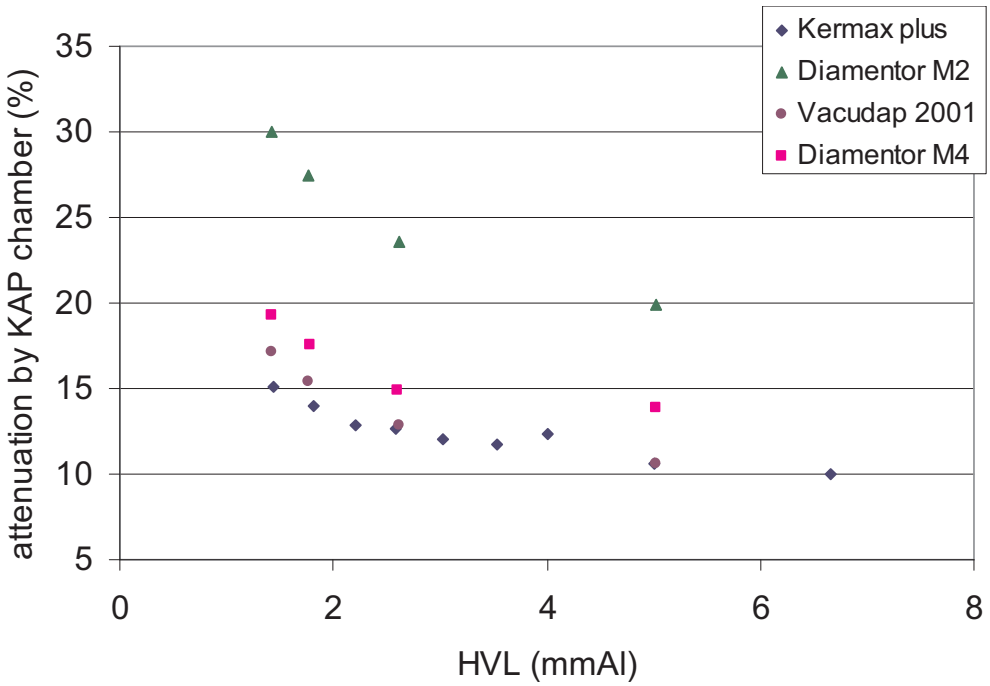

FIG. 10. Attenuation by KAP chambers calibrated in SSDLs.

3.2.3. Uncertainty

3.2.3.1. Evaluation of uncertainties

Section 6.7 and Appendices I and II of TRS No. 457 have comprehensive information about uncertainty evaluation. The SSDL participants have prepared their laboratory uncertainty budget according to those instructions and IAEA-TECDOC-1585. The countries involved in this activity were: Brazil, Cuba, the Czech Republic, Finland, Greece, IAEA, Thailand, and Vietnam. The budgets that were developed are presented in Appendix VIII.

Table 12 presents an explanation of the origin of each component. The uncertainty components listed were taken from different laboratories. As outlined in TRS No. 457, there is no 'universal' uncertainty budget and different contributions may have to be taken into account depending on the nature of the facility and on the type of instrument(s) involved in the calibration.

The uncertainty budget is an indication of the ability of a laboratory to perform measurements. The development of an uncertainty budget evolves with improvements in instrumentation and procedure quality within the installation. Comparison exercises are frequently used to assess laboratory capability to perform measurement and evaluate uncertainty. While it is relatively easy to identify the factors of uncertainty in one's measurements, it is more difficult to quantify these, highlighting the intrinsic lack of certainty about uncertainty.

TABLE 12. ORIGIN OF UNCERTAINTY BUDGET COMPONENTS FOR CALIBRATION OF RQR METERS

Uncertainty component		Type*	Origin of component
Symbol	Name		
	Measurements with reference chamber		
N_K	Calibration of reference chamber	B	From calibration certificate k=1
k_{stab}	Stability of ref. ion. chamber	A/B	Long term stability measurement SD over the mean or max. dev.
M_{raw}	Repeatability of the Ref. chamber	A	SD over the mean of the measurements (reference chamber)
k_s	Saturation/recombination correction	B	Usually negligible
k_{leak}	Leakage current	B	Less than 0.1% of the signal. Assess the U for two options: 1. subtract from signal 2. use the max. limit (0.1%)
k_{dist}	Chamber positioning	B	deviation of chamber position from the reference position
k_{elec}	Electrometer calibration	B	From electrometer certificate k=1
$k_{elec\text{-}res}$	Electrometer resolution	B	Can be combined with the raw readings of the chamber. Usually a small contribution (0.03%)
$k_{t,p}$	Air density correction for T and P	A	SD over the mean of the measurements
	T and P cal. factors	B	From thermometer and barometer certificates k=1
k_{Q,Q_0}	Difference in beam quality (from calibration laboratory)	B	Effect of any difference between the qualities of the beams at the SSDL and at the IAEA/PSDL.
k_{time}	Timing	A	If the relative timing uncertainty is expected to be larger than 0.1%, it should be included.
k_{fs}	Departure of the field size from the reference condition (inhomogeneity, uniformity)	B	Chamber size must be taken into account to evaluate component

TABLE 12. ORIGIN OF UNCERTAINTY BUDGET COMPONENTS FOR CALIBRATION OF RQR METERS (cont.)

Uncertainty component		Type*	Origin of component
Symbol	Name		
	Measurements with user's instrument (chamber and electrometer)		
M_{raw}	Repeatability of the user instrument	A	• Same origin as those from the reference chamber, but reflecting measurements with user chamber.
k_s	Saturation correction.	B	• These standard uncertainty contributions are summed in quadrature to obtain the combined standard uncertainty for the measurement.
k_{res}	Instrument resolution (user)	B	
k_{leak}	Leakage current(user)	A	• The expanded uncertainty is then obtained on multiplying the standard uncertainty by a coverage factor k.
k_{dist}	Deviation from ref. distance	B	
$k_{t,p}$	Air density correction for T and P	A	
	T and P cal. Factor	B	
k_{time}	Timing	A	
k_{elec}	Electrometer calibration	B	
$k_{elec-res}$	Electrometer resolution	B	

* Note: In theory, each process or component of uncertainty will contain both type A and B uncertainties [22]. However, when one type is considered negligible it was omitted from this table.

3.2.3.2. Evaluation of uncertainties for KAPs

Uncertainties in KAP calibrations were estimated by participants. Uncertainty (k=2) for KAP calibration in the SSDL was 3–5%. An example for uncertainty estimation is presented in Table 13.

TABLE 13. UNCERTAINTY EVALUATION FOR CALIBRATION OF KAP METER IN SSDL [23]. NOTE THAT DIFFERENT TYPE OF AIR KERMA METER WAS USED AS A REFERENCE INSTRUMENT.

	Relative uncertainty (%) $k = 2$	
Source of uncertainty	Reference KAP meter for the tandem method	Field KAP meter for the laboratory method
Calibration coefficient of the air kerma meter	1.6	3
Difference between the energy spectra used in the air kerma and KAP calibrations	1.2	1.2
Reading of the air kerma meter	0.8	1.2
Reading of the KAP meter	1.2	1.2
Distances d_K and d_A	0.5	0.8
Difference between the distances of $P_{KA,cal}$ and M_{ref} measurements	0.3	0.3
Aperture area A of the KAP collimator	1	1
Deviations from the applied air density corrections	1	1
Short-term instability of the x-ray beam	0.5	0.5
Inhomogeneities of the x-ray field	0.6	0.6
Estimated total relative uncertainty	3.0	4.1

3.3. EXPERIMENTAL COMPARISON OF RQR BEAMS BY MEANS OF IONIZATION CHAMBER EXCHANGE

3.3.1. General

This section provides a comparison of the calibration of two ionization chambers in RQR beams from various SSDLs. This comparison was conducted under activity 3 of the CRP. Seven SSDLs (Brazil, the Czech Republic, Cuba, Finland, Greece, Thailand and Vietnam) as well as the IAEA participated.

The main task of this comparison was to test TRS No. 457 in respect to the calibration of dosimeters and to investigate the suitability of radiation qualities available at SSDLs for the calibration of dosimeters used in conventional diagnostic radiology. The second task of this activity was to compare the calibration results among the participating laboratories.

The calibration of the ionization chambers was carried out in the participating SSDLs November 2007–2009. During this period, the initial calibration of the chambers, a recalibration and a final calibration were performed at the SSDL in Greece, in order to check the stability of the chambers.

3.3.2. Ionization chambers and calibration in terms of N_k

The IAEA provided two A3 Exradin ionization chambers (s/n XR 071832 and XR 072321). The chambers are spherical with 3.6 cc active volume. The reference point of the chamber is taken to be the geometric centre of the active cavity volume (sphere).

The chambers were connected to the SSDLs electrometer and the calibration coefficients at RQR beam qualities were obtained by cross calibration of these chambers against the SSDLs reference standards (substitution method).

In general, the SSDLs applied the following method for the calibration.

— The chamber for calibration was mounted at 1 m distance from the focal spot.
— A bias 300 V was set and some pre-irradiation was given.
— Leakage current was measured.
— Measurements for RQR qualities were performed.
— Measurements were also performed for reference dosimeter of the SSDL.
— The calibration coefficient is calculated by

$$N_K = \frac{K_{i,ref}}{M_{A3,cor}} = \frac{M_{ref} \cdot N_{ref}^{RQR5} \cdot k_{Q,ref} \cdot k_{Tp}}{M_{A3} \cdot k_{Tp}}$$

and is referred to 20.0°C and 101,325 kPa.

3.3.3. X ray apparatus and beam qualities

The X ray systems used in the participating SSDLs for the chamber calibrations are those described in Table 13. The reference instruments used are given in Table 14.

3.3.3.1. Specific calibration methods

The following calibration methods were applied by the participating SSDLs.

Electrometer use/mode

Brazil: Charge collected over 60 seconds (10 times)

Cuba: Charge collected over 10 to 30 seconds depending on Kerma rate (5 times) and the whole procedure repeated 2 times

TABLE 14. REFERENCE INSTRUMENTS USED IN CALIBRATIONS BY THE PARTICIPATING SSDLs

	Brazil	Finland	Greece	Vietnam	Thailand	Cuba
Chamber type/ serial number	Radcal 20X5-3/20647	PTW 23344/0947 and NE 2561/097 (RQR 9)	PTW 77337	Exradin A4 REF 92715	Exradin A5 REF 92724	Radcal 10X5-6/ 16376
Electrometer type/ serial number	Keithley 6517A/1138780	Keithley 6517/ 629300	PTW Unidos 20314	PTW Unidos	PTW Unidos	PTW Unidos
Chamber sensitivity (calibration sensitivity)	1.032E + 07 Gy/C		4E+08 C/Gy			
Measured air kerma/ air kerma rate	0.37 – 0.43 mGy/s	0.5 mGy/s	0.3 – 1.2 mGy/s		~0.31 mGy/s	0.3 – 1.2 mGy/s
Traceability of standards (calibration date and place)	PTB (2007-11-26)	PTW 23344 (02/06) PTB (RQR 2-7) NE 2561: (02/04) BIPM (BIPM 100-250)	PTW, (30/04/2005)	PTB (through Greece)	PTB (through Greece)	PTB (08/01/2008)

Finland: Charge collected over 5 sec (10 times) and the whole procedure repeated 5 times
Greece: Charge collected over 60 seconds (10 times)
Vietnam: Charge collected over 60 seconds (5 times) and the whole procedure repeated 3 times

Air Kerma rates:

Brazil: 0.37 – 0.43 mGy/s
Cuba: 0.31 – 1.19 mGy/s (depending on RQR quality)
Finland: 0.5 mGy/s
Greece: 0.3 – 1.2 mGy/s (depending on RQR quality)
Vietnam: ~ 0.5mGy/s
Thailand : 0.31 mGy/s
Therefore, all participants used about the same air kerma rates.

Method applied

All participants applied the substitution method. The deduced N_K values refer to reference environmental conditions. The calibrations were performed at 1 m distance from the X ray focus.

Formalism for the determination of N_K

Brazil:

$$N_K^{User} = \frac{\dot{K}_{a,Q}}{M_{corr}^{User}} \cdot k_{Q,Q_0},$$

where k_{Q,Q_0} is the difference in beam qualities from the PSDL and the SSDL.

Finland, Greece:

$$N_K = \frac{M_{ref} \cdot N_{ref} \cdot k_{Tp}}{M_K \cdot k_{Tp}}.$$

Calibration performed (reference point) at 1m distance from the tube focus.

Uncertainties

The uncertainty evaluation was based on the Guide to the expression of uncertainty in measurements (ISO 1995). The expanded U referred to 95% confidence level.

3.3.4. Results

Stability of measurements

Greece has calibrated the transfer chambers twice (initial calibration in October–November 2007 and recalibration in October 2008). The results, as well as the mean values and the differences between calibrations, are presented in Tables 15 and 16.

TABLE 15. CALIBRATION OF TRANSFER CHAMBER A3 EXRADIN XR 071832

			1/11/2007		6/10/2008		mean		% differ
Code	kV	HVL	Nk	u	Nk	u	Nk	u	
		mm Al	mGy/nC	%	mGy/nC	%	mGy/nC	%	
RQR2	40	1.44	8.160	2.50%	8.156	2.50%	8.1576	1.77%	–0.05%
RQR3	50	1.81	8.126	2.50%	8.126	2.50%	8.1259	1.77%	0.01%
RQR4	60	2.20	8.089	2.50%	8.089	2.50%	8.0890	1.77%	–0.01%
RQR5	70	2.58	8.093	2.50%	8.072	2.50%	8.0827	1.77%	–0.26%
RQR6	80	2.95	8.088	2.50%	8.074	2.50%	8.0812	1.77%	–0.17%
RQR7	90	3.49	8.088	2.50%	8.065	2.50%	8.0765	1.77%	–0.28%
RQR8	100	3.98	8.085	2.50%	8.070	2.50%	8.0774	1.77%	–0.18%
RQR9	120	4.98	8.096	2.50%	8.086	2.50%	8.0910	1.77%	–0.12%
RQR10	150	6.61	8.104	2.50%	8.075	2.50%	8.0893	1.77%	–0.36%

The uncertainty of the mean was calculated by

$$u = \frac{1}{\sqrt{\sum_1^n \frac{1}{u_i}}},$$

where u_i are the uncertainties at each calibration. The results showed that calibrations were consistent with each other. The differences were less than 0.5% at all RQR beams. This consistency also proves that the chambers' response had not been changed with time or due to their transportation in different countries.

TABLE 16. CALIBRATION OF TRANSFER CHAMBER A3 EXRADIN XR 072321

			25/10/2007		6/10/2008		mean		% differ
Code	kV	HVL	N_k	u	N_k	u	N_k	u	
		mm Al	mGy/nC	%	mGy/nC	%	mGy/nC	%	
RQR2	40	1.44	8.265	2.50%	8.298	2.50%	8.2815	1.77%	0.40%
RQR3	50	1.81	8.219	2.50%	8.256	2.50%	8.2376	1.77%	0.46%
RQR4	60	2.20	8.189	2.50%	8.203	2.50%	8.1958	1.77%	0.17%
RQR5	70	2.58	8.173	2.50%	8.173	2.50%	8.1726	1.77%	0.00%
RQR6	80	2.95	8.178	2.50%	8.172	2.50%	8.1753	1.77%	–0.08%
RQR7	90	3.49	8.161	2.50%	8.164	2.50%	8.1626	1.77%	0.03%
RQR8	100	3.98	8.168	2.50%	8.170	2.50%	8.1688	1.77%	0.03%
RQR9	120	4.98	8.161	2.50%	8.187	2.50%	8.1744	1.77%	0.32%
RQR10	150	6.61	8.174	2.50%	8.180	2.50%	8.1772	1.77%	0.07%

The mean value of the two calibrations will be used hereafter for the evaluation of the results from all participants.

The uncertainty U_{stab} for the stability of each instrument was derived from these data using the equation (at 95% confidence level)

$$U_{stab} = 2 \cdot \sqrt{\frac{\sum_{i}^{m} s_i^2}{m}}$$

with s_i being the standard deviation of the calibration coefficients for the radiation quality i and m being the total number of radiation qualities used for the stability check. The U_{stab} was calculated as 0.28% for the XR071832 and 0.36% for the XR072321 chamber.

Stability with ‘check source’

Since the above stability checks using X ray systems and RQR qualities include any variation on the X ray system output, the Cs137 irradiator (STS - OB6) was also used as a ‘check source’ for the stability of the chambers. The chambers were irradiated under fixed geometry and corrections for decay have been applied. The SD % of the measurements were 0.07% and 0.08% for the XR071832 and the XR072321 respectively.

Calibration coefficients

The calibration coefficients from all SSDLs for the two A3 Exradin chambers (XR 071832 and XR 072321) are presented in Tables 17 and 18.

TABLE 17. CALIBRATION COEFFICIENTS FROM ALL SSDLS: A3 EXRADIN XR 071832

		Greece				Finland				Brazil				Vietnam				Cuba			
Code	kV	HVL	N_k	u		HVL	N_k	u		HVL	N_k	u		HVL	N_k	u		HVL	Nk	u	
		mm Al	mGy/nC	%	k_Q	mm Al	mGy/nC	%	k_Q	mm Al	mGy/nC	%	k_Q	mm Al	mGy/nC	%	k_Q	mm Al	mGy/ nC	%	k_Q
RQR2	40	1.44	8.1576	1.77	1.009																
RQR3	50	1.81	8.1259	1.77	1.005	1.77	8.081	2.30	1.013	1.77	8.144	1.70	1.008	1.81	8.179	2.60	1.005	1.75	8.100	1.24	1.005
RQR4	60	2.20	8.0890	1.77	1.001									2.2	8.171	2.60	1.004				
RQR5	70	2.58	8.0827	1.77	1.000	2.77	7.980	2.30	1.000	2.6	8.083	1.70	1.000	2.58	8.141	2.60	1.000	2.60	8.058	1.24	1.000
RQR6	80	2.95	8.0812	1.77	1.000									2.95	8.167	2.60	1.003				
RQR7	90	3.49	8.0765	1.77	0.999									3.49	8.179	2.60	1.005				
RQR8	100	3.98	8.0774	1.77	0.999																
RQR9	120	4.98	8.0910	1.77	1.001	5.03	8.019	2.30	1.005	5.05	8.080	1.70	1.000					4.97	8.047	1.24	0.999
RQR10	150	6.61	8.0893	1.77	1.001																

TABLE 18. CALIBRATION COEFFICIENTS FROM ALL SSDLS: A3 EXRADIN XR 072321

		Greece				Finland				Brazil				Vietnam				Cuba			
			mean																		
Code	kV	HVL	N_k	u		HVL	N_k	u		HVL	N_k	u		HVL	N_k	u		HVL	Nk	u	
		mm Al	mGy/nC	%	k_Q	mm Al	mGy/nC	%	k_Q	mm Al	mGy/nC	%	k_Q	mm Al	mGy/nC	%	k_Q	mmAl	mGy/nC	%	k_Q
RQR2	40	1.44	8.2815	1.77	1.013									1.45	8.317	2.60	1.011				
RQR3	50	1.81	8.2376	1.77	1.008	1.77	8.157	2.30	1.014	1.77	8.252	1.70	1.008	1.8	8.254	2.60	1.004	1.75	8.161	1.24	1.006
RQR4	60	2.20	8.1958	1.77	1.003									2.19	8.236	2.60	1.001				
RQR5	70	2.58	8.1726	1.77	1.000	2.77	8.046	2.30	1.000	2.6	8.186	1.70	1.000	2.6	8.224	2.60	1.000	2.60	8.110	1.24	1.000
RQR6	80	2.95	8.1753	1.77	1.000									2.96	8.235	2.60	1.001				
RQR7	90	3.49	8.1626	1.77	0.999									3.51	8.269	2.60	1.005				
RQR8	100	3.98	8.1688	1.77	1.000																
RQR9	120	4.98	8.1744	1.77	1.000	5.03	8.082	2.30	1.005	5.05	8.167	1.70	0.998					4.97	8.080	1.24	0.996
RQR10	150	6.61	8.1772	1.77	1.001																

Figures 11–14 show the calibration results (calibration coefficients N_K and k_Q values) and the associated uncertainties (at 95% confidence level) for all beam qualities at each participating SSDL.

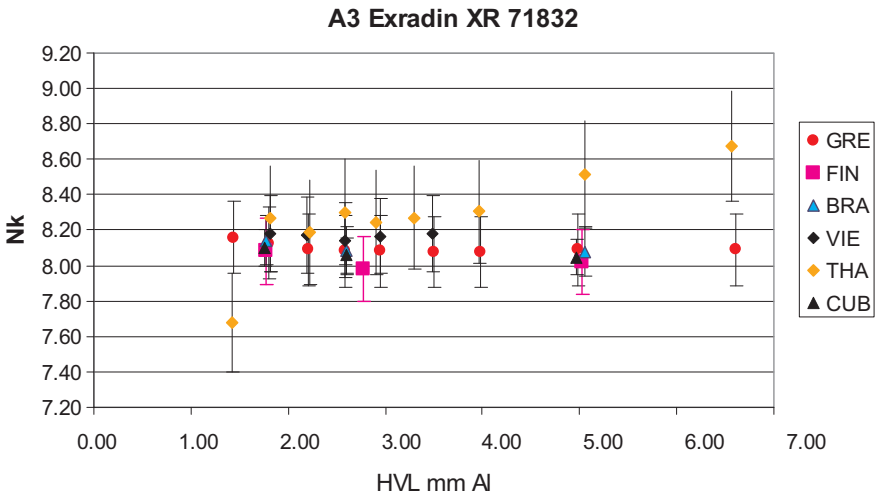

FIG. 11. The calibration coefficients N_K data RQR beam qualities of A3 Exradin XR 71832 chamber as deduced by the participating SSDLs. The error bars correspond to the expanded uncertainties at coverage factor of 2 (k=2, about 95% confidence level)

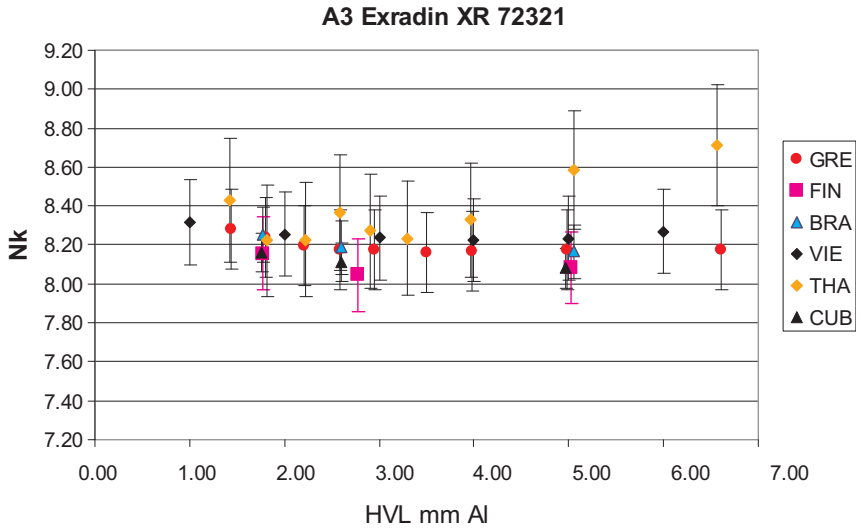

FIG. 12. The calibration coefficients N_K at RQR beam qualities of A3 Exradin XR 72321 chamber as deduced by the participating SSDLs. The error bars correspond to the expanded uncertainties at coverage factor of 2 (k=2, about 95% confidence level).

The k_Q values were calculated as the ratio of the N_K at a given quality to the N_K at RQR5.

The k_Q values for each chamber, SSDL and RQR quality are presented in Figures 13 and 14. For the clarity of the graphs, the uncertainties were not shown.

The HVL value at a certain RQR quality differed slightly between SSDLs. In order to evaluate the influence of this HVL difference, the following method was applied. The curve k_Q vs. HVL for GRE data was plotted, with k_Q to be the mean value of the two calibrations (initial and recalibration) of the XR 071832 A3 chamber. Then, the k_Q of the chamber was found by interpolation at the HVL value at each RQR of each SSDL. The difference between k_Q at GRE HVL and the k_Q at the SSDL HVL is a measure of the influence of HVL difference to the calibration coefficients. The influence of the different HVL values between SSDLs during calibrations was found to be insignificant.

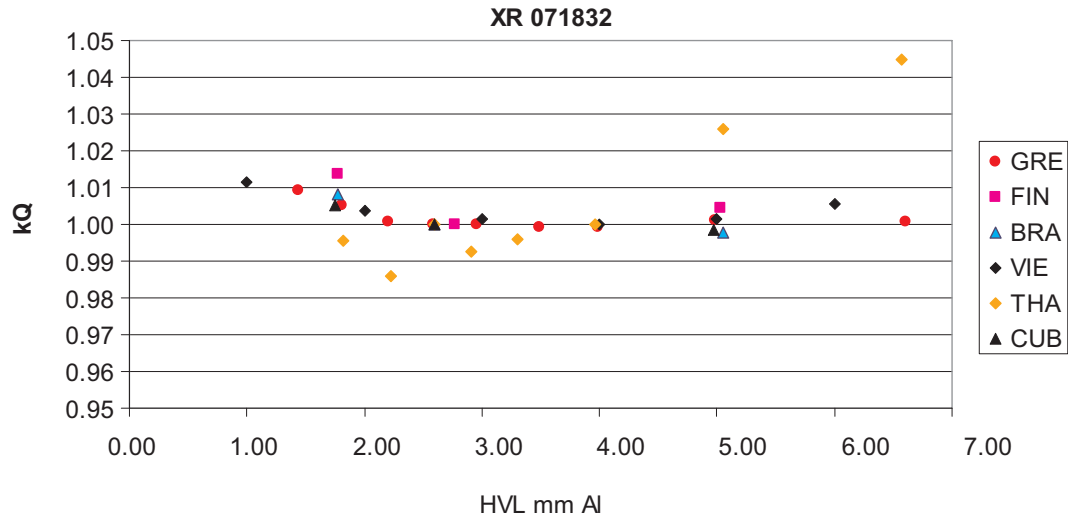

FIG. 13. The k_Q values at RQR beam qualities for the A3 Exradin XR 71832 chamber as obtained from the calibration at the participating SSDLs.

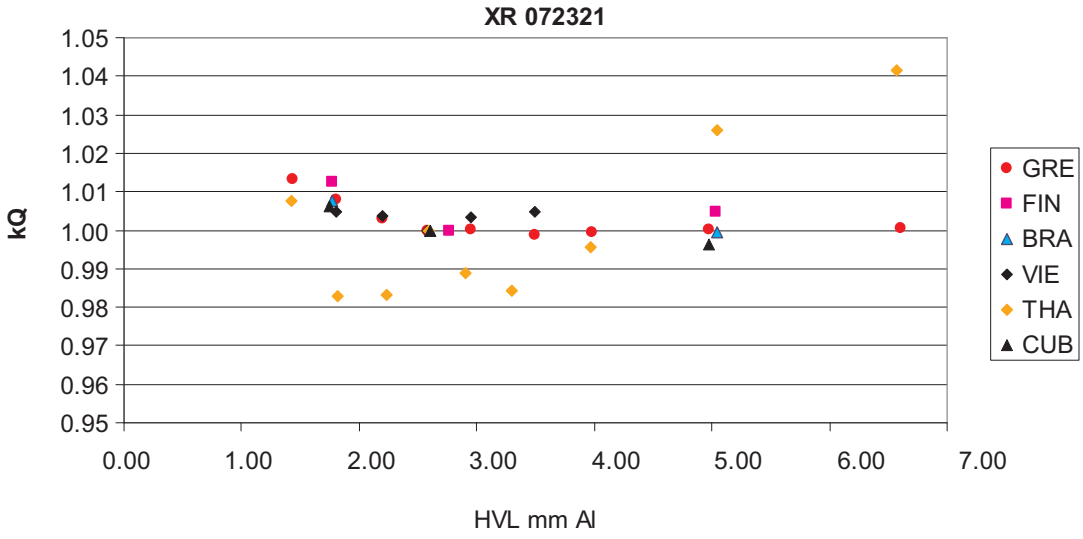

FIG. 14. The k_Q values at RQR beam qualities for the A3 Exradin XR 72321 chamber as obtained from the calibration at the participating SSDLs.

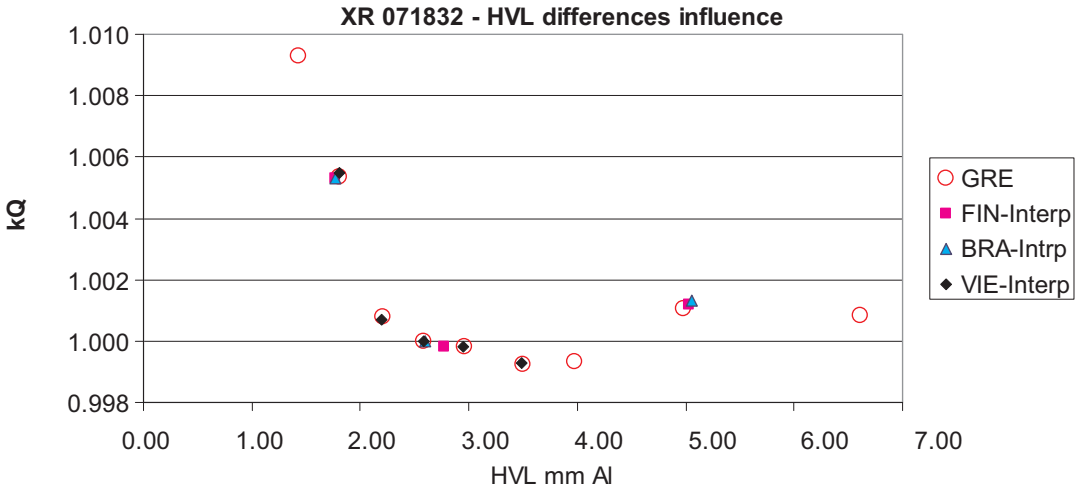

FIG. 15. The k_Q of the A3 Exradin XR 71832 chamber as deduced from interpolation at the actual HVL value at each RQR of each SSDL.

The N_K values from each participant at each RQR were compared to the mean value of all N_K (at the same RQR quality) and the ratio

$$\left(\frac{N_K^{SSDL}}{N_K^{mean}} \right)_{chamber}^{RQR}$$

was deduced.

Figures 16 and 17 present the above ratio for the two chambers at RQR3, RQR5 and RQR9.

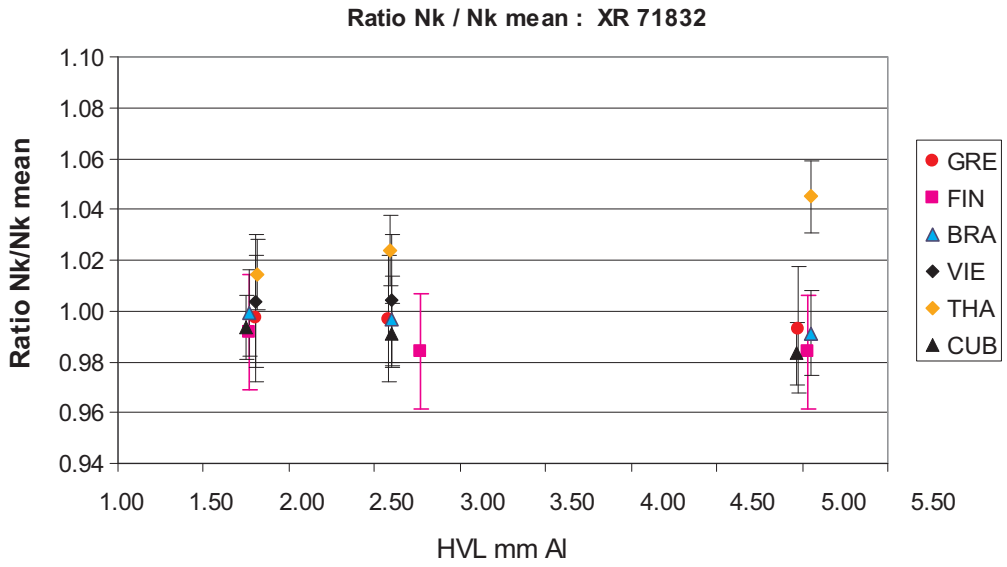

FIG. 16. The ratios of the measured to mean calibration coefficients at RQR3, RQR5 and RQR9 qualities, for A3 Exradin XR 71832 chamber.

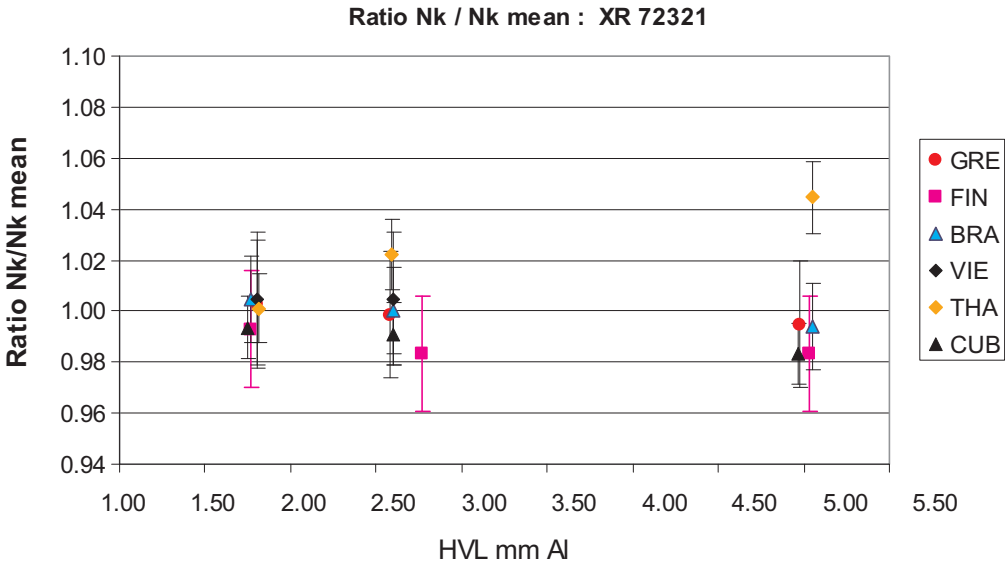

FIG. 17. The ratios of the measured to mean calibration coefficients at RQR3, RQR5 and RQR9 qualities, for A3 Exradin XR 72321 chamber.

At each RQR quality, the ratio of the calibration coefficient N_K and the mean N_{Kmean} (from all SSDLs) was calculated for each chamber. The ratios N_K/ N_{Kmean} indicate the total spread of the calibration coefficients normalized to $N_{Kmean.}$ These ratios are combined for all RQR qualities for all SSDLs and presented in Fig. 18.

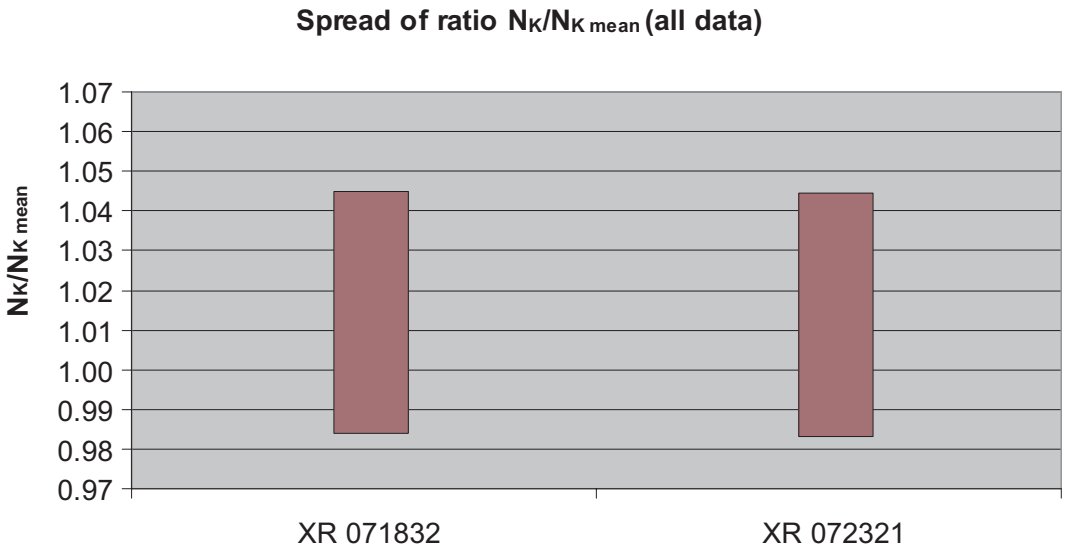

FIG. 18. Ratio N_K/N_{Kmean} over six centres for two chambers.

Each bar indicates the spread of N_K for all SSDLs.

ratio N_K/N_{Kmean}	XR 071832	XR 072321
min value	0.984	0.983
max value	1.045	1.045
Range	0.0612	0.0612
Mean	1.000	1.000
Median	0.997	0.999

The maximum spread is 6.1% for both XR071832 and XR 072321 chambers.

3.3.5. Conclusions

— The calibration procedures and results were consistent between participating SSDLs.
— The derived N_K coefficients at certain RQR from different SSDLs (except in one case) were within the relevant uncertainties.
— Any differences in HVL between SSDLs did not affect the N_K values significantly.

3.4. EXPERIMENTAL DOSIMETRY AUDIT OF RQR BEAMS BY MEANS OF TLDS

3.4.1. General

This section provides a comparison of the calibration of a TLD system in RQR beams at several SSDLs. This comparison was conducted under activity 6 of the CRP with six SSDLs (the Czech Republic, Greece, Thailand, Vietnam, Finland and Brazil) and the IAEA participating. The calibration of the TLDs was carried out in the participating SSDLs December 2006–2008. The participants irradiated the TLDs and specified the value of incident air kerma. The Czech Republic (NRPI) evaluated the results.

TL dosimeters were calibrated by the schedule in Table 19.

A major task of this activity was to test the usefulness of TLDs as a tool for global auditing of diagnostic radiology calibration capacity at SSDLs as well as to compare the calibration results amongst the participating laboratories.

TABLE 19. LIST OF INSTITUTIONS CALIBRATING THE TLDs IN THEIR RQR BEAMS (RQR-M BEAMS IN AUSTRIA)

Country	Date of calibration	Institution
Czech Rep.	XII.2006	National Radiation Protection Institute (NRPI)
Greece	VII.2007	Greece Atomic Energy Commission (GAEC)
Thailand	XII.2007	SSDL, Division of Radiation and Medical Devices
Vietnam	I.2008	Institute of Nuclear Science and Technology (INST)
Finland	V.2008	Radiation and Nuclear Safety Authority
Brazil	VII.2008	Center of Nuclear Technology Development (CDTN)
IAEA	III.2007, IV.2008	International Atomic Energy Agency

3.4.2. TLD system used and calibration in terms of N_k

The TLD system of the Czech Republic was used for the comparison. The participants each performed irradiation at their own SSDLs.

The TL detectors were TL chips of LiF:Mg,Cu,P[9] encapsulated in either a black plastic sachet or special plastic card. Three TL chips in the sachet or in the card represented one TL dosimeter. TRS No. 457 recommends using sachets for packing the dosimeters.However, larger variations in the response of the TL chips were observed when using sachets. Therefore, the cards were used as well.

A manual TL reader Harshaw 4500 was used for the detector readout. The detectors were heated in the reader by a planchet. Nitrogen was used to ensure an inert atmosphere during the readout. The time temperature profile used was as follows: preheat 130°C for 8 s followed by readout for 20 s, maximum temperature 240°C, temperature rate 10°C/s. Annealing of the detectors was performed before each use of the detectors. Annealing was performed in an oven, using 240°C for ten minutes, following by a rapid cooling to a room temperature by placing a support with detectors onto a large aluminium block.

The procedure followed was that given in TRS No. 457 (Appendix VIII: Field calibrations, IX.4-Calibration of TLDs) as this procedure applies well to calibration in the SSDL also.

The dosimeters were irradiated in a horizontal geometry, free in air, at a focus to dosimeter distance of 100 cm (in Thailand 150 cm). Free in air geometry was achieved by placing the dosimeters on an adhesive tape within a frame as shown in Fig 19. The applied incident air kerma was in the range 5–10 mGy. In the Czech Republic and Greece, calibration for all RQR beams was made. In Thailand, Finland and Brazil RQR3, RQR5 and RQR9 qualities were used. In Vietnam, RQR3, RQR5 and RQR7 qualities were used.

The TLD system was additionally calibrated twice in mammographic beams at the IAEA.

To correct each TL reading, individual sensitivity factors k_{si} were used. Several TLD batches were used in the calibrations. To correct for different TLD batch sensitivity and long term changes in sensitivity of the TLD reader, five reference TL detectors from each batch were reserved. During each calibration, these reference TL detectors were equally irradiated in the TLD laboratory of NRPI and then read out with dosimeters irradiated in the SSDL. The correction factor for different TLD batch sensitivity and changes in sensitivity of the reader k_{rs} was calculated as follows:

$$k_{rs} = \frac{\bar{M}_{Q,ref.batch}}{\bar{M}_{Q,batch}}$$

[9] This is a different composition from the TL material recommended in TRS No.457 (LiF:Mg,Ti) but was the one available at the coordinating centre.

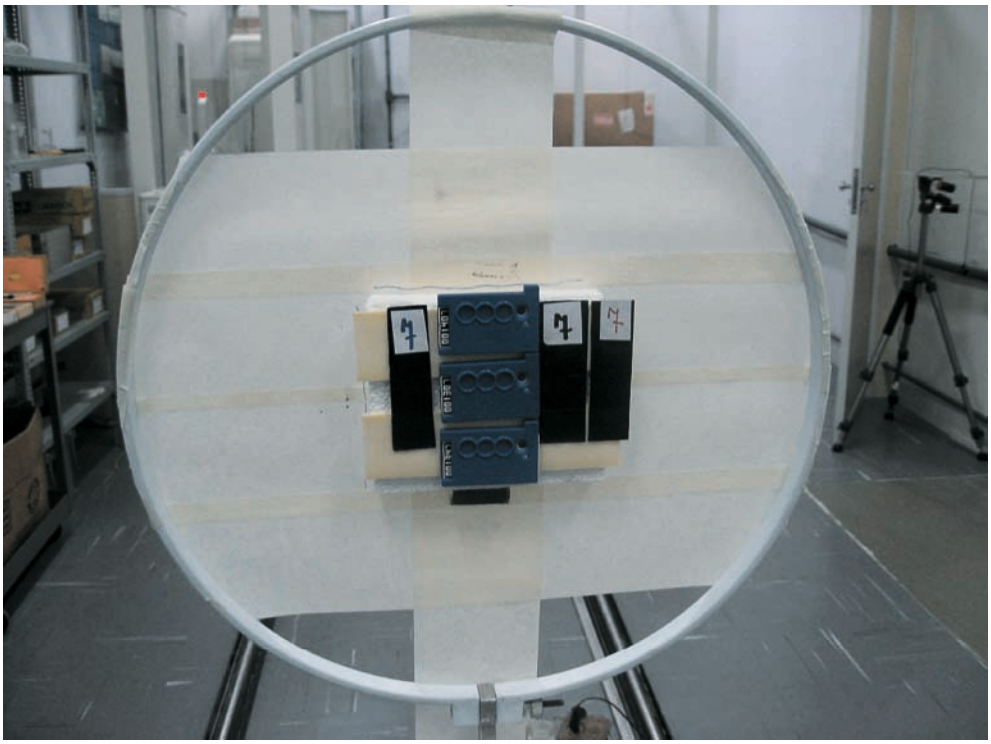

FIG. 19. TL dosimeters in a form of black plastic sachets and blue plastic cards, each containing 3 TL chips, during calibration in Brazil.

Where $M_{Q,ref.batch}$ is a mean corrected value of reference TL detectors from batch S576/05 during calibration in the Czech Republic (calibration in the Czech Republic using batch S576/05 is the reference for the audit). $M_{Q,batch}$ is a mean corrected readout value of reference TL detectors of any batch during any other calibration. This is a similar principle to using a monitor chamber during the calibration of chambers using the substitution method.

3.4.3. X ray apparatus and dosimetric system of the SSDLs

Table 20 shows the X ray systems that were used in the participating SSDLs for the TLDs calibrations.

TABLE 20. SSDL EQUIPMENT FOR TLD CALIBRATION

	Czech Republic	Greece	Thailand	Vietnam	Finland	Brazil
Unit	Seifert, Isovolt US2	Pantak 225HF	Siemens, Stabilipan	Pantak	Seifert, Isovolt HS	Pantak ISOVOLT 320/13 HS
Filtration	3 mm Be	1mmBe+5.2mm PMMA	2,5 mm Al	1 mm Be	1mmBe, 0.21 mm Al @ 60 kV	7 mm Be
Field size	7 cm	14.7 cm	15 cm	19 cm	14.5 cm	12 cm
Reference dosimeter — chamber	Exradin A3	PTW, W-77337	Standard imaging, A4 — Exradin	Standard imaging, A4 — Exradin	PTW 23344 (RQR3, RQR5), NE 2561 (RQR9)	Radcal 20X5-3/20647
Reference dosimeter — electrometer	Keithley 196	PTW, Unidos 10002	Wellhofer, Dosi 1	PTW, Unidos	Keithley 6571	Keithley 6517A/1138780

3.4.4. Results

The results of the calibration, expressed as calibration coefficients $N_{K,Qo}$ for beam quality RQR5, are shown in Table 21 (TL detectors in sachets) and in Table 22 (TL detectors in cards). Normalized values of $N_{K,Qo}$ for detectors packed in plastic sachets are shown in Fig. 20. The normalized value represents a ratio of $N_{K,Qo}$ of the participant and $N_{K,Qo}$ of the Czech Republic, which is the reference value for the audit. These values are the result of the audit. Normalized values of $N_{K,Qo}$ for detectors packed in plastic cards are shown in Fig. 21. Here, the normalized value represents a ratio of $N_{K,Qo}$ of the participant and mean $N_{K,Qo}$ of all participants (reference value from the Czech Republic is not available due to the reconstruction of the SSDL).

TABLE 21. COMPARISON OF CALIBRATION COEFFICIENTS OF THE TLDS FOR RQR5 BEAM QUALITY FOR THE USED TLD BATCHES — TL DETECTORS PACKED IN PLASTIC SACHETS

Country	Calibration coefficient $N_{K,Qo}$ (μGy/nC)			
	batch S343/03	batch S349/03	batch S576/05	batch S808/08
Czech Rep.			0.836	
Greece	0.846	0.751		
Thailand	0.836	0.822	0.805	
Vietnam	0.836	0.816	0.779	
Finland	0.812	0.789		0.842
Brazil		0.796	0.723	0.783

TABLE 22. COMPARISON OF CALIBRATION COEFFICIENTS OF THE TLDS FOR RQR5 BEAM QUALITY FOR THE USED TLD BATCHES – TL DETECTORS PACKED IN PLASTIC CARDS

Country	Calibration coefficient $N_{K,Qo}$ (μGy/nC)			
	Batch S343/03	Batch S349/03	Batch S576/05	Batch S808/08
Thailand	0.792	0.778	0.762	
Vietnam	0.782	0.767	0.736	
Finland	0.760	0.759		0.773
Brazil		0.739	0.696	0.740

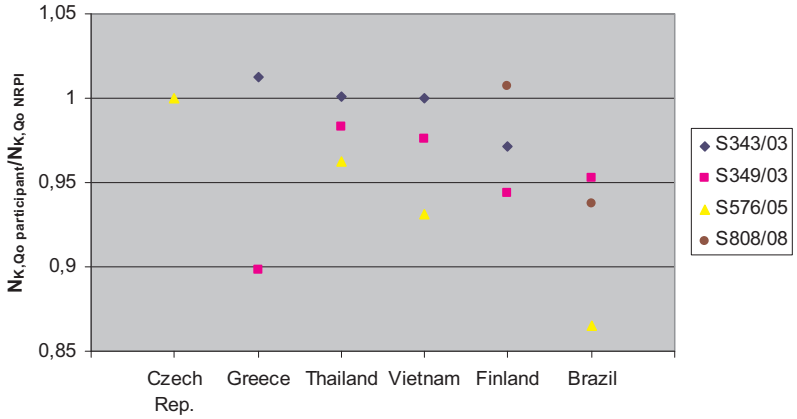

FIG. 20. Results of activity 6b — TLD audit of SSDL dosimetric equipment for RQR5 quality using different TLD batches — TL detectors packed in plastic sachets.

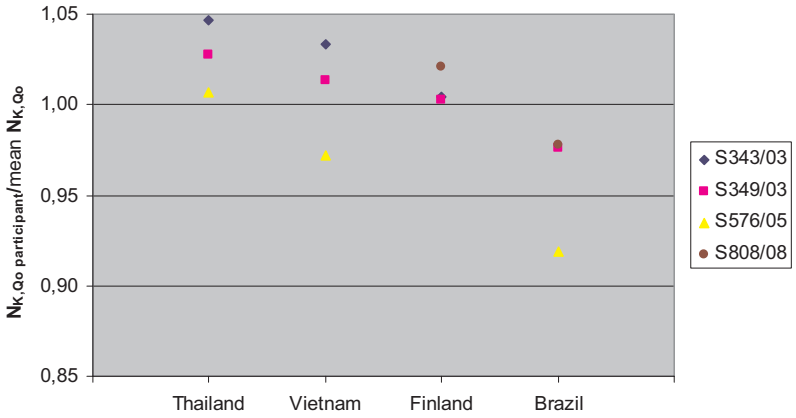

FIG. 21. Results of activity 6b — TLD audit of SSDL dosimetric equipment for RQR5 quality using different TLD batches — TL detectors packed in plastic cards.

Correction factors k_{Q,Q_0} for quality of the beam are shown in Fig. 22 for detectors packed in sachets and in Fig. 23 for detectors packed in the plastic cards. The results are not shown for each batch separately. Each correction factor k_{Q,Q_0} for the particular beam quality is calculated as a mean of k_{Q,Q_0} values for the TL batches used in the calibration.

The results of calibration in mammographic beams expressed as calibration coefficients N_{K,Q_0} for beam quality RQR-M2 are shown in Table 23 together with correction factors k_{Q,Q_0} for the quality of the beam. Only one batch of dosimeters was used. During the second calibration in April 2008, TL detectors were packed in sachets and in cards.

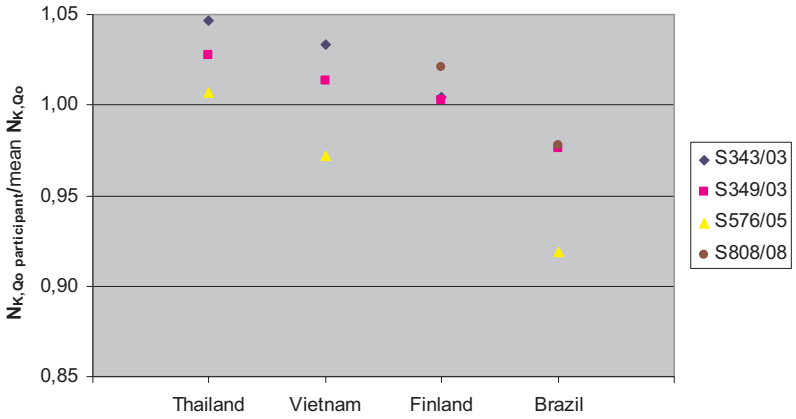

FIG. 22. Summary of correction factors k_{Q,Q_0} of the TLDs calibrated in RQR beams — TL detectors packed in plastic sachets.

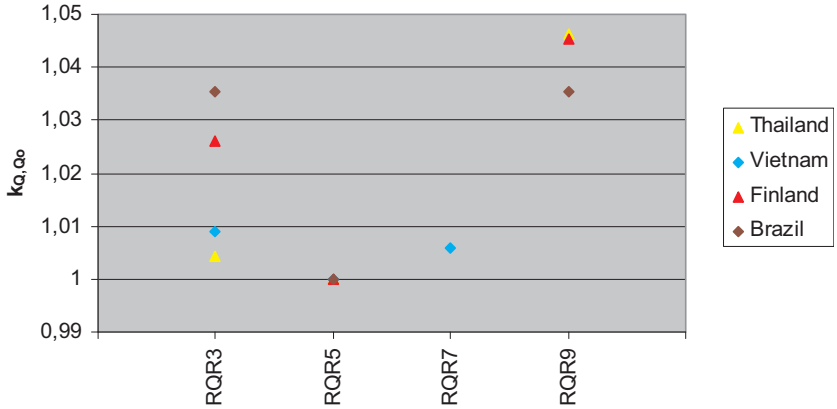

FIG. 23. Summary of correction factors k_{Q,Q_0} of the TLDs calibrated in RQR beams — TL detectors packed in plastic cards.

TABLE 23. COMPARISON OF CALIBRATION COEFFICIENTS AND CORRECTIONS FACTORS k_{Q,Q_0} OF THE TLDs IRRADIATED IN MAMMOGRAPHIC BEAMS AT THE IAEA

Beam quality	HVL (mm Al)	N_{K,Q_o} (μGy/nC)			k_{Q,Q_o}		
		III.07 (sachet)	IV.08 (sachet)	IV.08 (card)	III.07 (sachet)	IV.08 (sachet)	IV.08 (card)
RQR-M1	0,323				1.04	1.03	1.02
RQR-M2	0,358	1.09	1.02	1.03	1.00	1.00	1.00
RQR-M3	0,376				1.00	0.99	0.98
RQR-M4	0,411				0.95	0.97	0.95
Rh/Rh 25 kV	0,362				1.00	0.99	0.99
Rh/Rh 30 kV	0,444				0.91	0.92	0.94
Rh/Rh 35 kV	0,504				0.88	0.92	0.91
Rh/Rh 40 kV	0,548				0.87	0.89	0.86

The uncertainty analysis of the calibration coefficient N_{K,Q_o} for calibration in the SSDL beam is given in Table 24. Each component of the uncertainty given in the table has a different value for the different SSDL, TLD batch used and number of used TL detectors. To get a conservative estimate, the highest values were considered.

TABLE 24. UNCERTAINTY ANALYSIS OF THE CALIBRATION COEFFICIENT FOR CALIBRATION IN THE SSDL

Component of uncertainty	Values given in %	Source of the value of the uncertainty component
Kerma value given by SSDL	2.00	Value determined by SSDL
Field homogeneity	0.58	Value determined by SSDL, Report of the second RCM-CRP E2.10.06, July 2007
Distance	0.23	Value determined by SSDL, Report of the second RCM-CRP E2.10.06, July 2007
var. coeff. of mean TL response	1.10	Based on statistical analysis of the response variation
Reader and TLD batch sensitivity	1.48	Based on statistical analysis of the response variation
combined uncertainty	2.8	
expanded uncertainty (k=2)	5.6	

Note: Fading was not considered. TL detectors calibrated in RQR and reference TL detectors used for reader and batch sensitivity correction were annealed, irradiated and readout the same day. Fading dosimeters sent to Greece and those staying with reference dosimeters in NRPI have shown the same response. Energy dependence was not considered. RQR5 beams within SSDLs are considered the same.

3.4.5. Discussion and conclusions

The results of the calibration and the large value of expanded uncertainty do not indicate a possibility of using this TLD system for auditing SSDLs. It is clear that the 12% difference in calibration coefficients for two TLD batches (after the TLD batch sensitivity correction) received from calibration in Greece (Table 21) is higher than their expanded uncertainties. It indicates that some influencing factors were not included or were underestimated in the uncertainty analysis. For different batches, the results were systematically shifted, even though the different sensitivity of the batches was corrected by reference dosimeters. Therefore, it is better to use the same batch of

TLDs for calibration and for required dosimetric task if there is a sufficient amount of TLDs available. Then the reference dosimeters will be used just for correction of changing the sensitivity of the reader.

The systematic shift of the results is not the same for a given batch for all calibrations. For instance, the ratio of calibration coefficient of Greece and Finland is 1.04 for the batch S343/03 and 0.95 for the batch S349/03 (see Table 21). This ratio assessed from two ionization chamber measurement is 1.013 and 1.016 (see Tables 17 and 18 and Section 3.3.4).

The systematic difference between the results from different batches is similar for both types of packing of the TL detectors. It is therefore believed that the TL cards do not show any advantage. The problem is caused by different sensitivity of the batches. Even if the correction of the different sensitivity was made, a small systematic shift can be still observed. The packing of TL detectors in TL cards have the advantage of lower standard deviation of readout values.

It is thought that the standard deviation of the readout values from the calibrations, mainly for TLDs packed in the sachets, is underestimated. The standard deviation of the readout values was determined by an experiment, when individual detectors were placed in plastic cards and equally irradiated by reference irradiator Dosacus IR-1with a ^{90}Sr source. The irradiation was equivalent to air kerma 0.8 mGy in the RQR beam quality. TLDs were irradiated one day after annealing and readout one day after the irradiation. However, when calibrating (or using) the TLDs, the individual detectors are encapsulated in a plastic sachet. The process of packaging, poor geometry during irradiation (the sachets are not perfectly flat) and the higher possibility of damaging the TLDs may cause higher variations of the readout values. Moreover there are long intervals between annealing to irradiation and irradiation to readout during clinical use with the TLDs not stored in laboratory conditions. This can cause higher variations of the readout values as well.

TRS No. 457 recommends the use of TLDs for measurements on patients except in the case of mammography. TRS No. 457 strictly requires calibration of any dosimeter (not only TLDs) free in air, which it clearly describes. Emphasis is given to the proper use of the correction factor k_Q. However the transition from the irradiation free in air to irradiations on a phantom or on a patient is not discussed . It is this transition that brings the problem of energy dependence of a dosimeter since the backscattered radiation has a lower energy than that of the calibration RQR beams. This is even more evident for TLDs with a higher effective atomic number than air and can result in a significant error.

This problem can be solved by the calibration of TLDs directly on a phantom. In the SSDL, the entrance surface air kerma has to be calculated from incident air kerma and appropriate backscatter factor as recommended in TRS No. 457. The uncertainty of the calibration will be increased by the uncertainty of the backscatter factor, however the overall uncertainty of K_e estimation will be reduced, due to use of a more appropriate calibration coefficient N_{K,Q_0} and correction factors k_Q. Unfortunately, such a phantom is not defined in TRS No. 457. The phantom chosen for the calibration should correspond to patients (with respect to material and size). It could be 20 cm of water or PMMA because backscatter factors are known for these materials. Beam shape and beam size during calibration should correspond to beam shape and beam size in clinical conditions, but this is not possible in the SSDL.

3.4.6. Conclusions

— For the purpose of K_e measurement in general radiography, TLDs (including packaging) could be calibrated on a phantom instead of free in air. Such a phantom needs to be defined as it is not in TRS No. 457. It can be 20 cm of water or PMMA, because backscatter factors are known for these materials.
— The same batch of TLDs should be used for both the calibration and for clinical measurement. Even if a correction for different TL batch sensitivity is made, the results obtained from TL measurements using different batches are systematically shifted.
— During calibration, several irradiations (optimally three) for each beam quality should be used. During each irradiation at least one dosimeter should be exposed containing three TL detectors.
— The response variation and individual sensitivity of TL detectors should be checked under the same conditions as used clinically. This means that the same packaging, similar values of delivered kerma and same time schedule of the whole process (annealing — irradiation — readout) should be used.
— It should be verified, that TL dosimeter (TL detectors and packing) used for the direct patient measurements is not visible on the clinical image.

4. IMPLEMENTATION FOR CLINICAL MEASUREMENT

4.1. INTRODUCTION

TRS No. 457 was written for the practice of dosimetry for diagnostic radiology in both the calibration laboratory (as discussed in Section 3) and in the clinical environment. The latter case is discussed in this section. This practice of dosimetry relies on the use of dosimeters with a traceable calibration and might include the use of radiological phantoms to represent typical patients or the collection of relevant data associated with clinical procedures to determine radiation dose quantities. The use of TLDs for diagnostic radiology dosimetry is also covered in TRS No. 457. The implementation of the use of TLDs to patient dosimetry has not been examined within this CRP. The calibration of TLDs in hospitals is reported in Section 5.2.

The aim of this section is to record the results of the methodologies described in Chapter 8 of TRS No. 457 for the 5 designated modalities. This was done with both phantom and patient dose methodologies. In both cases it was necessary to examine all aspects of the described dosimetry, such as the nomenclature and units of dose quantities, the use and effectiveness of phantoms, the clarity of the described methodology and the effectiveness of the data recording instruments as suggested in TRS No. 457.

4.2. MATERIALS AND METHODS FOR TASKS

While in the standards laboratory strict protocols determine the methods and conditions of detector irradiation, the situation in clinical practice is quite dynamic. In order to apply TRS No. 457 dosimetry methodology in a uniform way at the various testing centres, it was necessary to utilize the written methodologies and specify the type of clinical examinations to be investigated. It was also important to examine the use of phantoms and the available instruments that could be expected to be found in typical clinical environments.

4.2.1. Clinical examinations protocol

Table 25 shows the examinations specified to be investigated for each section. These are amongst the most commonly performed examinations and also allow data from different centres to be more easily compared.

TABLE 25. SELECTED CLINICAL EXAMINATIONS

Modality	Examination(s)
General radiography	PA chest (Upright) AP abdomen
Fluoroscopy	Barium enema Barium meal (Gastro)
Mammography	Cranio-caudal (CC) projection
Computed tomography	Routine head (e.g. for stroke) single series Abdomen-pelvis — single series High-resolution chest — single series
Dental radiography	Bitewing — adult mode Orthopantogram (OPG) adult mode

4.2.2. Patient selection

Data was collected for all patients within the survey period. Patient height and weight was included where possible. A minimum of 20 patients was the aim, but as few as 10 patients for CT exams could be used, as these tend to use more standard protocols.

4.2.3. Phantoms recommended and survey possibility of fabrication

Certain phantoms have been recommended by TRS No. 457 for the five areas of clinical radiological dosimetry. The use of these phantoms was evaluated using the following criteria:

— The availability of the specified phantoms including the ability of various centres to have phantoms fabricated if possible;
— The usefulness and appropriateness of the phantom for clinical dosimetry as documented in TRS No. 457.

4.2.4. Instruments

Traceability is an important part of the CRP. Therefore participants were requested to give calibration information for all instrumentation used including the variety of dosimeters used including KAP meters.

In TRS No. 457 it is written that calibration coefficients for KAP meter can be used with HVL interpolation for total filtration up to about 3 mm Al. This may be possible with reasonable uncertainty if the clinically used total filtration is close to ones used in the calibration of KAP meter in the SSDL [19]. RQR radiation qualities, recommended for calibration of KAP meters, have a typical filtration between 2.5 mm Al and 5 mm Al. However if the total filtration is totally different to that used in calibration, it will give unacceptable errors to measurements even though calibration coefficients are used.

4.2.5. Formalism

Energy dependence is an important consideration in dosimetry. The use of the k_Q terminology is not familiar to most clinical centres. The CRP showed that in practice k_Q may be rolled into uncertainty in many cases. Alternatively, calibration factors can be determined at a number of different beam qualities by SSDLs, in addition to or along with k_Q. Further advice and examples on implementation of the beam quality correction is given in Appendix IV.

4.2.6. Uncertainty in clinical measurement

The determination of uncertainties is an important consideration in clinical dosimetry, and so uncertainties were determined for all measurements.

Contrary to the precisely standardized SSDL set-ups, measurement conditions in clinical systems are less consistent. This is especially the case if dose measurements are carried out on patients. In some installations, users may not be familiar with applying corrections (air density, k_Q) or these corrections cannot be applied due to technical reasons or beam qualities changing during an examination. Therefore, elevated uncertainties are expected.

The uncertainty budget varies to a great extent depending on the modality, the methodology, as well as the calibration of the instruments available. A discussion of uncertainties for general radiography, fluoroscopy, CT and mammography can be found in the corresponding chapters in TRS No. 457.

4.3. GENERAL RADIOGRAPHY

Dose management in general radiography, for both screen-film and digital detectors, necessitates knowledge of doses administered to patients. Therefore, different dose measurement possibilities are described in TRS No. 457. In quality control and inter-centre comparison of dose levels for typical examinations on standard sized patients, phantoms may be used. Nevertheless, phantom measurements cannot directly indicate doses to

TABLE 26. DOSIMETERS USED FOR GENERAL RADIOGRAPHY DATA COLLECTION

Participant	Dosimeter type	Ion chamber type	Calibration not older than 2 years
Austria	Unfors Xi and Unfors 565L	n. a.	Yes
Cuba	PTW Unidos E	PTW 77334	Yes
Czech Republic	not specified	Radcal 10X5-6-3	Yes
Hungary	Radcal 9015	not specified	no (3 years)
Rrepublic of Korea	Radcal 9095	not specified	Yes

patients for general radiography applications due to inherent limitations. This topic is discussed in the 'comparison of methods' section. In general radiography applications dosimetric values may be described in terms of incident air kerma, K_i, entrance surface air kerma, K_e, or kerma area product, P_{KA}. Table 26 summarizes the dosimetric equipment used by the participating centres. One centre used semiconductor devices, requiring a higher uncertainty for calibration corrections, the others ion chambers. During the data collection period no centre involved could collect data in terms of P_{KA} for general radiography of adult patients.

4.3.1. Clinical dose measurement methods

4.3.1.1. Measurement with phantoms

TRS No. 457 defines phantoms to be used to obtain dosimetric values for the most common X ray examinations in general radiography as:

— CDRH chest for posterior-anterior (anterior-posterior) chest imaging;
— CDRH abdomen/lumbar spine for anterior-posterior (posterior-anterior) abdomen and lumbar spine.

The methodology of determining incident air kerma, K_i, with phantom exposures is straight forward. It is crucial that exposure settings are used when exposing the phantoms as applied to the standard patient.[10]

Phantoms for general radiography dosimetry were available in all but one Member State participating in the clinical dosimetry exercise. From the latter, no data is included since exposure data from patients could not be reported. In all other participating countries, fabrication of CDRH phantoms — if not already available — was possible. One country provided data for different types of phantoms, CDRH and locally used PMMA slab phantoms since the latter were available for dose auditing and quality assurance procedures. Phantom and patient dose data from one country and one X ray facility from another participant were excluded because data integrity could not be verified. From one participating country, patient dosimetric data is reported whereas phantom data was withdrawn because it could not be compared to patient data, since the phantoms were not exposed with exposure parameters corresponding to a standard patient.

4.3.1.2. Measurement or calculation of dosimetric values from patient exposures

The methods used to determine *patient* doses in general radiology are:

— Calculation of entrance surface air kerma, $K_{e,}$ from incident air Kerma, K_i, determined using patient exposure parameters and tube output Y(d);
— Measurement of entrance surface air kerma, K_e, from TL dosimeters on patients;
— Measurement of kerma area product, P_{KA}, during patient exposures.

[10] Where entrance surface air KERMA, K_e, is the desired quantity K_i needs to be multiplied with the appropriate backscatter factor.

4.3.2. Uncertainty estimations in general radiography dosimetry

Table 27 shows estimations of typical uncertainties for clinical measurements (rounded to 2 significant digits). Typical expanded (2σ) uncertainties for clinical measurements will not be less than 10% (scenario A). In many instances, application of corrections for beam quality and individual patient diameter will not be feasible; in these cases expanded uncertainties up to approximately 25% should be considered. This will impact particularly on the comparison of doses to DRLs (dose reference levels). It should also be considered that by using incident air kerma (K_i) rather than K_e uncertainties in the backscatter factors can be avoided. Uncertainties are shown as examples provided appropriate care is taken during measurement and instruments are properly maintained and calibrated.

TABLE 27. EXAMPLE OF UNCERTAINTY BUDGETS FOR GENERAL RADIOGRAPHY DOSIMETRY

	Scenario A	Scenario B	Scenario C
Description of scenario	Determination of K_i using dose output measurements. Corrections applied	Practical method; as in scenario A. but no manual corrections applied	Measurement of P_{KA} during patient exposure, no corrections applied
Uncertainty (k=2) in dosimetric quantity due to			
Intrinsic error of dosimeter	3.2%	3.2%	3.2%
Calibration coefficient $NK_{Q,0}$	1.6%	1.6%	1.6%
Long term stability of dosimeter reading	1%	1%	1%
Difference in beam qualities between calibration and clinical use	1%	5%	20% (1)
Field size/field inhomogeneity	2%	2%	5%
Focus-skin-distance	4%	10%	—
Focus detector distance and scatter influence at dose output measurement	3%	3%	—
X ray output accuracy — at (patient exposure) (2) — at output measurement (3 exposures)	 5% 2.9%	 5% 2.9%	 —
air density correction: pressure temperature	 0.2% 0.5%	 2% 2%	 2% 5% (3)
Electromagnetic compatibility and humidity, other uncertainties estimated < 1% each	2%	2%	2%
combined expanded (2σ) uncertainty in K_i or P_{KA}	8.8%	14%	22%
Determination of K_e			
Backscatter factor	5%	20%	
expanded (2σ) uncertainty in K_e	10%	24%	

Notes:
(1) mostly due to added copper filters
(2) 5% in output accuracy assumed for converter generators. Should be replaced by appropriate tolerances (10 to 20%) if 6 or 12 pulse generators used [24]
(3) uncertainty due to temperature partly due to warming of chamber from tube housing and light field.

4.3.3. Comparison of methods

4.3.3.1. Comparison of methods using patient exposure data

The use of P_{KA} measurement will give additional information on beam collimation. Therefore, auditing P_{KA} will also monitor collimation practice. For some projections (skull) this may result in dose over estimations if poor collimation practice is done. In most general X ray procedures, this effect is a positive one in terms of patient dose estimation. In terms of the practical application of $P_{KA,}$ if a KAP meter is permanently installed in the X ray device, or can be installed for selected time intervals, there is the advantage that a significant number of patients are monitored, whereas TL dosimetry will only monitor a few patients under auditing conditions. Therefore, a change in exposure practice will be more difficult to detect.

4.3.3.2. Comparison of methods applying phantoms with methods using patient exposure data

If a radiological patient equivalent phantom is used, phantom doses are expected to be reasonably close to the corresponding average patient doses, if identical technique parameters were used.

A comparison of patient and phantom doses in participating Member States showed that in all but two cases patient doses were higher than the doses determined with the corresponding CDRH phantoms (Fig. 24). Patient selection criterion for this comparison was 70+/–10 kg (centre G provided patient data for 60 +/–10 kg and did not report phantom data consistent to patient data). The ratio between patient doses applied to standard sized patients, and the corresponding phantom doses ranged approximately 0.5–3 (Fig. 25).

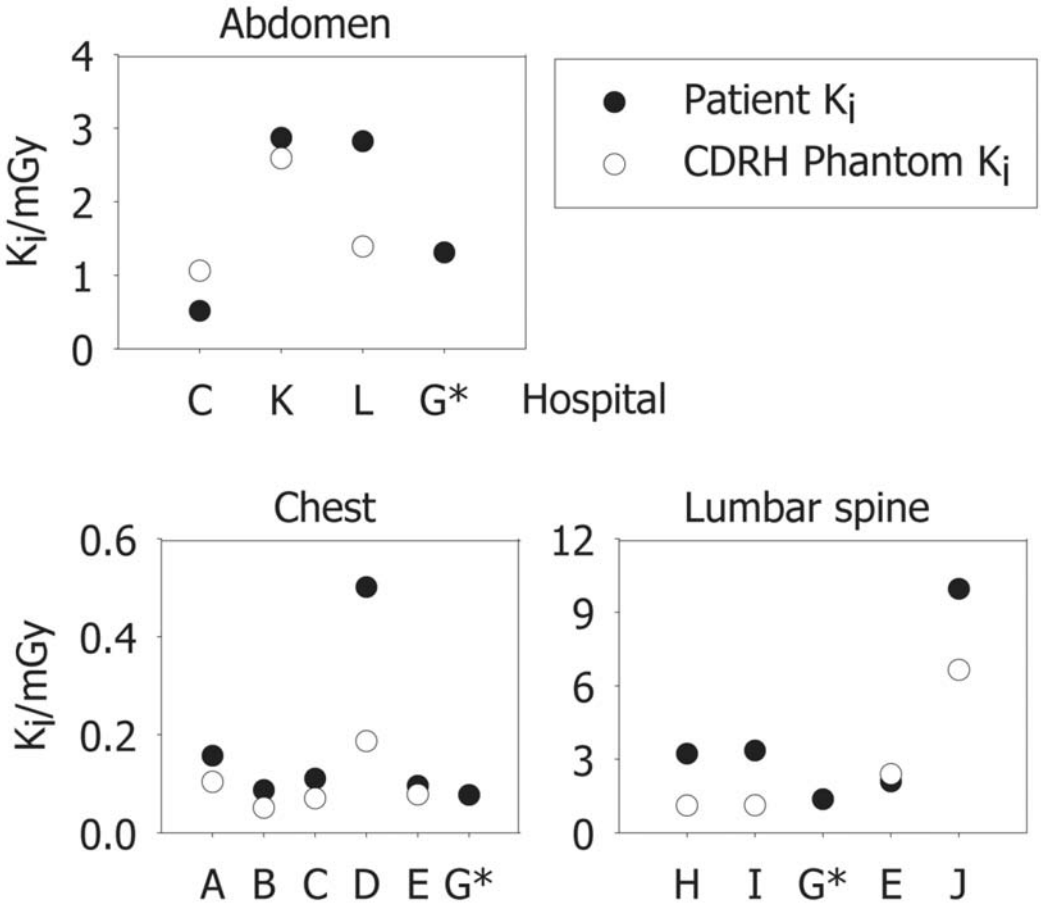

*FIG. 24. Average Patient doses compared to reported phantom doses. *: Standard patients in hospital G refer to 60+/–10 kg (others: 70+/–10). No phantom data available from hospital G.*

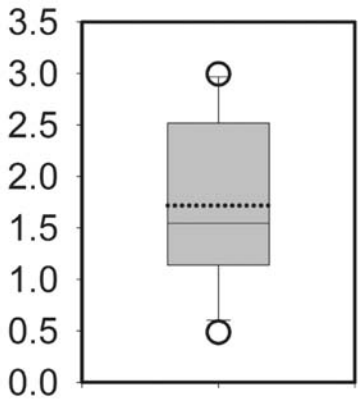

FIG. 25. Ratio of patient to phantom doses. Dotted line corresponds to mean value, solid line in the box to median.

4.3.4. Problems encountered

As stated previously, exposing the CDRH phantoms with appropriate technique factors was problematic to at least one institution. At this institution, a wide range of kV was applied for patients, and the selection of appropriate kV was not regulated by strictly defined technique charts. In this case, it is difficult to identify technique parameters (kV) to use with the phantom. However, if variation of kV is minimal the most often used setting could be taken for phantom exposures. Another issue was the choice of mAs when exposing the phantom. In cases where AEC is used on patients, the AEC must also be used when exposing the phantoms.

Unusually high discrepancies in patients and phantom dose ratios may indicate a potential for an optimization of procedures or review of dosimetric methodology. In one institution, the phantom and patient doses reported were different by a factor of 10.9 in one room and 22.9 in another, indicating an error in data collection (calculation of dose from dose output measurements in this case). In the case of unusually high differences in phantom and patient doses, re-evaluation of the dosimetric methods applied is recommended in the first instance.

4.3.5. Conclusions

In participating Member States, the availability of phantoms or the possibility of phantom production from PMMA and 99.5% purity grade aluminium, according to published specifications, was not found to be a major issue. Two centres completed fabrication of the CDRH phantoms. Details are given in Appendix III. Nevertheless, phantoms may be most useful in quality assurance programmess. Consistent determination of patient doses with phantoms is not the dosimetric method of choice, especially if dose data from patient exposures can also be collected. In cases where collection of patient doses, or data on patient exposures, from which retrospectively patient doses can be determined, is not feasible due to restrictions in equipment or educated personnel, phantom measurements provide rough guidance on doses delivered to patients which can be useful. Certain technical restrictions could be, for example, the use of an X ray system where post exposure mAs values are not displayed with AEC used. Care must be taken to expose the phantom with appropriate technique factors or AEC. It should be noted that the restrictions on recommending phantoms for patient dosimetry do not apply to CT and mammography as the situation is different.

Doses to phantoms can indeed provide useful data in general radiography. If care is taken on exposing the phantoms correctly, phantom dose values are valuable for dose intercomparison exercises. In this case, phantom doses are compared to phantom doses (optimally measured with identical phantoms) rather than interpreted as dose to a standard sized patient.

In situations where the use of consistent (patient physique dependent) technique charts for standard radiographic procedures is not in common practise, dose determination by phantoms are, to some extent, dependent on the radiographer exposing the phantom. In this case, a single phantom exposure provides only a snapshot and may not provide a representative dose value. The importance of providing and adhering to standardized exposure charts is also emphasised in this respect, applying to all standard exposures, taken with or without AEC.

Determination of P_{KA} with phantoms, while sometimes possible, is neither described nor advised in TRS No. 457. In installations with attached KAP meter, the use of a phantom is not necessary since patient exposure data

is collected in routine imaging. In systems without installed KAP meter, a dosimeter providing a Kerma reading is the recommended method since the phantoms cannot provide anatomical landmarks or choice of appropriate field size, which might introduce an additional source of uncertainty in the measurements.

According to TRS No. 457, individual patient thickness, t_p, should be determined for the indirect assessment of incident air Kerma and entrance surface air Kerma. In clinical situations, assessment of individual patient thickness may seem difficult because of workflow limitations. In this case it is recommended to use an average patient thickness value rather than relying on individual measurements. Scenario B in the uncertainty budget estimation (Table 27) accounts for this situation. It may be advisable to collect patient exposure data for a large number of patients and use an estimated patient thickness[11] rather than a determination with fewer patients where the individual patient thickness is measured. This is especially the case if staff workload is an issue.

TL dose audits on patients may be performed in Member States for feasibility reasons; however there is no other reason for favouring K_e over K_i in general radiography. Practice shows that the most feasible method for dose auditing should be recommended. In some situations this may be using TL dosimeters. In most situations however, where no appropriately calibrated KAP meter is installed, the determination of K_i from exposure factors and tube output[12] is recommended. When an appropriately calibrated KAP meter is installed its use is recommended.

4.4. FLUOROSCOPY

This section contains the results of the clinical measurements made for fluoroscopy equipment, including both phantom and patient measurements, as described in TRS No. 457. Five participating countries provided data including a total of 20 hospitals. Conclusions have been drawn from the measured data and from the experiences of the participants, resulting in the recommendations in Section 4.4.3.

4.4.1. Clinical dose measurement methods

4.4.1.1. Phantom measurements

These data comprised measurement of entrance surface air kerma rate at the surface of a defined phantom, according to the recommendations in TRS No. 457. The worksheets used to collect the data are given in TRS No. 457, Section 8.5.2.6, with the replacement of ‘ABC setting’ with ‘manual or auto mode setting’. A range of phantoms were used to carry out the entrance surface air kerma rate measurements, depending on local availability, with two participants using 185mm PMMA. One using 190 mm PMMA and one using 200 and 300 mm water.

For all but one set of measurements, ionisation chambers were used, with corrections made for temperature and pressure and the calibration coefficient of the chamber. One participant applied a beam quality correction factor but, for the other participant, no such data was available for the chambers being used. Where a PMMA phantom was used, backscatter factors (B_w/B_{PMMA}) were applied using values given in Appendix VII of TRS No. 457, which are taken from Petoussi-Henss *et al* [25]. The dosimeters used and correction factors applied at the various institutions are given in Table 28, and experimental details are summarized in Table 29.

Entrance surface air kerma rate was calculated for water phantom measurements as

$$\dot{K}_e = \bar{\dot{M}} N_{K,Q_0} k_Q k_{TP}$$

And for PMMA phantom measurements as

$$\dot{K}_e = \bar{\dot{M}} N_{K,Q_0} k_Q k_{TP} \frac{B_w}{B_{PMMA}}$$

where M is the mean dosimeter reading, $N_{K,Q0}$ is the calibration factor for the dosimeter at reference beam quality Q_0, k_Q is the beam quality correction factor, and k_{TP} the temperature/pressure correction factor.

[11] Use height and weight to calculate patient diameters retrospectively.

[12] As well as calculation of K_e if desired.

TABLE 28. DOSIMETERS AND CORRECTION COEFFICIENTS USED FOR PHANTOM MEASUREMENTS IN FLUOROSCOPY

Participant	1[a]	2[b]	3[c]	4[d]	5[e]
Dosimeter model	Radcal 10X5-60E	Radcal 9015	RTI R 100	Radcal 9095	Radcal 9015
Calibration traceable to	PTB	PTB	PTB	PTB	NPL
N_{Kqo}	Yes	Yes	Yes	Yes	Yes
k_{TP}	Yes	Yes	Yes	Yes	Yes
k_Q	No	No	No	Yes	No

[a] Czech Republic
[b] Finland
[c] Hungary
[d] Republic of Korea
[e] UK

TABLE 29. SUMMARY OF EXPERIMENTAL DETAILS FOR PHANTOM MEASUREMENTS IN FLUOROSCOPY

Participant	1[a]	2[b]	3[c]	4[d]	5[e]
Phantom	190 mm PMMA	200/300 mm water	185 mm PMMA	200/300 mm water	185 mm PMMA
Focus-to-intensifier distance (mm)	960–1020	Not specified	800–1000	830–1000	980–1200
Focus-to-chamber distance (mm)	510– 580	720–800	420–700	360–740	620–1000
Filtration	none added	4 mm Al + 0.1 mm Cu	3.5 mm Al	not specified	None added
Range of tube voltage (kV)	68–83	76–110	70–120	70–110	66–97
Range of tube current (mA)	0.92–4.1	0.4–5.9	1.3–15.6	0.6–2.9	0.3–128
Range of field-size (mm)	90–150[1]	380[2]	160–300[1] 120–205[2]	350[2]	150–410

[a] Czech Republic[1]Circular field
[b] Finland[2]Rectangular field
[c] Hungary
[d] Republic of Korea
[e] UK

No problems were encountered with either the measurement or calculation of the entrance surface air kerma rate. In addition, there were no discrepancies between the procedures followed and TRS No. 457, although there may have been some confusion in what was meant by field size on the data collection sheets. This should have been intensifier field size rather than collimated field size, as this is an important variable affecting entrance surface air kerma rates.

The uncertainty budget for the measurements was calculated by each participant in accordance with TRS No. 457 recommendations. An example of this analysis for 1 participant is given in Table 30.

TABLE 30. UNCERTAINTY ANALYSIS FOR PHANTOM MEASUREMENTS IN FLUOROSCOPY

Influence quantity	Uncertainty (%) k=1
Intrinsic error, N_{KQ0}, k_Q	2.06
Radiation quality	0.5
Kerma rate	2.5
Air pressure	0.04
Temperature	0.07
Distance from focus	1.4
Electromagnetic compatibility	1.5
Operating voltage	1.2
Long term stability	1
BSF[1]	2.5
Relative expanded uncertainty (k=2)	9.72%

[1] Measurement with PMMA phantom

4.4.1.2. Patient measurements

These measurements were made with KAP meters either attached to the X ray tube housing or integral to the equipment. Three of the participants collated patient measurements for the same equipment used in the above phantom measurements. The range of equipment used for patient measurements is detailed in Table 31. Most KAP meters used were calibrated in situ, and correction coefficients were applied to the measured readings in accordance with TRS No. 457. These factors and the calibration details are given in Table 32. The calibration of KAP meters in hospitals is discussed in greater detail in Section 5.1 of this publication. The collected KAP data was mainly taken from barium studies; however data from a small number of interventional cardiac studies were also included. Fluoroscopy equipment parameters were recorded for all examinations.

TABLE 31. X RAY EQUIPMENT USED FOR PATIENT MEASUREMENTS IN FLUOROSCOPY

Participant	Equipment
Finland	Philips Velara 80/Multidiagnost Eleva (2)
Hungary	Siemens Axion Artis Philips Multidiagnost 4
Republic of Korea	Hitachi TU-300
UK	Siemens Sireskop (2) Siemens Axion Artis Siemens Angiostar + Philips Allura Philips MD4 Philips MD3 Philips Telediagnost GE Advantix DBS Toshiba EPS30

TABLE 32. KAP METERS AND CORRECTION COEFFICIENTS USED FOR PATIENT MEASUREMENTS IN FLUOROSCOPY

Participant	1[a]	2[b]	3[c]	4[d]
KAP meter	Not specified	Diamentor M4-KDK	Diamentor M2	Diamentor M2 or integral KAP meter
Calibration traceability	PTB	PTB	Not specified	NPL
N_{PKAQo}	Yes	Yes	Yes	Yes
k_{TP}	Yes	Yes	Yes	Yes
k_Q	No	No	No	No

[a] Finland
[b] Hungary
[c] Republic of Korea
[d] UK

Air kerma area product was determined according to the equation

$$P_{\mathrm{KA}} = M N_{P_{\mathrm{K}A},Q}\, k_Q\, k_{\mathrm{TP}}$$

where M is the KAP meter reading.

4.4.2. Uncertainty estimations for dosimetry in fluoroscopy

In Table 33, examples of estimations of typical uncertainties for dosimetric measurements in fluoroscopy are shown. Typically, uncertainties (2σ) in the range 10–30% are observed in clinical dosimetric measurements for fluoroscopy examinations if appropriate care is taken and instruments are properly maintained and calibrated.

TABLE 33. ESTIMATIONS OF TYPICAL UNCERTAINTIES FOR DOSIMETRIC MEASUREMENTS IN FLUOROSCOPY

	Scenario 1	Scenario 2	Scenario 3
Description of Scenario	Determination of entrance surface air kerma rate $\dot{K}_e$ using a phantom, applying air density and beam quality correction	Determination of entrance surface air kerma rate $\dot{K}_e$ using a phantom, no manual corrections applied	Measurement of P_{KA} during patient fluoroscopy, P_{KA} chamber calibrated in situ
Uncertainty (k=2) in dosimetric quantity due to			
Intrinsic error of dosimeter	3.2%	3.2%	3.2%
Calibration coefficient $NK_{Q,0}$	1.6%	1.6%	1.6%
Long term stability of dosimeter reading	1%	1%	1%
Difference in beam qualities between calibration and clinical use	3%	6%	20% (1)
Field size/field inhomogeneity	2%	2%	5%
Distance measurements and correction	4%	4%	—
Scatter radiation	3%	3%	—

TABLE 33. ESTIMATIONS OF TYPICAL UNCERTAINTIES FOR DOSIMETRIC MEASUREMENTS IN FLUOROSCOPY (cont.)

Description of Scenario	Scenario 1 Determination of entrance surface air kerma rate $\dot{K}_e$ using a phantom, applying air density and beam quality correction	Scenario 2 Determination of entrance surface air kerma rate $\dot{K}_e$ using a phantom, no manual corrections applied	Scenario 3 Measurement of P_{KA} during patient fluoroscopy, P_{KA} chamber calibrated in situ
Kerma rate	5%	5%	5%
In situ calibration of P_{KA} chamber	—	—	7.5%
difference in table attenuation compared to in situ calibration point due to varying beam hardness (under couch systems)	—	—	8%
air density correction:			
pressure	0.2%	2%	2%
temperature	0.5%	2%	5% (2)
Electromagnetic compatibility and humidity, other uncertainties estimated < 1% each	2%	2%	2%
Backscatter factors	5%	5%	—
combined expanded (2σ) uncertainty in K_i or P_{KA}	10%	12%	24%

(1) Mostly due to added copper filters.
(2) Uncertainty due to temperature partly due to warming of chamber from tube housing.

4.4.3. Comparison of methods

The variation in air kerma rate measurements between participants may best be illustrated by a box plot as shown in Figure 26. For each hospital, a range of entrance air kerma rates was obtained, varying with intensifier field size setting and, where available, fluoroscopy mode (e.g. pulsed or continuous). In the figure, each data set illustrates the range of data (from different field size settings) for a specific hospital and fluoroscopy mode. The different hospitals are indicated by different letters, and the different fluoroscopy modes by different numbers e.g. B1 is hospital B, continuous fluoroscopy, and B2 is hospital B, pulsed fluoroscopy. The extent of each box represents data between the 1st and 3rd quartiles of the data set, with lines extending to maximum and minimum values. The mean value of the data set is represented by a cross and the median value by a horizontal line. A wide variation in entrance air kerma rates can be seen between hospitals. As expected, pulsed fluoroscopy settings give lower air kerma rates than continuous settings for the same hospital.

For the patient measurements, KAP can be compared according to examination type, for barium meal and barium enema examinations, as illustrated in Figures 27 and 28.

This data also shows considerable variation between hospitals, as is commonly observed for patient dose data. The high values at hospital F1 is likely to be due to the relatively high number of radiographs taken during these examinations, particularly in barium enemas, which are included in the total KAP. The data from hospital H8 is from old equipment which has since been replaced.

In order to compare the two methods, paired data is required comprising both phantom and patient measurements on the same equipment. Three participants have obtained such data for a total of 11 hospitals. Figures 29 and 30 show mean KAP values from patient data plotted against the corresponding air kerma rate measurement made using a phantom (either 18.5 cm PMMA or 200 mm water) for barium meal and barium enema examinations. On these graphs, each dot represents a single hospital.

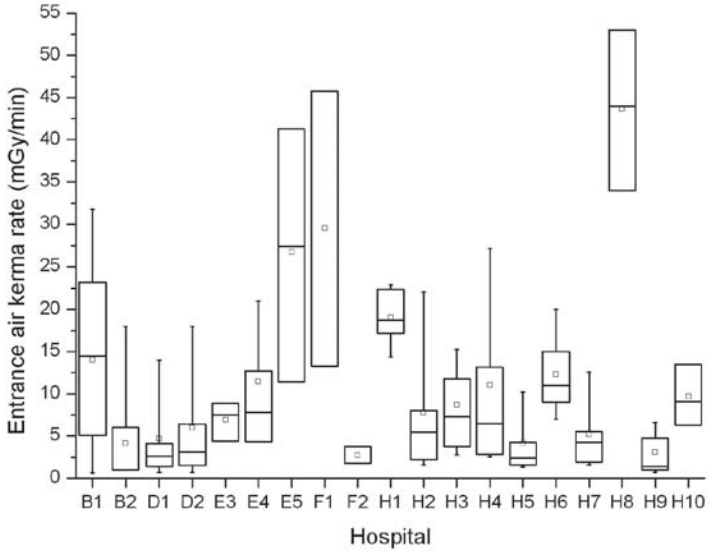

FIG. 25. Ratio of patient to phantom doses. Dotted line corresponds to mean value, solid line in the box to median.

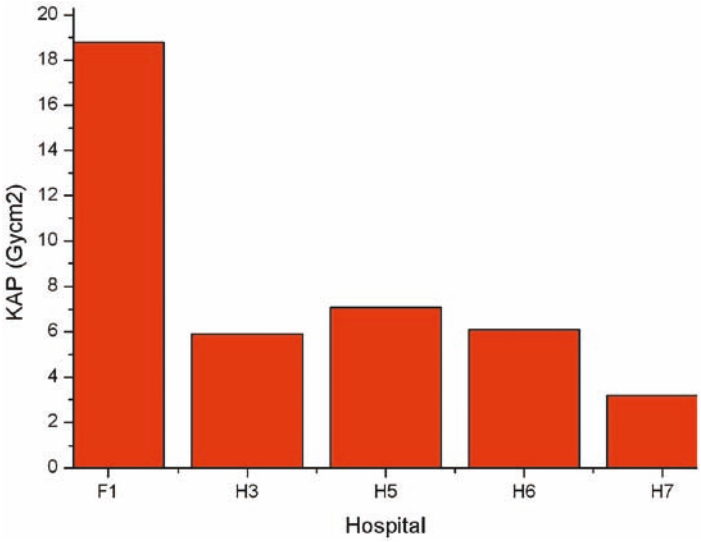

FIG. 27. KAP for barium meal exams.

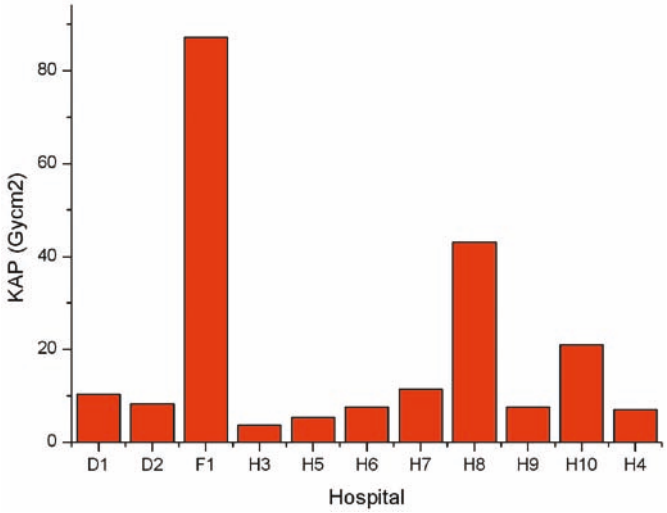

FIG. 28. KAP for barium enema.

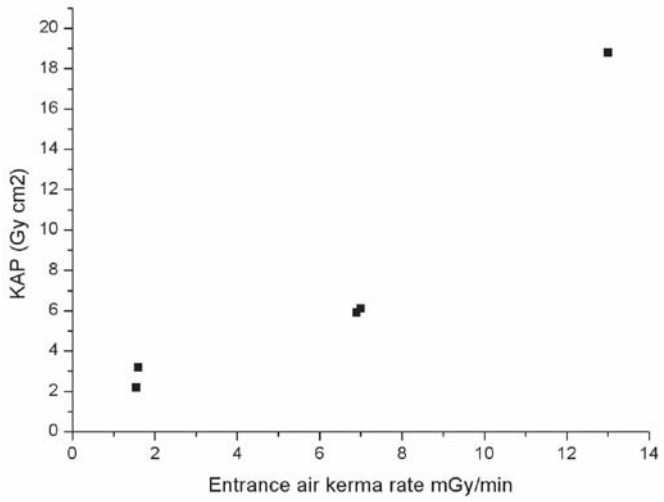

FIG. 29. Patient vs. phantom measurements for barium meal examinations.

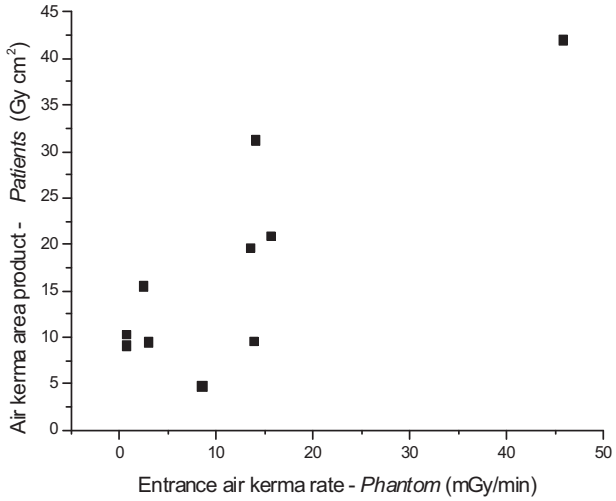

FIG. 30. Patient vs. phantom measurements for barium enema examinations.

The graphs show that although it is not possible to derive a numerical relationship between phantom and patient data, there appears to be is a clear trend between the two sets of data; particularly for the high dose outliers. Although the equipment air kerma rate is only one of many factors affecting patient dose, a high value for this parameter is very likely to correspond to above average values of patient KAP. This is well illustrated by hospital H8 where the high patient KAP values during a barium enema examination corresponded to high values of phantom entrance air kerma measurements. The results led to engineers being called in to look at the equipment. The equipment was near the end of its working life and was consequently taken out of use.

The trend between phantom and patient measurements is seen more clearly for barium meal examinations than for barium enema examinations. This is likely due to the fact that, for the latter, patient measurements generally include a greater number of image acquisitions, in particular high dose radiographic images. For some examinations, such as cardiac studies, image acquisition accounts for a large percentage of the total dose and, in these cases, TRS No. 457 phantom measurements of fluoroscopy entrance air kerma rate alone will not be indicative of the patient dose. Similar measurements of entrance air kerma per acquisition would be needed to provide a more complete assessment of the dose characteristics of the equipment.

4.4.4. Conclusions

The experiences of the participants, and analysis of their results, lead to the following recommendations; although it would be desirable to confirm these with a more extensive set of measurements from a greater range of

clinical centres. The dosimetry methods given in TRS No. 457 for clinical measurements are, in general, easily applicable to the majority of diagnostic X ray departments. Difficulties, if encountered, are likely to be due to the lack of availability of KAP meter equipment or lack of suitable staff to collate the patient technique data. Both these points relate to patient, rather than phantom measurements.

— The observed trends in patient air kerma product generally follow the trends in phantom air kerma rate measurement, where the choice of fluoroscopy program has been taken into account and there is not a significant contribution to dose from radiographic images. This indicates that where patient measurements are not possible, knowledge of the programs used clinically, along with incident air kerma rates measured with a phantom for these settings may be used as a broad indicated assessment of patient dose.
— Where clinical protocols involve a large number of image acquisitions, in addition to fluoroscopy, the entrance surface air kerma per acquisition will often be a more important predictor of patient dose than fluoroscopy air kerma rate. The methodology for phantom measurements should thus be extended to include similar measurements for acquisition settings, particularly if patient measurements are not possible.
— Paediatric dosimetry is an important area not specifically addressed in TRS No. 457. Entrance air kerma rates could be determined with 10 and 15 cm water phantoms to assess paediatric patient protocols.

4.5. MAMMOGRAPHY

TRS No. 457 consistently suggests two kinds of mammography dose measurements, using either phantoms or measurements with patients. Participants for this activity were Cuba, Czech Republic, Finland, Greece, Republic of Korea and Hungary, the latter was also the activity coordinator. The evaluation of the results and experiences are based on six sets of phantom measurements and 8 sets of measurements on patients, using 8 mammography X ray units.

4.5.1. Clinical dose measurement methods

4.5.1.1. Phantom measurements

The phantom recommended in TRS No. 457 for determination of incident air kerma, K_i, and the mean glandular dose is a 45 mm thick, homogeneous PMMA phantom. This phantom simulates a standard breast of 50 mm thickness and 50% glandularity. It can be concluded that all of the participants used the phantom recommended in TRS No. 457. However, the manufacturer and the serial number of the phantoms were not recorded on the datasheets. Six sets of phantom measurements were performed using the preferred method in TRS No. 457, where the air kerma was measured in the absence of the phantom using a suitable ionisation chamber. The Czech Republic tested both alternative methods in TRS No. 457, using the additional entrance surface air kerma as measured by TL dosimeters on the same mammography X ray equipment. A comparison of these experiences can be found in Section 4.5.2.

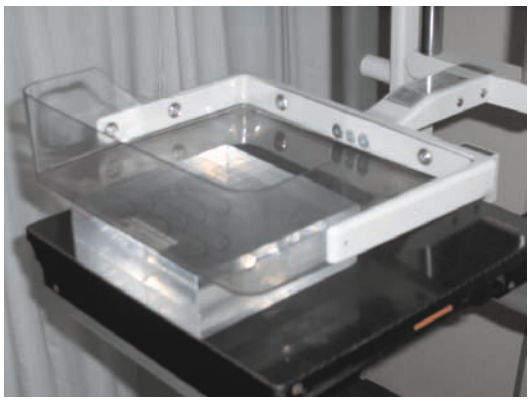

FIG. 31. Determination of tube loading.

Set-up of the two phases of phantom measurement is shown in Figures 31 and 32. Following the instructions in TRS No. 457, determination of tube loading was performed with the compression plate lowered down onto the phantom. For the measurement of the incident air kerma at the relevant tube loading, the phantom was removed, the compression plate was lifted up to halfway between the focal spot and the breast support table, and the radiation detector was positioned at the mammographic reference point.

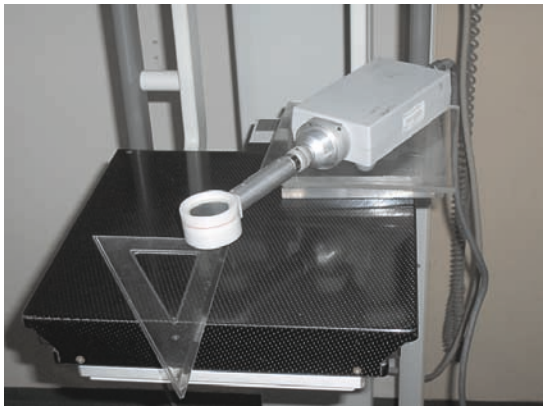

FIG. 32. Measurement of incident air kerma at the relevant tube loading.

The type of the mammography X ray equipment and the film screen combinations used can be found in Table 34.

TABLE 34. THE MAIN CHARACTERISTICS OF THE MAMMOGRAPHY X RAY EQUIPMENT

Identification	Equipment	Screen film
A	Performa MGF-110 Alpha RT	Fuji UM-MA HC Agfa Mamoray HD
B	Hologic LoRad Selenia	digital
C	Instrumentarium Alpha RT	Fuji UM-MA HC Agfa Mamoray HD
D	Siemens Mammomat 3000	Agfa Mamoray Agfa Mamoray
E	Siemens Mammomat 3000	no data
F	Agfa DM 1000	no data
G	Alpha 1 Ausonics	Fast screen KODAK Slow Chinese screen
H	Siemens Mammomat 2	Fast screen KODAK Slow Chinese screen

Table 35 shows that for the standard breast phantom, the exposure is typically performed with a Mo/Mo target filter combination, under automatic exposure control and at 28 kV tube voltage. However, the tube loadings under the AEC cover a twofold range. This wide range can be explained by other technical factors, such as the different focus to film distance and the AEC settings, etc. All X ray machines were equipped with post-exposure indication of the tube loading. Phantom measurements were not performed by G and H.

Dose measurements typically were performed with ionization chambers. The type of ionization chamber, the calibration factors, $N_{k,Q0}$, and the reference conditions of calibration are summarized in Table 36.

TABLE 35. EXPOSURE CONDITIONS FOR THE STANDARD BREAST PHANTOM

Identification	A	B	C	D	E	F
Target/filter	Mo/Mo	Mo/Mo	Mo/Mo	Mo/Mo	Mo/Mo	Mo/Mo
kV	28	29	27	28	28	30
AEC	yes	yes	yes	Yes	yes	yes
mAs	54	53	59	36	61	77

TABLE 36. CALIBRATION FACTOR OF THE IONIZATION CHAMBERS AND THE REFERENCE CONDITIONS OF CALIBRATION

Identification	Dosimeter	Beam quality HVL	P_0 (kPa)	T_0 (^oC)	$N_{k,Q0}$
A	Radcal 9010 10X5-6M	0.26 mm Al	101.3	22	1 mGy/reading
B	Radcal 9095	0.24 mm Cu	101.3	22	1.0643
C	Radcal 9015 10X5-6M	0.37 mm Al	101.3	20	1.003 mGy/reading
D	Radcal 9015 10X5-6M	0.37 mm Al	101.3	20	1.003 mGy/reading
E	Radcal 9015 10X5-6M	0.37 mm Al	101.3	20	1.003 mGy/reading
F	Radcal 9015 10X5-6M	0.37 mm Al	101.3	20	1.003 mGy/reading
G	PTW 77334	0.32 mm Al	101.3	20	18.19 mGy/nC
H	PTW 77334	0.32 mm Al	101.3	20	18.19 mGy/nC

Table 37 shows the recalculated dosimeter readings corresponding to the tube loading recorded under AEC, M_{auto}, the correction factor for temperature and pressure, k_{TP}, the factor which corrects for differences in the response of the dosimeter at calibration quality Q_0 and the quality Q of the clinical X ray beam, k_Q, as well as the incident air kerma, K_i. Only one participant (B) performed beam quality correction of calibration factor of the dosimeter. The maximum correction factor for temperature and pressure, k_{TP}, is 1.010. The average incident air kerma is 7.342 mGy with the range of 4.222–12.47 mGy.

TABLE 37. CALCULATED DOSIMETER READINGS, CORRECTION FACTORS AND THE INCIDENT AIR KERMA

Identification	A	B	C	D	E	F
M_{auto}	7.863	7.78	6.601	4.168	4.818	12.74
k_{TP}	1.002	0.999	1.007	1.010	1.000	1.000
k_Q	1	1.02	1	1	1	1
K_i	7.335	8.45	6.760	4.222	4.818	12.47

The conversion coefficient $c_{D_{G50},K_{i,PMMA}}$ for the measured half value layer and the 50 mm standard breast thickness and 50% glandularity that is simulated by the 45 mm PMMA phantom, converts the incident air kerma to the PMMA phantom to the mean glandular dose for the standard breast using the equation:

$$D_G = c_{D_{G50},K_{i,PMMA}} \cdot s \cdot K_i$$

For the Mo/Mo target/filter combination s=1. See Table 8.6 in TRS No. 457.

Table 38 shows the most important exposure conditions, the measured half-value layers, $HVL_{meas.}$, the interpolated conversion coefficients, $c_{D_{G50},K_{i,PMMA}}$, and the mean glandular dose, D_G, to the standard breast with the expanded uncertainties can be found.

TABLE 38. THE EXPOSURE CONDITIONS, THE MEASURED HALF-VALUE LAYERS, HVL_{MEAS}, AND THE INTERPOLATED CONVERSION COEFFICIENTS

Identification	A	B	C	D	E	F
Target/filter	Mo/Mo	Mo/Mo	Mo/Mo	Mo/Mo	Mo/Mo	Mo/Mo
kV	28	29	27	28	28	30
HVL_{meas}	0.346	0.368	0.369	0.381	0.341	0.365
$c_{D_{G50},K_{i,PMMA}}$	0.200	0.202	0.210	0.215	0.198	0.208
D_G (mGy)	1.47	1.7	1.42 ± 0.12	0.91 ± 0.12	0.95 ± 0.08	2.59 ± 0.22

4.5.1.2. Patient measurements

The mean glandular dose for a given patient is estimated using knowledge of the selected exposure parameters and compressed breast thickness as well as the measured X ray tube output. Tables 39–41 contain the results of patient dose measurements. Each of these tables contain the average, the minimum and maximum value as well as a correlation coefficient, indicating the correlation between the compressed breast thickness and the given parameter of either tube voltage, tube loading or mean glandular dose. Every second data line in the table contains the data for the subgroup of patients where the compressed breast thickness is exactly 50 mm. The third data line summarizes again the results of phantom measurements.

The compressed breast thicknesses covered a range of 20–75 mm for almost every mammography unit. The average compressed breast thickness value for the mammography units is in the range of 40.8–60 mm and the overall average is 49 mm.

The most typical target/filter combination is Mo/Mo; however, some of the mammography X ray projections are performed using Mo/Rh target/filter combinations.

Generally, the tube voltage values cover a wide range, but C used only one kV value, that is 27 kV. It can be seen that mammography units G and H use higher tube voltage ranges, 30–35 kV and 29–35 kV, respectively for each unit. Only poor correlation can be found between the tube voltages and the compressed breast thickness.

The average mAs values range from 41 mAs to 100 mAs. G and H reported that the mammography units examined applied constant 80 or 100 mAs because the mammography units are not equipped with automatic exposure control. Consequently, in the case of G and H, there is no reason to find correlation between the breast thicknesses and the tube loadings. At the other mammography centres, a strong correlation was found with the exception of the Republic of Korea ($r = 0.46$).

The average incident air kerma values are the smallest for C, D, E and F units participating in the CRP, while the highest are for G and H units. Generally the doses cover a wide range; the values from G and H cover only a two-fold range. The fact is that those mammography units use constant tube loading with the kV values altering only a little amount. By evaluating the correlation coefficients between the compressed breast thickness and the incident air kerma, it can be conclude, that the two values are correlated, but this correlation is weak in case of the mammography G and H units.

TABLE 39. THE RESULTS OF PATIENT DOSE ESTIMATIONS; NUMBER OF PARTICIPANTS, TARGET/FILTER COMBINATION AND TUBE VOLTAGE

Participant	N	Target/filter	Tube voltage (kV)			
			Average	Min.	Max.	Corr.
C	40	Mo/Mo	27	27	27	—
	5	Mo/Mo	27	27	27	—
	(Phantom)	Mo/Mo	27	—	—	—
D	19	Mo/Rh	27.4	26	29	0.79
	1	Mo/Rh	27	—	—	—
	(Phantom)	Mo/Mo	28	—	—	—
G	33	Mo/Mo	32	29	35	0.56
	2	Mo/Mo	35	35	35	—
	(Phantom)		—	—	—	—
G*	20	Mo/Mo	33	30	35	0.55
	4	Mo/Mo	34.3	33	35	—
	(Phantom)	—	—	—	—	—
H	19	Mo/Mo	28	25	30	0.68
	2	Mo/Mo	26	25	28	—
	(Phantom)		—	—	—	—
H*	57	Mo/Mo	28.3	28	30	0.57
	12	Mo/Mo	28	28	28	—
	(Phantom)		—	—	—	—
B	12	Mo/Mo and Mo/Rh	27.7	24	32	0.99
	1	Mo/Mo	30	—	—	—
	(Phantom)	Mo/Mo	29	—	—	—
A	18	Mo/Mo and Mo/Rh	27	24	29	0.56
	2	Mo/Mo	27	26	28	—
	(Phantom)	Mo/Mo	28	—	—	—

G* and H* are MLO projections.
Note: Every second data line in the table contains the data for the subgroup of patients where the compressed breast thickness is exactly 50 mm.

TABLE 40. THE RESULTS OF PATIENT DOSE ESTIMATIONS; BREAST THICKNESS AND TUBE LOADING

Participant	Breast thickness (mm)				Tube loading, $P_{It,pat}$ (mAs)			
	Average	Min.	Max.	Corr.	Average	Min.	Max.	Corr.
C	47	20	70	1	45	12	90	0.88
	50	—	—	—	40	32	51	—
	50	—	—	—	59	—	—	—
D	53	29	85	1	41	17	110	0.83
	50	—	—	—	30.7	—	—	—
	50	—	—	—	36.1	—	—	—
G	41	35	55	1	100	100	100	—
	50	—	—	—	100	100	100	—
	—	—	—	—	—	—	—	—
G*	43	38	50	1	100	100	100	—
	50	—	—	—	100	100	100	—
	—	—	—	—	—	—	—	—
H	51	35	60	1	80	80	80	—
	50	—	—	—	80	80	80	—
	—	—	—	—	—	—	—	—
H*	53	45	60	1	80	80	80	—
	50	—	—	—	80	80	80	—
	—	—	—	—	—	—	—	—
B	40.8	20	70	1	73.9	49	104	0.46
	54	—	—	—	104	—	—	—
	50	—	—	—	53.27	—	—	—
A	60	35	75	1	82	51	122	0.81
	50	—	—	—	54.5	51	58	—
	50	—	—	—	54	—	—	—

G* and H* are MLO projections.
Note: Every second data line in the table contains the data for the subgroup of patients where the compressed breast thickness is exactly 50 mm.

TABLE 41. THE RESULTS OF PATIENT DOSE ESTIMATIONS; INCIDENT AIR KERMA AND MEAN GLANDULAR DOSE

Participant	Incident air kerma, K_i (mGy)				Mean glandular dose, D_G			
	Average	Min.	Max.	Corr.	Average	Min.	Max.	Corr.
C	5.28	1.2	10.5	0.89	1.04	0.56	1.88	0.65
	4.6	3.68	5.87	—	0.88	0.7	1.12	—
	6.76	—	—	—	1.42	—	—	—
D	3.39	1	11.3	0.86	0.66	0.35	1.61	0.72
	2.2	—	—	—	0.5	—	—	—
	4.22	—	—	—	0.91	—	—	—
G	16.3	11.5	21.5	0.64	4.96	3.81	6.47	0.66
	21.56	21.56	21.56	—	6.47	6.47	6.47	—
	—	—	—	—	—	—	—	—
G*	18.1	12.7	21.4	0.64	5.45	3.95	6.42	0.65
	20.6	18.3	21.4	—	6.19	5.5	6.42	—
	—	—	—	—	—	—	—	—
H	13.1	8.96	16.1	0.78	4.06	2.87	4.83	0.79
	10.98	8.96	13	—	3.52	2.87	4.17	—
	—	—	—	—	—	—	—	—
H*	13.6	12.8	16.08	0.76	4.29	4.1	4.82	0.83
	13.04	13.04	13.04	—	4.17	4.17	4.17	—
	—	—	—	—	—	—	—	—
B	10.6	5.35	20	0.92	2.42	1.58	3.71	0.49
	18.63	18.63	18.63	—	3.71	—	—	—
	8.45	—	—	—	1.7	—	—	—
A	9.1	3.9	13.9	0.83	1.68	0.87	2.47	0.76
	6.75	6.3	7.2	—	1.26	1.17	1.35	—
	7.34	—	—	—	1.47	—	—	—

G* and H* are MLO projections.
Note: Every second data line in the table contains the data for the subgroup of patients where the compressed breast thickness is exactly 50 mm.

The mean glandular dose values are in the range of 0.66–5.45 mGy and the correlation with the compressed breast thickness is moderate.

4.5.2. Comparison of methods

The comparison of the two methods for patient dose estimation in mammography can be made by analyzing the data in Tables 39–41.

For mammography unit C, the average compressed breast thickness involved in the study was 47 mm. This value is very close to the compressed breast thickness of 50 mm which is simulated by the 45 mm thick PMMA phantom. The glandularity is assumed to be 50%. By comparing the technical parameters of mammographic exposure, it can be concluded that the exposures are made on constant tube voltage and this tube voltage is the same that was applied for the phantom measurements. The target/filter combination was also the same. However, the tube loadings are very different; the tube loading is about 30% higher for the phantom measurements. In addition, the same tendency can be found in the incident air kerma and mean glandular dose values. In the subgroup of patients for which the compressed breast thickness is exactly 50 mm, the difference between the dose values for the patients and the phantom are more significant. This may suggest that the composition of the breast for the participants is significantly less than 50% glandular tissue.

For D, the average compressed breast thickness of the patients is also very close to the compressed breast thickness of 50 mm which is simulated by the 45 mm thick PMMA phantom. The tube voltages used were the same, but while the patient exposures are made by using Mo/Rh target/filter combination, the phantom exposures were made by using the Mo/Mo target/filter combination. The tube loadings for the patient exposures are about 14% higher. The incident air kerma and mean glandular dose values are about 25% and 38% higher for the phantom which can be caused by the different target/filter combination.

For B, the average breast thickness and the average tube voltage are less than the breast thickness of 50 mm, representing the phantom used and the tube voltage applied by the phantom measurements. Surprisingly, the average value of the tube loading, the incident air kerma and the mean glandular dose are higher for the patients. The differences are significant, 39%, 25% and 42% respectively.

For A, the average compressed breast thickness is about 20% higher than the compressed breast thickness of 50 mm, which is simulated by the 45 mm thick PMMA phantom. Consequently, the tube loading and the estimated dosimetric values are also higher for the patients.

The partners from G and H units did not perform phantom measurements, so comparison of the results cannot be made.

If a radiological patient equivalent phantom is used, phantom doses are expected to be close to the corresponding average patient doses if identical technical parameters are used. However, it is well known that breast glandularity varies between women and also generally decreases with age [26–29]. According to our study, the phantom measurement cannot directly indicate the doses to patients. However, it can be pointed out that the dose values for the phantom are between the minimum and maximum values of the appropriate patient dose values.

4.5.3. Conclusions

The experiences of the participants and analysis of their results lead to the following conclusions and recommendations:

- The methods in TRS No. 457 are clear and applicable in a clinical environment. Their widespread use can promote the evaluation of temporal tendency in patient medical exposures and the comparison of patient dose measurement results among different countries and regions worldwide.
- To improve the confidence level of patient dose estimation, it is suggested to increase the number of patients involved in the dose measurements.
- The use of software for the calculation of mean glandular dose is encouraged.

4.6. COMPUTED TOMOGRAPHY

This section contains the results of the hospital measurements made for computed tomography (CT) equipment; including both phantom and patient measurements. There have been six participating countries for these data including a total of 20 hospitals.

4.6.1. Clinical dose measurement methods

Common to both phantom and patient dose measurement methods is the use of the k_Q formalism, to correct readings from a chamber calibrated at one standard beam quality to the beam quality actually used. This approach has proven unfamiliar to most clinical participants and more guidance may be required from SSDLs to users to enable such factors to be derived and applied correctly. It is particularly important that the use, or lack of use, of a k_Q factor is accounted for when assessing the uncertainty budget for the measurements.

4.6.1.1. Phantom measurements

These data comprised measurement of computed tomography air kerma indices in air ($C_{a,100}$) and in head and body phantoms (C_W), according to the recommendations of TRS No. 457. The worksheets used to collect the data are based on those given in TRS No. 457, Section 8.7.2.5. A range of Perspex phantoms were used to carry out the weighted CTDI measurements, depending on local availability, including RADCAL and WELLHOFER phantoms. The X ray equipment for which measurements were carried out is detailed in Table 42.

TABLE 42. X RAY EQUIPMENT USED FOR PHANTOM MEASUREMENTS IN CT

Manufacturer	Model	Number of slices
GE	Lightspeed	4
	Lightspeed Plus (2 scanners)	4
	Lightspeed VCT (3 scanners)	64
Philips	Brilliance 40 (2 scanners)	40
	MX8000	16
Siemens	Tomoscan AV	1
	Somatom Plus (2 scanners)	1
	Somatom AR.HP	1
	Somatom AR.SP	1
	Somatom Sensation 4	4
	Somatom Sensation 16	16
	Somatom Sensation 64 (2 scanners)	64
Toshiba	Asteion	4
	Aquilion	16
	Aquilion	64

For all measurements, Radcal ionisation chambers were used with corrections made for temperature and pressure and the calibration coefficient of the chamber. For a few centres, the dosimeter used is not routinely calibrated in terms of dose length, and so an extra factor for chamber length needed to be incorporated into the calculation of $C_{a,100}$ and C_W. No participants applied a beam quality correction factor even though reference conditions for calibration of the chambers varied considerably. The dosimeters used and correction factors applied at the various institutions are given in Table 43.

For measurements made in air, the CT air kerma index is calculated using[13]:

$$C_{\mathrm{a},100} = \frac{10}{nT} M\, N_{P_{\mathrm{KL}},Q_0} k_Q\, k_{\mathrm{TP}}; \quad {}_{\mathrm{n}}C_{\mathrm{a},100} = \frac{C_{\mathrm{a},100}}{P_{\mathrm{I}t}}$$

For the measurement in Perspex head and body phantoms, the CT air kerma indices are calculated using:

$$C_{\mathrm{PMMA},100,\mathrm{p}} = \frac{10}{NT} \bar{M}_{\mathrm{p}}\, N_{P_{KL},Q_0} k_Q\, k_{\mathrm{TP}}$$

$$C_{\mathrm{W}} = \frac{1}{3}\left(C_{\mathrm{PMMA},100,\mathrm{c}} + 2\, C_{\mathrm{PMMA},100,\mathrm{p}}\right) \text{ and } {}_{n}C_{\mathrm{W}} = \frac{C_{\mathrm{W}}}{P_{\mathrm{It}}}$$

where M is the mean dosimeter reading, P_{IT} the total tube loading for that slice during measurement, nT is the total width of the irradiated slice in mm, $N_{PKL,Q0}$ is the calibration factor for the dosimeter at reference beam quality Q_0, k_Q is the beam quality correction factor, and k_{TP} the temperature/pressure correction factor.

TABLE 43. DOSIMETERS AND CORRECTION COEFFICIENTS USED FOR PHANTOM MEASUREMENTS IN CT

Participants	1[a]	2[b]	3[c]	4[d]	5[e]	6[f]
Dosimeter model	PTW Unidos	RADCAL 2025-3CT	RADCAL 10X5-3CT	RADCAL 9015	RADCAL 20X6-3CT	RADCAL
Dosimeter serial no.	1100	4014	8929	91-0059	26-0774	
Calibration traceability	PTB	PTB	PTB	PTB	PTB	NPL
N_{PKLQo*} used	Yes	Yes	Yes	Yes	Yes	Yes
k_{TP} used	Corrected by instrument	Yes	Yes	Yes	Yes	Yes
k_Q <> 1	No	No	No	No	No	No

[a] Austria
[b] Greece
[c] Finland
[d] Hungary
[e] Republic of Korea
[f] UK

The uncertainty budget for the measurements was calculated by each participant in accordance with TRS No. 457. An example of this analysis for 1 participant is given in Table 44.

Some of the points noted by participants during the measurements included some difficulty experienced in selecting axial mode, as required for the CT air kerma indices. Confusion may well have arisen from the advice given in TRS No. 457 to make CTDI measurements for all clinically used settings, which are now generally helical, while the CT air kerma measurement protocol is specifically for axial scans (see Appendix VII). Difficulties were also experienced in locating the position of maximum peripheral dose. The lack of availability of beam quality correction factor is noted, but should not be a problem provided that calibration of the ionisation chamber was

[13] The factor of 10 in the formulae for $C_{\mathrm{a},100}$, $C_{\mathrm{PMMA},100,\mathrm{c}}$ and $C_{\mathrm{PMMA},100,\mathrm{p}}$ takes into account the use of a dosimeter calibration in mGy.cm and a slice thickness specified in mm.

TABLE 44. UNCERTAINTY BUDGET FOR CT DOSE CALCULATIONS

Influence quantity	Uncertainty (%) k=1
Intrinsic error $N_{K,Q}$	2.0
Radiation quality	0.5
Kerma rate	0.5
Direction of radiation incidence	1.0
Air pressure	0.5
Temperature and humidity	0.5
Electromagnetic compatibility	1.5
Field size/field homogeneity	1.0
Operating voltage	1.2
Long term stability of user's instrument	0.5
Precision of reading	0.6
Precision of tube loading indication	1.0
Precision of chamber/phantom positioning in the centre of the gantry	0.3
Uncertainty of 1mm in phantom diameter and 0.5 mm in depth of measurement bores	0.35
Uncertainty in chamber response for in-phantom measurements (C_w only)	3.0
Relative combined standard uncertainty (k=1) for $C_{a,100}$	3.5
Relative expanded uncertainty (k=2) for $C_{a,100}$	7.0
Relative combined standard uncertainty (k=1) for C_W	4.6
Relative expanded uncertainty (k=2) for C_W	9.2

carried out at a similar beam quality to that found in CT, and that the chamber energy response is fairly flat. This should however be reflected in the uncertainty budget.

4.6.1.2. Patient measurements

Patient dose is specified in terms of C_{VOL} and $P_{KL,CT}$ using phantom measurements in conjunction with individual patient scan parameters, rather than making any direct measurement of patient dose. There has been much discussion over recent months, within the scientific community, regarding the appropriateness of these quantities for assessing patient dose. A review of CT dose quantities, particularly in relation to multi-slice scanner technology, may be desirable, but is beyond the scope of this publication. $P_{KL,CT}$ is the quantity most closely related to patient dose and can be used for comparing techniques and setting diagnostic reference levels. However, it does not represent, and was never intended to represent, an 'actual' dose to the patient. The phantoms used in its derivation are not good representations of human size and anatomy. Work has been carried out to relate actual organ doses and effective dose to $P_{KL,CT}$, as well as to ${}_nC_a$. C_{VOL} represents the part of $P_{KL,CT}$ relating to the scanner and technique selection, rather than the scan length. It is now customary for CT scanners to display a value for $P_{KL,CT}$ at the end of a scan (usually referring to it as dose length product or DLP) and this may be particularly useful as it will often (although not always, depending on scanner type) include the effects of any dose modulation that has been used. However, the accuracy of such displayed values should always be checked, by either using the calculation methods in TRS No. 457 or a direct measurement method.

All of the participants collated patient measurements for at least some of the equipment used in the above phantom measurements. The range of examinations for which data was collected included: head, chest and abdomen studies, and exposure parameters were recorded for each examination. Some of the data sets contain relatively small numbers of patients and may, in some cases, represent standard protocol rather than patient specific data.

$_nC_{VOL}$ is calculated using $_nC_{VOL} = {}_nC_W\,(NT/l)$ and $P_{KL,CT}$ is calculated using $P_{KL,CT} = {}_nC_{VOL}\,l\,P_{It}$ or, more simply $_nC_w\,NT\,P_{It}$.

where NT is the total width of the irradiated slice in mm, l is the couch movement per rotation in mm, and P_{IT} is the total mAs for the scan.

Some of the common issues arising for the participants included problems in obtaining good sets of patient data, presumably due to lack of access to or time on the scanners, and potential difficulties in interpreting the information provided on the scanner console (e.g. mAs or effective mAs). It was also noted that the worksheets reference the quantity $_nC_{vol}$ rather than C_{vol}, although both are discussed in TRS No. 457. The latter quantity would seem to be more appropriate as a patient related dose quantity as it includes the actual mAs used in the examination.

One important additional point relates to the calculation of the dose parameters on the Siemens Sensation 64 scanner, which uses a flying focus to obtain the full 64 slices from 32 × 0.6mm detector rows. In order to calculate the dosimetric quantities correctly, for any scan using a nominal 64 × 0.6 acquisition it is necessary to use 32 × 0.6 in all equations involving NT as this is the correct total width of the irradiated slice.

4.6.2. Comments on methods

4.6.2.1. Phantoms

Figure 33 shows the $_nC_{a,100}$ data for the 3 GE scanners at 120 kV for each available slice width, and Figs 34 and 35 show the corresponding $_nC_w$ data for head and body modes respectively.

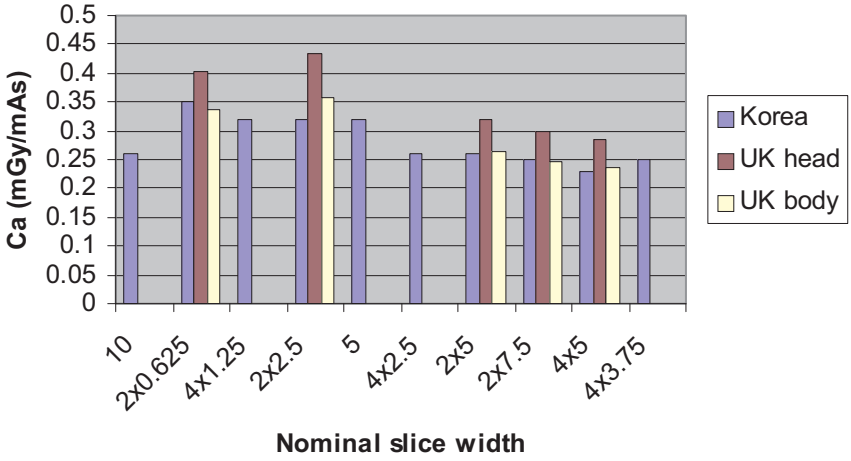

FIG. 33. $_nC_{a,100}$ at each institution.

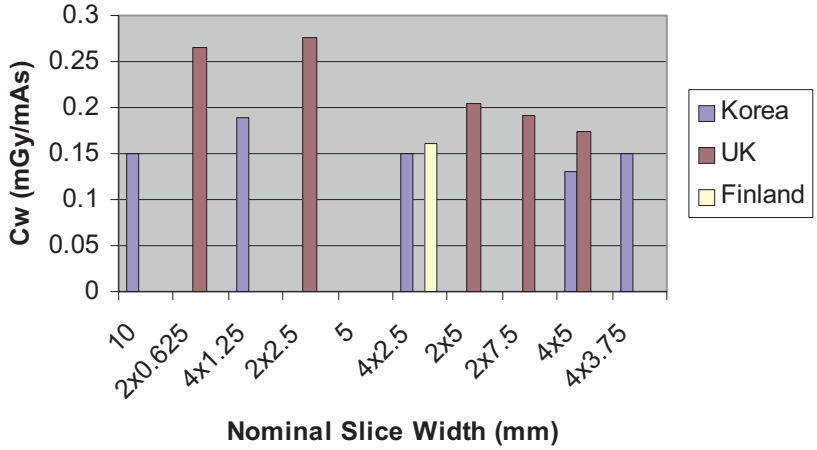

FIG. 34. $_nC_w$ (head mode).

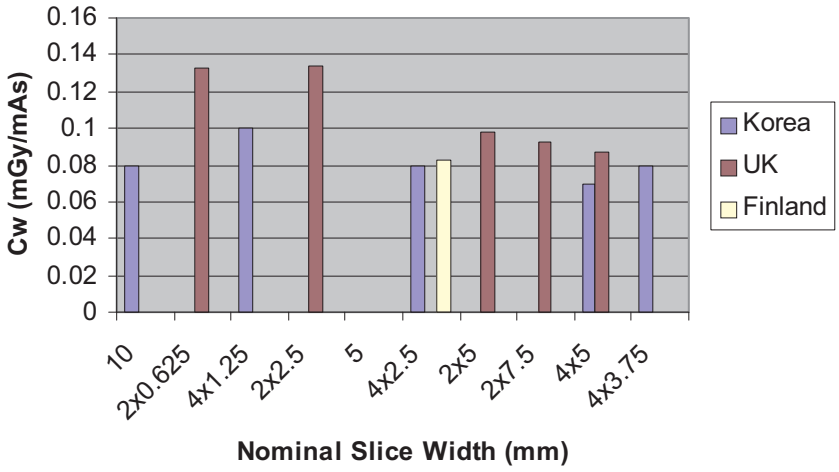

FIG. 35. ${}_nC_w$ (body mode).

The ${}_nC_a$,100 data shows good agreement between the Korean data and the head filter data from the UK, whereas the ${}_nC_w$ data shows the UK data to be significantly higher than that from the Republic of Korea and Finland. More data would be required to investigate the causes of this, although it may be due to slight differences in beam quality between the scanners.

A sample of the patient data is presented in Figs 36 and 37, giving calculated values of $P_{KL,CT}$ for head and abdomen examinations respectively.

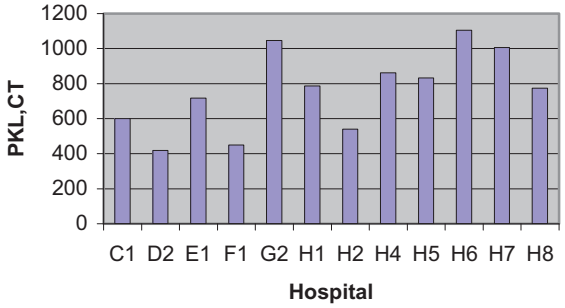

FIG. 36. Patient dose measurements for head examinations.

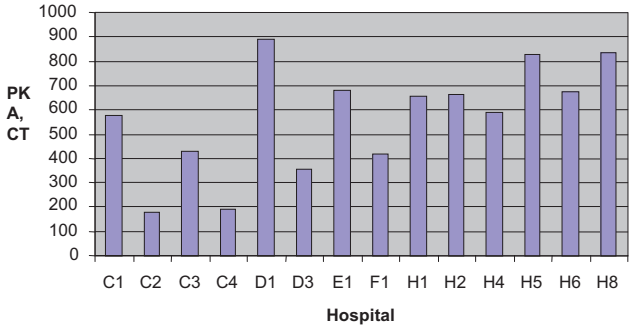

FIG. 37. Patient dose measurements for abdomen examinations.

The data shows a wide spread in value particularly for the abdomen examinations. This can be explained by a consideration of the technique factors reported by the various hospitals. For the abdomen examinations there were wide differences in the scan length used, with some hospitals scanning the abdomen and pelvis. This highlights the need, when comparing data, to classify examinations in terms of the clinical indication rather than the anatomical area alone. For head examinations, the scan length was less variable, but reported mAs values varied by a factor of nearly 2, which has a corresponding effect on $P_{KL,CT}$.

No data was obtained for scanner displayed values of $P_{KL,CT}$ so a comparison of calculated and displayed values could not be carried out. This would be a useful additional exercise to carry out.

4.6.3. Conclusions

The experiences of the participants, and analysis of their results, lead to the following conclusions:

— The dosimetry methods given in TRS No. 457 for clinical measurements can, in general, be easily applicable to the majority CT scanners in diagnostic X ray departments. However, difficulties, if encountered, are likely to be due to problems in correctly interpreting the technique information reported by the scanners as this can vary widely between manufacturers, particularly for 3rd generation scanners. It may be necessary to adapt the data collection forms for the scanner in use.
— Phantom measurements alone, while a good tool for quality assurance, are not an adequate predictor of patient dose as they take no account of technique settings which can vary hugely between different sites, even for the same scanner.
— More detailed instructions for applying TRS No. 457 to different scanner types are given in Appendix VII.
— Scanner displayed values of dose length product may be most useful for data collection, although their accuracy needs to be verified by measurement and, for paediatric patients, the phantom size to which they relate should be known.
— Extra care needs to be taken in the interpretation of data when the dosimeter used has not been calibrated in terms of mGy cm, and also when making measurements on equipment using a flying focal spot.

4.7. DENTAL

Although dental radiographs deliver small doses to patients, their frequency is high. Therefore, especially when collective dose to the population is assessed, dosimetry of dental imaging is important. In commissioning and regular testing of dental radiographic equipment, dose has to be determined. Dental imaging modalities considered in TRS No. 457 are intraoral (bite wing) projections and OPG (panoramic radiographs).

4.7.1. Clinical dose measurement methods

4.7.1.1. Intraoral radiography

For intraoral radiography, the adopted dose quantity is the incident air Kerma, K_i. Three Member States; Austria, the Czech Republic and the United Kingdom, participated in the activity. All centres reported on the feasibility and encountered difficulties, if any, 2 centres reported values measured at 11 clinical systems. Of these, five systems in one Member State were measured with TLDs, the others with electronic dosimeters (see Table 45).

TABLE 45. DOSIMETRIC EQUIPMENT AND NUMBER OF SYSTEMS FOR BITEWING INTRAORAL RADIOGRAPHY SYSTEMS

Center	Dosimeter	Calibrated within 2 years of measurement	Number of systems
B	PMX-1/D	yes	5
C	Radcal 9010	yes	1
C*	TLD	Lab calibration	5

From all systems measured with ion chamber systems, at least 2 settings were reported, (molar — big, molar — small, and if three reported, also molar medium). Where TLDs were used, only one program was measured (upper molar, in Fig. 38 plotted pooled with molar medium data).

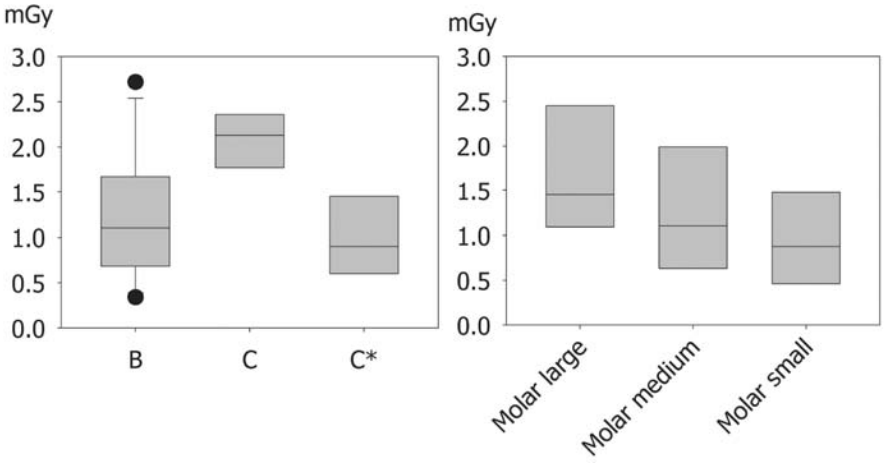

FIG. 38. K_i for intraoral radiographs by Member State (left) and by protocol.

4.7.1.2. Panoramic radiography (OPG)

In OPG systems, KERMA length product is the measured quantity of choice. Placing the chamber perpendicular to the slit collimator defines the lengths in the measured P_{KL} as the slit widths on the ion chamber. By multiplying the measured P_{KL} with the slit height at the chamber — focus distance the KERMA area product is derived.

Three Member States participated in this survey. Table 46 shows instrumentation and number of OPG systems from which measurements were reported. In all cases, a program corresponding to an average adult patient full OPG was used. One Member State also provided data from one OPG system measured with TLDs.

TABLE 46. INSTRUMENTATION USED AND NUMBER OF SYSTEMS INCLUDED

Center	Dosimeter	Chamber	Calibrated within 2 years of measurement	Systems
A	PTW Unidos M	CT Chamber	Yes	1
B	Radcal	CT Chamber	Yes	5
C	Radcal 9015	10X5-3CT	Not reported	4
C*	TLD		Lab calibration	1

Dose data is shown in Fig. 39.

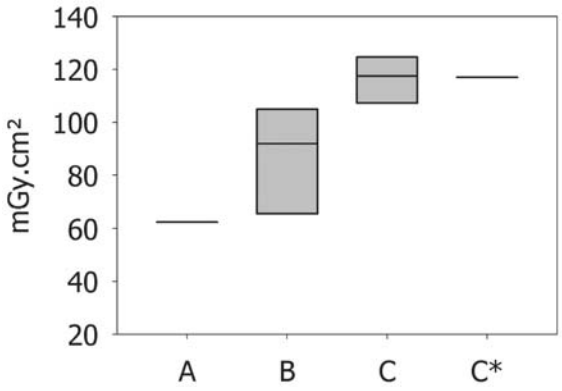

FIG. 39. P_{KA} values reported for OPG.

4.7.2. Comment on methods

4.7.2.1. Intraoral radiography

In intraoral radiography incident air kerma K_i is measured. Alternatively, P_{KA} could be determined by multiplying K_i with the beam area measured on a film.

In case positioning of the detector of the dosimeter at the centre of the exit of the spacer/director cone is not practical or feasible, it may be positioned at a larger distance. In this case the entries 'focus to tip distance' and 'focus to detector distance' in the worksheet provided on page 221 in TRS No. 457 are different and an inverse square law distance correction has to be performed. This may also be considered if the reference point of the chamber does not coincide with the chamber entrance wall. Distances from the focal spot to the spacer cone/director tip are very short.

If a chamber is used, positioning the ionisation chamber of the dosimeter using a tripod may be considered to keep sufficient distance from backscattering objects.

4.7.2.2. Panoramic radiography (OPG)

Measurement of kerma length product with a horizontally orientated P_{KL} chamber, normally used for dosimetry with CT systems, eliminates the otherwise often extremely tedious task of measuring dose with a small sensor in front of the slit collimator, especially if the position of the beam chances slightly during the rotation. Therefore, using a (small) dose sensor on a panoramic X ray machine has the potential of introducing both, type A and B uncertainties due to misalignment or incomplete sensor coverage. This is eliminated with the method described in TRS No. 457. Care has to be taken that the chamber is calibrated for appropriate dental beam qualities (RQR5), or that the appropriate correction factor is applied if calibration has been carried out at CT beam qualities (RQT).

After measuring the kerma length product with the horizontally aligned chamber, kerma area product is calculated by multiplying the P_{KL} measured with the beam height. According to TRS No. 457, when measuring the beam height, it may happen that the height is measured directly at the collimator. Usually, the reference line of the cylindrical chamber is the centre. In this case, the beam height will be measured in a slightly larger distance to the focal spot than the dose length product. A distance correction could easily be applied by multiplying the beam height measured with a distance correction factor

$$k_d = \frac{d_{fc}}{d_{fh}}$$

where d_{fc} represents the distance of focal spot to chamber reference line (centre of chamber), and d_{fh} the distance of focal spot to film used for measuring beam height. Alternatively, this correction factor can be applied to P_{KA} if P_{KA} is calculated according to TRS No. 457 as measured P_{KL} multiplied by the measured beam height. Note that this correction is not according to the inverse square law as for dose, since the quantity corrected is a dose length product. As an alternative method to measure P_{KA} for a panoramic examination, a KAP ionization chamber can be fixed on the X ray beam collimator to give a direct measure of P_{KA} [30, 31]. However, this method is not in general clinical usage and the method was not investigated in this study.

Note: there is no guidance on uncertainties here since that provided in TRS No. 457 is in enough detail.

4.7.3. Conclusions

In panoramic installations kerma area product determination provides a simple and accurate dose measurement methodology and is therefore recommended. P_{KA} is calculated by multiplying the measured kerma length product by the beam height, which may not necessarily be equal to the slit collimator height. Thus, measurement of the beam height is recommended:

— A distance correction of the beam height to the focus — centre of chamber distance can be used to reduce uncertainties.
— Using dose measurements, applying a small detector as may be required by local QC procedures is not recommended for two reasons. The first has been explained in the previous section; the second is that this methodology would not take the beam widths into account and can therefore not provide indication on dose to the patient.

4.8. CONCLUSIONS

4.8.1. Summary of use of TRS No. 457 dosimetry formalism

The dosimetry formalism as described in TRS No. 457 was initially unfamiliar to most clinical participants, particularly the use of instrument calibration coefficient and beam quality correction coefficient.

The calculation methodologies for CT were found to not be applicable to all scanner types, particularly 3rd generation multi-slice helical scanners. This led to much confusion in extracting and processing the required patient data.

4.8.2. Summary of instrumentation issues

The importance of calibrating all clinical dosimetry devices must be emphasized. This includes removable KAP meters in their clinical environment, integral KAP meters with console display and displayed CT scanner dosimetric values. All detectors, especially those that are solid state, must comply with IEC 61674. This is especially important with mammography and CT detectors.

4.8.3. Summary of phantom results

Availability of phantoms was not seen to be a major issue amongst participants, although some countries did not have the appropriate phantoms for general radiography. TRS No. 457 procedures were found to be straightforward to follow in all cases.

Since the standard size of adults varies between regions, it is not possible to have any phantom that represents standard conditions worldwide.

It is preferable that phantoms should be standardized with a quality system, country or region.

Phantoms should be exposed under standard exposure conditions, preferably under automatic exposure conditions.

4.8.4. Summary of patient data collection processes

It should be well understood that patient dose is not accurately determined by the use of phantoms.

Successful collection of valid data requires capable radiographers under the supervision of a medical physicist with a speciality in diagnostic radiology. These conditions are often not possible, leading to difficulties in data collection.

Availability of dosimetric instrumentation, especially KAP meters, has been an issue for some participants.

Access to patient data was also an issue as not all participants had clinical involvement.

4.8.5. Comparison of phantom and patient methodologies

Phantoms will ideally simulate a standard patient. However, this ideal is rarely possible when considering the difference in size of people between geographical regions in the world.

In general, it was concluded that actual patient measurements are preferable for assessing patient doses, with phantom measurements being more useful for quality assurance purposes. However, where patient measurements are not possible, phantom measurements may give some broad indication of typical dose levels provided that clinically used techniques are used.

4.8.6. Concluding summary

— Determination of patient dose has significant impact if the results are compared to dose reference levels. This will inform the process of optimization along with relevant image quality data.
— Phantoms ideally will simulate a standard patient. However this ideal is rarely possible.
— The importance of the use of k_Q values for dosimetry measurements in reducing the uncertainty budget must be emphasised to users.
— It is of great importance that users are aware of the consequences of the dosimetric method on uncertainty budget when making dosimetric measurements, particularly when compiling DRL data.
— It is recognized that care will be needed in paediatric dosimetry in order to achieve an acceptable uncertainty budget.
— Recommended future work is in paediatric dosimetry and the consequent uncertainty budget.
— Mammography dosimetry will benefit from the use of software systems. These are available in the current mammography QC documents.
— In mammography, the HVL value should be measured as part of the dosimetry method. If tabulated values are used, the effect on the uncertainty should be calculated.

5. FIELD CALIBRATIONS

5.1. KAP METERS

Field KAP meters used in clinical practice should be calibrated in situ whenever possible as this is the only way thata clinically relevant radiation environment, including scattering, can be incorporated in the calibration. One major example of this is the effect of patient couch on beam attenuation being dependant on the geometry used in clinical practice. However, in cases when field KAP meters need to be calibrated in a standards laboratory, a correction factor for clinical setting should be measured and applied [23].

5.1.1. General

Field calibrations of KAP meter were performed in three countries using the two different methods described in TRS No. 457, namely the diagnostic dosimeter method with the knowledge of the beam area at the reference plane and the reference KAP meter method, also referred to as the tandem method. The results were collected on calibration forms. The methods were evaluated and the results compared. The following problems were encountered. The comparison was based on the fact that the calibration of reference KAP chambers was completed according to TRS No. 457 with only RQR radiation qualities. Toroi et al. [19] demonstrated that HVL can not be used as a single radiation specifier for interpolation with KAP meters. However, interpolation based on two radiation quality specifiers is difficult and in some cases impossible, if only calibration with RQR radiation qualities is available (see also Section 3.1.2.4 and Appendix V). Another problem was that the methods were used in different geometries, e.g. one in the under coach and the other in the over coach situation, while further difficulties relating to the automatic operation of the clinical X ray machines.

While the instructions for calibration described in TRS No. 457 were found to be quite clear, some X ray equipment operational issues were not expected. The ideal use of manual operation of X ray tube conditions is not always feasible with modern fluoroscopy equipment. It is sometimes necessary to drive the automatic exposure control (AEC) to the clinically used tube voltage and filtration settings with the use of additional ‘patient equivalent’ filtration positioned close to the image receptor in order to contribute as little scatter as possible to the calibrating detector. In under couch set-ups, the mounting of filter material is more difficult as it can not be positioned on the table requiring the use of a specially designed table or a device to attach the filtration directly to the image receptor. In some units, only fluoroscopy mode can be used, where both dose rate and tube voltage is set by the AEC according to patient attenuation. To achieve and cover the range of clinically used radiation qualities,

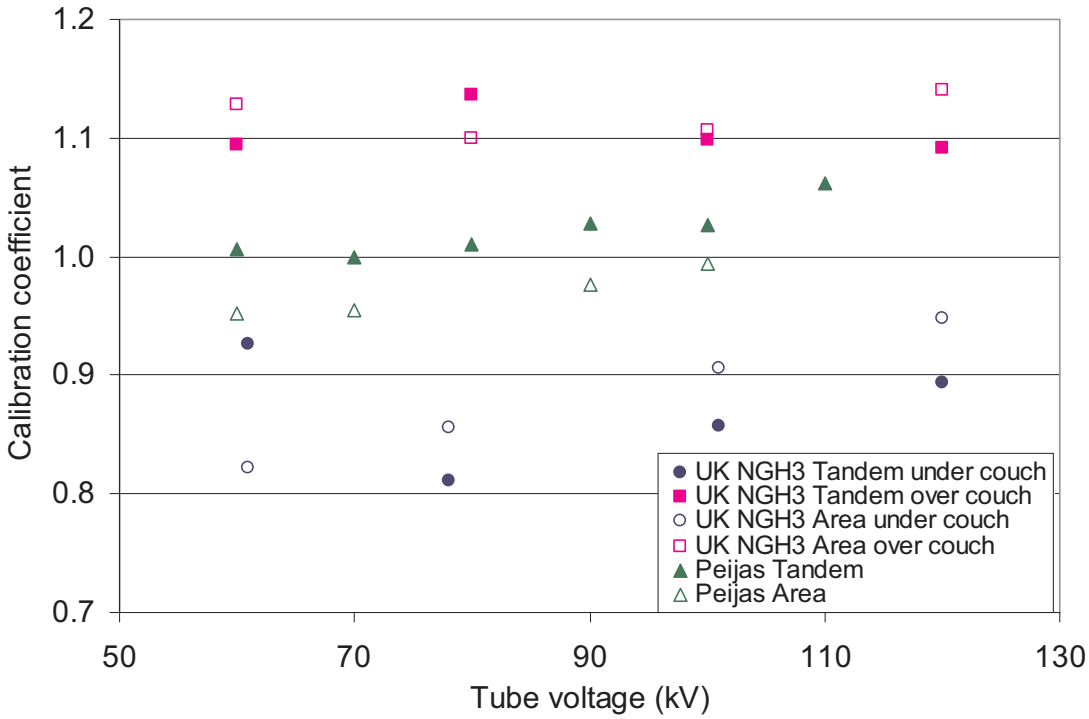

FIG. 40. Calibration coefficients for field KAP meters calibrated using two methods. The open symbols are for the diagnostic dosimeter or area and the closed symbols for the reference KAP or tandem method.

different thicknesses of attenuator are needed. Another difficulty is that fluoroscopy and cine modes may operate in different ways and this may also have an effect on the calibration procedure.

Some results are given in Fig. 40. As a result of the calibration it can be concluded that, even though the radiation qualities were not exactly the same in calibration and in clinical measurements, results for the two methods were typically within 6%.

Generally, it has been noticed that the calibration coefficients using the diagnostic dosimeter method are slightly lower compared to the results using the reference KAP meter method. This effect may be attributed to the extra focal radiation producing a dose outside the intended (collimated) field not accounted by the diagnostic dosimeter method, but measured by the reference KAP meter.

Field size dependence is an important issue in KAP measurements. As most KAP chambers are not checked in a laboratory, minimal checks should be done in the clinical situation. When using the calibration with reference KAP meter there should not be large changes with different field sizes because the whole beam passes through both chambers [23].

5.1.2. Field KAP calibration using diagnostic dosimeter

In this method the field KAP meter is calibrated using a diagnostic dosimeter. This method is usually more useful for hospital staff because a suitably calibrated dosimeter is usually available. However, choosing the right calibration coefficient for the air kerma meter is sometimes difficult due to AEC operation. In some cases the determination of irradiation area and measurement distance can also be difficult and uncertain. New digital systems may require the estimation of field size from softcopy images which may add additional uncertainty to the measurement. Significantly, in this method, errors arising from inhomogeneity in the field will increase the uncertainty of the calibration [32].

5.1.3. Field KAP calibration using reference KAP meter

In order to do this calibration, it is necessary to have a reference KAP meter calibrated for the incident beam, as detailed in TRS No. 457 and discussed in Section 3.2.2.1. The main difficulty with this calibration method is the

radiation quality dependence of the KAP meters. As previously mentioned, at least two radiation quality specifiers (tube voltage, total filtration or HVL) should be used [23]. Sometimes if a KAP meter has a poor measurement resolution, a large field size is needed to reach higher KAP values. In this method, the field size is limited by the size of the reference KAP meter and very high exposures may be needed.

In conclusion, this method was found to be easy to perform but a large range of radiation qualities were needed to calibrate the reference KAP meter. This would be considerably simplified if a KAP meter was available with minimal energy dependence [21]. This method is also useful if a suitable KAP meter can be used for calibrating several systems.

5.1.4. Correction measurement for field KAP meter calibrated at laboratory for transmitted beam conditions

When a KAP chamber is attached close to the tube housing, the scattering properties are totally different to those at the surface of the patient [33, 23]. This was studied in the CRP and it was observed that KAP meters calibrated for transmitted beam, at a certain reference distance, give increased readings for identical beam exposures when moved in close proximity with the tube housing. This was investigated with measurements of KAP meters mounted next to the tube housing, M(0), and 30 cm from the mounting, M(30). Correction factors k(30,0) were calculated as

$$k(30,0) = M(30)/M(0)$$

Typically the correction factors were between 0.90 and 0.95. Based on this result, it can be concluded that additional measurement should be performed at the clinical site to determine appropriate correction factors. In some cases one KAP meter may be used for several X ray units, therefore, correction factors should be derived for each unit.

It should be emphasised that the calibration of field instruments should be done in situ and that the above case of external calibration of field KAP meter at a SSDL should be discouraged.

5.1.5. Uncertainties for clinical scenarios

TRS No. 457 does not give an example of uncertainty estimation for field KAP calibrations. Participants of this CRP project estimated the uncertainties for their calibrations and these varied from 5% to 25%. One example for uncertainty estimation is given in Table 47.

5.1.6. Other issues

The energy dependence of KAP meters probably results from the KAP meter wall materials. One possible solution for the reference KAP meter is to use chambers with improved materials. One such example is the Radcal Patient Dose Calibrator (PDC).

For calibration of an under couch KAP meter, the advice in TRS No. 457 is to put the reference detector on the table top to include its beam attenuation. Although not investigated in the this CRP, the KAP position for this calibration has been further considered with a suggestion that the reference detector should be positioned some distance above the table top up to reduce the contribution of the undefined scattered radiation from the table to the reference chamber. Although the scatter from the table top enters the patient in an under couch set-up, the KAP should be calibrated for primary radiation.

5.1.7. Conclusions

- Field KAP meters should be calibrated (or cross calibrated*) against an appropriately calibrated diagnostic dosimeter or reference KAP meter in situ in the clinical environment.
- If this is not possible, the field KAP can be calibrated at the SSDL. However, a system specific correction factor should be determined and used.

TABLE 47. EXAMPLE OF UNCERTAINTY ESTIMATION FOR CALIBRATING FIELD KAP METER (FROM TOROI ET AL. [23])

Source of uncertainty	Relative uncertainty (%) $k = 2$		
	Tandem method	Beam area method	Laboratory method with unit-specific corrections[a]
Calibration coefficient N_{ref} of the reference meter	3.0[b]	3.0	4.1[b]
Difference between the radiation qualities in the calibration and use of the reference meter	3–5	1	4–6
Reading M_{ref} of the reference meter	1.5	1.5	
Reading M_{field} of the field KAP meter	1.5	1.5	
The ratio of the readings $M(d)/M(d_0)$ of the field KAP meter			2
Deviations from the applied air density corrections (two different chambers)	2	2	
Field area A measured from the film		4	
Distance correction factor $(d_K/d_A)^2$		2	
Effects of field inhomogeneities		4	
Short-term instability of the x-ray beam			1.5
Stray radiation and other uncontrolled factors of the method	1.5	1.5	1.5
Total relative uncertainty	5.4–6.7	7.5	6.4–7.8

— Calibrations should be done with clinically relevant beams that reflect the particular usage of the system (see Appendix V).
— The user requesting a calibration for a KAP meter should specify if the instrument is to be used to measure incident or transmitted radiation.
— The user requesting a calibration for a KAP meter should specify beam qualities that reflect the conditions of usage.

*Term to be applied if a hospital and not an SSDL person is responsible for the calibration.

5.2. CALIBRATION OF TLDS

5.2.1. General

This section provides a comparison of the calibration of a TLD system using clinical beams similar to RQR5 with calibration in an SSDL as described in Section 3.4, and reports on a comparison of calibrations for three clinical beams in four hospitals (the Czech Republic, Austria, Greece and Thailand). The main task of this comparison was to test the accuracy and feasibility of calibration of a TLD system in clinical beams.

5.2.2. Comparison of calibration coefficients from SSDL and hospital

5.2.2.1. Calibration of TLDs in terms of N_k in clinical beams similar to RQR5

TLDs were calibrated using two medical X ray units in clinical beams similar to RQR5. The first X ray unit was a general radiography Siemens Polydoros unit with an HVL of 2.71 mm Al at 70 kV. The second unit was a

dental Prostyle Intra unit with an HVL of 2.31 mm Al at 70 kV. For both units, TLDs were calibrated free in air in three different geometries: horizontal beam, vertical beam without shielding and vertical beam with shielding of backscattered radiation from the table. Horizontal beam geometry corresponds to the calibration geometry in SSDL; the beam was directed towards a window, therefore no scattered radiation was present at the irradiation site. However, this geometry may be difficult to realize in an X ray room in a hospital. In vertical beam geometry, TLDs were placed 30–40 cm above the table, small amounts of scattered radiation from the table and from a floor could reach the TLDs. In vertical beam + shielding geometry, a protective lead apron was laid on the table. It is believed scattered radiation is attenuated in the shielding before reaching the TLDs, therefore the geometry should be equivalent to the geometry in the SSDL.

The TLD system used was the same as described in Section 3.4. TL detectors were packed only in the sachets.

5.2.2.2. Results

A summary of calibration coefficients both from SSDLs and from hospital is shown in Fig. 41. The section 'SSDL' in the figure represents the calibrations at SSDLs (see Chapter 3.4, Fig. 19). The sections 'horizontal beam', 'vertical beam + shielding' and 'vertical beam' in the figure represent calibrations in the hospital. The maximum difference between the values is 16%, which an unexpectedly high value according to the assessed uncertainties.

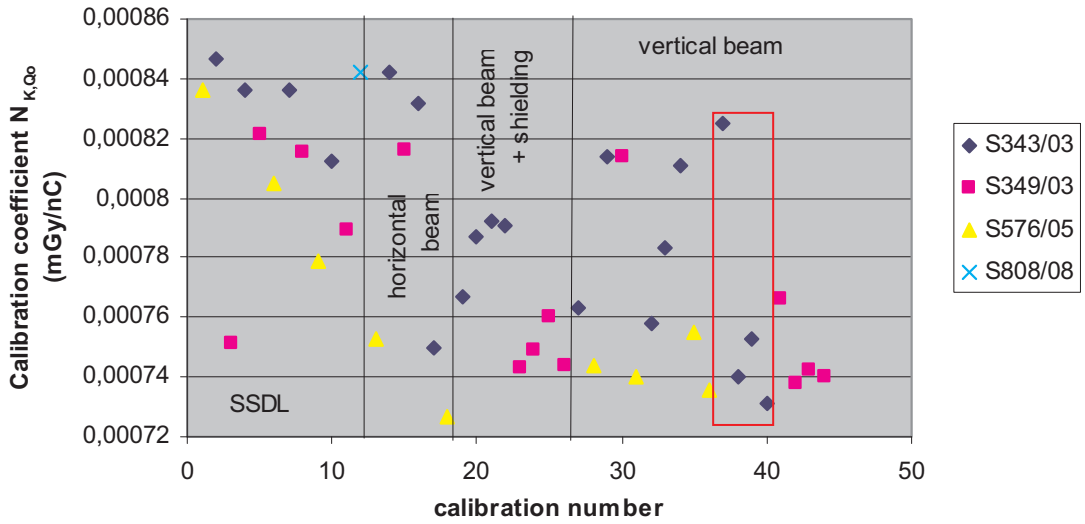

FIG. 41. Summary of calibration coefficients (RQR5 at SSDLs and clinical beams equivalent to RQR5) with respect to geometry and TLD batch (values in the red box are discussed in Chapter 5.2.2.3).

The uncertainty analysis of the calibration coefficient N_{K,Q_o} for calibration in clinical beam is given in Table 48. Each component of the uncertainty given in the table has a different value for the different geometries, the TLD batch used and the number of TL detectors used. To get a conservative estimate, the highest values were considered.

5.2.2.3. Discussion and conclusions

When evaluating the calibration coefficients from calibration of TLDs in the hospital, the same problems with correction of batch sensitivity occurred, that have already been commented on in Section 3.4 (calibrations at SSDL). The problem with reproducibility of readout values (see Section 3.4) is relevant here as well. It can be demonstrated by the values in the red box in Fig. 41. The maximum difference between the highlighted values is 13%. The expanded uncertainty of the calibration coefficient given in Table 48 is not relevant in this case, since the majority of components of the uncertainty can be eliminated. These dosimeters were annealed simultaneously, irradiated and read out under the same conditions, one after the other. Only the standard deviation of the mean response of the TL dosimeter and X ray unit stability can cause the difference. Expanded uncertainty (k=2) caused by these two effects should be, in this particular case, only 2.4%.

TABLE 48. UNCERTAINTY ANALYSIS OF THE CALIBRATION COEFFICIENT FOR CALIBRATION IN HOSPITAL

Source of uncertainty	Values given in %	Source of the value of the uncertainty component
Kerma value given by hospital	4.05	Value determined by hospital
Field homogeneity	0.80	Based on measured field homogeneity of the X ray machine
Distance	1.40	Based on uncertainty of focus to chamber/TLD distance measurement
X ray unit stability	0.31	Based on statistical analysis of the tube output variation
Mean TL response variation	1.48	Based on statistical analysis of the TL response variation
Reader and batch sensitivity	1.48	Based on statistical analysis of the TL response variation
combined uncertainty	4.8	
expanded uncertainty (k=2)	9.7	

The largest variations among the values of $N_{K,Qo}$ are for the horizontal geometry. This was expected as it is difficult to set up accurately for this geometry in the hospital X ray room. At the hospital, the vertical geometry with shielding of backscattered radiation seems to be the most accurate and relevant to SSDL geometry.

The comparison has shown that there are not significant differences in values of calibration coefficients received from calibration at SSDLs and at hospitals. However, the possibility of some mistakes during calibration in the hospital is larger than during calibration at the SSDL. The uncertainty of the calibration coefficient received from calibration in the hospital is greater, mainly due to the larger uncertainty of incident air kerma determination and the less accurate set up of the dosimeters in the beam. The comparison was only made for RQR5 quality and clinical qualities similar to RQR5.

5.2.3. TLD audit of clinical dosimetric equipment

5.2.3.1. Calibration in clinical beams

To check the reliability of the calibration of TLDs in clinical beams and the reliability of postal use of TLDs as well, a comparison of TLD calibrations made in hospitals in the Czech Republic, Greece, Austria and Thailand was performed. The participants irradiated just the TLDs and specified the value of incident air kerma. The Czech Republic evaluated the results.

Radiation qualities used for the audit were as follows:

Beam 1: 120 kV no added filtration: (as typical for adult thorax examination).
Beam 2: 70 kV with 1 mm Al and 0.2 mm Cu added filtration: (as typical for paediatric examination.
Beam 3: 70 kV no added filtration (as typical for fluoroscopy).

The TLD system used was the same as described in Section 3.4. TL detectors were packed only in sachets.

5.2.3.2. X ray apparatus and dosimetric system of the hospitals

The following X ray systems were used in the participating hospitals for the TLD calibrations.

TABLE 49. CLINICAL EQUIPMENT

	Czech Republic	Greece	Austria	Thailand
X ray unit	Siemens Polydoros	Philips, Medio 50 CP	Philips, Optimus 80	Trex Medical System HFQ
Filtration	2,95 mm Al	2,5 mm Al	2,5 mm Al + 0,35 mm collimator	0,7 mm Al + 2 mm Al
Reference dosimeter — chamber	90X5-6	Inovision 96035 B	PTW, SF chamber TM 34060-2	Exradin A5
Reference dosimeter — electrometer	Radcal 9010	Inovision 35050 A	PTW, Unidos M	Wellhofer Dose 1

5.2.4. Results

The results of the audit are shown in Tables 50 and 51. To assess incident air kerma from TLD measurement, calibration coefficient $N_{K,Qo}$ and correction coefficients $k_{Q,Qo}$ were both taken from calibrations in RQR beams and clinical beams (vertical geometry + shielding). For each clinical beam quality of the participant, the appropriate value of the correction factor $k_{Q,Qo}$ was used according to the HVL specified by the participating hospital. The results of the audit are expressed as a ratio of incident air kerma assessed from TLD measurement and incident air kerma specified by participating hospital.

TABLE 50. COMPARISON OF INCIDENT AIR KERMA DETERMINED BY TLDs ($K_{i,TLD}$) AND INCIDENT AIR KERMA GIVEN BY THE PARTICIPATING HOSPITALS ($K_{i,hosp.}$), USING TLD CALIBRATION COEFFICIENTS FROM CALIBRATION IN RQR BEAMS

	HVL in hospital beams (mm Al)			$K_{i,TLD}/K_{i,hosp.}$		
Beam	1	2	3	1	2	3
Austria	5.02	5.16	2.78	1.23	1.24	1.22
Greece	4.71	5.24	2.83	1.22	1.26	1.20
Thailand	5.88	5.75	3.21	1.55	1.31	1.35
Czech Republic	4.81	4.94	2.71	1.08	1.11	1.00

TABLE 51. COMPARISON OF INCIDENT AIR KERMA DETERMINED BY TLDS AND INCIDENT AIR KERMA GIVEN BY THE PARTICIPATING HOSPITALS, USING TLD CALIBRATION COEFFICIENTS FROM CALIBRATION IN CLINICAL BEAM SIMILAR TO RQR5

	TLD/K_i		
Beam	1	2	3
Austria	1.10	1.11	1.09
Greece	1.10	1.13	1.07
Thailand	1.39	1.17	1.21

The uncertainty analysis of incident air kerma determination by means of TLD is given for calibration in an SSDL (Table 52) and in clinical beams (Table 53)

TABLE 52. UNCERTAINTY ANALYSIS OF THE INCIDENT AIR KERMA DETERMINATION BY MEANS OF TLDS USING CALIBRATION COEFFICIENT FROM CALIBRATION IN SSDL

Component of uncertainty	Values given in %	Source of the value of the uncertainty component
Calibration coefficient	2.79	See Table 24
Field homogeneity	0.80	Generally not known — depends on the accuracy of TLDs placement in the centre of the field and field homogeneity of individual units, value taken from Table 24
Distance	1.40	Generally not known — depends on accuracy of TLDs placement in correct distance and accuracy of measurement of the distance, value taken from Table 24
X ray unit stability	0.31	Generally not known — different for each unit, value taken from Table 24
var. coeff. of mean TL response	1.91	Based on statistical analysis of the response variation
Reader and batch sensitivity	1.48	Based on statistical analysis of the response variation
Uncertainty of K_{Q,Q_o}	2.00	Based on differences in k_Q,Q_o factors, see Figs 22 and 23
Combined uncertainty	4.5	
Expanded uncertainty (k=2)	9.0	

TABLE 53. UNCERTAINTY ANALYSIS OF THE INCIDENT AIR KERMA DETERMINATION BY MEANS OF TLDS USING CALIBRATION COEFFICIENT FROM CALIBRATION IN CLINICAL BEAMS

Component of uncertainty	Values given in %	Source of the value of the uncertainty component
calibration coefficient	4.84	See Table 48
Field homogeneity	0.80	Generally not known — depends on accuracy of TLDs placement in the centre of the field and field homogeneity of individual units, value taken from Table 48
Distance	1.40	Generally not known — depends on accuracy of TLDs placement in correct distance and accuracy of measurement of the distance, value taken from Table 48
X ray unit stability	0.31	Generally not known — different for each unit, value taken from Table 48
var.coeff. of mean TL response	1.91	Based on statistical analysis of the response variation
reader and batch sensitivity	1.48	Based on statistical analysis of the response variation
uncertainty of K_{Q,Q_o}	2.00	Based on differences in k_{Q,Q_o} factors, see Figures 22 and 23
Combined uncertainty	6.0	
Expanded uncertainty (k=2)	12.0	

5.2.4.1. Discussion and conclusions

The results of the audit are very inconsistent. Austria and Greece are in good compliance with each other. However, the values of incident air kerma determined by TLD differ by approximately 23% from the values given by Austria and Greece (if TLD calibration in RQR beams is used). As can be seen in Fig. 41, such a discrepancy cannot be explained by different conditions during irradiation in either the SSDL or the hospital. Or by differences in treatment of the TLD. The TLDs irradiated in the hospitals could detect some backscattered radiation, but not to the extent indicated. Results from comparison in Thailand are even more puzzling, particularly for beam 1, but the trend is similar as that for Austria and Greece. Better compliance was achieved when using calibration coefficients for TLDs obtained from clinical beam calibrations. For this purpose calibration coefficients from irradiation in vertical geometry + shielding of backscattered radiation was used.

In the case of clinical dose measurements using TLDs calibrated in clinical beams (the worst case according to its uncertainty), the expanded uncertainty of the incident air kerma or entrance surface air kerma will be approximately 12%. This is a larger uncertainty than the example in TRS No. 457 where 10% uncertainty for TLD measurement (Section 8.3.4.) and 12% uncertainty for direct measurement of entrance surface air kerma using TLDs (Section 8.4.3.4) is indicated. However, the achieved accuracy of K_i assessment by means of TLDs is in compliance with the requirement of European Commission Recommendations for patient dosimetry in diagnostic radiology using TLD (EUR 19604 EN). In this recommendation, expanded uncertainty (k=2) for TLD measurements on patients should not exceed 25%.

However, the results of the audit indicate the existence of some unsolved problems within the TLD system; working procedures or mistakes during irradiation. According to the results of the TLD audit in clinical beams and TLD audit in SSDL (see Section 3.4), the calibration in SSDL should be preferred.

5.2.5. Conclusions

— Calibration of TLDs in an SSDL is preferable as it is difficult to achieve a high level of standardization during calibration in clinical beams,which is clear based on the results of the audit of clinical beams.
— Care should be taken to avoid backscatter in the calibration of TLDs as this can cause difficulty in some environments.
— During the calibration in clinical beams, incident air kerma should be checked by a reference dosimeter at the beginning, during and after the TLD irradiation. Moreover, some monitor chamber (could be the reference dosimeter) placed directly on the table under the TLDs should be used for each irradiation.

6. CONCLUSIONS AND RECOMMENDATIONS

6.1. CONCLUSIONS

Radiation qualities used to calibrate instrumentation for diagnostic radiology dosimetry at calibration facilities should reflect the clinical situations as closely as possible in order to give accurate calibration. Two cases have been identified where the divergence between clinically used beam and calibration beam qualities, as defined by IEC and advocated in TRS No. 457, induce remarkable uncertainty; namely dosimetry of mammography and KAP meters. It should be noted however that even in 2008, new beam qualities were available for mammography from PTB that have closed this gap for mammography.

For the clinical implementation of TRS No. 457, one conclusion is that the code of practice needs to be more extensively implemented in the clinical environment. This would require the involvement of clinical staff by the participants. There is also a need to upgrade the dosimetry for CT in the light of the increasing X ray beam widths used in emerging multi-detector CT (MDCT) scanners and cone beam scanners. The specific nature of CT dosimetry also demands that dosimetry record sheets be customised to specific CT machine types.

In addition, a major outcome of the current CRP was the realization that there is great benefit in SSDL personnel understanding more of the clinical applications of the dosimeters they calibrate and similarly that clinical physicists also benefit from a more rigorous understanding of instrument calibration. The following statements then summarize this finding.

— SSDL personnel should be aware of the clinical needs associated with the instruments that are calibrated.
— Medical physicists working in the clinical environment should be aware of the importance of instrument calibration including a substantial knowledge of metrological practice.
— All personnel should understand the operation of the instruments.

6.2. RECOMMENDATIONS

6.2.1. General recommendations relating to TRS No. 457 implementation

(1) The calibration coefficient of the dosimeter is strictly valid only for the reference conditions specified in the certificate. As the measurement conditions in the hospital usually do not match the reference conditions, the user should investigate the effect of influence quantities on the measurement and apply corrections if necessary. Examples of influence quantities are the beam quality (changes with tube voltage and filtration), air kerma rate (various for radiography and fluoroscopy), ambient pressure and temperature etc. Their effect on the dosimeter response can be experimentally determined and it can usually be assessed from the information provided by the manufacturer. IEC 61674 sets the requirements on the dosimeter performance under different conditions.

(2) The interpolation[14] of the calibration coefficient of the reference instrument at an SSDL, relative to radiation beam quality, may be unavoidable in some cases. The uncertainties of these interpolations shall be included in the overall uncertainties of the measurements.

(3) TRS No. 457 states that the lead aperture thickness for calibration should be 2 mm. This thickness is sufficient for RQR beams but it may not be enough for beams with a higher filtration like the RQT beams. Apertures of 2.5–3.0 mm thick are recommended for such cases. The criterion of 0.1% transmission will then be met.

(4) It is stated in TRS No. 457 that RQR beam qualities should be verified by measuring the air kerma or air kerma rate with and without an aluminium attenuation layer of the thickness given in column 3 of Table 6.2. This ratio (K_{HVL}/K_0), should be in the range 0.485–0.515. This is equivalent to the ratio of measured to nominal HVL values that lie within 0.957–1.044.

(5) SSDLs can calibrate dosimeters in a limited number of beam qualities. They should provide guidance to end users on correction factors when dosimeters are used in qualities that deviate from those used for the calibration. An example is interpolation between two tube voltages or two filtrations.

(6) Difficulties in the availability of the calibration of high voltage dividers at PSDLs were observed. It is suggested that accredited commercial companies be more involved in the calibrations of dividers.

(7) Developments in fluoroscopy, especially in paediatric interventional techniques, require the use of copper filters and thus qualities that are not covered by RQR beams used for calibrations of KAP meters. SSDLs have limited possibilities of using other suitable beams with the higher filtration. One possible option is to expand radiation qualities available at SSDLs by using filtrations of RQR and RQT qualities with adjusted high voltages. Based on the work of Toroi et al. [19], it is probably easiest to do interpolations of calibration coefficients using tube voltage and fixed filtration. See Appendix V for some examples using aluminium filtrations close to 3 mm Al and 5 mm Al and Al with added copper filtration.

(8) The SSDLs should provide information to end users on beam qualities used for KAP calibrations. The user should request calibrations in beam qualities that best match the beams used in the hospital and clearly state it in the calibration request.

[14] Extrapolation of calibration coefficients for values outside of the range of measured data is not considered valid.

(9) SSDLs should reference IAEA-TECDOC-1585 for estimation of calibration uncertainties. The uncertainty budget should reflect the practice in the laboratory. SSDLs should be aware that having lower uncertainty does not guarantee a better calibration.

(10) Calibrations of P_{KL} and KAP meters are done using small fields. The SSDL should investigate the accuracy of the determination of the K_{air} at the point of measurement for the beam qualities and dosimeters used.

(11) Because field size is an important factor influencing KAP measurements, SSDLs should perform measurements using different field sizes. The calibration certificate shall clearly state the field sizes used by the SSDL. The effect of field size on KAP calibration should be included in the uncertainty budget.

(12) Present mammography equipment uses more and more varied target/filter combinations (Mo/Rh, Rh/Rh, W/Rh and W/Al [34] and W/Ag [35]). Various studies show that changes in X ray spectra will influence mainly the response of the semiconductor detectors. Thus, the calibration of mammography dosimeters in Mo/Mo beams may not be sufficient for some detectors in some clinical beams. It is suggested that PSDLs put effort into development of the new mammography standards. At the same time, the laboratories that possess such beams should study the effect of spectral changes on the dosimeter performance and give guidance to the users on selection of the dosimeter and correction factors.

(13) Conditions during clinical mammography measurements of the incident air kerma, K_i, differ from those of calibration since the detector is 45 mm above a cassette table and a compression paddle is present in the beam. Therefore the spectrum (beam quality) may differ from the one during the calibration. There may also be scattered radiation incident on the detector. The user should investigate the influence of these effects on the results of measurements and apply appropriate corrections if needed.

(14) SSDLs should primarily develop procedures to calibrate kV meters in terms of PPV as guided. Due to the fact that existing kV meters display in kVp, kV_{mean} and kV_{max}, SSDLs should also provide calibration in terms of these variables as requested.

(15) SSDLs should have adequate ways to estimate the high voltage applied to the X ray tube during calibration. It is highly recommended to measure and monitor the high voltage applied to the X ray tube invasively. In line with IEC recommendations [36], X ray kV descriptors such as maximum peak voltage and mean peak voltage should be acceptable as well as PPV. If necessary, non invasive measure of X ray tube voltage using descriptors acceptable to IEC can be used, with care.

(16) Comparisons of ionization chambers should be organized following ISO Guide 43. SSDLs should follow their calibration procedures and the pilot laboratory should carry out stability measurements of the transfer chambers using a suitable source (e.g. Cs-137).

(17) In principle, TLDs can be used for auditing SSDLs provided a suitable TLD system is selected and an achievable uncertainty is assessed.The results of the current project indicate that successful implementation is non trivial and becomes more difficult when extended to the clinical environment.

(18) A team effort is required for successful collection of valid dosimetric patient data consisting of capable radiographers under the supervision of a medical physicist with a specialty in diagnostic radiology.

(19) As described in TRS No. 457, determination of patient dose has significant clinical impact when the results are compared to appropriate dose reference levels. Together with relevant image quality data, this will inform the process of optimization.

(20) SSDLs should make efforts to clarify for the users the role of k_Q values in demonstrating detector energy dependence (e.g. appendix to certificate, training events etc).

(21) It is of great importance that users are aware of the consequences of the dosimetric method on the uncertainty budget when making dosimetric measurements, particularly when compiling DRL data.

(22) TRS No. 457 can be applied to paediatric dosimetry. However, particular care will be needed in using appropriate phantoms and assessing the uncertainly budget.

(23) Mammography dosimetry implementation will benefit from the use of software (such as Excel). Programs are available in the current IAEA mammography QC publications.

(24) In mammography, the HVL value should be measured as part of the dosimetry method. If tabulated values are used, the effect on the additional uncertainty should be determined.

(25) Modalities delivering potentially high risk procedures require priority in dose determination and monitoring.

(26) Field KAP meters should be calibrated (or cross-calibrated) against a reference KAP meter or air kerma dosimeter in situ in the clinical environment. If this is not possible the field KAP can be calibrated at the SSDL, however a system specific correction factor should be determined and used.

(27) Cross-calibrations should be done with clinically relevant beams that reflect the particular usage of the system. This is of particular importance for meters with strong energy dependence such as KAP meters and some solid state devices.
(28) The user requesting a calibration for a KAP meter should specify if the instrument is to be used to measure incident (reference KAP meter) or transmitted radiation (field KAP meter), noting that the latter would be a rare event.
(29) The user requesting a calibration for a KAP meter should clearly specify the beam qualities that reflect the conditions of usage.
(30) Care has to be taken when applying phantom dosimetry data to patient dose situations.

6.2.2. Recommendations for future review of areas in TRS No. 457

(1) Instruction on the establishment of beam qualities should give more attention to the effect of focal spot size and the proximity of apertures to the beam shape.
(2) Guidance should be given on the methods of tube alignment and the necessary measurement of beam kerma profiles.
(3) Table 6.4 in Chapter 3 in TRS No. 457 should have the tissue weighting factors for effective dose updated in line with ICRP 103.
(4) To improve the confidence level of patient dose estimation, it is suggested to increase the number of patients involved in the dose measurements.
(5) CT scanner type specific data sheets should be used to assist in data collection as described in Appendix VII.
(6) Table 8.6. in Chapter 8 of TRS No. 457 should have updated s factors for new mammographic target/filter combinations. Recently published data is available by Dance et al.[37].

6.2.3. Recommendations for future action

(1) Where possible, worksheets for clinical dosimetry should be put into Excel format using macro functions.
(2) Work should continue on the evaluation of (i) the use of PPV by SSDLs and (ii) the benefits of the use of PPV in the clinical environment.
(3) The implementation of calibration procedures for CT chambers and KAP meters should be carefully reviewed and related to the clinical tasks performed by these detectors.
(4) Future work in paediatric dosimetry, including the use of relevant phantoms and consequent uncertainty budgets, is required.
(5) Further work is needed to extend the CT procedures to accommodate new developments in CT technology.
(6) Manufacturers should be encouraged to comply with the IEC 60601-2-43 [38] requirement to include KAP meters (or calculation) in equipment used for fluoroscopy; especially high dose rate equipment.
(7) Attention should be given in determining organ doses and associated uncertainties. Some examples include skin, eye, and foetus.

Appendix I

RESEARCH ACTIVITIES

TABLE 54. PARTICIPANTS IN ACTIVITIES 1–3

Activities 1–3	RQR	RQR-M	RQT	Uncertainty budgets	Comparison of calibrations
Brazil	X			X	X
Cuba	X		X	X	X
Czech Republic	X		X	X	X
Finland	X		X	X	X
Greece	X	X	X	X	X
IAEA	X	X		X	X
Thailand	X			X	X
Vietnam	X		X	X	X

TABLE 55. PARTICIPANTS IN ACTIVITY 4

Activity 4		GR	Fluoro	Mammo	CT	Dental
Austria	Phantom	X			X	X
	Patient	X			X	X
Brazil	Phantom	X				
	Patient					
Cuba	Phantom					
	Patient	X		X		
Czech Republic	Phantom	X	X	X	X	
	Patient	X	X	X	X	X
Finland	Phantom		X	X	X	
	Patient		X	X	X	
Greece	Phantom	X	X	X	X	
	Patient	X	X	X	X	X
Hungary	Phantom	X	X	X	X	
	Patient	X	X	X	X	X
Korea, Republic of	Phantom	X	X	X	X	
	Patient	X	X	X	X	
Thailand	Phantom	X				
	Patient	X				

TABLE 55. PARTICIPANTS IN ACTIVITY 4 (cont.)

Activity 4		GR	Fluoro	Mammo	CT	Dental
UK	Phantom		X		X	
	Patient		X		X	
Vietnam	Phantom					
	Patient					

TABLE 56. PARTICIPANTS IN ACTIVITIES 5–7

Tasks 5a&b, 6a, b & c & 7	Calib SSDL KAPs	Calib clinical KAPs	TLD calib. SSDL check	TLD calib. Clinical check	TLD for patient dosimetry	Practical peak voltage
Austria		X		X	X	X
Brazil			X*		X	X
Cuba	X		X*			
Czech Republic	X	X	X	X	X	
Finland	X	X	X*	X	X	X
Greece	X	X	X	X	X	X
Hungary		X		X	X	
IAEA			X*			
Korea, Republic of		X				
Thailand	X	X	X	X	X	X
UK		X				
Vietnam	X	X	X	X	X	X

* second run.

TABLE 57. TEMPLATE TIME FRAME

Activity	2005		2006			2007				2008			
	IV	I	II	III	IV	I	II	III	IV	I	II	III	IV
Activity 1 Setting-up of calibration beam qualities at SSDLs	X	X	X	X	X	X	X	X	X	X	X		
Activity 2 Development of calibration procedures including the uncertainty budget	X	X	X	X	X	X	X	X	X	X	X		
Activity 3 Comparison of calibrations				X	X	X	X	X	X	X	X	X	
Activity 4 Evaluation of measurement procedures in hospitals													
(a–i) report on phantoms and dosimeters available and capabilities to manufacture phantoms.		X				X	X	X					
(a–ii) implementation of measurement procedures in hospital incl. measurements with patients			X	X	X	X	X	X	X				
— report measurements in hospitals					X	X	X	X	X	X	X		
Activity 5 Calibration of KAP meters													
— at SSDLs						X	X	X	X	X	X	X	
— at hospitals							X	X	X	X	X	X	
Activity 6 TLD dosimetry audit for diagnostic measurements						X	X	X	X	X	X	X	
Activity 7 Practical peak voltage						X	X	X	X	X	X	X	
Activity 8 RCMs	X						X						X
Activity 9 Preparation of the publication					X		X			X	X	X	X

Appendix II

PROJECT ACTION PLAN

The nominated activities were:

Activity 1. Setting-up calibration beam qualities at SSDLs

SSDLs to set-up calibration beam qualities according to TRS No. 457.

Activity 2. Development of calibration procedures including the uncertainty budget

SSDLs to develop their laboratory procedures and establish the uncertainty budget for the calibration of user dosimeters as described in Appendix 2 of TRS No. 457.

Activity 3. Comparison of calibrations

Calibration comparison to be organized for participating SSDLs. This will include the calibration of a selected instrument in selected beam qualities. The comparisons are indicated in Table 54.

In addition to the common assignments, the following tasks are to be performed by the coordinator:

— Follow-up on the intercomparison run (participants confirm to the coordinator that they received the chamber);
— Performing the reference calibrations of instruments;
— Collection of data and first evaluation of results.

Activity 4. Evaluation of measurement procedures in hospitals

A number of tasks were set in this activity*:*

(1) Research the feasibility of adopting and implementing the procedures described in TRS No. 457 into clinics.
(2) Report on availability of phantoms and dosimetric instrumentation.
(3) Research the possibility of fabrication of phantoms necessary for clinical measurement. Provide phantoms to clinics. Report on the possibility of providing phantoms to local hospitals outside the project in order to enable dosimetric measurements.
(4) Create the uncertainty budget for each type of dosimetric estimation, including the dose estimation from patient data.
(5) Inter-comparison of measurements obtained with different methods as detailed further. Identify possible problems and pitfalls in setup and measurement procedures.

TABLE 58. SCHEDULE FOR COMPARISONS

Sep/Oct07	Nov07	Dec07	Feb08	Mar/Apr08
GRE initial calibration.	FIN	IAEA-VIE	IAEA-BRA	GRE (re-calibr.)
Apr08	Jun08	Jul08	Sep08	Nov08
CZ	IAEA Calibration	IAEA-THA	IAEA-CUB	IAEA-GRE Final analyse.

— General radiography *(Austria to coordinate)*
Patient dose methods:
A. Calc from K_e from K_i + pat. parameters
B. K_e from K_i from phantom
C. K_e from $K_{S,TLD}$ on patient(**activity 6c**)
D. K_e from TLD on phantom; (**activity 6c**)
E. P_{KA} for patient, (if possible)
F. P_{KA} for phantom, (if possible). NOT NEEDED
• Coordinator to compare methods A–D and E–F.
General comments of tasks and phantom and instrument survey information
Uncertainty tables for A–F.
— Fluoroscopy *(UK to coordinate)*
Patient dose methods: –
G. (rate) K_e from phantom
H. P_{KA} for patient + patient data
• Coordinator to compare methods or data trends G–H
General comments of tasks and phantom and instrument survey information
Uncertainty tables for G–H.
— Mammography *(Hungary to coordinate)*
Patient dose methods: –
I. calc MGD from K_i + pat. Parameters
J. K_i from phantom
• Coordinator to compare methods I-J
General comments of tasks and phantom and instrument survey information
Uncertainty tables for I-J.
— CT *(UK to coordinate)*
Patient dose methods: –
K. $C_{a,100}$ + C_w head & body,
L. calc P_{KL} from pat. Parameters; compare to console value P_{KL}
• Coordinator to collate results K–L.
General comments of tasks and phantom and instrument survey information
Uncertainty tables for K–L.
— Dental *(Austria to coordinate)* COMPLETED
Dose methods: –
M. calc from K_i for bit wing
N. P_{KL} for OPG unit
O. Use KAP meter for P_{KA} for OPG unit and for bit wing.
• Coordinator to collate and compare results M–O
General comments of tasks and instrument survey information
Uncertainty tables for M–O.

Activity 5. Calibration of KAP meters

5a) Participants will calibrate their KAP meter for various diagnostic radiation qualities at SSDLs and provide a calibration certificate of a typical calibration and the uncertainty budget. Participants will report the feasibility of TRS No. 457 and on their experience with the long term stability of KAP meters (repeated calibrations).

5b) Participants will calibrate the fixed KAPs (or displays of KAPs) of clinical X ray equipment or portable KAPs on-site (at clinics). Both calibration methods described in TRS No. 457 should be used (1. air kerma × area and 2. use of a reference KAP meter that has been calibrated at SSDLs). Use over and under couch set-ups.

Participants will at least report the change of KAP value after calibration (5a) and the results of comparison of the two calibration methods used (5b). Form for reporting detailed information is to be provided by the activity coordinator.

Activity 6. TLD dosimetry audit for diagnostic measurements

6a) Testing of methodology for TLDs calibration

Participants will calibrate their TLDs at RQR beams (at SSDL) and at the clinical beams (at hospital). Participants will irradiate and evaluate their TLDs, describe calibration procedures (including photos), compare calibration factors from RQR and clinical beams, and assess the uncertainty budget. Participants will send their results to the Czech Republic. The Czech Republic will analyse them. Optimal detailed procedure for the calibration of TLDs for purposes of clinical patient dose measurements will be recommended.

This activity is time independent of activities 6b–6d. Each participant in this activity studies their own TLD system, while in activities 6b–6d the Czech TLD system is used.

6b) TLD audit of SSDL dosimetric equipment

The TLD audit is organized by the Czech Republic. The Czech Republic will provide instruction sheets, datasheets and TLDs. Participants will irradiate TLDs according to the instruction sheet and send them to the Czech Republic. The TLDs will be evaluated by the Czech Republic. Conclusions will be drawn from the exercise.

The TLD audit includes only SSDLs and is integrated with activity 3 to allow TLD and ionisation chamber irradiation at the same time if possible (see timeline — Table 57). For practical reasons however it will be convenient to start with this activity in advance (not strictly waiting till activity 3 has taken place) because all further TLD activities (6b, 6c) must follow activity 6a. Therefore Thailand and Vietnam will be asked to participate sooner.

SSDLs will be asked to irradiate the TLDs to the specified range of air kerma. K_{air} using radiation qualities RQR3, RQR 5, and RQR9.

Data and instruction sheets (practical guidance, timetable, the value of the delivered incident air kerma, method of dose determination) are already available.

6c) TLD audit of clinical dosimetric equipment

The TLD audit is organized by the Czech Republic. The Czech Republic will provide instruction sheets, data sheets and the TLDs. Participants will irradiate TLDs according to the instruction sheet and send them to the Czech Republic. The TLDs will then be evaluated by the Czech Republic. Conclusions will be drawn from the exercise.

One hospital per country is going to participate in the audit. This activity will proceed once activity 6b is successfully achieved.

The participants in hospitals will be asked to irradiate the TLDs to the specified range of incident air kerma using clinical radiation qualities described below.

Radiation qualities clinically described as:

(1) 1. 120 kV no added filtration: (as typical for adult thorax examination);
(a) 2. 70 kV with 1 mm Al and 0.2 mm Cu added filtration: (as typical for paediatric examination);
(b) 3. 70 kV no added filtration (as typical for fluoroscopy).

6d) TLD measurement on PMMA phantoms and patients

The TLD audit is organized by the Czech Republic. The Czech Republic will provide instruction sheets, datasheets and the TLDs. Participants will irradiate TLDs according to the instruction sheet and send them to the Czech Republic. The TLDs will then be evaluated by the Czech Republic. Conclusions will be drawn from the exercise.

This activity will not progress until activities 6b and 6c are completed successfully. It should be done in conjunction with the general radiography section in activity 4 to enable the comparison of results from TLDs, ionization chambers and patient exposure data collection.

The participants will irradiate TLDs on PMMA slab phantom under clinical conditions corresponding to patient irradiation. The participants will provide the detailed description of the irradiation conditions and will

estimate entrance surface air kerma delivered to the TLD. The K_e estimated by the participants will be compared with K_e estimated from the TLD measurement.

If consistent results are obtained from phantom measurements, the TLD irradiation will be repeated on a group of patients undergoing some frequent clinical examinations. Clinical X ray examinations suitable for the patient measurements will be specified in data sheets.

6e) Summary of methodology for TLDs calibration

Analysis of the results of activities 6b, 6c and possibly 6d, along with the uncertainty budgets that have been calculated, will allow a summary of the use of TLDs for radiological dosimetry. From this, an optimal detailed procedure for the TLD use at a clinical practice could be recommended.

Activity 7 Practical peak voltage

Measurement procedures of tube potential in diagnostic X ray equipment will be tested with as a wide range of generator technologies (voltage ripple) when possible. kVp, kVp average and practical peak voltage values will be compared. Estimation of variation in kVp readings of different kVp measuring devices will be made with SSDL and clinical systems. The uncertainty budget for measuring tube potential will also be estimated. Testing will be carried out of the procedures described in TRS No. 457: Verification of the concept of practical peak voltage and its applicability in kVp determination.

Appendix III

SURVEY ON CALIBRATION CAPABILITY FOR DIAGNOSTIC RADIOLOGY RADIATION DETECTORS CONDUCTED IN 2008

Responses came from 38 SSDLs, representing 37 different countries.

Question 1: Do you currently have the facility to calibrate diagnostic radiology X ray dosimeters?

19 Yes; however of these ,4 were excluded for the following reasons:

1 site — One was from a commercial company with SSDL status. Their work was not typical of most country SSDLs and in fact they recorded 4,881 detectors calibrated in 2007 using IEC beams and TRS No. 457.

2 sites — They were actually doing protection detector calibrations with ISO 4037 (one other centre also mentioned ISO 4037 but (correctly) did not check yes to question 1.

1 site — It was clear from the comments that they were not carrying out calibration or had the facility for diagnostic X ray dosimeters.

Further analysis of the **15 Yes** responses to Q1 showed that 3 did not perform calibrations in 2007 due to technical problems. In addition, 2 other centres checking **No** to Q1 did perform calibrations in 2007. Of the **15 Yes** responses, 11 stated they followed TRS No. 457, with 1 following a national protocol and 2 others using other protocols. All but one centre used some IEC beams (this other centre is being followed up — in case they are using ISO 4037), with 2 centres using additional non IEC mammography beam qualities. The breakdown of the IEC beam usage was:

13 RQR, 5 RQA, 4 RQR-M, 3RQA-M, 3 RQT.

For question 1 there were **19 No** responses, with **13** of these indicating they plan to have a facility within 3 years.

Question 4: For a typical detector calibration, how many calibration points are performed? The mean of all who gave a non zero answer was 3.6, with a 1–7 point range.

Question 5: How many diagnostic detectors did you calibrate in 2007? The mean of all who gave a non zero answer was 23.9, with a range of 5–60 detectors. The total number of detector calibrated was 335.

Summary: Analysis of the returned survey forms showed:

— Replies were received from 38 SSDLs representing 37 different countries.
— Currently. 15 SSDL sites have the facility to make diagnostic X ray, with a further 13 indicating they plan to have a facility in 3 years time.
— Of the 15 sites above, 11 follow TRS No. 457 while 13 use IEC beam qualities.
— There is a large range in the activity of diagnostic radiology calibrations with 5–60 detectors calibrated in a year, with a total of 335 detectors for 2007. The one commercial facility registered as an SSDL on the other hand calibrated 4,881 detectors in the same period.
— At some facilities there is some confusion about what is meant by diagnostic radiology dosimetry calibration. This should not include activities of calibration for protection purposes using ISO 4037 beam qualities. Instead the publication TRS No. 457 and appropriate IEC beam qualities should be used.

ACKNOWLEDGEMENTS

We would like to thank the countries that participated in the survey. Argentina; Bangladesh; Brazil; Bulgaria; Canada; China; Cuba; Cyprus; Czech Republic; Finland; Georgia; Germany; Greece; Guatemala; Islamic Republic of Iran; Ireland; Israel; Latvia; Macedonia; Madagascar; Norway; Pakistan; Peru; Philippines; Poland.

Appendix IV

INTERPOLATION OF USER CALIBRATION COEFFICIENTS FROM DATA OF CALIBRATION CERTIFICATE.

Based on the guidance in TRS No. 457, the calibration laboratories have two options to present the calibration coefficients for a dosimeter (word *dosimeter* is used here in a general sense, covering all kinds of air kerma meters used in diagnostic radiology). The first option is to present the calibration coefficient for the reference beam quality in combination with the k_Q values for all the radiation qualities used in calibration. k_Q describes the response of the dosimeter relative to beam spectral characteristics; i.e. relative to beam quality specifier. The second option is to present the calibration coefficients for all the radiation qualities used in the calibration. The examples of presentation of the calibration coefficients are included in TRS No. 457 and printed in Tables 59 and 60.

TABLE 59. RADIATION QUALITY DEPENDENT INFORMATION TO BE PROVIDED IN THE CALIBRATION CERTIFICATE FOR LABORATORIES STATING $N_{K,Q0}$ AND K_Q

Radiation quality	Added filtration* mm Al	HVL mm Al	Air kerma rate mGy/min	$N_{K,Q0}$ Gy/C	k_Q	Relative expanded uncertainty, $N_{K,Q0}$ k_Q (k=2) %
RQA 2	6.5	2.24	0.59		1.012	0.77
RQA 3	12.5	3.80	0.63		1.018	0.77
RQA 4	18.5	5.35	0.60		1.012	0.77
RQA 5**	23.5	6.75	0.62	$5.341*10^5$	1.000	0.77
RQA 6	28.5	8.10	0.60		0.995	0.77
RQA 7	32.5	9.18	0.63		0.990	0.77
RQA 8	36.5	10.09	0.64		0.987	0.77
RQA 9	42.5	11.52	0.62		0.988	0.77
RQA 10	47.5	13.36	0.63		0.991	0.77

* Additional filtration to filtration obtained during the establishment of RQR qualities (see Section 6.5.2).
** This beam is generally selected as the reference radiation quality for attenuated beams in general radiography.

TABLE 60. RADIATION QUALITY DEPENDENT INFORMATION TO BE PROVIDED IN THE CALIBRATION CERTIFICATE FOR LABORATORIES STATING $N_{K,Q}$

Radiation quality	Added filtration* mm Al	HVL mm Al	Air kerma rate mGy/min	$N_{K,Q}$ Gy/C	Relative expanded uncertainty, $N_{K,Q}$ (k=2) %
RQA 2	6.5	2.24	0.59	$5.405*10^5$	0.77
RQA 3	12.5	3.80	0.63	$5.437*10^5$	0.77
RQA 4	18.5	5.35	0.60	$5.405*10^5$	0.77
RQA 5	23.5	6.75	0.62	$5.341*10^5$	0.77
RQA 6	28.5	8.10	0.60	$5.314*10^5$	0.77
RQA 7	32.5	9.18	0.63	$5.288*10^5$	0.77

TABLE 60. RADIATION QUALITY DEPENDENT INFORMATION TO BE PROVIDED IN THE CALIBRATION CERTIFICATE FOR LABORATORIES STATING $N_{K,Q}$ (cont.)

Radiation quality	Added filtration* mm Al	HVL mm Al	Air kerma rate mGy/min	$N_{K,Q}$ Gy/C	Relative expanded uncertainty, $N_{K,Q}$ (k=2) %
RQA 8	36.5	10.09	0.64	$5.272*10^5$	0.77
RQA 9	42.5	11.52	0.62	$5.277*10^5$	0.77
RQA 10	47.5	13.36	0.63	$5.293*10^5$	0.77

* Additional filtration to filtration obtained during the establishment of RQR qualities (see Section 6.5.2).

Interpolation of k_Q factor or calibration coefficient for the user beam quality is illustrated with three examples. In the first two examples (Figs 42 and 43) the linear interpolation is made for a specific HVL value. In the third example the calibration coefficient of a reference KAP meter is interpolated relative to tube voltage (Fig. 44). Figure 45 shows thatthe calibration coefficient for a field KAP meter is considered relative to the range of accuracy and tube voltage. The calibration coefficients of KAP meters are presented here relative to tube voltage based on findings in the study of Toroi et al. [19].

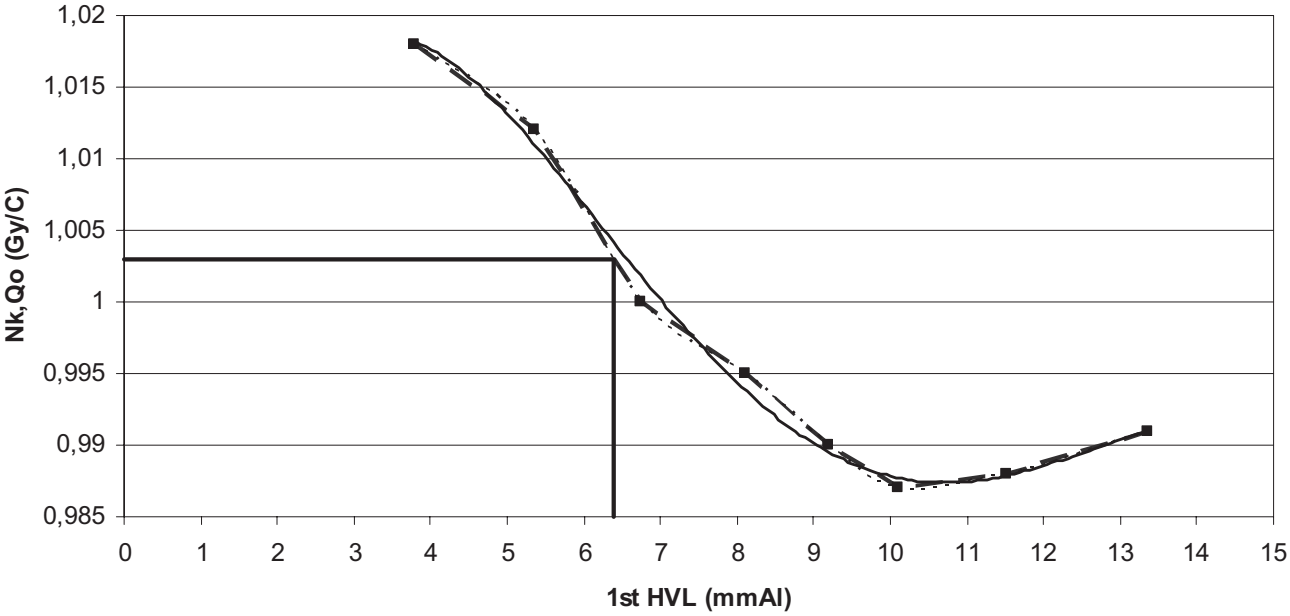

FIG. 42. k_Q factor from Table 59 plotted relative to the first half value layer. Dotted line: smoothed curve through the points. Broken line: linear fit to the points. Continuous curve: fourth order polynomial fit. Lines indicate the determination of the k_Q value for HVL 6.4 mm Al.

In Fig. 42, the data of k_Q from Table 59 is plotted as a graph. The difference of k_Q values between the smoothed curve and the linear fit is negligible. The polynomial curve represents the shape of the curve most clearly, although not going through all the values of k_Q. In this example, the user can make an interpolation from the linear fit, with minor (less than 0.3%) increased contribution to the measurement uncertainty budget. Interpolation can be made either by eye and liner from the figure or, for example, using the TREND function in an Excel sheet.

The linear interpolated value a linear regression:

$$k_Qy = (k_Qb-k_Qa)/(HVLb-HVLa)*HVLx + k_{Qa}$$

TABLE 61. EXAMPLE OF LINEAR INTERPOLATION OF CALIBRATION COEFFICIENT FOR HVL VALUE OF 6.9 MM AL. THE DATA FROM TABLE 59.

	HVL (mmAl)	k_Q
Values from the calibration certificate	5.35	1.012
Interpolated values for user HVL value	6.9	1.007
Values from the calibration certificate	6.75	1.000

$N_{K,y} = (N_{K,b}-N_{K,a})/(HVL_b-HVL_a)*HVL_x + N_{K,a}$

For HVL of 6.9 mmAl the result is:

$N_{K,y} = (1.000-1.012)/(6.75-5.35) * 6.9 + 1.012 = 1.007$

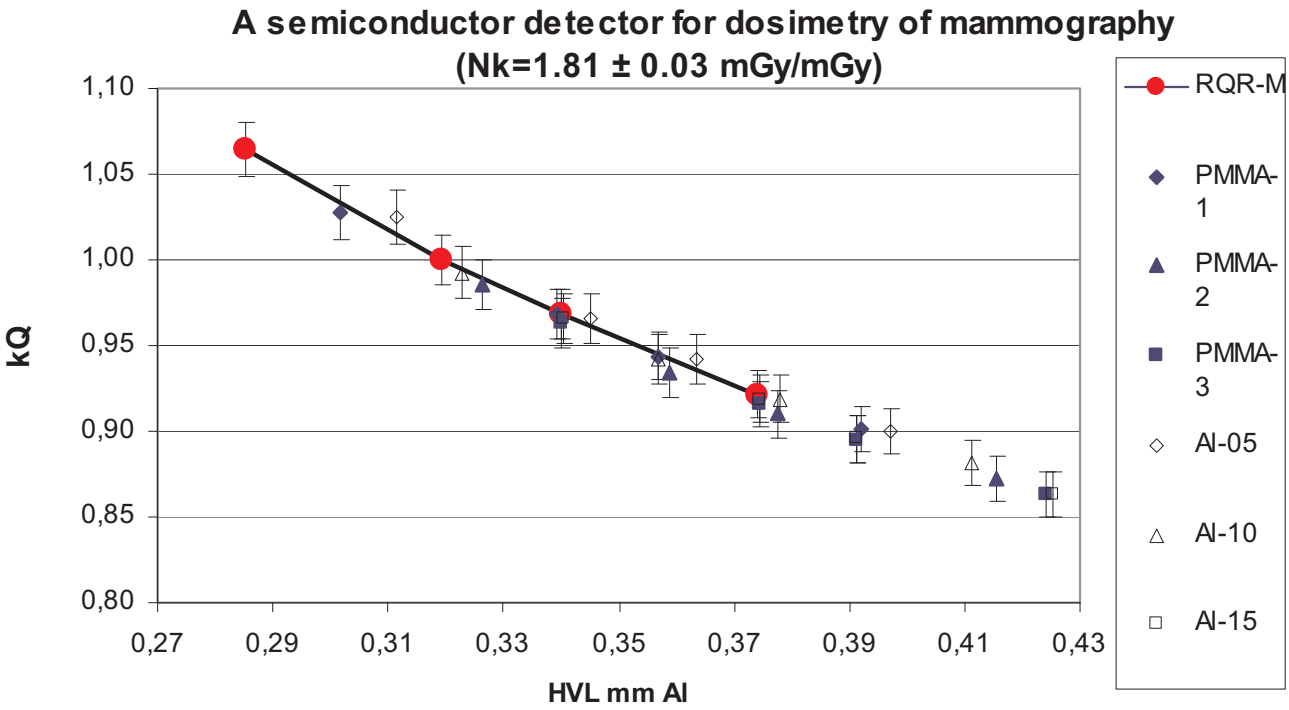

FIG. 43. k_Q factor for radiation qualities of mammography. RQR-M qualities are the standard qualities used at the calibration laboratory. PMMA-1, PMMA-2 and PMMA-3 are radiation qualities with additional filtration of 1 mm, 2 mm and 3 mm. Al-05, Al-10 and Al-15 are radiation qualities with additional filtrations of 0.05 mm, 0.10 mm and 0.15 mm aluminium. The line is a linear fit through points of RQR-M qualities (data provided by Costas Hourdakis).

In an example, only the k_Q factors for RQR-M qualities are assumed. The user needs to interpolate the k_Q factor for a radiation quality with HVL value of 0.311 mm Al. It can be seen in Fig. 43 that the k_Q value obtained by linear interpolation produces a k_Q value close to 1.02. A more accurate value can be calculated by linear regression from the k_Q values in the calibration certificate:

TABLE 62. EXAMPLE OF k_Q VALUES FOR MAMMOGRAPHY BEAM QUALITIES

Radiation quality	Tube voltage (kV)	Filtration	HVL (mm Al)	k_Q
RQR-M1	25	30 μm Mo	0.2851	1.064
RQR-M2	28	30 μm Mo	0.3193	1.000
RQR-M3	30	30 μm Mo	0.3400	0.969
RQR-M4	35	30 μm Mo	0.3741	0.922

TABLE 63. EXAMPLE OF LINEAR INTERPOLATION OF k_Q

	HVL (mm Al)	k_Q
Values from the calibration certificate	0.2851	1.064
User HVL and the interpolated value of k_Q	0.311	1.016
Values from the calibration certificate	0.3193	1.000

$$k_{Q,y} = (k_{Q,b}-k_{Q,a})/(HVL_b-HVL_a)*HVL_x + k_{Q,a}$$

For HVL of 0.311 mm Al the result is:

$$k_{Q,y} = (1.064-1.000)/(0.2851-0.3193) * 0.311 + 1.012 = 1.016$$

Assuming that the filtration of the user radiation quality corresponds with RQR-M + 0.01 mm Al, the measured k_Q value is 1.025. The error made by the user using the data for RQR-M qualities and interpolation is (1.025-1.016)/1.025 *100% = 0.9%; which is negligible.

Figures 44 and 45 present the calibration coefficients for a KAP meter. The results for RQR radiation qualities are those available from the calibration certificate expressed according to TRS No. 457. In addition to the values for RQR radiation qualities, radiation qualities with other additional beam filtrations are also presented.

If the chamber type specific data of k_Q is provided by the manufacturer, similar dissections can be made from that data.

Figure 44 presents the calibration coefficients for a comprehensive calibration of a reference KAP meter. For interpolation of the calibration coefficients for X ray beam quality of 65 kV and 4 mm Al filtration, the values of 3 mm Al and 5 mm Al filtrations are used. First the calibration factors are interpolated for 65 kV tube voltage for 3 mm Al and for 5mm Al filtrations and in the second step, the interpolation is performed for filtration of 4 mm Al at 65 kV tube voltage. Similarly the values for other tube voltages can be interpolated. Figure 44 shows calibration coefficients plotted with smoothed curves, but reliable results with minor error can be interpolated in this example by the linear fit for each interval of tube voltage. The calibration coefficients are presented relative to tube voltage [19].

In Fig. 45, a ‘window’ of calibration coefficients of about +– 10% is presented as a function of tube voltage allowing an optimized selection of radiation quality for a field KAP meter when a single calibration coefficient needs to be used. The range of radiation qualities should be consistent with the clinical use of X ray equipment and the importance of the reliable dosimetry for the specific examination types that are performed with that specific X ray equipment.

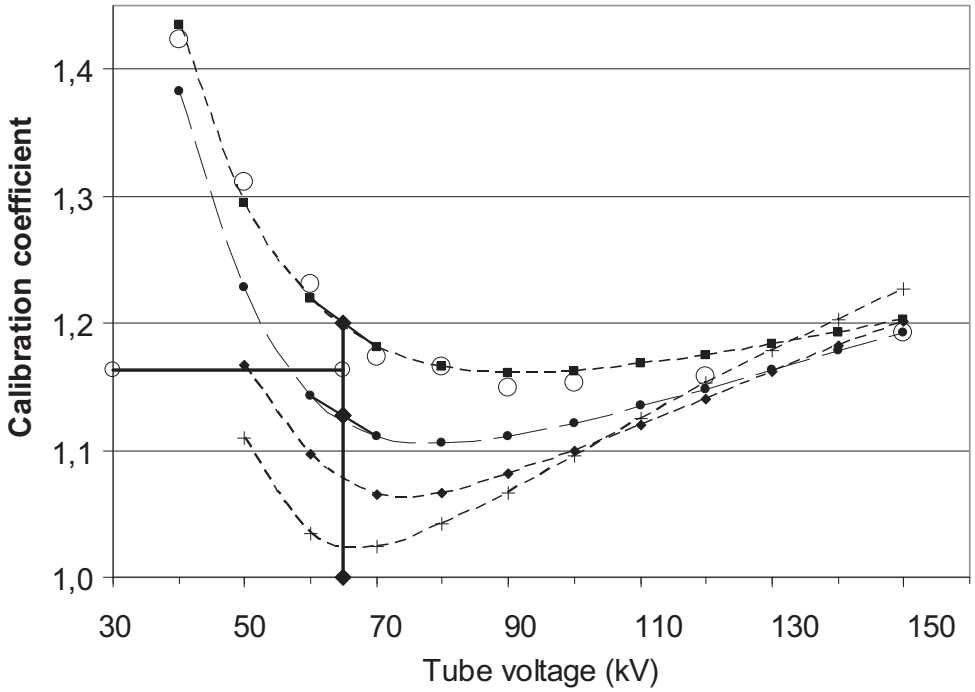

FIG. 44. Interpolation of calibration coefficients of a reference KAP meter (incident radiation). The calibration coefficients are plotted relative to tube voltage for different tube filtrations; RQR radiation qualities (large circles), 3 mm Al (squares), 5 mm Al (closed circles), 4 mm Al + 0.1 mm Cu (diamonds) and 4 mm Al + 0.2 mm Cu (crosses). Linear interpolation of the calibration coefficient for quality of 65 kV and 4 mm Al is presented by thick lines. In this example, the calibration coefficients are presented in arbitrary units.

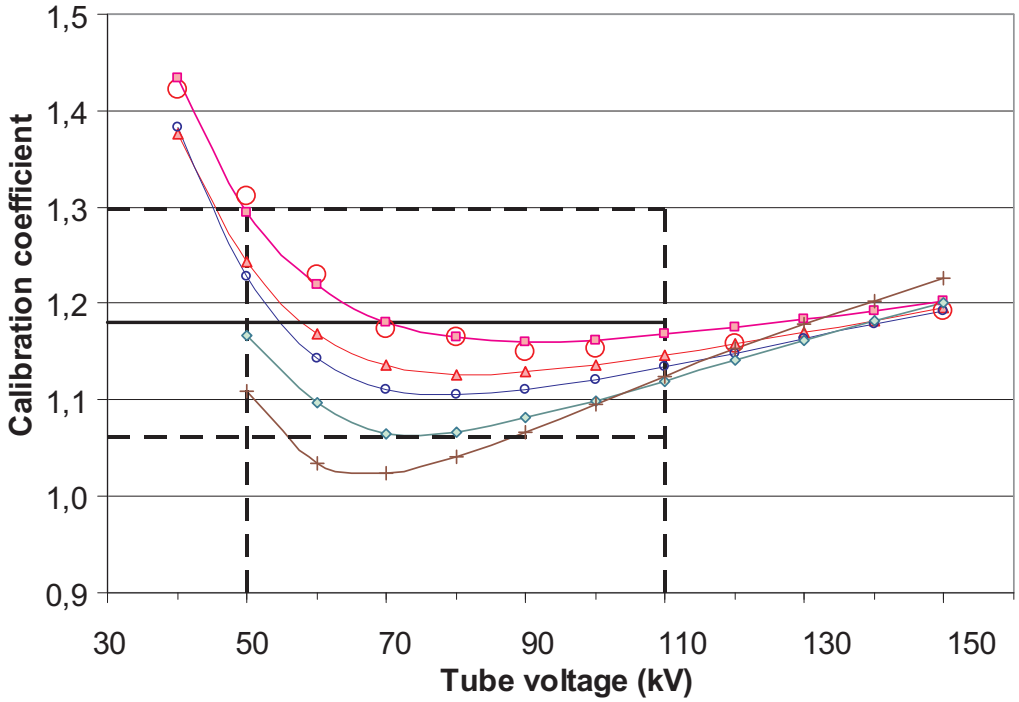

FIG. 45. Calibration coefficient of a field KAP meter relative to tube voltage for different tube filtrations, 3 mm Al (squares), 4 mm Al(triangles), 5 mm Al (circles), 4 mm Al + 0.1 mm Cu (diamonds) and 4 mm Al + 0.2 mm Cu (crosses). Continuous line: the selected calibration coefficient. Broken line: Range for +/– 10% relative to value of the calibration coefficient and the range of tube voltage under consideration (50 kV to 110 kV).

Appendix V

CLINICAL OPERATION CONDITIONS AND SUGGESTED CALIBRATION BEAM QUALITIES FOR KAP

KAP meters are increasingly recognized as beneficial in the measurement of patient dose for a large range of clinical diagnostic radiological procedures; including radiographic and fluoroscopic. Their use in complex and higher risk procedures is well established particularly in interventional angiography and procedural work. The use of KAP meters should be particularly supported in paediatric examinations. In order to calibrate the KAP meter most effectively, beam qualities should be defined that correspond to their clinical use.

V.1. CLINICAL EXAMINATION CONDITIONS

V.1.1. Interventional examinations

Interventional procedures include both cardiac and a large range of non-cardiac procedures. The beam qualities used for these procedures are usually specified by preset conditions supplied by the manufacturer. While the operator may have some choice in machine operation through the selection of appropriate preset conditions, the choice is often limited in terms of the tube voltage and the filtration to be used. Moreover, the actual radiographic exposure (again in terms of the tube voltage, filtration and tube current used) is under automatic control which varies according to patient conditions; notably the patient attenuation or thickness. An example of such an equipment set-up has been given by Lin [39] showing the operation configuration for a Siemens unit. In the case shown in Fig. 46, the filtration includes added copper of 0.2–0.9 mm operating typically at tube voltages of 60–80 kV. It is understood that this is also typical for other manufactures of modern interventional equipment, with Philips using 0.4–0.9 mm Copper in fluoroscopy mode. In order to know the actual beam conditions used clinically, it is necessary to know the thickness of the patient for each X ray projection or find the information in the DICOM header. Such an analysis has been completed by Balter for GE equipment; with similar studies for other equipment [40].

In summary, it is clear that potentially a large range of tube voltages and filtrations may be used clinically, ranging from no added copper to 0.9 mm added copper. It is also evident that the filtrations used may be different for cine and fluoroscopy modes for some units[39].

V.1.2. Procedural examinations

Most procedural work including Barium meals, vascular mobile C-arms, urology etc., do not use added copper with a typical total filtration of 4 mm Al for a tube voltage range of 60–110 kV.

V.1.3. Paediatric examinations

Cardiac examinations for paediatric patients using a Siemens bi-plane unit has recently been investigated [40]. Other paediatric examinations are similar to those described in Section 3.1.2.4, but typically with lower tube voltages

V.2. SUGGESTED BEAM QUALITIES FOR KAP CALIBRATION

When KAP meter is calibrated in a clinical situation, range of clinically used radiation qualities should be covered [19]. All available filtrations should be used with a range of tube voltages and interpolation within this tube voltage range is possible.

If a KAP meter is calibrated in a laboratory, clinically used radiation qualities are not necessarily available. RQR standard radiation qualities are generally used for calibration of diagnostic meters and proper calibration

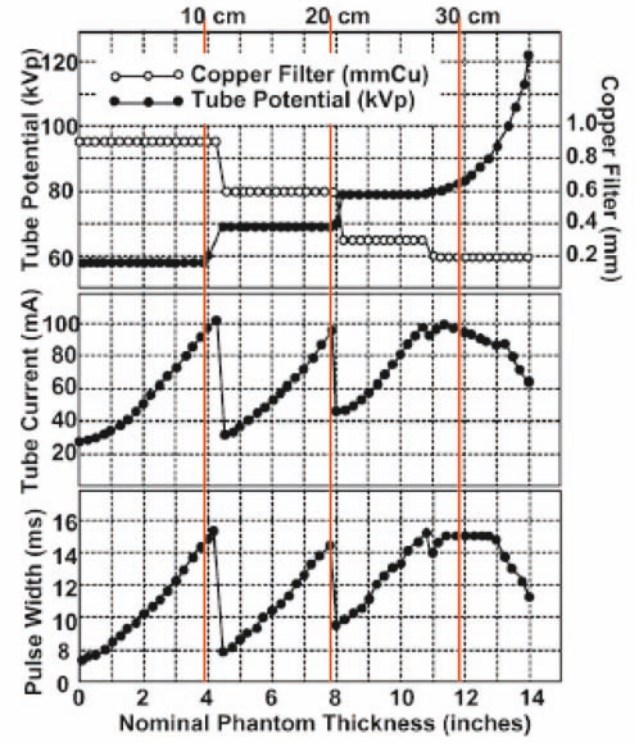

FIG. 46. Various imaging parameters as functions of phantom thickness. Note that the PMMA thicknesses in 10, 20, and 30 cm are shown covering all three graphs [39].

coefficients are selected based on HVL. For KAP meters, interpolation based on HVL is not reliable [19]. Interpolation based on two radiation quality specifiers is difficult with RQR qualities with variation filtration and tube voltage. In this case, it would be more convenient to do KAP calibration with fixed filtration and range of tube voltages, consistent with the clinically used radiation qualities.

Clinically used aluminium filtrations could be covered quite well with filtrations of 3 mm and 5 mm Al. However, a curve with RQR radiation qualities could be used as a rough estimation for this filtration range with an error typically less than 10% [19]. The largest problem occurs in radiation qualities with added copper filtrations. If these radiation qualities are simulated, the thickness of aluminium is not essential. Typically values can be simulated with 3.5–4.5 mm Al. The selection of the thickness of copper filtration is more complex. Typically, used copper filtrations can be covered with 0.1 mm and 0.2 mm Cu. If calibration capabilities for all possible copper filtration from 0.1 mm to 0.9 mm is needed, interpolation based on filtration with fixed tube voltage could possible be used. In this case, copper filtration of 0.6 mm and 0.9 mm could be included and interpolation based on those four different copper filtrations could be performed. Tube voltage range could be covered with selected values for example 50 kV, 70 kV, 90 kV and 120 kV.

Appendix VI

FABRICATION OF PHANTOMS

Additional information can be found in the AAPM report 31 [41].

CDRH chest: The chest phantom consists of 25.4 × 25.4 cm pieces of type 1100 alloy aluminium and PMMA (clear acrylic) with a 19 cm air gap. The exact configuration of aluminium, PMMA and air gap is detailed in Fig. 47. Clinical testing of the phantom has shown it to be equivalent to a 23 cm patient for the PA chest projection [42].

CDRH abdomen/lumbar spine: The abdomen and lumbar spine phantom consists of 25.4 × 25.4 cm pieces of PMMA 16.95 cm thick in the soft tissue region and 0.46 cm of aluminium (type 1100 alloy) and 18.95 cm PMMA for the spinal region. The exact configuration of aluminium and PMMA is detailed in Fig. 48. Clinical testing of the phantom has shown it to be equivalent to a 21 cm patient for the AP abdomen and lumbar spine projections [43].

There are currently two CT dosimetry phantoms commonly in use. The head phantom consists of a 16 cm diameter PMMA cylinder 15 cm in length. The body phantom consists of a 32 cm diameter PMMA cylinder 15 cm in length. Both phantoms have 8 surface dosimeter holes and one central dosimeter hole with removable acrylic rods or alignment rods, however only 4 are used for dosimetry and the other 4 are not needed and are there for historical purposes. The exact configuration of the head phantom is shown in Fig. 49.

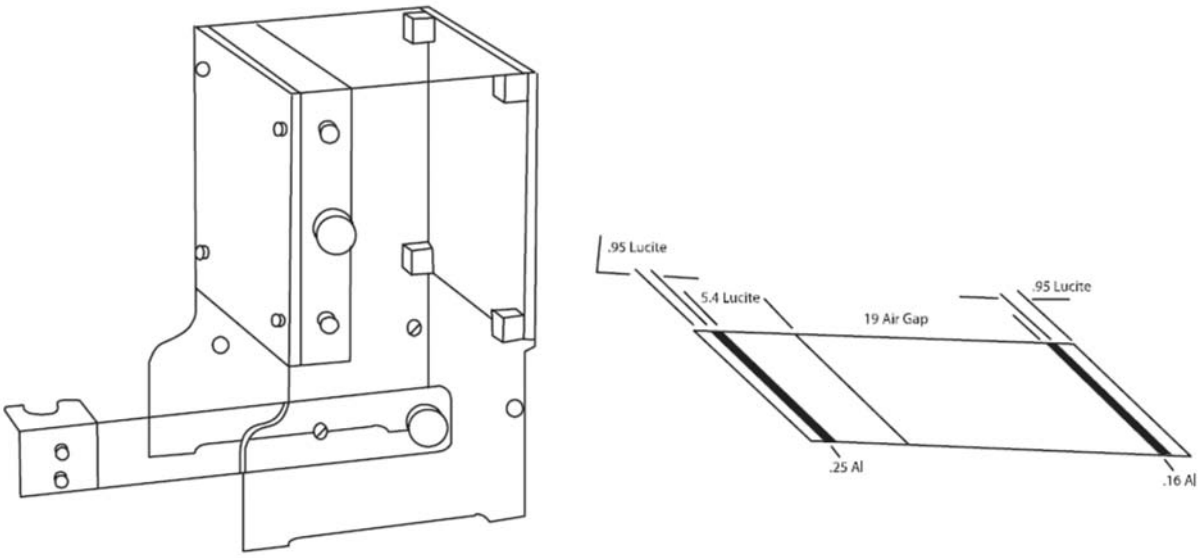

FIG. 47. CDRH patient equivalent and aluminium (LucAl standard chest phantom (all dimensions in cm)[41].

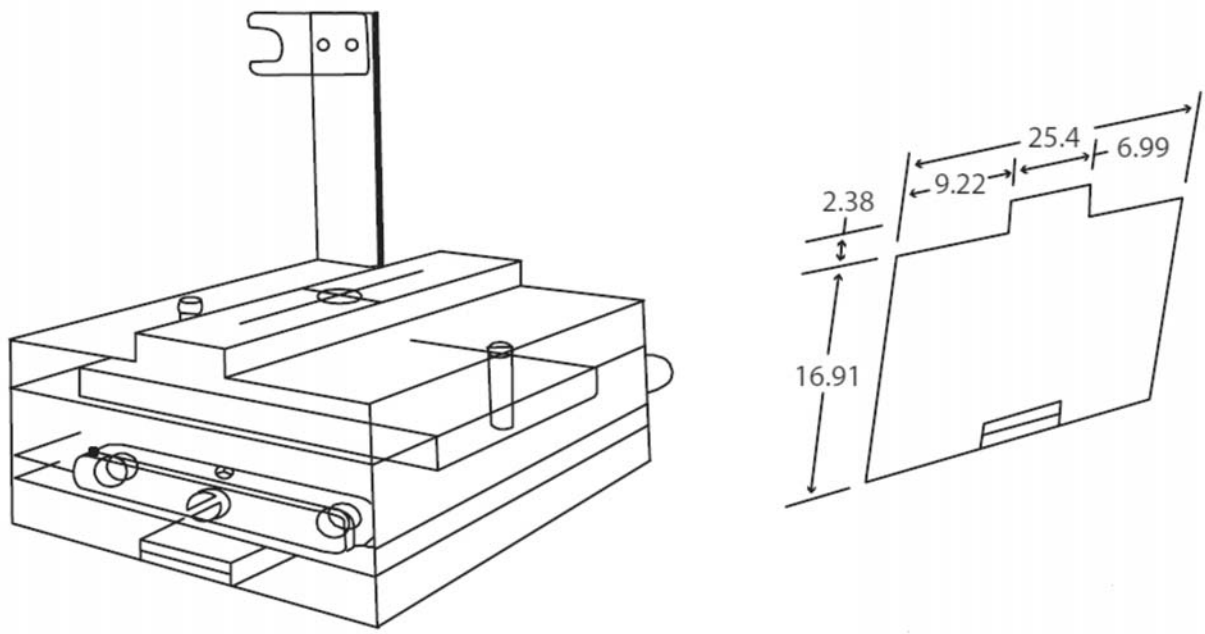

FIG. 48. CDRH patient equivalent PMMA (Lucite) and aluminium (LucAl) standard abdomen and lumbo-sacral spine phantom (all dimensions are in cm)[41].

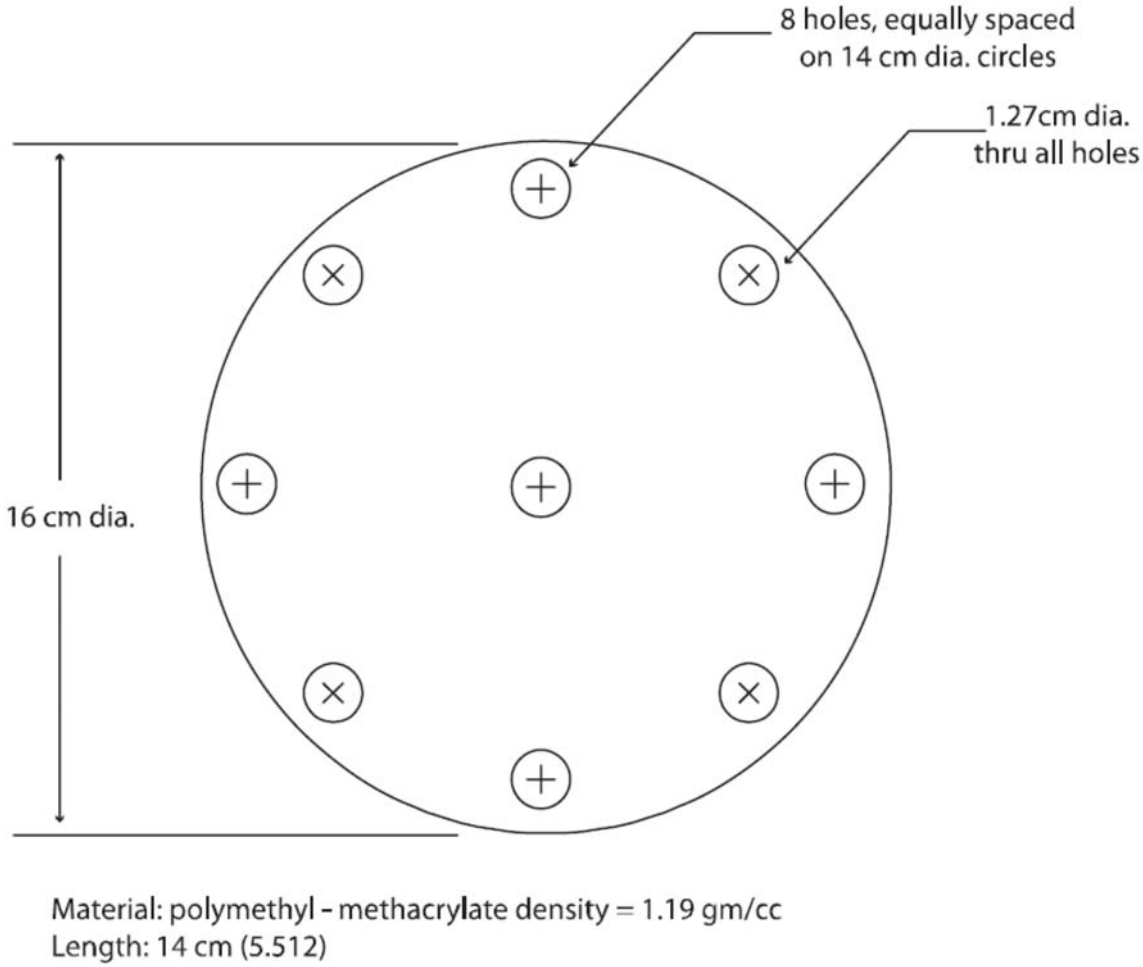

FIG. 49. Head CT phantom [41].

Appendix VII

CT WORKED EXAMPLES

VII.1. NOMENCLATURE

Confusion may easily arise from the multiple uses of P_{IT}, which can be used to represent tube current — time product (mAs) per rotation, effective mAs (i.e. mAs corrected for pitch), and total mAs per scan. In the following examples, P_{IT} has been used solely for mAs per rotation, with P_{IT}*eff* used to denote effective mAs and P_{IT}*tot* to denote total mAs. It should be noted that, for helical scanners utilizing effective mAs, C_{VOL} may be redefined as follows.

$$C_{VOL} = {}_{n}C_{VOL}\ P_{IT}$$

where P_{IT} is mAs per rotation for scan

$$C_{VOL} = {}_{n}C_{w}\ P_{IT}eff$$

where P_{IT}*eff* is effective mAs per slice for scan

$P_{KL,CT}$ may be defined in a number of equivalent ways, as further illustrated and the equation of choice will depend on the information provided on a particular scanner.This varies between manufacturers and sometimes even between scanner models. The equations considered to be of most general use are given in bold, and a series of worked examples are then given for different scanner types. It should be noted that, as scanner technology is rapidly changing, some of the specific comments made regarding displayed information may become outdated, although the general principles and considerations will remain the same.

$$\begin{aligned} P_{KL,CT} &= \mathbf{C_{VOL}\ x\ scan\ length} \\ &= \boldsymbol{nl}\ \mathbf{C_{VOL}} \\ &= nl\ {}_{n}C_{w}\ P_{IT}eff \\ &= nl\ {}_{n}C_{w}\ P_{IT}\ NT/l \\ &= NT\ {}_{n}C_{w}\ nP_{IT} \\ &= \boldsymbol{NT\ {}_{n}C_{w}\ P_{IT}tot} \\ &= l\ {}_{n}C_{VOL}\ P_{IT}tot \end{aligned}$$

where l is the distance moved by the couch per scanner rotation in cm, n is the number of rotations, P_{IT}*eff* the effective mAs per slice, P_{IT} the mAs per rotation, P_{IT}*tot* the total mAs per scan (= mA x total scan time) and NT is the nominal slice width (N simultaneously obtained slices of thickness T cm). The units of $P_{KL,CT}$ are mGy cm

VII.2. WORKED EXAMPLE FOR A SINGLE SLICE SCANNER

Single slice scanners tend to use values of mAs per rotation rather than effective mAs and do not use the concept of pitch, specifying instead an incremental table movement together with nominal slice width. The total number of slices can be used to determine the scan length or the total mAs. The data collection sheet for single slice scanners should have a single column for ‘nominal slice width’, a column for ‘couch increment’ and a column for ‘total number of slices’. The sheets should not have columns for ‘pitch’ or effective mAs.

Example

Patient has an abdomen CT scan with the following technique settings:

10 mm slice width

15 mm increment
200 mAs / rotation
25 slices
${}_nC_w$ has been measured as 0.008 mGy/mAs

Using $C_{VOL} = {}_nC_{VOL}\, P_{IT}$ and ${}_nC_{VOL} = {}_nC_w NT/l$, for this scan

$$C_{VOL} = 0.008 \times \frac{1 \times 10}{15} \times 200 = 1.07 \text{ mGy}$$

Using $P_{KL,CT} = nl\, C_{VOL}$,
$P_{KL,CT} = 25 \times 1.5 \times 1.07 = 40$ mGycm
Or, using $P_{KL,CT} = NT\, {}_nC_w\, P_{IT}tot$
$P_{KL,CT} = 1.0 \times 0.008 \times (200 \times 25) = 40$ mGycm

VII.3. WORKED EXAMPLE FOR A SIEMENS MULTI SLICE SCANNER

Siemens multi slice scanners specify pitch and effective mAs. A value of total mAs for the scan is usually provided post exposure. Care should be taken that, if calculating $P_{KL,CT}$ for a 64 slice setting, the correct irradiated slice thickness is used in the equations e.g. 32 × 0.6 mm rather than 64 × 0.6 mm, as discussed in Section 4.4. Data collection sheets for these scanners should have columns for ‘effective mAs’ and ‘total mAs’.

Example

Patient has an abdomen CT scan with the following technique settings:

4 × 5 mm slices
Pitch 1.2
220 effective mAs
total mAs 9240
${}_nC_w$ has been measured as 0.008 mGy/mAs

Using $C_{VOL} = {}_nC_w\, P_{IT}eff$, for this scan
$C_{VOL} = 0.008 \times 220 = 1.76$ mGy
Using $P_{KL,CT} = NT\, {}_nC_w\, P_{IT}tot$

$$P_{KL,CT} = \frac{4 \times 5}{10} \times 0.008 \times 4240 = 68 \text{ mGycm (divisor of 10 is to convert mm to cm)}$$

VII.4. WORKED EXAMPLE FOR A PHILIPS MULTI SLICE SCANNER

Philips multi slice scanners also use pitch and effective mAs, but do not generally give a value of total mAs. The total scan length should be used to derive $P_{KL,CT}$. The data collection sheets for these scanners should correspondingly have columns for ‘total scan length’ and ‘effective mAs’.

Example

Patient has an abdomen CT scan with the following technique settings:

4 × 5 mm slices

Pitch 1.2
220 effective mAs
scan length 400mm
$_nC_w$ has been measured as 0.008 mGy/mAs

Using $C_{VOL} = {}_nC_w P_{IT}eff$, for this scan
C_{VOL} = 0.008 x 220 = 1.76 mGy
Using $P_{KL,CT} = C_{VOL} \times$ scan length

$$P_{KL,CT} = \frac{400}{10} \times 1.76 = 70 \text{ mGycm (divisor of 10 is to convert mm to cm)}$$

Appendix VIII

UNCERTAINTY BUDGETS PRESENTED BY PARTICIPANTS

This appendix presents the uncertainty budgets sent by participants. For reasons of simplicity, country related data are presented in alphabetical order.

Brazil

Chief investigator: M.M.O. Ramos
Research team: J.G.P. Peixoto, L. Baptista
Institution: Laboratório Nacional de Metrologia das Radiações Ionizantes — LNMRI
Instituto de Radioproteção e Dosimetria, Caixa Postal 37750
CEP.: 22780-160, Rio de Janeiro — RJ

Uncertainty budget for calibration of RQR meters at LNMRI/IRD — Brazil

Component of uncertainty		Type of uncertainty (A/B)	Distribution	Standard uncertainty (%)*
Symbol	Name			
Measurements with reference chamber				
N_K	Calibration of reference chamber (RQR)	B	Normal	0.385
k_{stab}	Stability of ref. ion. chamber	B	Rectangular	0.18
M_{raw}	Repeatability of the ref. ion. chamber	A	Normal	0.04
k_s	Saturation correction.	B	Rectangular	0.08
k_{leak}	Leakage current	A	Normal	0.05
k_{dist}	Deviation from ref. distance	B	Rectangular	0.04
k_{elec}	Electrometer calibration (K6517A)	B	Normal	0.05
$k_{elec-res}$	Electrometer resolution (K6517A)	B	Rectangular	0.02
$k_{t,p}$	Air density correction for T and P	A	Normal	0.03
	T and P cal. factors	B	Rectangular	0.60
$k_{Q,Q°}$	Difference in beam quality (from calibration laboratory)	B	Rectangular	0.09
Measurements with user's instrument (chamber and electrometer)				
M_{raw}	Repeatability of the user instrument (5)	A	Normal	0.40
k_s	Saturation correction.	B	Rectangular	0.08
k_{res}	Instrument resolution (user)	B	Normal	0.10
k_{leak}	Leakage current	A	Normal	0.05
k_{dist}	Deviation from ref. distance	B	Rectangular	0.04
$k_{t,p}$	Air density correction for T and P (5)	A	Normal	0.03
	T and P cal. factors	B	Rectangular	0.60
N_K^{user}	Calibration coefficient of user diagnostic dosimeter	Combined Expanded (k = 2)		0.70 1.4

Cuba

Chief investigator: G.W. Salas
Institution: Centro de Protección e Higiene de las Radiaciones — CPHR
20 No. 4113 e/ 41 y 47 Playa, Ciudad de la Habana — Cuba

Uncertainty budget for calibration of RQR meters at CPHR — SSDL in Cuba

Component of uncertainty		Type of uncertainty (A/B)	Distribution	Standard uncertainty (%)
Symbol	Name			
Components influencing only secondary standard				
Nk	Calibration of the secondary standard (at PSDL)	B	Normal	0.39
	Long term stability of secondary standard	B	Rectangular	0.20
	Electrometer calibration	B	Rectangular	0.14
	Scale reading	A	Normal	0.01
	Different energy spectrum	B	Rectangular	0.20
	Air density	B	Rectangular	0.10
	Leakage	B	Rectangular	0.01
Components influencing only instrument calibrated				
	Scale reading	A	Normal	0.01
	Recombination loss	B	Rectangular	0.01
	Leakage	B	Rectangular	0.01
	Air density	B	Rectangular	0.10
Components influencing both instruments				
	Positioning of the chambers	B	Rectangular	0.12
	Field inhomogeneity	B	Rectangular	0.14
	X ray output	B	Rectangular	0.29
N_K^{user}	Calibration coefficient of user diagnostic dosimeter	Combined Expanded (k = 2)	Normal Normal	1.24

Note: that the uncertainty budget is estimated for calibration of working standard. For calibration of user dosimeter it is used the working standard that lead to uncertainty of 1.58 (K=2).

Czech Republic

Chief investigator: I. Horáková
Institution: National Radiation Protection Institute

Uncertainty budget for calibration of air kerma RQR meters at National Radiation Protection Institute, Czech Republic

Component of uncertainty		Type of uncertainty (A/B)	Distribution	Standard uncertainty (unit*)
Symbol	Name			
N_K	Calibration coefficient of reference ion. chamber	B	Normal	1,9%
	Stability of ref. ion. chamber	A	Normal	0,4%
M_0^{ref}	Leakage of ref. ion. chamber	A	Normal	0,1%
m_0	Leakage of monitor chamber	A	Normal	0,05%
M^{ref}/m^{ref}	Ratio of ref. ion. chamber response to monitor chamber response	A	Normal	0,04%
	Measuring capacitance	B	Normal	0,15%
	Electrometer calibration	B	Normal	0,001%
M^{user}/m^{user}	Ratio of user diagnostic dosimeter response to monitor chamber response	A	Normal	0,05%
	Resolution of user diagnostic dosimeter	B	Rectangular	0,01%
M_0^{user}	Leakage of user diagnostic dosimeter	A	Normal	0,15%
m_0	Leakage of monitor chamber	A	Normal	0,05%
$k_{T,\,p}$	Temperature and pressure	A	Normal	0,5%
	Chamber positioning	B	Rectangular	0,2%
	Field inhomogeneity	B	Rectangular	0,25%
	Difference in beam quality	B	Rectangular	0,25%
N_K^{user}	Calibration coefficient of user diagnostic dosimeter	Combined Expanded (k = 2)	Normal Normal	2,06% 4,1%

Finland

Chief investigator: A. Kosunen
Research team: P. Toroi
Institution: STUK Radiation and Nuclear Safety Authority, Finland.
Calibration of the working standard with RQR-radiation qualities

Uncertainty budget for calibration of RQR-radiation qualities at STUK Finland

Component of uncertainty		Type of uncertainty (A/B)	Distribution	Standard uncertainty (%)
Symbol	Name			
Measurement of air kerma with secondary standard				
N_k	Air kerma calibration coefficient for secondary standard	A	Normal	0.3
		B	Rectangular	0.25
	Constancy of air kerma-calibration coefficient	B	Rectangular	0.29
	Measurement of ionization current	A	Normal	0.2
		B	Rectangular	0.06
	Uniformity differences of radiation beams	B	Rectangular	0.29
	Differences in energy spectra of radiation beams	B	Rectangular	0.58
	Leakage current	B	Rectangular	0.12
	Air temperature and pressure correction	B	Rectangular	0.08
	Correction for recombination			
	Correction for polarity			
	Stem effect			
Measurement of air kerma with working standard				
	Positioning to calibration distance	B	Rectangular	0.12
	Uniformity differences of radiation beams (difference in chamber size)			
	Air temperature and pressure correction	B	Rectangular	0.08
	Display accuracy	A	Normal	0.3
N_K	Calibration coefficient of working standard	Combined Expanded (k = 2)	Normal Normal	0.91 1.82

Calibration of the user diagnostic meter with RQR radiation qualities

Uncertainty budget for calibration of RQR radiation qualities at STUK Finland				
Component of uncertainty		Type of uncertainty (A/B)	Distribution	Standard uncertainty (%)
Symbol	Name			
Measurement of air kerma with working standard				
N_k	Air kerma calibration coefficient for working standard	A	Normal	0.47
		B	Rectangular	0.78
	Constancy of air kerma-calibration coefficient	B	Rectangular	0.29
	Measurement of ionization current	A	Normal	0.2
		B	Rectangular	0.06
	Uniformity differences of radiation beams			
	Differences in energy spectra of radiation beams			
	Leakage current	B	Rectangular	0.12
	Air temperature and pressure correction	B	Rectangular	0.08
	correction for recombination			
	correction for polarity			
	stem effect			
Measurement of air kerma with user diagnostic meter				
	Positioning to calibration distance	B	Rectangular	0.12
	Uniformity differences of radiation beams (difference in chamber size)	B	Rectangular	0.17
	Air temperature and pressure correction	B	Rectangular	0.08
	Display accuracy	A	Normal	0.5
N_K^{user}	Calibration coefficient of user diagnostic dosimeter	Combined Expanded (k = 2)	Normal Normal	1.13 2.26

Uncertainty budget for calibration of KAP meters at STUK Finland				
Component of uncertainty		Type of uncertainty (A/B)	Distribution	Standard uncertainty (%)
Symbol	Name			
1. KAP rate measurement with working standard				
Measurement of air kerma with working standard				
N_k	Air kerma calibration coefficient for secondary standard	A	Normal	0.47
		B	Rectangular	0.78
	Constancy of air kerma-calibration coefficient	B	Rectangular	0.29
	Measurement of ionization current	A	Normal	0.20
		B	Rectangular	0.06
	Leakage current	B	Rectangular	0.12
	Air temperature and pressure correction	B	Rectangular	0.08
	Positioning	B	Rectangular	0.29
	Uniformity differences of radiation beams (difference in chamber size)	B	Rectangular	0.29
A	Uncertainty for area measurement	B	Rectangular	0.58
		Combined	Normal	1.21
2. KAP measurement with diagnostic KAP meter				
M	Positioning of KAP meter	B	Rectangular	0.09
	Repetition	A	Normal	0.30
	Reading	B	Rectangular	0.03
	Leakage current	B	Rectangular	0.23
k_t	Temperature correction	B	Rectangular	0.14
K_p	Ambient pressure correction	B	Rectangular	0.58
	Repeatability of X ray machine	B	Rectangular	0.25
t_s	Radiation time	B	Rectangular	0.12
		Combined	Normal	0.76
N_K^{user}	Calibration coefficient of user diagnostic dosimeter	Combined Expanded (k = 2)	Normal Normal	1.43 2.87

Greece

Chief investigator: C.J. Hourdakis
Institution: Greek Atomic Energy Commission
Ionizing Radiation Calibration Laboratory, Attiki, Athens

Uncertainty budget for Calibration of dosemeters used in diagnostic radiology at Greek Atomic Energy Commission's Calibration Laboratory

Component of uncertainty		Type of uncertainty (A/B)	Distribution[1]	Standard uncertainty (unit)[2]
Symbol	Name			
Measurements of K_{air} by SSDL reference chamber				
	Nk from calibration laboratory	B	R	1.30%
	Nk stability	A	G	0.50%
	Electrometer accuracy	B	R	—[4]
	Scale reading/resolution	B	R	0.10%
	Uniformity of X ray beam	B	R	0.58%[5]
	Difference in beam quality (from calibration laboratory)	B	R	0.50%
	Positioning in distance	B	R	0.23%
	$k_{P,T}$ — Temperature and pressure[3]	A	G	0.17%
		B	R	0.08%
	Electrometer builtin timer	A	G	—[4]
		B	R	—[4]
	Leakage current	B	R	0.00%
	Chamber recombination losses	B	R	0.00%
	Shutter timer accuracy	B	R	0.05%
	Shutter timer reproducibility	A	G	0.05%
Measurements with user's dosemeter				
	Electrometer accuracy	B	R	0.50%
	Scale reading/resolution	B	R	0.10%
	Uniformity of X ray beam	B	R	0.58%[5]
	Positioning in distance	B	R	0.23%
	$k_{P,T}$ — Temperature and pressure[3]	A	G	0.17%
		B	R	0.08%
	Electrometer Built in timer	A	G	0.50%
		B	R	
	Leakage current	B	R	0.00%
	Chamber recombination losses	B	R	0.00%

Uncertainty budget for Calibration of dosemeters used in diagnostic radiology at Greek Atomic Energy Commission's Calibration Laboratory

Component of uncertainty		Type of uncertainty (A/B)	Distribution[1]	Standard uncertainty (unit)[2]
Symbol	Number			
	Shutter timer accuracy	B	R	0.05%
	Shutter timer reproducibility	A	G	0.05%
	Reproducibility of Kair measurements with SSDL instrument	A	G	0.5%
	Reproducibility of measurements with USER instrument	A	G	0.5%
	Field size for KAP or KLP calibration	B	R	3.00%

Remarks:

[1] Type A uncertainties follow a Gaussian distribution (indicated as G) and type B uncertainties follow a rectangular (indicated as R)

[2] All uncertainties are expressed as percentages at 1 SD.

[3] The uncertainty of the kP,T factor is calculated, taking into account the calibration factors of the thermometer and barometer, the difference of temperature between the thermometer and point of measurement and any variation of temperature during the measurements.

[4] Included in calibration factor. Otherwise, the instrument specifications used.

[5] Chamber size is taken into account (non uniform chamber irradiation because of field inhomogeneity).

[6] This budget is used for all type of calibrations in terms of K_{Air}, P_{KL}, P_{KA}. The fields are evaluated accordingly.

[7] The values presented in last column are typical values. User's instrument performance characteristics may change the above values significantly.

Coverage factor = 2, Confidence level 95% (2 SD).

Thailand

Chief investigator: S. Srimanoroth
Institution: SSDL Bangkok, Division of Radiation and Medical Devices
Department of Medical Sciences (DMSC), Nonthaburi

Uncertainty budget for calibration of air kerma RQR meters at SSDL Bangkok, Division of Radiation and Medical Devices, Thailand.

Component of uncertainty		Type of uncertainty (A/B)	Distribution	Standard uncertainty (%)
Symbol	Name			
	Calibration Uncertainty of Standard	B	Rectangular	1.761
	X ray reproducibility	B	Rectangular	0.01
	X ray ripple	B	Rectangular	0.08
	Standard measurement			
	ratio of measurement	A	Normal	0.0254
	— reading display	B	Rectangular	0.024
	— temperature scale	B	Rectangular	0.0097
	— temperature cavity dev.	B	Rectangular	0.0335
	— pressure factor	B	Rectangular	0.0285
	— chamber position	B	Rectangular	0.05
	User measurement			
	— ratio of measurement	A	Normal	0.0282
	— reading display	B	Rectangular	0.024
	— temperature scale	B	Rectangular	0.0097
	— temperature cavity dev.	B	Rectangular	0.067
	— pressure factor	B	Rectangular	0.0285
	— chamber position	B	Rectangular	0.1
	Calibration coefficient of user diagnostic dosimeter	Combined Expanded (k = 2)		1.7687 3.54

Uncertainty budget for calibration of PPV meters at SSDL Bangkok, Division of Radiation and Medical Devices, Thailand.

Component of uncertainty		Type of uncertainty (A/B)	Distribution	Standard uncertainty (%)
Symbol	Name			
	Reproducibility of waveform measurement	A	Normal	0.0216
	Reproducibility of test reading	A	Normal	0.056
	Deviation of waveform reading	B	Rectangular	0.5774
	X ray divider certificate	B	Rectangular	0.01
	Deviation of divider calibration factor	B	Rectangular	0.008
	X ray HV ripple	B	Rectangular	0.08
	Oscilloscope certificate	B	Rectangular	0.06
	Calibration coefficient of user diagnostic dosimeter	Combined Expanded (k = 2)		0.5892 1.18

Vietnam

Chief investigator: Tran Ngoc Toan
Institution: Atomic Energy Commission's SSDL

Component of uncertainty		Type of uncertainty (A/B)	Distribution	Standard uncertainty (%)
Symbol	Name			
Measurements of K_{air} by SSDL reference chamber				
N_k	Calibration coefficient of reference ionization chamber	B	Rectangular	1.0
k_{stab}	Stability of reference ion. chamber	A	Normal	0.30
k_{elec}	Electrometer accuracy	B	Rectangular	0.20
$k_{elec\text{-}res}$	Scale reading/resolution	B	Rectangular	0.10
k_{unif}	Field inhomogeneity of X ray beam	B	Rectangular	0.20
k_{qual}	Difference in beam quality (from calibration laboratory)	B	Rectangular	0.30
k_{pos}	Positioning in distance	B	Rectangular	0.30
$k_{P,T}$	Air density correction for temperature and pressure	A	Normal	0.20
		B	Rectangular	0.10
M_0^{ref}	Leakage of reference ion chamber	A	Normal	0.01
m_0^{ref}	Leakage of monitor chamber	A	Normal	0.01
M^{ref}/m^{ref}	Ratio of reference chamber reading to monitor chamber reading	A	Normal	0.05
k_{shut}	Shutter timer reproducibility	A	Normal	0.05
Measurements with user's dosimeter				
k_{stab}	Stability of user ion chamber	A	Normal	0.30
k_{elec}	Electrometer accuracy	B	Rectangular	0.20
$k_{elec\text{-}res}$	Scale reading/resolution	B	Rectangular	0.10
k_{unif}	Field inhomogeneity of X ray beam	B	Rectangular	0.20
$k_{P,T}$	Air density correction for temperature and pressure	A	Normal	0.20
		B	Rectangular	0.10
M_0^{user}	Leakage of user ion chamber	A	Normal	0.02
m_0^{ref}	Leakage of monitor chamber	A	Normal	0.01
M^{user}/m^{user}	Ratio of user chamber reading to monitor chamber reading	A	Normal	0.06
k_{shut}	Shutter timer reproducibility	A	Normal	0.05
k_{pos}	Positioning in distance	B	Rectangular	0.30
N_k^{user}	Calibration coefficient of user air kerma (exposure) meter	Combined Expanded (k=2)		1.32 2.64

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Annex I

SSDL SURVEY FORM

INTERNATIONAL ATOMIC ENERGY AGENCY

Dosimetry and Medical Radiation Physics Section

Division of Human Health

THE IAEA/WHO NETWORK OF SSDLs

SURVEY ON CALIBRATION CAPABILITY FOR DIAGNOSTIC RADIOLOGY RADIATION DETECTORS

Could you please assist the IAEA by filling out the following survey *. The gathered data will be used to help plan the implementation of diagnostic radiology calibration standards throughout the Member States using TRS No. 457.

SSDL identification (optional): Country: _____ City: _____ Date: _____

1. Do you currently have the facility to calibrate diagnostic radiology X ray dosimeters? **YES☐/NO☐**
   If NO, do you plan to have the above in the next 3 years? **YES☐/NO☐**

2. Do you follow TRS No. 457 protocol (http://www-pub.iaea.org/MTCD/publications/PDF/TRS No. 457_web.pdf) or another?
   ☐ TRS No. 457 **☐ National protocol** **☐ Other**
   (if not TRS No. 457, please indicate: _____)

3. What beam qualities do you offer?

   ◦ *IEC beam qualities* **YES☐/NO☐** Please specify: _____

   RQR _____(i.e. 2-10) **RQA _____** **RQR-M _____**

   RQA-M _____ **RQT _____** **Other _____**

   ◦ *Non IEC beam qualities* **YES☐/NO☐** Please specify: _____

Target	Tube voltage (kV)	Filter (mm)	HVL (mm Al)
Example: Rh	25	0.025 Rh	0.351

4. For a typical detector calibration how many calibration points are performed? _____

5. How many diagnostic detectors did you calibrate in 2007? _____

6. Additional comments?_____
_____
_____

* Please send the form (e-mail, fax or mail) to:
Dosimetry and Medical Radiation Physics Section
Division of Human Health, International Atomic Energy Agency
Wagramer Strasse 5, PO Box 100, A-1400
Vienna, AustriaTel: 43 1 2600 21653
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E mail: dosimetry@iaea.org

Annex II

RQR BEAM QUALITIES ESTABLISHED BY THE PARTICIPATING SSDLs

RQR2	kV	Added filtration	1st HVL	2nd HVL	h	1stHVL meas/IEC	K_{HVL}/K_0	h meas/IEC	h differ
IEC	40		1.42		0.81				
Brazil	40.0	2.36	1.43	1.81	0.79	1.007	0.500	0.978	–0.02
Cuba	40	2.50	1.37	1.74	0.79	0.96	0.490	0.98	–0.02
Czech Republic	40	2.55	1.40	1.71	0.82	0.986	0.496	1.012	0.01
Finland	40	2.59	1.42	1.74	0.82	0.99	0.500	1.01	0.01
Greece	40	2.65	1.44	1.72	0.836	1.011	0.504	1.032	0.03
Thailand	40	2.40	1.43	1.79	0.79	1.007	0.501	0.975	–0.02
Vietnam	40	2.51	1.45	1.77	0.82	1.021	0.507	1.012	0.01
IAEA	40	2.424	1.411	1.76	0.802				

RQR3	KV	Added filtration	1st HVL	2nd HVL	h	1stHVL meas/IEC	K_{HVL}/K_0	h meas/IEC	h differ
IEC	50		1.78		0.76				
Brazil	50.0	2.41	1.80	2.47	0.73	1.013	0.500	0.963	–0.03
Cuba	50	2.5	1.75	2.25	0.78	0.98	0.494	1.03	0.02
Czech Republic	50	2.55	1.79	2.32	0.77	1.006	0.501	1.013	0.01
Finland	50	2.59	1.77	2.30	0.76	0.993	0.498	1.01	0.00
Greece	50	2.22	1.81	2.27	0.796	1.015	0.505	1.048	0.04
Thailand	50	2.40	1.80	2.39	0.75	1.011	0.504	0.987	–0.01
Vietnam	50	2.51	1.80	2.40	0.75	1.011	0.508	0.986	–0.01
IAEA	50	2.424	1.765	2.328	0.758				

RQR4	kV	Added filtration	1st HVL	2nd HVL	h	1stHVL meas/IEC	K_{HVL}/K_0	h meas/IEC	h differ
IEC	60		2.19		0.74				
Brazil	60.1	2.60	2.20	3.10	0.71	1.005	0.500	0.959	–0.03
Cuba	60	2.6	2.10	2.90	0.72	0.96	0.487	0.97	–0.02
Czech Republic	60	2.755	2.21	2.94	0.75	1.009	0.503	1.014	0.01
Finland	60	2.78	2.16	3.00	0.74	0.985	0.496	1.00	0.00
Greece	60	2.60	2.20	3.02	0.730	1.005	0.502	0.987	–0.01
Thailand	60	2.60	2.21	3.034	0.73	1.009	0.502	0.986	–0.01
Vietnam	60	2.51	2.19	2.96	0.74	1.000	0.496	1.000	0.00
IAEA	60	2.670	2.162	3.610	0.725				

RQR5	kV	Added filtration	1st HVL	2nd HVL	h	1stHVL meas/IEC	K_{HVL}/K_0	h meas/IEC	h differ
IEC	70		2.58		0.71				
Brazil	70.0	2.66	2.54	3.63	0.70	0.983	0.501	0.983	–0.01
Cuba	70	3.0	2.60	3.62	0.72	1.01	0.503	1.01	0.01
Czech Republic	70	2.835	2.58	3.65	0.71	1.000	0.500	1.000	0.00
Finland	70	3.10	2.62	3.68	0.70	1.014	0.504	0.99	–0.01
Greece	70	2.70	2.58	3.63	0.711	1.000	0.500	1.002	0.00
Thailand	70	2.85	2.63	3.72	0.70	1.019	0.506	0.986	–0.01
Vietnam	70	2.51	2.60	3.66	0.71	1.008	0.487	1.000	0.00
IAEA	70	2.851	2.553	3.610	0.707				

RQR6	kV	Added filtration	1st HVL	2nd HVL	h	1stHVL meas/IEC	K_{HVL}/K_0	h meas/IEC	h differ
IEC	80		3.01		0.69				
Brazil	80.1	2.81	2.94	4.47	0.66	0.978	0.501	0.956	–0.03
Cuba	80	3.0	2.92	4.22	0.69	0.97	0.492	1.00	0.00
Czech Republic	80	2.895	2.99	4.34	0.68	0.993	0.497	0.986	–0.01
Finland	80	3.02	2.98	4.30	0.70	0.991	0.497	1.01	0.01
Greece	80	2.85	2.96	4.40	0.672	0.982	0.497	0.973	–0.02
Thailand	80	2.90	3.016	4.401	0.69	1.002	0.502	1.0	0.0
Vietnam	80	2.51	2.96	4.35	0.68	0.983	0.487	0.985	0.02
IAEA	80	3.132	3.020	4.369	0.680				

RQR7	kV	Added filtration	1st HVL	2nd HVL	h	1stHVL meas/IEC	K_{HVL}/K_0	h meas/IEC	h differ
IEC	90		3.48		0.68				
Brazil	90.1	3.01	3.49	5.31	0.66	1.003	0.499	0.968	–0.02
Cuba	90	3.5	3.55	5.12	0.69	1.02	0.506	1.01	0.01
Czech Republic	90	3.235	3.47	5.01	0.69	0.997	0.500	1.015	0.01
Finland	90	3.25	3.48	5.09	0.68	1.001	0.500	1.01	0.00
Greece	90	3.00	3.49	5.12	0.682	1.004	0.501	1.003	0.00
Thailand	90	3.0	3.47	5.14	0.67	0.997	0.498	0.985	–0.01
Vietnam	90	3.00	3.51	5.16	0.68	1.009	0.498	1.000	0.00
IAEA	90	3.355	3.516	5.170	0.680	1.100	n/a	1.000	0.00

RQR8	kV	Added filtration	1st HVL	2nd HVL	h	1stHVL meas/IEC	K_{HVL}/K_0	h meas/IEC	h differ
IEC	100		3.97		0.68				
Brazil	100.1	3.32	4.04	6.17	0.65	1.017	0.499	0.962	–0.03
Cuba	100	3.5	3.91	5.82	0.67	0.98	0.496	0.99	–0.01
Czech Republic	100	3.54	3.99	5.80	0.69	1.005	0.501	1.015	0.01
Finland	100	3.37	3.94	5.81	0.68	0.993	0.498	1.00	0.00
Greece	100	3.20	3.98	5.87	0.678	1.003	0.502	0.997	0.00
Thailand	100	3.25	4.00	5.924	0.68	1.008	0.505	1.0	0.0
Vietnam	100	3.00	4.00	5.88	0.68	1.008	0.489	1.000	0.00
IAEA	100	3.475	3.962	5.892	0.673	0.998	n/a	0.990	–0.01

RQR9	kV	Added filtration	1st HVL	2nd HVL	h	1stHVL meas/IEC	K_{HVL}/K_0	h meas/IEC	h differ
IEC	120		5.00		0.68				
Brazil	120.1	3.61	5.09	7.45	0.68	1.018	0.499	1.005	0.00
Cuba	120	4.0	4.97	7.26	0.68	0.99	0.499	1.00	0.00
Czech Republic	120	3.95	5.01	7.16	0.70	1.002	0.501	1.029	0.02
Finland	120	3.82	5.03	7.28	0.69	1.005	0.501	1.01	0.01
Greece	120	3.55	4.98	7.40	0.673	0.996	0.499	0.989	–0.01
Thailand	120	3.67	5.03	7.49	0.67	1.006	0.502	0.985	–0.01
Vietnam	120	3.30	5.00	7.35	0.68	1.000	0.489	1.000	0.00
IAEA	120	3.968	5.019	7.370	0.681	1.004	n/a	1.001	0.00

RQR10	kV	Added filtration	1st HVL	2nd HVL	h	1stHVL meas/IEC	K_{HVL}/K_0	h meas/IEC	h differ
IEC	150		6.57		0.72				
Brazil	150.1	4.11	6.56	9.44	0.70	0.999	0.500	0.966	–0.02
Cuba	150	5.0	6.73	9.17	0.73	1.02	0.507	1.01	0.01
Czech Republic	150	4.64	6.53	8.99	0.73	0.994	0.498	1.014	0.01
Finland	150	4.45	6.60	9.16	0.72	1.005	0.501	1.00	0.00
Greece	150	4.25	6.61	9.49	0.696	1.005	0.502	0.967	–0.02
Thailand	150	4.20	6.60	9.37	0.70	1.004	0.502	0.972	–0.02
IAEA	150	4.793	6.625	9.309	0.712	1.008	n/a	0.989	–0.01

Annex III

RQT BEAM QUALITIES ESTABLISHED BY THE PARTICIPATING SSDLs

RQT8	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	100	RQR8 + 0.2 mm Cu	6.9		
Cuba	100	3.5 mm Al + 0.2 mm Cu	6.80	0.99	0.495
Czech Republic	100	0.21 mm Cu	6.9	1.003	0.501
Finland	100	3.38 mm Al + 0.2 mm Cu	7.17	1.039	
Greece	100	3.20 mm Al + 0.20 mm Cu	7.05	1.021	0.506
Thailand	100	RQR8 + 0.20 mm Cu	7.068	1.024	0.508
Vietnam	100	3.0 mm Al + 0.20 mm Cu	6.89	0.999	0.506

RQT9	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	120	RQR9 + 0.25mmCu	8.4		
Cuba	120	4.0 mm Al + 0.25 mm Cu	8.4	1.00	0.499
Czech Republic	120	0.255 mm Cu	8.4	1.004	0.501
Finland	120	3.79 mm Al + 0.25 mm Cu	8.68	1.033	
Greece	120	3.55 mm Al + 0.25 mm Cu	8.41	1.001	0.500
Thailand	120	RQR9 + 0.25 mm Cu	8.611	1.025	0.508
Vietnam	120	3.3 mm Al + 0.25 mm Cu	8.38	0.998	0.507

RQT10	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	150	RQR10 + 0.3mmCu	10.1		
Cuba	150	5.0 mm Al + 0.3 mm Cu	10.11	1.00	0.500
Czech Republic	150	0.315 mm Cu	10.1	1.001	0.500
Finland	150	4.38 mm Al + 0.3 mm Cu	10.43	1.033	
Greece	150	4.25 mm Al + 0.3 mm Cu	10.39	1.028	0.509
Thailand	150	RQR10 + 0.3 mm Cu	10.477	1.037	0.513

Annex IV

RQA BEAM QUALITIES ESTABLISHED BY THE PARTICIPATING SSDLs

RQA2	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	40		2.2		
Czech Republic	40	4.13	2.2	1.000	0.503
IAEA	40	6.482	2.24	1.018	
Thailand	40	6.40	2.26	1.027	0.51
Vietnam	40	2.51 + 4	2.19	0.995	0.508

RQA3	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	50		3.8		
Czech Republic	50	10	3.7	0.974	0.495
IAEA	50	12.424	3.79	0.997	
Thailand	50	12.40	3.88	1.021	0.507
Vietnam	50	2.51 + 10	3.8	1.000	0.506

RQA4	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	60		5.4		
Czech Republic	60	16	5.3	0.981	0.493
IAEA	60	18.670	5.40	1.000	
Thailand	60	18.60	5.43	1.006	0.508
Vietnam	60	2.51 +16	5.38	0.996	0.504

RQA5	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	70		6.8		
Czech Republic	70	21	6.7	0.985	0.493
IAEA	70	23.851	6.82	1.003	
Thailand	70	22.22	7.0	1.029	0.51
Vietnam	70	2.51 +21	6.8	1.000	0.501

RQA6	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	80		8.2		
Czech Republic	80	26	8.0	0.976	0.490
IAEA	80	29.132	8.17	0.996	
Thailand	80	28.90	8.39	1.023	0.508
Vietnam	80	2.51 + 26	8.19	0.999	0.502

RQA7	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	90		9.2		
Czech Republic	90	30	9.0	0.978	0.494
IAEA	90	33.355	9.29	1.010	
Thailand	90	33.0	9.49	1.032	0.51
Vietnam	90	3.0 + 30	9.19	0.999	0.506

RQA8	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	100		10.1		
Czech Republic	100	34	10.0	0.990	0.497
IAEA	100	37.475	10.22	1.012	
Thailand	100	35.55	10.61	1.050	0.507
Vietnam	100	3.0 + 34	10.1	1.000	0.504

RQA9	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	120		11.6		
Czech Republic	120	40	11.3	0.974	0.492
IAEA	120	43.968	11.65	1.004	
Thailand	120	40.0	11.89	1.025	0.509
Vietnam	120	3.3 + 40	11.6	1.000	0.5–7

RQA10	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	150		13.1		
Czech Republic	150	45	13.1	0.985	0.495
IAEA	150	49.76	13.31	1.016	
Thailand	150	45.0	13.66	1.043	0.509

Annex V

RQR-M BEAM QUALITIES ESTABLISHED BY THE PARTICIPATING SSDLs

RQR-M1	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	25		0.28		
Greece	25	0	0.285	1.018	0.507

RQR-M2	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	28		0.31		
Greece	28	0	0.318	1.025	0.513

RQR-M3	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	30		0.33		
Greece	30	0	0.338	1.024	0.510

RQR-M4	kV	Added filtration	1st HVL	1stHVL meas/IEC	K_{HVL}/K_0
IEC	35		0.36		
Greece	35	0	0.378	1.050	0.515

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