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Declaration

This thesis is my own work and has not been examined or submitted for examination in any other university.

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Finally, my warmest thanks and appreciation to my wife, Penina and my daughter, Trish for their prayers, encouragement and patience.

Dedication

I dedicate this work to my father, Gabriel Masinza Chiloli and mum, Philomena Malangalanga. May this work fulfil their incessant prayers and original aspirations expressed early in my lifetime.

Abstract

Kenya has about 3,000 X-ray and 30 Computed Tomography (CT) units. Most of these units (80%) are not calibrated. The remainder are calibrated by use of transfer standards whose calibration status is unknown, hence a broken international traceability chain. Many hospitals and clinics use different radiation qualities and standards, some of which may be unsuitable due to non-calibration after many years of use or replacement of major parts. Where calibrations were actually performed, the test equipment are either not calibrated or sometimes sent out of the country for the service at a considerable cost. For this reason, there is need to establish this capability in the country to bridge this gap.

The objective of the study was to develop reference X-ray radiation beam qualities (RQR) at the Secondary Standards Dosimetry Laboratory (SSDL), KEBS. RQR represent the beam that is incident on a patient when undergoing diagnostic medical examinations and provides diagnostic dosimetry traceability that is presently lacking in Kenya.

Air kerma rates were determined using a one litre reference free in air ionization chamber calibrated at the Primary Standards Dosimetry Laboratory at PTB in Germany. By determination of the Half Value Layers (HVL), narrow series beam qualities meeting the ISO 4037 part 1 criteria were established using high purity aluminium filters placed in the beam. Various RQR were then established using a 30 cm³ Xradin A4 chamber to determine air kerma, HVL and homogeneity coefficient for each beam quality setting.

The results obtained compared to the IEC 61267 criteria and were evaluated by use of statistical mean and percentage standard deviations of the measurands, interpolation, ICRP developed formulae and conversion factors taking into account the effect of temperature and pressure to obtain the corrected values of charge produced in ionization chambers.

Compared to the ISO 4037 criteria, the interpolated HVL values were found to be in agreement within the $\pm 5\%$ tolerance. The developed reference radiation beams were found to be within the $\pm 3\%$ allowable limits. RQR beam parameters were adjusted by addition of filtration and tested to comply with the IEC 61627 standard criteria. Sources of measurement uncertainties (resolution, calibration, position from tube focus and standard deviation) were identified and estimated. The main source of uncertainty (0.58 %) during the calibration process was found to be due to the ionization chamber positioning set-up.

The established narrow series (N-series) were found to comply with the ISO 4037 requirements within $\pm 4\%$. Subsequent RQR beam parameters established were found to be in agreement with the standard values in IEC 61267 within $\pm 1\%$, within the permissible tolerance limits of $\pm 3\%$ for both the homogeneity coefficient and first HVL. All reference radiations were reproduced with success within the IEC tolerance limits. Therefore the SSDL at KEBS can calibrate transfer standards and provide an unbroken chain of traceability.

CONTENTS

Declaration	i
Acknowledgments	ii
Dedication	iii
Abstract	iv
List of Abbreviations, Symbols and Acronyms	ix
List of Tables	x
List of Figures	xi
CHAPTER ONE	1
INTRODUCTION	1
1.1 Ionizing Radiation Metrology	1
1.2 Utility of X-rays in Medicine	2
1.3 Quality Assurance and Dose Management	3
1.4 The Secondary Standards Dosimetry Laboratory at KEBS	4
1.6 Statement of the Problem	8
1.7 Objectives of the Study	10
1.7.1 Main Objective	10
1.7.2 Specific Objectives	10
1.8 Justification and Significance of the Study	10
CHAPTER TWO	12
LITERATURE REVIEW	12
2.0 Introduction	12
2.1 Optimization of X-ray Imaging Systems	12
2.2 Patient Exposures in Radiology	15
2.3 Bone and Soft Tissue Interactions with Photons	20
2.4 Effective Doses in Radiology	22
CHAPTER THREE	25
THEORETICAL FRAMEWORK	25
3.0 Introduction	25
3.1 X-ray Beam Characteristics	25

3.1.1	X-ray Beam Quantity	25
3.1.2.	X-ray Beam Quality	28
3.2	Interaction of X-rays with Matter	29
3.2.1	Classical Scattering	29
3.2.2	Compton Scattering	29
3.2.3	Photoelectric Effect	30
3.2.4	Pair Production	31
3.2.5	Photodisintegration	32
3.3	Radiation Quantities and Units	32
3.3.1	Quantities that Describe Radiation Beam	32
3.3.1.1	Fluence	32
3.3.1.2	Fluence Rate (Flux)	32
3.3.1.3	Energy Fluence	33
3.3.1.4	Energy Fluence Rate	33
3.3.2	Quantities that Describe Deposited Energy	34
3.3.2.1	Kerma	34
3.3.2.2	Absorbed Dose	35
3.3.3	Exposure	35
3.3.4	Absorbed Dose and Kerma Relation	36
3.3.5	Kerma in Air, Dose and Exposure	37
CHAPTER FOUR		39
MATERIALS AND METHODS		39
4.0	Introduction	39
4.1	Conditions for Air Kerma Reference Rate Measurement	40
4.2	Set Up of X-ray System at KEBS	40
4.3	Verification of Beam Qualities	42
4.4	Inverse-Square Law and X-ray Tube Focus Positioning	43
4.5	Beam Profile, Symmetry and Flatness	45
4.6	Inherent Tube Filtration	46
4.7	Establishment of Air Kerma Reference and Dose Rates for ISO Beam Qualities	46
4.8	Determination of Half Value Layer (HVL)	50

4.8.1	First and Second HVL and Homogeneity Coefficient	50
4.8.2	HVL Measurement Set-Up	51
4.8.3	HVL Measurement Procedure	53
4.9	Establishment of RQR Beam Qualities for the Range 40 kV to 150 kV	54
4.10	Uncertainty Analysis	56
CHAPTER FIVE		57
RESULTS AND DISCUSSION		57
5.0	Introduction	57
5.1	Determination and Verification of Beam Qualities	57
5.2	Inverse-Square Law and X-ray Tube Focus Positioning	59
5.3	Beam Profile, Symmetry and Flatness	60
5.4	Air Kerma Reference and Dose Rates for ISO Beam Qualities	61
5.4.1	Conversion Coefficients	62
5.4.2	Correlations Between the Air Kerma Rates (K_{air}) and Monitor Chamber	66
5.5	Evaluation of Uncertainties	70
CHAPTER SIX		72
CONCLUSIONS AND RECOMMENDATIONS		72
6.0	Conclusions	72
6.1	Recommendation for Further Work	75
REFERENCES		76
APPENDICES		80
App I:	PTB Calibration Certificate	80
App II:	ISO 4037 Narrow Beam Qualities Established at KEBS SSDL	86
App III:	X-Ray Set Up	93

List of Abbreviations, Symbols and Acronyms

AAPM	:	American Association of Physicists in Medicine
BIPM	:	Bureau International des Poids et Mesures
FACs	:	Free-Air Ionization Chambers
FCD	:	Focus to Chamber Distance
$H^*(10)$:	Ambient Dose Rate (Operational Quantity)
HVL	:	Half Value Layer
IAEA	:	International Atomic Energy Agency
IEC	:	International Electrotechnical Commission
IPEM	:	Institute of Physics and Engineering in Medicine
ISO	:	International Organization for Standardization
K_{air}	:	Air Kerma
K_{air}	:	Air Kerma Rate
K_{PT}	:	Correction Factor for Temperature and Pressure
KEBS	:	Kenya Bureau of Standards
SCD	:	Source to Chamber Distance
PSDL	:	Primary Standards Dosimetry Laboratory
PTB	:	Physikalisch-Technische Bundesanstalt
Q	:	Electronic Charge
Q_{corr}	:	Charge (corrected for temperature and pressure)
RQR	:	Reference X-Ray Radiation
SSDL	:	Secondary Standards Dosimetry Laboratory
STP	:	Standard Temperature and Pressure
TRS	:	Technical Report Series
UNSCEAR	:	United Nations Scientific Committee on the Effects of Atomic Radiation
UoN	:	University of Nairobi
WHO	:	World Health Organization

List of Tables

Table 4.1:	Characterization of radiation qualities (RQR)	40
Table 5.1:	Modifications made in X-ray qualities	58
Table 5.2:	The final ISO 4037 narrow spectrum series established at KEBS SSDL	58
Table 5.3:	X-ray focus positioning	59
Table 5.4:	Conversion coefficients for the ISO Narrow Beam Qualities	63
Table 5.5:	Analysis of beam quality characteristics for the ISO 4037 narrow series	63
Table 5.6:	Reference RQR beams established at KEBS SSDL	68
Table 5.7:	Comparison of first half value layer results from selected countries	69
Table 5.8:	Parameters to characterize the uncertainty in the Xradin A4 ionization chamber calibration procedure in the RQR reference radiation beam establishment	71
Table AII.1:	The N40 Beam Quality Characteristics	86
Table AII.2:	The N60 Beam Quality Characteristics	87
Table AII.3:	The N80 Beam Quality Characteristics	88
Table AII.4:	The N100 Beam Quality Characteristics	89
Table AII.5:	The N120 Beam Quality Characteristics	90
Table AII.6:	The N150 Beam Quality Characteristics	91
Table AII.7:	The N200 Beam Quality Characteristics	92

List of Figures

Figure 2.1:	Effective patient doses calculated from individual scan parameters with the 16 effective radius correction to women and men.	16
Figure 2.2:	Variation in mass attenuation coefficients for photoelectric absorption and Compton scattering in bone and soft tissue with photon energy.	20
Figure 3.1:	Effect of tube current on X-ray spectrum at constant kV.	26
Figure 3.2:	Effect of filtration on X-ray spectrum.	28
Figure 3.3:	Schematic diagram of classical scattering.	29
Figure 3.4:	Compton scattering of an incident photon of energy $h\nu$ and momentum p to energy $h\nu'$ and momentum p' .	30
Figure 3.5:	Schematic diagram of photoelectric effect	31
Figure 3.6:	Schematic diagram of pair production.	31
Figure 4.1:	KEBS SSDL X-ray system set up.	41
Figure 4.2:	Illustration of the 'mis-positioning' of the tube focus in relation to the wall mounted ruler.	45
Figure 4.3:	An illustration of first and second HVL.	51
Figure 4.4:	HVL measurement set up at SSDL, KEBS.	52
Figure 4.5:	Attenuation curve for the beam RQR 6 expressed as the ratio of $K(d)$ behind a filtration of thickness, d , to the air kerma, K_0 , of the un-attenuated beam.	55
Figure 5.1:	Plot of charge against Source to Chamber Distance.	60
Figure 5.2:	X-ray Beam Profile at SCD of 100 cm and aperture diameter of 5 cm.	61
Figure 5.3:	Monitor chamber (PTW UNIDOS E) stability during air kerma measurements for ISO 4037 qualities.	66
Figure 5.4:	Correlations between air kerma rates and monitor chamber for N40 to N200.	67
Figure A3.1:	Pictures of X-ray setup.	93
(a)	A picture of the X-ray equipment layout at KEBS.	93
(b)	Visual display for temperature monitoring.	93
(c)	The PTW UNIDOS and UNIDOS E electrometers for charge display.	93
(d)	The MP1 control console for settings of kilovoltage (kV) and current (mA).	93

CHAPTER ONE

INTRODUCTION

1.0 Background of the Study

The X-ray beam parameters required to describe a beam quality are: inherent tube filtration, beam uniformity; beam field size, 1st and 2nd half value layers (whose ratio is referred to as the homogeneity coefficient), energy spectrum and peak voltage. This chapter provides background information on the X-ray beam qualities and covers some aspects of need for radiation metrology and the use of X-rays with emphasis on diagnostic radiology, states the problem in the study, enumerates the study objectives and outlines the justification and significance of the study.

1.1 Ionizing Radiation Metrology

Ionizing radiation metrology is the basis to achieve reliability of dose measurements as applied in individual dosimetry for workers occupationally exposed to radiation, in patients submitted to radio-diagnostics or radiotherapy and in environmental monitoring. The aim of reliable measurements is to establish or assert the radiological protection procedures in order to avoid or minimize the harmful biological effects of ionizing radiations.

The use of reliable radiation detectors is a requirement to get a high level radiation metrology. It therefore means that detectors must be properly calibrated and must also comply with the performance requirements that are set by national and international standards. There are many international standards that establish the characteristics to guarantee that detectors are adequate to be used for specific purposes (ISQ 4037-1, 1996).

X-ray dosimetry is expected to be done with radiation dosimeters that were type tested and calibrated in X-ray representative beams. X-ray reference radiations are defined by parameters such as high voltage peak of the equipment, half value layer (HVL), homogeneity coefficient (HC), and energy spectra, among others. Metrology laboratories maintain a metrological coherence among their X-ray beams through the adoption of the internationally established reference radiations. The International Electro technical Commission (IEC), International Organization for Standardization (ISO) and American Association of Physics in Medicine (AAPM) have published standards for radio-diagnostic, radioprotection and radiotherapy areas, respectively. These include the ISO 4037 series, the IEC 61627 standards and the Report no. 74 on quality control in diagnostic radiology (Oliveira *et al.*, 2007).

For diagnostic dosimetry, the IEC has established standards for forty (40) reference beams: nine (9) radiation qualities for conventional radio-diagnostic (RQR), nine (9) aluminium attenuated beam qualities to simulate the presence of a patient (RQA), 3 copper attenuated beam qualities (RQC), three (3) computed tomography radiation qualities and sixteen (16) mammography qualities.

1.2 Utility of X-rays in Medicine

The discovery of X-rays in 1895 by W.C. Röntgen has enabled the display of human internal anatomical structures and revolutionized the field of medicine. Since then, the use of X-rays has contributed to the diagnosis and treatment of many diseases thereby helping to improve the health of people all over the world. Medical imaging systems have developed from simple units used to image specific anatomical sites to systems that can visualize the whole body, obtain information concerning functional aspects of specific organs and even yield information about organ and tissue chemistry. Nowadays, medical imaging equipment is

taking advantage of modern digital technology and has become a symbol of 'high technology' (IAEA, 2007).

1.3 Quality Assurance and Dose Management

Radiologists constantly face the dilemma of trying to minimize patient exposure whenever possible, while still using exposures that are high enough to produce images of good enough quality as to be able to provide a proper diagnosis. Quality assurance provides a framework for achieving this goal. The basic strategy for quality assurance in diagnostic radiology was formulated by WHO and involves various activities, including managerial and technical activities (WHO, 1982). The International Basic Safety Standards for Protection against Ionizing Radiation and for the Safety of Radiation Sources (IBSS) provide requirements to establish a quality assurance programme for medical exposures (IAEA, 1996).

It is necessary that a quality assurance programme in diagnostic and interventional radiology include image quality assessment, film rejection analysis, patient dose evaluation, measurements of physical parameters of the radiation generators, etc. Various quality control tests are thus needed to ensure that the radiology machines are working properly (IAEA, 2007). The IBSS also requires that guidance levels be established to provide guidance on what is achievable with current good practice. The levels should be derived from the data provided from wide scale surveys. In aspects geared towards health and protection of individuals against the dangers of ionizing radiation in relation to medical exposures, it is necessary to conduct extensive dose measurements and establish diagnostic reference levels comparable to the international guidance levels (IAEA, 1996).

1.4 The Secondary Standards Dosimetry Laboratory at KEBS

The SSDL at KEBS was established as part of the IAEA/WHO network in June 2007. The laboratory has a 20 Ci Caesium-137 gamma and a 40-250 kV X-ray protection level calibration systems. These systems can calibrate radiation detection equipment (survey meters, ionization chambers, alarms (beepers) and dosimeters) used for radiation protection. In the year 2009, the laboratory completed a bilateral gamma and X-ray beam inter-comparison study with the primary laboratory at the National Institute of Science and Technology (NIST) of the USA, where calibration factors of the chambers for the inter-comparison were found to be within 0.9 % agreement (O'Brien *et al.*, 2010).

It is expected that the development of reference radiation beam qualities (RQR) will increase the scope of the services provided by the laboratory to include performance assessment of clinical X-ray systems and provide the much needed traceability of measurements. This will be an important contribution towards reduction of patient and personnel doses in diagnostic radiology arising from equipment whose performance characteristics are undesirable or are unknown following major repairs and parts replacement.

1.5 The Narrow Series (N-series) X-ray Beam Qualities

Radiation quality is a measure of the penetrative power of an X-ray beam, usually characterized by a statement of the tube potential and the HVL. The narrow beam qualities (N-series) for the X-ray requirements have been set by the International Organization for Standardization through the ISO 4037 series of standards. Once established for an X-ray equipment in a calibration laboratory, air kerma reference rates need to be determined from time to time because of their dependence on the tube current (mA) and voltage (kV).

The full characterization of X-ray beams is based upon measurement of the photon fluence spectrum. However, due to unavailability of a X-ray spectrum analyser and in practice, an X-ray beam can be characterized by measurement of the first and second HVL in order to obtain a qualitative description of the diagnostic X-ray field. The X-ray tube voltage should be measured in terms of the practical peak voltage, preferably with an invasive device or, alternatively, with a non-invasive one. This is because the readings obtained in this manner would reflect the actual kilovoltage (kV) output of the X-ray tube. HVL measurements are usually performed with ionization chambers (IAEA, 2007).

The quality of a filtered X radiation is characterized by the mean energy, E , of a beam, expressed in kilo-electron volts (keV); resolution, expressed in percent; HVL (air kerma), expressed in millimeters of Al or Cu and homogeneity coefficient, h . In practice, the quality of the radiation obtained depends primarily on the high-voltage across the X-ray tube, the thickness and nature of the total filtration, and the properties of the target (IEC, 1994).

The International Committee on Radiological Units (ICRU) recommends that the characterization of radiation quality of X-ray beams used for medical imaging by the utilization of a combination of various parameters, including first and second half-value layer, HVL_1 and HVL_2 , the ratio of HVL_1 and HVL_2 , (homogeneity coefficient), the tube voltage (kV), and the total filtration. In most cases a combination of three of these parameters will suffice for characterization. The radiation intensity is also an important characteristic of an X-ray tube, including filtration (ICRU, 2005).

In order to ensure the production of the reference radiation in conformance with the given specifications in ISO 4037 and IEC 61267 international standards, an X-ray installation has to comply with certain conditions. The tube target must be made of tungsten and the inclination

angle set at 45°. Additionally, the HVL measurements must be performed using filters of aluminium or copper of high purity of 99.999% (IEC, 1994).

Dosimetry is an area of increasing importance in diagnostic radiology. There is a realization amongst health professionals that the radiation dose received by patients from modern X-ray examinations and procedures can be at a level of significance for the induction of cancer across a population, and in some unfortunate instances, in the acute damage to particular body organs such as skin and eyes (UNSCEAR, 2008).

The fundamental safety objective, as stated in the Fundamental Safety Principles, is to protect people and the environment from harmful effects of ionizing radiation. This objective has to be achieved without unduly limiting the operation of facilities or the conduct of activities that give rise to radiation risks. Therefore, the system of protection and safety aims to assess, manage and control exposures to ionizing radiation so that radiation risks and health effects are reduced to the extent reasonably achievable. Thus protection has to be equated to both equipment parameters and human operational factors (IAEA, 1996).

The formulation and measurement procedures for diagnostic radiology dosimetry have recently been standardized through an international code of practice which describes the methodologies necessary to address the diverging imaging modalities used in diagnostic radiology. Common to all dosimetry methodologies is the measurement of the air kerma from the X-ray device under defined conditions. To ensure the accuracy of the dosimetric determination, such measurements need to be made with appropriate instrumentation that has a calibration that is traceable to a standards laboratory.

Dosimetric methods are used in radiology departments for determination of patient dose levels to allow examinations to be optimized and to assist in decisions on the justification of

examination choices. Patient dosimetry is important for special cases such as for X-ray examinations of children and pregnant patients. It is also a key component of the quality control of X-ray equipment and procedures (Meghziene *et al.*, 2010).

Ionization chambers are the most frequently used dosimetry systems in diagnostic radiology. They have various geometries depending on the application. The two main types of ionization chamber geometries in use in SSDLs are plane parallel and spherical ionization chambers. An electrometer is used in conjunction with an ionization chamber to collect the charge generated in the gas (typically air) within the sensitive volume of the chamber.

For accurate dosimetry, the ionization chamber and electrometer must be sent for a calibration at a Primary Standards Dosimetry Laboratory (PSDL) in terms of air kerma in X-ray beams of known qualities. In the case of X-rays used in diagnostic radiology, the beam quality is specified in terms of the peak value of the high voltage applied across the X-ray tube (peak voltage), and the first HVL, expressed in millimeters of aluminium (mm Al).

A quality control program for X-ray equipment used for diagnostic procedures, necessary for providing adequate diagnostic information at acceptable levels of the patient and staff exposure, includes the measurement on a routine basis of various parameters that affect the performance characteristics of X-ray systems. One of these parameters is the beam quality, specified in terms of the HVL. Minimum HVL limits are recommended to ensure that the lowest energies in the unfiltered spectrum are removed (IEPM, 2005).

This study is aimed at contributing towards the protection of patients and radiation workers in diagnostic radiology in Kenya by establishing diagnostic reference radiation beam qualities (RQR) at the SSDL in KEBS. RQR are X-ray beams from diagnostic equipment and incident on a patient undergoing general radiography, fluoroscopy and dental examination and are

realized by means of a tungsten anode X-ray tube. They are important in providing laboratory calibration of transfer standards that are used by service providers to check the exposure characteristics of clinical diagnostic systems.

It is expected that, in this way, the calibration capability of the SSDL will be enhanced to offer standardized, quality assured calibration services to clinical diagnostic X-ray services providers to keep patient doses low, support quality assurance programmes through transfer calibrations and re-establish equipment exposure characteristics upon major repairs or parts replacement. The laboratory was established in the year 2007 and has been admitted to the IAEA/WHO network of SSDLs in support of the safe applications of ionizing radiation in industry and medicine (Hourdakis, 2007).

1.6 Statement of the Problem

At present, calibrations of diagnostic X-ray equipment in Kenya do not assure quality and are not traceable to national or international standards. Many hospitals and clinics use different radiation qualities and standards, some of which may lead to overexposure of patients. The rapid increase in dosimetry applications in diagnostic and interventional radiology calls for a calibration system of beam qualities RQR that can provide traceability of measurements which are transferable to clinical X-ray systems and result in optimized radiological protection of patients. Recently (2007) KEBS, in collaboration with the IAEA set up a SSDL facility that has the capacity to establish, maintain and provide the needed traceability in protection level, diagnostic and interventional radiology radiation equipment. However, only the protection level capability is set up, necessitating the verification and development of diagnostic calibration capabilities. Realization of that capacity requires the development, maintenance and transfer RQR beam qualities that are in conformance with ISO 4037 and

IEC 61267 standards and that demonstrate increased scope in the calibration of clinical X-ray systems and optimized radiological protection of patients.

There is, therefore, the need for the SSDL at KEBS to develop RQRs that conform to the ISO 4037 and IEC 61267 criteria so as provide the necessary traceability required and ensure that the diagnostic equipment used in Kenya do not injure patients in the course of diagnosis.

1.7 Objectives of the Study

1.7.1 Main Objective

The main objective of this study was to develop reference diagnostic X-ray radiation beam qualities (RQR) through beam parameter analysis in order to increase the scope of SSDL services at KEBS in calibration traceability of diagnostic X-ray systems used in hospitals.

1.7.2 Specific Objectives

- a) To measure and verify the Air Kerma rate (K_{air}) reference values (the ISO 4037 narrow series N40 to N200 radiation protection qualities) by determining the first and second half value layers (HVL) on the installed Hopewell X-ray system.
- b) To develop and establish, in accordance with IEC 61267, RQR for the diagnostic X-ray range 40 kV to 150 kV at the SSDL in KEBS.
- c) To determine, for each RQR quality, the first and second half value layer and the homogeneity coefficient and test and compare to the criteria set in IEC 61267.
- d) To document a method for establishing an X-ray calibration facility for both radiation protection and diagnostic radiology.

1.8 Justification and Significance of the Study

Radiation exposures resulting from medical radiological procedures constitute the largest part (above 90%) of the population exposure. Over half these doses emanate from exposures from artificial X-ray radiation. There is a need to control these doses through optimization of outputs of X-ray imaging systems. It is generally recognized that even a 10% reduction in patient dose is a worthwhile objective for optimization (IAEA, 2007).

The ultimate aim of patient dosimetry with respect to X-rays used in medical imaging is to determine dosimetric quantities for the establishment and use of guidance levels (diagnostic reference levels). It is essential, therefore, to standardize the procedures for the dose measurement in the diagnostic X-ray clinics and establish a reference system within Kenya that can address the component of equipment performance (ICRP, 1991).

Owing to the increased demand for dosimetry measurements in diagnostic and interventional radiology, it has become important to provide traceability of measurements in this field. At present, the manner in which calibrations of diagnostic X-ray equipment is performed in Kenya is not coordinated and the equipment are not traceable due to lack of such a link. The absence of a standardized approach to these measurements has led to the possibility that many diagnostic X-ray facilities use different radiation qualities and standards, some of which may be unsuitable. Quality control can only work satisfactorily if correct calibrations and measurements are made.

Kenya has over 3,000 diagnostic facilities spread in some 2000 medical centres. These equipment are from different manufacturers based in different countries. Despite this variation, X-ray exposures for purposes of diagnosis need to be averagely the same for the same examinations. Additionally, whenever major repairs and parts replacement (commonly the X-ray tube) occurs, it is important to re-establish equipment performance parameters before deploying it on patients.

CHAPTER TWO

LITERATURE REVIEW

2.0 Introduction

The international standards organization namely, ISO and IEC prepare and publish standards that specify criteria and requirements for the characterization of X-radiation beams used to test and calibrate X-radiation detectors (ISO 4037-1) and related electrical and electronic requirements (IEC 61627).

The IAEA Technical Reports Series 457 is an international code of practice for dosimetry in diagnostic radiology. It provides procedures for establishment of specific diagnostic radiology radiation qualities in order to calibrate instruments and to use these calibrated instruments to perform dosimetric procedures in clinical practices based on the application of ISO and IEC standards (IAEA, 2007).

This chapter discusses various studies that establish a basis for the need for traceability conducted elsewhere in an attempt to establish calibration capabilities based on international standards and to apply the requirements on existing clinical systems.

2.1 Optimization of X-ray Imaging Systems

Radiological imaging is a process by which the attenuation of an X-ray beam traversing a part of a human body is recorded in a medium for later medical interpretation of potential pathology or injury (radiography) or displayed in real-time on a monitor for functional assessment (diagnostic fluoroscopy) or intervention (interventional radiology) (IAEA, 2007).

X-ray units do not always produce the same quality X-rays, in terms of output air kerma, for a given kV. This may be due to incorrect calibration, age of apparatus, drift, waveform of output beam and other causes. Unless the beam quality is known, dose measurements and tests on radiographic recording systems may be invalid. It has been shown that a difference of 10 kV can affect the patient integral dose by 20-40 % (Behrman and Yasuda, 1998).

The IAEA/WHO Network, through SSDLs designated by the Member States, provides a direct linkage of national dosimetry standards to the international measurement system (SI: *Système International*). Through the proper calibration of field instruments by the SSDLs, these measurements are traceable to the PSDLs and the Bureau International des Poids et Mesures (BIPM). The Network has proven to be of value in improving national capabilities for instrument calibration and the level of awareness of the need for better accuracy and traceability (IAEA, 2007).

The key requirement in optimization for diagnostic medical exposures is to ensure that the quality of the image is adequate for diagnosis, but this must be balanced against the need to keep the dose as low as reasonably practicable (Martin, 2008). The choice of X-ray tube potential and so beam energy is a crucial component of optimization for reduction of patient doses. If it is too high, the image contrast may be too poor, to allow a diagnosis, but if it is too low, the radiation dose to the patient could be unnecessarily high. Experiences from many countries have established that different optimal tube potentials are used for different examinations. 120 and 140 kV are used for CT-examinations, 90-100 for radiography of thicker parts (e.g. lateral lumbar spine), 70-80 kV for abdomen and pelvis, 50-60 kV X-rays are used for thinner less attenuating regions, such as arms, feet and hands (Kiljunen, 2008). The choice of the range is dictated by the part of the body to be examined, the level of detail required by the radiologist and radiation protection considerations.

Since the majority of low energy X-ray photons will be heavily absorbed in superficial tissues and contribute little to the image, diagnostic X-ray beams are filtered using thin sheets of metal (2.5 – 4.0 mm of aluminium) to reduce the proportion of low energy photons. Filtration requirements may be specified in terms of the thickness of the filters, or HVL.

It has been observed that while digital techniques in radiology can reduce patient doses, they also have the potential to significantly increase them. This is because of the high number of patients requiring X-ray services, non-calibration of the systems and the various techniques employed. The X-ray digital technology is advancing rapidly and will soon affect hundreds of millions of patients. If careful attention is not paid to the radiation protection issues of digital radiology, medical exposure of patients will increase significantly without concurrent benefit (Vano and Fernandez, 2007).

A study to review diagnostic radiological equipment performance and resultant patient dose values was undertaken in Ghana in which equipment survey data was taken from 10 X-ray rooms across 7 individual hospitals in order to establish basic equipment performance levels against IPEM standards. The results established a baseline level of equipment acceptability and allowed entrance surface dose (ESD) values to be verified and calculated using proprietary software (Ward *et al.*, 2009).

A marked range of performance variation between the radiographic equipment was found. The tube and generator performance were acceptable in about 20% of the sample. However, the wide range of Entrance Surface Dose (ESD) values highlighted that a prioritized approach is needed to address areas of investigation and non-compliance, especially where values exceed basic safety standards. The results serve as an example of how standardization of technique and equipment calibration could contribute to optimization (Ward *et al.*, 2009).

In a study of the patient dosimetric optimization of various X-ray machines at the Radiology Department at Kenyatta National Hospital in Kenya, Korir et al, 2007, concluded that the entrance surface doses (ESD) for 189 randomly selected patients from three different X-ray rooms in most diagnostic procedures in adults exceeded the international limits for entrance surface dose reference levels. The study established that the entrance surface doses ranged from 0.33 mGy to 143 mGy for various exposure examinations against international guidance limits of 0.4 to 40 mGy. The X-ray units were tested for quality control performance. They failed with respect to kVp accuracy, focal spot size and total filtration tests. The major causes were attributed to positioning, underexposure and overexposure resulting from incorrect beam qualities (Korir *et al.*, 2007).

2.2 Patient Exposures in Radiology

Diagnostic radiology generally refers to the imaging and analysis of images obtained using X-rays. These include plain radiographs (e.g. chest X-rays), images of the breast (i.e. mammography), images obtained using fluoroscopy (e.g. with barium meal or barium enema) and images obtained by devices using computerized reconstruction techniques such as Computed Tomography (CT). In addition to their use for diagnosis, interventional or invasive procedures are also performed in hospitals (UNSCEAR, 2008).

Medical ionizing radiation sources provide by far the largest contribution to the population dose from artificial sources and most of this contribution comes from diagnostic X-rays (above 90%). One of the reasons for this situation is the large number of X-ray examinations performed every year. Approximately 3.6 billion diagnostic (3.1 medical and 0.5 dental) X-ray examinations are undertaken annually in the world (UNSCEAR, 2008).

Three quarters of all examinations occur in countries accorded health care level II, which account for only one quarter of the world population. Only 1% arises from the lower healthcare levels III and IV, which include one fifth of the world population. However, most growth in medical radiology is in developing countries where facilities and services are often lacking. Health care Level II refers to the availability of one physician for every 1,000 – 2,999 people, Level III to between 3,000 and 10,000 people per physician and level IV to less than one physician for every 10,000. This means that the developed world has by far more per capita distribution of X-ray facilities than those in emerging economies, which are now witnessing a steady growth in this sector (UNSCEAR, 2008).

The typical highest organ doses in projection radiography range from 1–20 mGy, but an increasing part of medical radiation exposure is due to X-ray procedures. Organ doses in this range are generally below the level required to produce deterministic effects. However, all X-ray procedures may give rise to stochastic effects. The likelihood and severity of skin injury depends on the dose delivered to a particular portion of skin (IAEA, 2007).

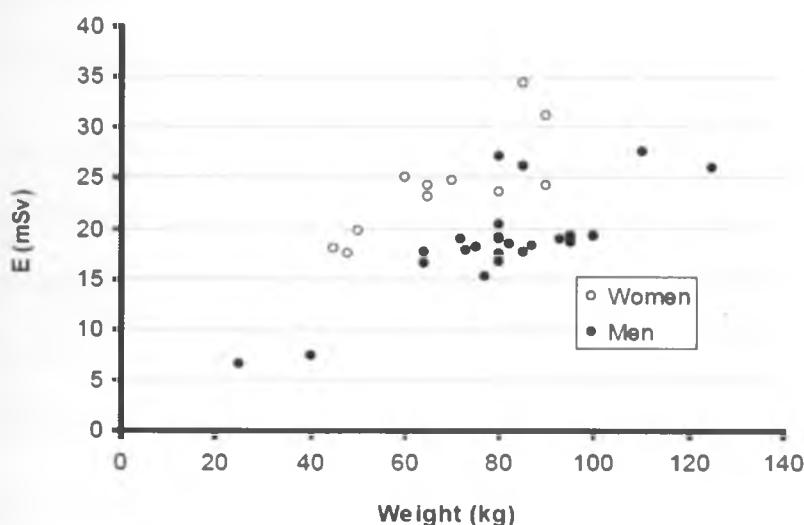


Figure. 2.1: Effective patient doses calculated from individual scan parameters with the effective radius correction to women and men (Source: Kijlunen, 2008).

Wide variations in patient dose for the same type of X-ray examination have been evident from various international dose surveys. Results have shown the variation of mean doses, from a factor of 3 for an anteroposterior lumbar spine to a factor of 23 for chest X-ray. The reasons for these dose variations are complex, but, in general low tube potential, high mAs and low filtration were identified as the root causes. In the dosimetry of medium energy X-rays, which are generated using X-ray tube voltages between 40 and 150kV, ionization chambers are routinely used to assess their performance and it is important that they are calibrated using standard X-ray fields (Johnston and Brennan, 2000).

The use of X-rays in emerging economies including Kenya is increasing year by year and is deemed to increase further, as a large part of the world that had little access to X-ray diagnosis endeavour to obtain them. Surveys in conventional radiology show differences in patient doses of up to a factor of 20, or even higher, among hospitals of the same country, for the same radiological examination and for average sized patients. This clearly demonstrates the need for reduction of unnecessary exposure (IAEA, 2007). This may be done by ensuring that measurements in radiation protection and radiation safety for the assessment of external and internal exposure are made by use of reliable measurement instruments and methods.

First steps on establishment of guidance levels in diagnostic radiology have been undertaken in a number of countries in 2006-7 and this has brought about the existence of a basis for patient exposure reduction. Exposure of paediatric patients and pregnant women are receiving special attention in the recent coordinated research project (CRP) of the IAEA because of higher implied risks of exposure (IAEA, 2007).

The types and number of interventional radiological procedures are rapidly increasing, as the benefits for the patient can be dramatic, in some countries their number doubles every 3-4 years (Amis *et al.*, 2007). In such procedures often, doses received are high, and in some

cases, radiation injuries can occur and have been reported in some cardiac and also in non-cardiac procedures such as angioplasty, radiofrequency ablation and stent implantations. Often, these procedures have to be repeated for the same patient. In addition, more and more physicians (non-radiologists and non-cardiologists) with no education in radiation protection are involved in these procedures. Digital radiology has the potential for reducing patient exposure, but ironically, has often led to substantially increased exposure due to lack of or inappropriate calibration (Amis *et al.*, 2007).

In diagnostic radiology, optimization of protection is essential to achieve the benefit (early detection, reduction in false positive and false negative diagnosis and with great impact on reducing mortality) with the lowest radiation exposure.

Basically there are two methods of patient dose reduction; those associated with the equipment and software and those involving the selection of imaging techniques by the operators. Any dose reduction on patients will also diminish the dose received by the occupationally exposed staff and the public. For the latter two, additional protection is provided by structural and ancillary shielding (Jensen and Lindborg, 1981).

In 2001, a study by the American Association of Physicists in Medicine (AAPM) demonstrated that patient dosimetry and evaluation of image quality are basic aspects of any quality control (QC) program in diagnostic radiology and traceability is assured when measurements on equipment are complying with national or international standards (Coffey *et al.*, 2001). Further, the study found that image quality must be adequate for diagnosis and obtained with reasonable patient doses. The study recognized that though no dose limits apply to medical exposure to patients, diagnostic reference levels (DRLs) or reference values (RVs) have been proposed by the ICRP.

The study found that the implementation of digital radiography techniques can entail an increase in patient radiation doses if a strict QC program is not launched in parallel. One of the main causes for the increase is the wide dynamic range of the digital imaging systems, which allows overexposure with no adverse effect on image quality. In addition, the lack of specific training in the new digital techniques for some radiographers and the lack of well established methods to audit patient doses in digital systems can worsen the problem of patient exposure (Coffey *et al.*, 2001).

Typically, for conventional screen-film radiography, systematic overexposure is readily apparent because of elevated film blackening. This is not the case with digital techniques, and implementation of continuous patient dose monitoring instead of isolated annual evaluations will help to improve patient protection by avoiding systematic overexposures for long periods (Vano and Fernandez, 2007).

In routine state X-ray inspection programs in New Jersey in the United States of America, the inspectorate focussed on measurement of X-ray machine parameters such as kVp and mAs, timer accuracy, collimation, etc using ionization chambers and digital meters calibrated in standard X-ray beams. These measurements were related to two indicators of performance: image quality and entrance skin exposure (ESE). Five years of data have been gathered. Both ESE and image quality were checked and the inspectors conducted an audit of the facility's quality assurance program. It was found that entrance skin exposure (ESE) decreased by 34% for lumbar spine, 46% for chest, and 66% for foot X-ray procedures. Image quality has improved by 22%. Quality improvement initiatives were extended to the larger dental X-ray community. Through outreach and information sharing, stakeholders were instructed in the factors that affect patient radiation exposure and image quality and were encouraged to take actions to improve in these areas (Lipoti, 2008).

A number of studies in Kenyan hospitals have emphasized the need for routine quality assurance and/ control (QA/QC) programs on diagnostic radiation equipment and ionizing radiation facilities, for compliance with safety and regulatory requirements. These studies demonstrated that safety audits, and effective implementation are essential to assure the safety of radiation users (Owino, 2001; Muchina, 2006; Chumba, 2007).

2.3 Bone and Soft Tissue Interactions with Photons

Important tissue interaction processes for photons within the energy range of diagnostic X-rays (10 – 150 keV) are photoelectric absorption and Compton scattering. More of the contrast between tissues is due to the photoelectric effect for which the probability of interaction increases rapidly with atomic number.

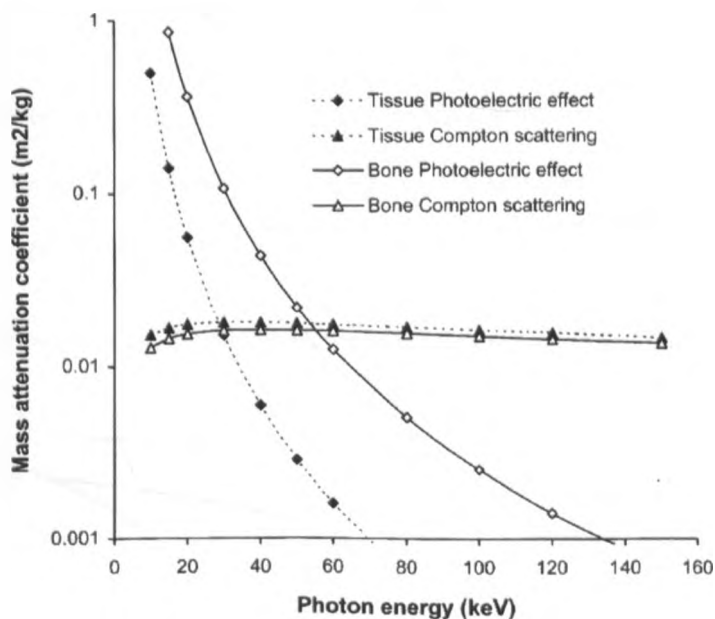


Figure 2.2: Variation in mass attenuation coefficients for photoelectric absorption and Compton scattering in bone and soft tissue with photon energy (Source: Martin, 2008)

In the diagnostic energy range, the number of photons interacting through photoelectric effect decreases with photon energy, while the number of interactions by Compton scattering is almost independent of energy (Martin, 2008). (See Fig 2.2).

As a result, for lower energy X-ray beams, the proportion of photoelectric interactions is higher and so the image contrast will be better, but few X-rays are transmitted through the body. Therefore, higher radiation intensities are required to produce images and the radiation doses to patients are greater. The contrast from higher energy X-ray beams will be poorer, but more photons will be transmitted through the body and reach the image receptor. Thus the amount of radiation required to produce an image, and so the dose given to a patient, will be lower (Martin, 2008).

The number and complexity of medical procedures using X-rays is steadily increasing. As a result, the doses from medical exposures now make up the largest dose to the population in some developed countries (UNSCEAR, 2008). Key developments include the change from film to digital radiography, the increasing sophistication of interventional radiology allowing more complex procedures, the speed and facilities available with the multi-slice computed tomography scanners that have extended the range of applications (Martin, 2008).

It is further observed that while digital techniques in radiology have the potential to reduce patient doses, they also have the potential to significantly increase them. This is a technology that is advancing rapidly and will soon affect hundreds of millions of patients. If careful attention is not paid to the radiation protection issues of digital radiology, medical exposure of patients will increase significantly without concurrent benefit (Vano and Fernandez, 2007).

A study by (Vano and Fernandez, 2007) demonstrated that patient dosimetry and evaluation of image quality are basic aspects of any quality control (QC) program in diagnostic

radiology and traceability is assured when measurements on equipment are complying with national or international standards. Further, the study found that image quality must be adequate for diagnosis and obtained with reasonable patient doses. It is widely known that even though no dose limits apply to medical exposure to patients, diagnostic reference levels (DRLs) or reference values (RVs) have been proposed by the International Commission on Radiological Protection (ICRP, 1991).

Typically, for conventional screen-film radiography, systematic overexposure is readily apparent because of elevated film blackening. This is not the case with digital techniques, and implementation of continuous patient dose monitoring instead of isolated annual evaluations will help to improve patient protection by avoiding systematic overexposures for long periods (Vano and Fernandez, 2007).

The benefits of diagnostic imaging are immense and have revolutionized the practice of medicine. The increased sophistication and clinical efficacy of imaging have resulted in its dramatic growth over the past quarter century. Since the population dose is expected to increase on the basis of the higher number of radiological examinations performed today, it is recommended that before equipment that uses ionizing radiation in a procedure is introduced, there should be general agreement that the benefits exceed the risks and that an attempt has been made to reduce the potential risks as low as practicable through calibration and dose measurements (Amis *et al.*, 2007).

2.4 Effective Doses in Radiology

Most physicians have difficulty assessing the magnitude of exposure or potential risk. Effective dose provides an approximate indicator of potential detriment from ionizing

radiation and should be used as one parameter in evaluating the appropriateness of examinations involving ionizing radiation (Johnston and Brennan, 2000).

Standard radiographic examinations have average effective doses that vary by over a factor of 1000 (0.01–10 mAs). Computed tomographic examinations tend to be in a more narrow range but have relatively high average effective doses (approximately 2–20 mAs), and average effective doses for interventional procedures usually range from 5–70 mAs. Average effective dose for most nuclear medicine procedures varies between 0.3 and 20 mAs. These doses can be compared with the average annual effective dose from background radiation of about 3 mAs (Mettler *et al.*, 2008).

A research coordinated by IAEA found that patients in developing countries often need to have X ray examinations repeated so that doctors have the image quality they need for useful medical diagnosis. The study also found that the quality of X-ray images improved up to 16 percentage points in Africa, 13 % points in Asia and 22 % points in Eastern Europe. At the same time, patient dose reductions ranging from 1.4% to 85% were achieved overall. These improvements are directly attributed to the introduction of a QA/QC programme with emphasis on equipment exposure parameters calibration. The purpose of QC testing is to detect changes that may result in a clinically significant degradation in image quality or a significant increase in radiation exposure (Muhogora *et al.*, 2008).

The beam quality has a major impact on patient dose and a somewhat smaller impact on the quality of the final image. Beam quality will change as the X-ray tube ages due to deposition of target material on the inside of the tube window and to roughening of the target track. This measurement should be made at least annually and whenever the X-ray tube or collimator is replaced or serviced (Coffey *et al.*, 2001).

The quality of an X-ray beam can be characterized by the X-ray spectrum, measured by using spectrometers based on scintillation counters, germanium or silicon detectors, or by crystal diffraction. These techniques, however, require considerable expertise and are time-consuming. Therefore, it is recommended that the quality of X-ray beams used for medical imaging be characterized by a combination of various parameters (HVL1, HVL2, the ratio of HVL1 to HVL2 (i.e. homogeneity coefficient), the tube voltage and the total filtration) (ICRU, 2005). In most cases, the quality of an X-ray beam can be adequately specified by means of the combined information on tube voltage, HVL1, and HVL2, or the tube voltage, HVL1, and total filtration (ICRP, 1991).

Despite all the effort to optimize radiography in recent years, doses for similar examinations in different hospitals still vary substantially. A reduction can be achieved by carrying out optimization through the performance of regular equipment quality assurance and periodic patient dose surveys to ensure that lower dose levels are maintained (IAEA, 2007).

The X-ray spectrum is defined as the energy distribution of the radiation produced in an X-ray exposure. In a study exploring the effects of key factors affecting X-ray spectra namely; generator type, peak tube potential, and filtration, it was found that: (i) Different generator types are characterized by the amount of ripple in the kilo-voltage waveform. (ii) As peak tube potential increases, the HVL increases nearly linearly; radiation output increases by approximately the square of the tube potential. (iii) Filtration materials with ($Z < 42$) produce similar spectra, with only slight variations in efficiency (Nickoloff and Berman, 1993).

CHAPTER THREE

THEORETICAL FRAMEWORK

3.0 Introduction

Voltage (kV), biasing current (mA) and filtration (millimetres of aluminium) affect the output of an X-ray tube. This chapter discusses the effect of these key inputs and provides a general background on X-ray spectrum and beam characteristics (quality and quantity), the interaction of X-ray with matter and the absorbed dose and kerma relation.

3.1 X-ray Beam Characteristics

X-ray beam can be described by its quality and or its quantity. Each of these characteristics is discussed separately in the following sections.

3.1.1 X-ray Beam Quantity

The X-ray beam quantity is the X-ray intensity (number of photons per unit area per unit time) or the radiation exposure; and is affected by the change in any of the following factors: Milliampere seconds, kilovoltages (kVps) and distance and filtration. Milliampere second (mAs) is the product of X-ray tube current by the time of exposure, it controls the number of electrons accelerated towards the anode. If the current is doubled, twice as many electrons will flow from the cathode to the target, and hence twice as much X-ray photons will be produced. Thus, X-ray quantity is directly proportional to the mAs (ICRP, 1996).

Thus:

$$\frac{I_1}{I_2} = \frac{mAs_1}{mAs_2} \quad (3.1)$$

Where I_1 is the X-ray intensity that is produced when a current mAs_1 , is applied on the tube, and I_2 is the X-ray intensity that is produced when current mAs_2 is applied on the X-ray tube. Thus increasing X-ray tube current will also increase X-ray quantity with the same ratio (see Figure 3.2)

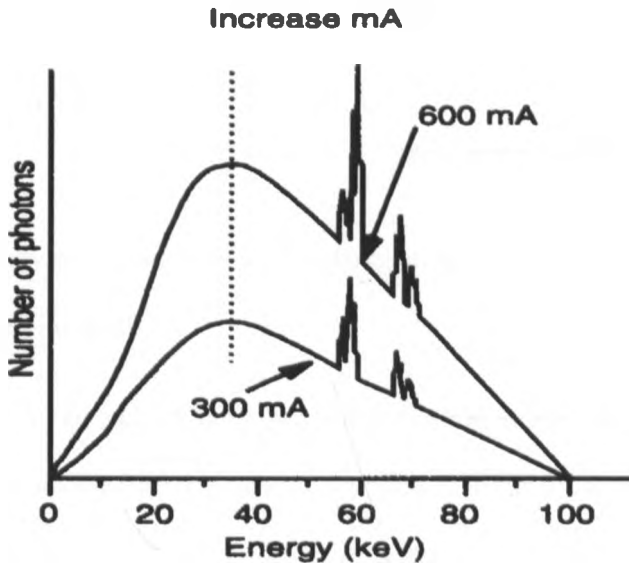


Figure 3.1: Effect of tube current on X-ray spectrum at constant kV. There is a marked change of quantity but no change of quality Source: (Bushberg *et al.*, 2001).

Increase in the applied voltage will increase the probability of Bremstrahlung interaction and hence more X-ray photons will be produced. It was found that X-ray quantity is approximately proportional to the square ratio of the applied voltage (ICRP, 1996)

$$\frac{I_1}{I_2} = \left(\frac{kVp_1}{kVp_2} \right)^2 \quad (3.2)$$

Where I_1 is the intensity of the beam produced when kVp_1 voltage is applied on the tube and I_2 is the intensity of the beam when kVp_2 voltage is applied on the tube. Any change in the potential will affect both the amplitude and the position of the X-ray spectrum. The area

under the curve increases with the square of the factor by which kVp is increased and the relative distribution of emitted X-ray photons shifts to the right (higher energies) (Ahmed, 2007). Thus for the same mAs increasing applied voltage will increase X-ray beam quantity.

The intensity of X-rays is inversely proportional to the square distance from the target (inverse square law); and thus:

$$\frac{I_1}{I_2} = \left(\frac{d_2}{d_1} \right)^2 \quad (3.3)$$

Where I_1 is the intensity of the beam when a distance d_1 is used and I_2 is the intensity of the beam when a distance d_2 is used.

Any material that lies in the path of the X-ray beam is called a filter. There are two types of filtration; inherent and added filtration. The X-ray tube housing for example is an inherent filter material. Any added material to the beam is called added filtration. Filtration reduces the X-ray quantity by selectively removing low energy X-ray photons that do not add any information to the image for diagnosis and hence improves the X-ray beam quality.

Thus filtration has the following effects on the X-ray beam:

- Change in the X-ray spectrum shape (Figure 3.2)
- The peak of the spectrum shifts towards higher energies
- The maximum energy remains unchanged
- The minimum energy shifts towards higher energies

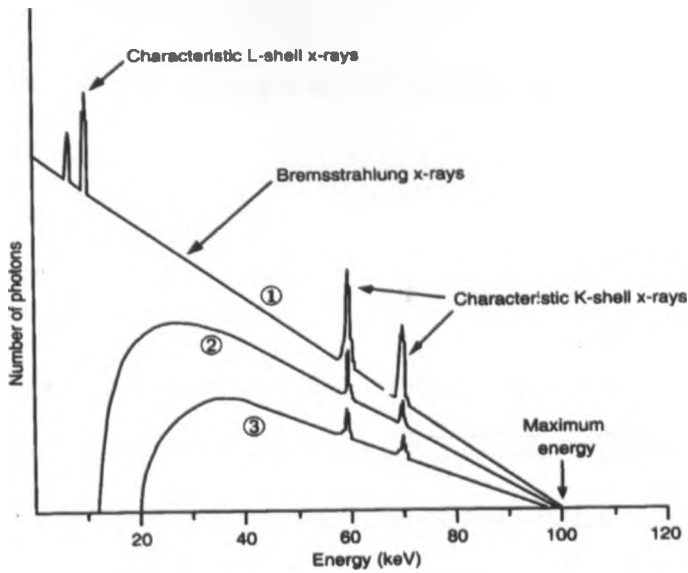


Figure 3.2: Effect of filtration on X-ray spectrum. There is a change in quantity and quality as spectrum shifts to higher energy; 1- spectrum out of anode, 2- after window tube housing (inherent filtration) and 3- after additional filtration. Source: (Bushberg *et al.*, 2001).

3.1.2. X-ray Beam Quality

The X-ray quality is a measure of the penetrating ability of the X-ray beam and it is measured by the HVL of the beam. HVL is the thickness of a substance needed to reduce the intensity of the beam into half of its original value. X-ray beam quality is affected by the applied voltage (kVp), the HVL the target material and the filtration.

The kVp controls the speed of the accelerated electrons and therefore controls the energy of the produced X-rays and the half value layer. The atomic number of the target material affects both the number and the effective energy of the X-rays. When the atomic number of the target is increased, the spectrum is shifted to the right. Increase of total filtration will increase the beam quality by removing low energy photons.

3.2 Interaction of X-rays with Matter

X-ray photons may interact with matter via any of the following five interaction processes described below:

3.2.1 Classical Scattering

In this interaction (Figure 3.3), the incident photon suffers change in its direction but not in wavelength. This kind of interaction is sometimes called coherent scattering. There are two types of coherent scattering, namely; Thomson scattering and Rayleigh scattering. Thomson scattering involves one electron in the interaction whereas in Rayleigh scattering the interaction happens with the whole atom. As this kind of interaction does not involve energy loss and hence no ionization of the atom and only a very small percentage of the radiation undergo coherent or classical scattering, this interaction never plays any important role in diagnostic radiology (Bushberg *et al.*, 2001).

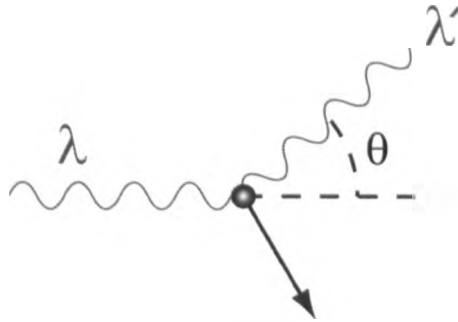


Figure 3.3: Schematic diagram of classical scattering Source: (Bushong, 1994).

3.2.2 Compton Scattering

In this interaction (Figure 3.4) a high energy photon strikes a free electron in the target and ejects it; the photon changes its direction and loses some of its energy as a kinetic energy given to the ejected electron. The scattered photons produce noise to the image, and cannot

be completely removed by the use of grids (Curry *et al.*, 1984). The scattered radiation increases the dose to the patient and staff, and contributes nothing to the diagnostic information.

The change in the wavelength of the scattered photon is given by:

$$\lambda' - \lambda_0 = \frac{c}{\nu'} - \frac{c}{\nu_0} = \frac{h}{m_0 c} (1 - \cos \theta) \quad (3.4)$$

Where λ is the wavelength of the incident photon, λ' is the wavelength of the scattered photon and θ is the scattering angle of the photon (Hendee and Ritenour, 1992).

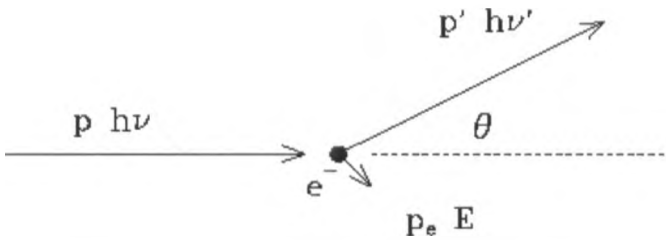


Figure 3.4: Compton scattering of an incident photon of energy $h\nu$ and momentum p to energy $h\nu'$ and momentum p' . The electron is initially at rest and acquires energy E and momentum P_e . Source: (Curry *et al.*, 1984).

3.2.3 Photoelectric Effect

In this interaction (Figure 3.5), the incident photon ejects an electron from the atom by giving it energy, which leaves the atom in an ionized state with an electron vacancy that is filled immediately by an electron from a higher energy level accompanied by an emission of characteristic radiation. The kinetic energy of the ejected electron is the difference between the binding energy and the incident photon energy (Knoll, 2010).

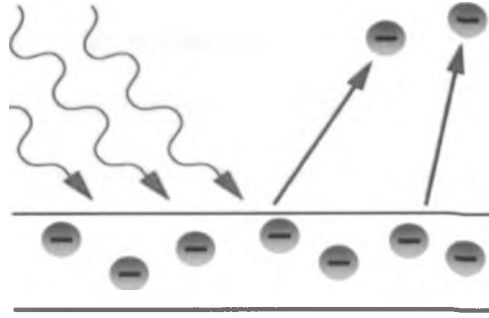


Figure 3.5: Schematic diagram of photoelectric effect

3.2.4 Pair Production

In this interaction (Figure 3.6) a photon with a high energy interacts with the nucleus where the photon disappears and in its place an electron positron pair appears. For this interaction to take place, the energy of the incident photon must be at least 1.02 MeV. This is because the total rest mass of the electron positron pair is about $1.02 \text{ MeV}/c^2$ (Knoll, 2010). Because of its high energy, this interaction is not important in diagnostic radiology.

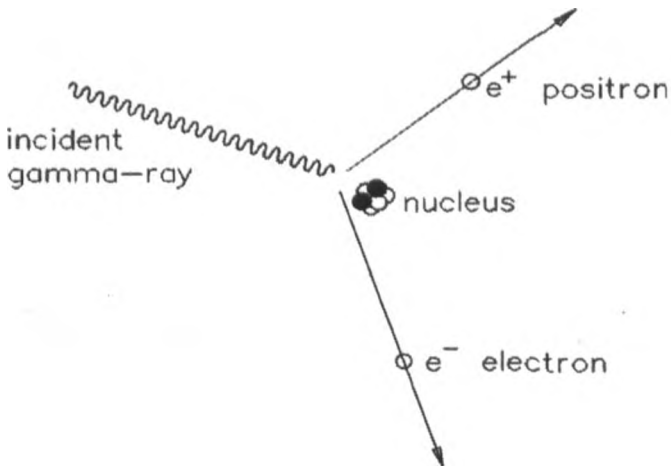


Figure 3.6: Schematic diagram of pair production.

3.2.5 Photodisintegration

In this interaction the incident photon has energy greater than 10 MeV and hence it interacts directly with the nucleus and split it in parts with emission of neutrons. Because of the high photon energy required for this interaction this interaction does not occur in diagnostic X-ray and as such plays no role (Curry *et al.*, 1984).

3.3 Radiation Quantities and Units

There are two types of radiation quantities; those that describe radiation beam itself and those that describe the amount of energy deposited in tissue or matter by a beam of radiation. The former are fluence, fluence rate (flux), energy fluence and energy fluence rate while the latter are kerma, absorbed dose and exposure.

3.3.1 Quantities that Describe Radiation Beam

3.3.1.1 Fluence

The fluence (Φ) of a beam of radiation that contains photons can be described by specifying the number of particles (dN) that cross an area (da) perpendicular to the beam, Thus,

$$\Phi = \frac{dN}{da} \quad (3.5)$$

The SI unit of the fluence is m^{-2} (Hart *et al.*, 1994).

3.3.1.2 Fluence Rate (Flux)

The fluence rate (Φ) of a beam describes the number of particles (dN) that cross a unit area (da) perpendicular to the beam per unit time (dt) (Hart *et al.*, 1994).

$$\phi = \frac{d^2 N}{dadt} = \frac{d\Phi}{dt} \quad (3.6)$$

The SI unit of flux is $\text{m}^{-2}\text{s}^{-1}$ (Hart *et al.*, 1994).

3.3.1.3 Energy Fluence

The energy fluence (Ψ) of a beam is the amount of radiation energy (dE_{beam}) passing through a unit area (da).

$$\Psi = \frac{dE_{\text{beam}}}{da} \quad (3.7)$$

The SI unit of energy fluence is MeV/m^2 (Hart *et al.*, 1994).

In the case of monoenergetic photons with energy $h\nu$, where h is the Plank constant and ν is the radiation frequency, equation 3.7 can be written as:

$$\Psi = \frac{dN \cdot h\nu}{da} \quad (3.8)$$

3.3.1.4 Energy Fluence Rate

The energy fluence rate (ψ) of a beam is the amount of radiation energy carried by a beam crossing a unit area (da) perpendicular to the beam per unit time (dt) (Hart *et al.*, 1994).

$$\psi = \frac{d\Psi}{dt} \quad (3.9)$$

The SI unit of energy fluence rate is $\text{MeV}/(\text{m}^2 \cdot \text{s})$ (Hart *et al.*, 1994).

3.3.2 Quantities that Describe Deposited Energy

The amount of energy a beam deposits in matter such as tissue relates to the amount of damage caused by the beam. The transfer of energy from a radiation beam to a medium can occur in a single stage for direct ionizing radiation or in two stages for indirect ionizing radiation, such as photons. When a photon interacts with a matter, it gives all or part of its energy to an electron of the matter. The electron then gives its energy to the medium via excitation or ionization.

3.3.2.1 Kerma

The kinetic energy released from ionizing radiation per unit mass is called Kerma and is measured in J/Kg or Gray (Gy) (Ahmed, 2007).

$$K = \frac{d\bar{E}_{tr}}{dm} \quad (3.10)$$

Where $d\bar{E}_{tr}$ is the average energy transferred from indirect ionizing radiation, to the medium. If the incident beam is a mono-energetic beam, kerma is given by:

$$K = \Phi \left(\frac{\mu}{\rho} \right) E_{tr} \quad (3.11)$$

Where $\frac{\mu}{\rho}$ is the mass attenuation coefficient of the medium for the incident beam energy, Φ is the fluence where the kerma occurred and $\Phi \left(\frac{\mu}{\rho} \right)$ is number of interactions per unit mass.

3.3.2.2 Absorbed Dose

Although the incident photon transfers all or part of its energy to an electron at a point, not all the transferred energy is given to the medium. As such, the absorbed dose may be defined as:

$$D = \frac{d\bar{E}_{ab}}{dm} \quad (3.12)$$

Where $d\bar{E}_{ab}$ is the average energy imparted by charged particles to the medium. The unit of absorbed dose is the same as that of kerma J/Kg or Gy (Shapiro, 2002).

3.3.3 Exposure

We cannot sense radiation directly so we have to detect a quantity that it effects. Radiation ionizes the atoms of the medium that it passes through. The medium that has been used to quantity radiation is air. The amount of ions produced by a certain beam of photons in a sample of air is called exposure (Shapiro, 2002). The exposure is defined as the number of electric charges dQ that is produced per unit mass of air (dm).

$$X = \frac{dQ}{dm} \quad (3.13)$$

The unit of exposure is Coulomb per kilogram (C/Kg).

If the average energy required to produce one ion pair in air is W_{air} and the charged particle energy released per unit mass of air is $\Psi \left(\frac{\mu_{en}}{\rho} \right)_{air}$, where Ψ is the energy fluence and

$\left(\frac{\mu_{en}}{\rho} \right)_{air}$ is the mass energy absorption coefficient of air, which is defined as:

$$\left(\frac{\mu_{en}}{\rho}\right) = \left(\frac{\mu}{\rho}\right) \frac{\bar{E}_{ab}}{h\nu}$$

Thus, the total charge produced per unit mass of air (exposure) is given by:

$$X = \Psi \left(\frac{\mu_{en}}{\rho} \right)_{air} \left(\frac{e}{W_{air}} \right) \quad (3.14)$$

Where e is the charge of the electron.

3.3.4 Absorbed Dose and Kerma Relation

Part of the incident photon energy is transferred to an electron at a point, but not all the transferred energy is given to the medium. Part of the electron energy is irradiated away as Bremsstrahlung. The absorbed dose is the amount of energy actually retained in the medium. Because the length of the electron tracks may be appreciable, kerma and absorbed dose do not take place at the same location. The absorbed dose (D) is given by:

$$D = \left(\frac{d \bar{E}_{ab}}{dm} \right) = \Phi \left(\frac{\mu}{\rho} \right) \bar{E}_{ab} \quad (3.15)$$

Where $d \bar{E}_{ab}$ is the part of the average kinetic energy transferred to electrons that contributes to ionization excluding the energy loss by bremsstrahlung. $d \bar{E}_{ab}$, \bar{E}_{ab} depends on the photon energy and the absorbent medium. Equation 3.15 can also be written as:

$$D = \Phi \left(\frac{\mu}{\rho} \right) \bar{E}_{ab} = K(1 - g) \quad (3.16)$$

Where g is the fraction of energy that is lost to bremsstrahlung. For low energy photons g is very small and hence $\bar{E}_{ab} = \bar{E}_{rr}$ and therefore, kerma = dose.

If the dose to air is measured at a certain point, it is very simple to calculate dose to any other material in the same place and subject to the same energy fluence. The ratio of the dose to any two different materials subject to the same energy fluence is given by:

$$\frac{D_1}{D_2} = \frac{\psi(\mu/\rho)_1}{\psi(\mu/\rho)_2} = \frac{(\mu/\rho)_1}{(\mu/\rho)_2} \quad (3.17)$$

This means that the ratio of the absorbed doses is equal to the ratio of the mass attenuation coefficients of the two materials.

3.3.5 Kerma in Air, Dose and Exposure

From equation 3.14, energy fluence can be given by:

$$\Psi = \left[\frac{X}{\left(\frac{\mu_{en}}{\rho} \right)} \right] \left(\frac{W_{air}}{e} \right) \quad (3.18)$$

From the definition of kerma, $K_{air} = \Psi \left(\frac{\mu_{en}}{\rho} \right)_{air}$ (3.19)

If we substitute from equation 3.18 into equation 3.19, we obtain

$$K_{air} = X \left(\frac{W_{air}}{e} \right) \left[\frac{\left(\frac{\mu_{en}}{\rho} \right)}{\left(\frac{\mu_{en}}{\rho} \right)_{air}} \right] = \frac{X \left(\frac{W_{air}}{e} \right)}{(1-g)} \quad (3.20)$$

Where $\left[\frac{(\mu_{en}/\rho)}{(\mu_{en}/\rho)} \right] = 1/(1-g)$ and g is the fraction of electron energy lost in

Bremsstrahlung. As stated before, g is very small for diagnostic radiography range, and as such air K can be given by:

$$K_{air} = X \left(\frac{W_{air}}{e} \right) \quad (3.21)$$

Thus kerma can directly be calculated from exposure, using equation 3.16. Since the effect of Bremsstrahlung is negligible in the diagnostic radiography range, and using equation 3.16, one can note that absorbed dose is equal to kerma and hence absorbed dose may be given by:

$$D_{air} = X \left(\frac{W_{air}}{e} \right) \quad (3.22)$$

Thus the exposure at a point, P can be determined through

$$X_p = M_p N_k \quad (3.23)$$

Where N_k is the calibration coefficient for the standard reference chamber and M_p is the chamber reading. The air Kerma in air, therefore, at point P is given by :

$$K_{air} = 0.876 \frac{D_{air}}{R} X_p \quad (3.24)$$

CHAPTER FOUR

MATERIALS AND METHODS

4.0 Introduction

The development and analysis of X-ray beam parameters requires various preparatory stages that lay a foundation on which they are built. This chapter discusses the methods, activities and processes necessary for RQR beam parameter analysis and their comparison to the criteria set by IEC 61627 and ISO 4037 standards. It explains how air kerma reference rates are determined using a one litre (1000 cm³) standard ionization chamber and outlines the set-up of X-ray system used in the study.

Further, the inverse-square law is investigated for this set-up and the true X-ray focus position determined. Additionally, the ISO narrow beam qualities were verified by way of measurement of air kerma rates using both the 1000 cm³ and 30 cm³ ionization chambers through the determination of half value layers (HVL1 and HVL2) and the homogeneity coefficient. The RQR beam qualities for the range 40 kV to 150 kV are considered established if the additional filtration, HVL and the homogeneity coefficient meet the criteria of IEC 61627 and ISO 4037.

In order to conform to standard requirements of ISO 4037-1, a permanent filtration of 3.5 mm of aluminium was placed between filter wheel and monitor chamber (F2 in Figure 4.1). The standard provides that inherent filtration can be adjusted to a maximum of 4 mm of aluminium, which corresponds to a HVL of 2.75 mm of aluminium (ISO, 1996).

4.1 Conditions for Air Kerma Reference Rate Measurement

Standard RQR are described by the set of parameters given below in Table 4.1.

- an emitting tungsten target;
- an X-ray tube voltage adjusted to the values given in column 2 of Table 4.1;
- an adjusted total filtration of the X-ray source assembly;
- the first half-value layer as given in column 3 of Table 4.1
- the homogeneity coefficient within ± 0.03 to that given in column 4 of Table 4.1

These radiation qualities represent the beam incident on the patient in diagnostic radiography.

They were realized by means of a tungsten anode X-ray tube.

Table 4.1: Characterization of radiation qualities (RQR) (Source: IEC 61627: 2005)

Standard Radiation Quality	X-ray tube voltage (kV)	First Half-Value Layer (mmAl)	Homogeneity Coefficient
RQR 2	40	1.42	0.81
RQR 3	50	1.78	0.76
RQR 4	60	2.19	0.74
RQR 5	70	2.58	0.71
RQR 6	80	3.01	0.69
RQR 7	90	3.48	0.68
RQR 8	100	3.97	0.68
RQR 9	120	5.00	0.68
RQR 10	150	6.57	0.72

4.2 Set Up of X-ray System at KEBS

The measurement set up at the KEBS SSDL is as shown in Figure 4.1 below. The X-ray apparatus consists of X-ray housing (h), the diaphragm / aperture wheel holder (D1), the filter wheel holder (F1) and the tube inherent (permanent) filter holder (F2). The HVL filter holder (D2), with a diaphragm, was used for the HVL measurements. The HVL holder was placed approximately midway between the tube and the detector. The diaphragm D1 was remotely

adjusted from the control unit. The filters F1 were selected from the control unit (Comet MPI controller) while the F2 was permanently installed.

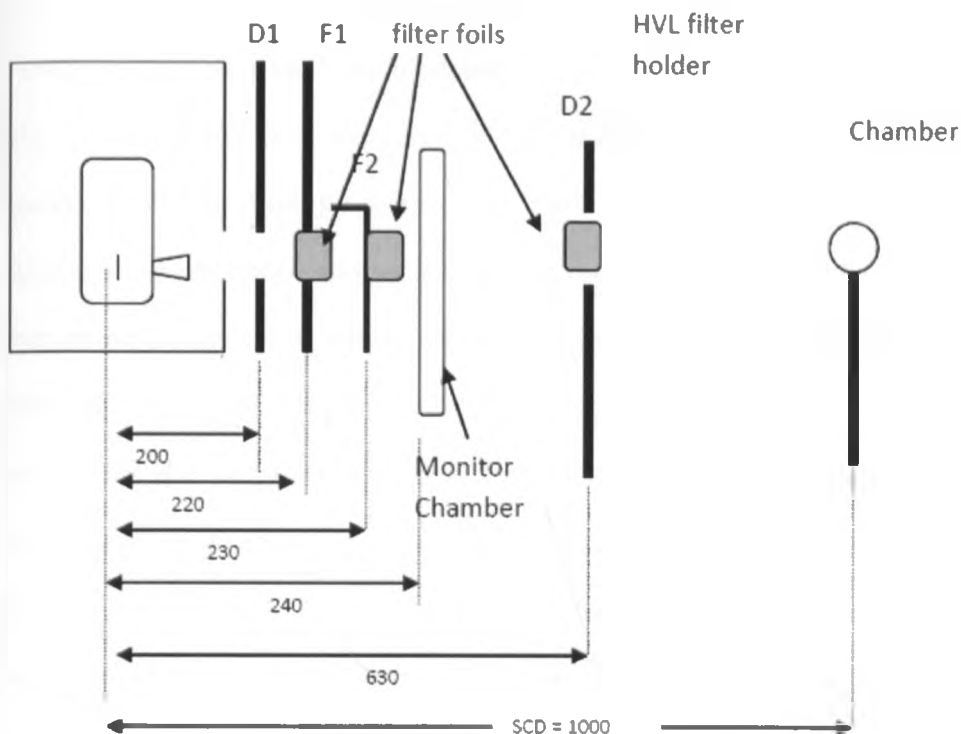


Figure 4.1: KEBS SSDL X-ray system set up (all distances are in mm).

X-ray radiation beams were generated by a constant potential Hopewell 225 kV X-ray machine. Air kerma measurements were performed with the reference 1000 cm³ PTW ionization chamber (Model 32002), calibrated in the PSDL at PTB Germany and whose metrological parameters are known (See appendix I). This is a standard chamber that is normally used for radiation protection measurements and which has very flat energy response of approximately $\pm 4\%$ over the 40 to 150 kV range and thus remained stable during the whole measurement exercise. The chamber was connected to a PTW UNIDOS electrometer

and was considered as the standard instrument since it was traceable to the primary international standard dosimeter at PTB.

The PTW transmission monitor chamber was connected to a UNIDOS E electrometer and was useful in monitoring the X-ray beam stability during the measurement period. The model 32002 spherical chamber is designed for the measurement of ionizing radiation in the protection level range from 0.1 mAs/h to 0.3 Sv/h. Superior features make the chambers suitable as standard chambers for calibration purposes. This is achieved by the thin layer of aluminium on the inner wall surface, which provides for an increased photoelectric yield to compensate for the absorption of soft X-rays. It fulfils the requirement for excellent reproducibility and long-term stability of the sensitive volume. The spherical construction ensures a nearly uniform response to radiation from every direction. This is achieved by the thin layer of aluminium on the inner wall surface, which provides for an increased photoelectric yield to compensate for the absorption of soft X-rays. The outer chamber diameter is 140 mm.

4.3 Verification of Beam Qualities

The X-ray system installed at the KEBS Secondary Standards Dosimetry laboratory was supplied with a filter wheel and aluminium filter foils of varying thicknesses so as to be used in the development and establishment of narrow series (N-series) beam qualities from N40 to N250, complying with the criteria in ISO 4037-1 standard. However, verification and confirmation was not done at the time of installation and commissioning. As part of the preparatory phase of this work, the X-ray beam qualities, N40, N60, N80, N100, N120, N150 and N200, had the kilo voltage (kV) and the necessary additional filters combinations verified through HVL measurements. The results were then compared to the limits in the ISO 4037

standard. The A4 EXTRADIN ionization chamber connected to PTW UNIDOS electrometer was placed at 100 cm distance from focus.

The 5 cm in diameter D1 diaphragm and the HVL diaphragm of 6.5 cm in diameter was used. The filters (Al) for the HVL measurements (called as HVL filters hereinafter) were placed in the appropriate filter holder, at 50 cm from the focus. For each beam quality, three sets of measurements were performed: one without any HVL filter, one with HVL filter just thinner than the expected HVL value and one with HVL filters just thicker than the expected HVL value. For each set of measurements three successive charge readings were taken, which were then corrected for temperature and pressure.

The HVL value was calculated from the interpolation of the values measured in these three sets using excel spread sheets. The calculated HVL values were compared to the values in the international standard ISO 4037. When the percentage (%) difference between the measured and the ISO values was more 5%, the additional filtration was adjusted adequately, in order to get an HVL not more than 5% of the ISO value.

4.4 Inverse-Square Law and X-ray Tube Focus Positioning

The inverse square law in X-ray radiation exposure is stated as (from equation 3.3):

$$\frac{I_1}{I_2} = \left(\frac{d_2}{d_1}\right)^2 \quad (4.1)$$

Where I is intensity and d is distance (radius) of the measurement point from the source.

If we consider an X-ray equipment with the tube focus (the zero '0' position on the wall ruler) is mis-positioned by an amount x cm, then any Focus to Chamber Distance (FCD) is

considered as (FCD + x). If the source intensity is Q_0 and at the experimental 100 cm mark is Q , then equation 4.1 becomes

$$\frac{Q_0}{Q} = \left(\frac{FCD + x}{100 + x} \right)^2 \quad (4.2)$$

Evaluating equation 4.2 above yields

$$\sqrt{\frac{Q_0}{Q}} = \frac{1}{100 + x} \cdot FCD + \frac{x}{100 + x} \quad (4.3)$$

where;

x is the mis-positioning of the focus measured in centimetres,

Q_0 is the intensity (charge) of the source at the origin measured in Coulomb,

Q is the intensity (charge) at the 100 cm position measured in Coulomb,

FCD is the focus to chamber distance measured in centimetres.

Equation (4.3) above is a linear equation of the form $y = mx + c$. Charge measurements were made for various Focus to Chamber distances (FCD) (Table 5.3). The plot in Figure 5.1 was used in determining the value of x , which gives the true position of the tube focus. This is essential in determining the true FCD and the uncertainty due to positioning.

This mis-positioning could be attributed to some reasons: Firstly, the inaccurate positioning of the wall ruler comparing to the “true” position of the source (X-ray tube focus). The Inverse Square Law (ISL) applies only for a point source. The dimensions of the X-ray tube focus were 1.2 mm fine and 4.0 mm standard focus. This fact causes a “theoretical” mis-placement of the “true” source from the “phenomenon” point source as shown in Figure 4.2.

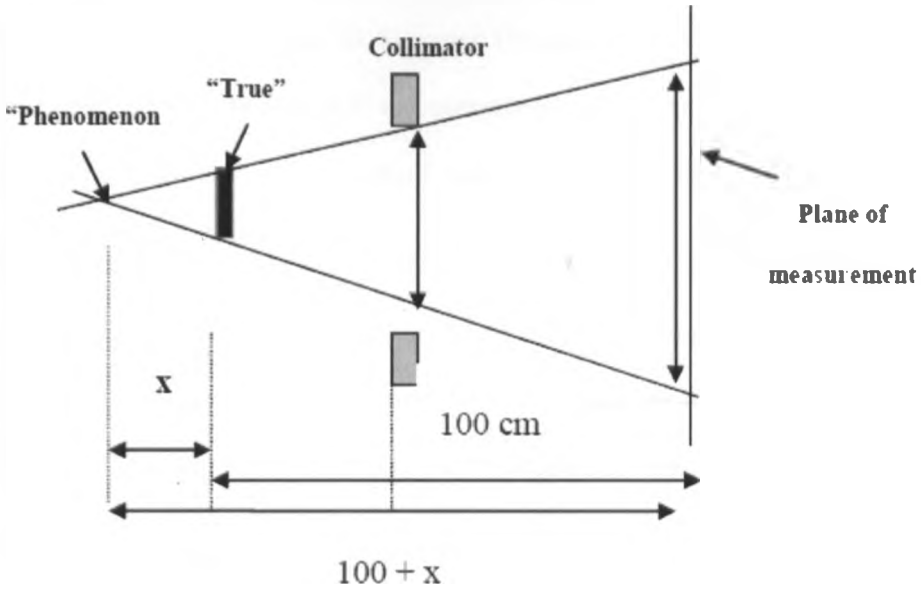


Figure 4.2: Illustration of the ‘mis-positioning’ of the tube focus in relation to the wall mounted ruler.

Secondly, the change of the photon energy spectrum due to attenuation in air in relation to the energy response of the chamber and the uncertainties of the test method itself contribute the rest. If the inverse square law (ISL) is verified, then the air kerma values in distances other than those where the measurement were taken could be calculated using interpolation of the experimental data according to the ISL. In any case, this mis-positioning of the source is considered during the uncertainty calculations.

4.5 Beam Profile, Symmetry and Flatness

The beam symmetry is a measure of the shifting of the profile in respect to the central axis. The 95% flatness region is the distance between the points (left & right) corresponding to 95% of maximum intensity (at 0 cm). A beam profile captures and displays the spatial intensity profile of a beam at a particular plane transverse to the beam propagation path.

The beam profiles were taken for the 5 cm in diameter apertures (D1, see Figure 4.1). The profile was taken with the 5 cm in diameter D1 aperture. An EXTRADIN A3 ionization chamber connected to PTW UNIDOS electrometer was used to scan the beam in a horizontal plane (left to right) in steps of 1 cm at the Focus to Chamber Distance (FCD) of 100 cm.

For each step the charge was measured in 10 sec in the Integration Current Mode (Low Range) of the Electrometer. The tube voltage and current settings were 120 kVp and 15 mA respectively. The field size was measured as the distance between the points (left & right) corresponding to 50% (0.5) of maximum intensity (at 0 cm). The resulting profile is as shown in Figure 5.2.

4.6 Inherent Tube Filtration

The support device with the permanent filtration of 3.5 mm of aluminium was placed between filter wheel and monitor chamber (F2 in Figure 4.1). According to ISO 4037 the inherent filtration should be adjusted to 4 mm Al, which corresponds to an HVL of 2.75 mm Al (ISO 4037 Part 1, 1996).

4.7 Establishment of Air Kerma Reference and Dose Rates for ISO Beam Qualities

The one litre reference ionization chamber (PTW LS01) and the electrometer (PTW UNIDOS) were used to obtain the absolute air kerma (\dot{K}_{air}) rate measurements in ISO narrow beam qualities i.e. N40, N50, N60, N70, N80, N90, N100, N120, and N150. A PTW flat monitor chamber (MC) with its electrometer (PTW UNIDOS E) was used, in order to monitor tube output stability. This chamber was permanently affixed onto the X-ray system.

A high voltage of +400V was applied to the anode of the PTW LS01 ionization chamber (Serial No. 0243) connected to a PTW UNIDOS electrometer (Serial No. 20706). The Integrate Current Low mode (ICL) was selected for obtaining measurements. This is because the settings would enable the measurement of both the leakage current and as well as the low ionization currents flowing in the chamber to a greater precision. The monitor chamber (transmission type) was connected to the electrometer PTW - UNIDOS E to ascertain X-ray tube output stability.

Using a laser positioning system, the chamber was adjusted so as to lie on the central beam axis (CBA) with the geometric centre of the sensitive volume of the ionization chamber placed at 200 cm and used as the reference point of measurement. The 5 cm diameter aperture was selected on the aperture wheel in order to create the desired narrow beam geometry. The Electrometer – Chamber system was left to stabilise overnight to achieve electronic equilibrium. An indeterminate exposure time in the X-ray equipment was selected each time since this equipment can keep emitting radiation for long periods of time without stopping. Only the shutter was used to stop the rays leaving the tube window.

The reference Focus to Chamber Distance (FCD) was set at 200 cm for reference air kerma. The cumulative charge time of 60 seconds was used for the LS01 PTW – UNIDOS system and to 10 seconds for the Monitor Chamber (MC) – UNIDOS E system. The MC system helps to monitor the stability of the tube output.

The chamber was pre-irradiated, nulled (zeroing or re-setting) and leakage current measurements made. These values are required to be as low as possible with the the limit of $\pm 0.05\%$ observed. Additionally, these values were monitored so that they must fluctuate about a mean result and must not increase or decrease monotonically. For each beam quality,

the air kerma rate \dot{K}_{air} and the Monitor Chamber readings were recorded for various mA settings from 2.5 mA to 20 mA.

The air kerma rate (\dot{K}_{air}) was calculated for each beam quality & mA setting using the equation 4.4 (IAEA TRS 457, 2007):

$$\dot{K}_{air} = Q \cdot k_{PT} \cdot N_K^{quality} \quad (4.4)$$

Where;

Q is the charge collected by the ionization chamber in coulombs,

$N_K^{quality}$ is the calibration coefficient from the PSDL (PTB, Germany), which is the calibration coefficient of the dosimeter in terms of the air kerma ($N_k = 25.45 \mu\text{Gy/nC}$).

k_{PT} is the temperature and pressure correction factor relating the temperature and pressure during measurement to the standard temperature and pressure (s.t.p.) and is given by;

$$k_{PT} = \left(\frac{273.2 + T}{273.2 + T_o} \right) \left(\frac{P_o}{P} \right) \quad (4.5)$$

where

P_o is the reference standard pressure expressed in hPa (mbars),

T_o is the reference standard temperature expressed in oC,

T is the mean temperature during measurement,

P is the prevailing pressure during measurement.

In this work, the reference pressure $P_o = 1013.25$ hPa and the reference temperature $T_o = 20^\circ$ C were used.

The monitor chamber (MC) reading was corrected for temperature and pressure using equation 4.5.

$$Q_{MC_{corrected}} = Q_{MC} k_{PT} \quad (4.6)$$

where;

$Q_{MC_{corrected}}$ is the charge collected by the monitor chamber and corrected for temperature and pressure,

Q_{MC} is the actual charge from the monitor chamber as read from the electrometer,

k_{PT} is the temperature and pressure correction factor relating the temperature and pressure.

The air kerma rate values (in $\mu\text{Gy}/\text{min}$) were then correlated to the $Q_{MC_{corrected}}$ readings (in nC/10 sec) as shown in Figures 5.4 (a – g).

The corresponding temperature and pressure correction factors were obtained using equation 4.5. By using a calibration coefficient ($Nk = 25.45 \mu\text{Gy}/\text{nC}$) obtained from the PSDL (PTB, Germany), the mean air kerma rates were determined. Consequently, applying the conversion factors in Table 4.6, dose rates were calculated. The same procedure was repeated for the defined ISO 4037 standard kV values of 40, 60, 80, 100, 120, 150 and 200. The results obtained are in Table 5.5 (a-g).

4.8 Determination of Half Value Layer (HVL)

The half-value layer (HVL) is the thickness of specified material (in this case, aluminium) that will reduce the air-kerma rate of a narrow beam of radiation to one-half its initial value. The second half value layer (HVL₂) is the additional thickness of the absorber that attenuates the air-kerma rate to 25% of its initial value. The contribution of all scattered radiation, other than any which might be present initially in the beam concerned, is deemed to be excluded since the geometry of measurement was that of a narrow beam (i.e the diameter of the beam was just sufficient to irradiate the detector completely and uniformly). In this work, the field size was 42.7 cm diameter at the source to chamber distance of 100 cm. An aperture of 5 cm diameter restricted this beam.

HVL is a beam quality specifier that, together with tube voltage and total filtration, is used to characterize diagnostic X-ray spectra. HVL₁ and HVL₂ on the central axis was determined by the attenuation measurements of a stationary X-ray tube using a PTW ionization chamber and high purity (99.9 %) 1 mm thick aluminium foils stacked together to minimize layers of air in between them. The output stability of the X-ray tube was monitored by a PTW transmission ionization chamber.

4.8.1 First and Second HVL and Homogeneity Coefficient

The ratio between HVL₁ and HVL₂ is termed the homogeneity coefficient, h (equation 4.7):

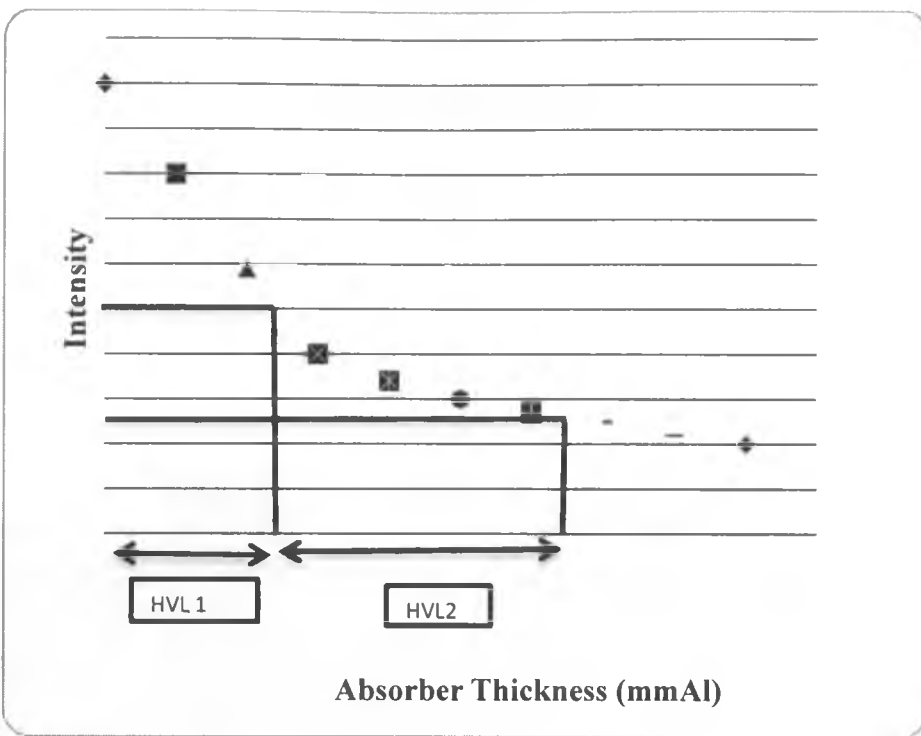


Figure 4.3: An illustration of first and second HVL.

$$h = \frac{HVL_1}{HVL_2} \quad (4.7)$$

The value of h gives an indication about the width of the X-ray spectrum. Its value lies between 0 and 1 with higher values indicating a narrower spectrum. Typical values of h for beams used in diagnostic radiology are between 0.7 and 0.9 (IEC 61627, 2005).

4.8.2 HVL Measurement Set-Up

The half value layer (HVL) measurement set-up used in the Secondary Standards Dosimetry Laboratory, does not differ from that used in a diagnostic X-ray clinic. The geometry of measurement was that of a narrow beam (see Figure 4.4). The diameter of the beam was adjusted so as to be just sufficient to irradiate the detector completely and uniformly. An

unnecessarily large cross-sectional area of the beam is likely to produce additional scattered radiation that will contribute to the recorded signal. The aperture in the beam limiting diaphragm should be just large enough to produce the smallest beam covering the measuring chamber (see Figure 4.4).

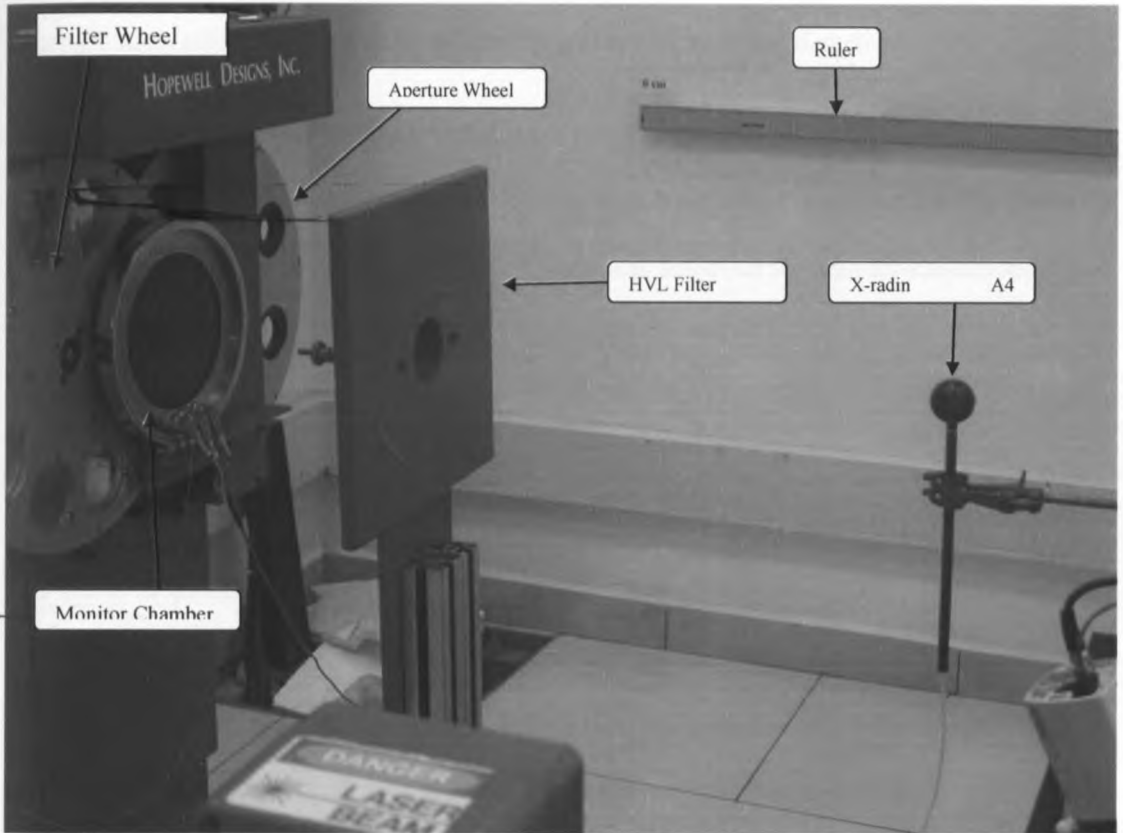


Figure 4.4: HVL measurement set up at SSDL, KEBS.

In this work, a 5 cm aperture was used so that it produces a beam of diameter of 42.7 cm at the focus to chamber distance of 100 cm. This ensured that the chamber was uniformly irradiated and possible scatter radiation is kept to a minimum.

To avoid differences in air kerma rate measurements recorded by the XRADIN A4 ionization chamber, caused by variations in the output of the X-ray tube, the monitor chamber (MC) was used to keep track of any such occurrences. To this end, the readings of ionization chamber (XRADIN A4) were normalized with respect to the readings of the monitor chamber. The monitor is fixed in the beam such that its readings are independent of the presence and the thickness of the absorber. This was achieved by locating the absorbers approximately equidistant from the monitor chamber and from the detector.

Since the temperature and pressure in the room varies over the period of the measurement, a correction for this influence was applied. Precautions were taken to ensure that the variation did not exceed 1 °C as required (ISO, 1996). This was achieved through air-conditioning and monitoring of the room temperature during measurement.

4.8.3 HVL Measurement Procedure

The initial measurements of air kerma rate were made in the absence of any absorber and this measurement was repeated as the last measurement after having measured the air kerma for absorbers of various thicknesses. The air kerma rate was then measured for several absorber thicknesses close to 50 % of the value of the air kerma rate measured initially without any absorber. For the second HVL (HVL_2) measurements of air kerma rate were made by placing additional aluminium filters until the values were close to 25% of the initial air kerma without any absorber.

The measured values of the air kerma rate for various absorbers were plotted against the absorber thickness on a semi-logarithmic scale. The HVL values were derived by

interpolation from the graph. Three measurement points around the actual HVL thickness were sufficient for the linear interpolation.

4.9 Establishment of RQR Beam Qualities for the Range 40 kV to 150 kV

The radiation qualities (RQR) were established according to the international standard IEC 61267. These radiation qualities represent the beam incident on the patient in general radiography, fluoroscopy and dental applications. They were realized by means of a tungsten anode X-ray tube at the KEBS SSDL.

The air kerma rate, K_0 , of the un-attenuated beam were determined as well as air kerma rate, K_d behind an aluminium filtration of thickness, d . Air kerma measurements were performed using a 30 cm³ Shonka-Wyckoff Spherical Chamber, Xradin ionization chamber, model A4, which has an ionization collection efficiency of 99.8%. The chamber is constructed of durable C552 Shonka air-equivalent plastic, providing excellent conductivity. The chamber was placed at a Focus to Chamber (FCD) distance of 100 cm.

A plot of the attenuation curve was made by using a linear scale on the abscissa for the attenuation layer thickness and a logarithmic scale on the ordinate for the attenuation factor. A transparent rectangular template, of height and width of which, both in the respective units of the diagram, are given by a factor of four and by the first HVL of the standard radiation quality to be realized multiplied by $(1 + 1/h)$, respectively, where h is the homogeneity coefficient of the standard radiation quality (IEC, 1994).

An auxiliary horizontal line on the template was made, dividing it into two parts of equal size, and another vertical line at a distance from the left edge of the template corresponding to

the first HVL. This template was then positioned on the attenuation curve in such a way that the edges of the template are parallel to the axes of the diagram and that the upper left and the point of intersection of the two auxiliary lines coincide with points on the attenuation curve (see Fig 4.5 below).

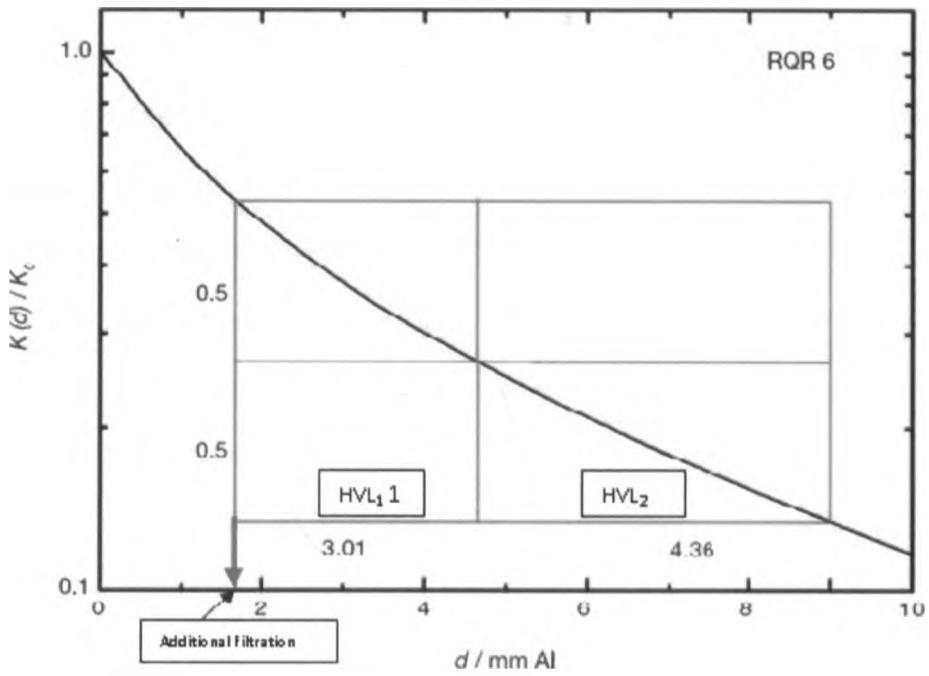


Figure 4.5: An example of the attenuation curve for the beam RQR 6 expressed as the ratio of the air kerma, $K(d)$ behind a filtration of thickness, d , to the air kerma, K_0 , of the un-attenuated beam.

The difference between the position of the left edge of the template and the ordinate gives the amount of additional filtration required to establish the radiation quality RQR. The next step was to add the additional filtration determined above. Finally, the HVL achieved with the modified filtration was verified by measuring the air kerma rate with and without an aluminium attenuation layer of the thickness given in column 3 of Table 4.1. The desired

radiation quality was considered established when the ratio of air kerma (rate) values lay between 0.485 and 0.515 as defined in the IEC 61627 standard.

4.10 Uncertainty Analysis

Uncertainty analysis important in order to assess the quality of a measurement or calculation, facilitate quantitative comparison of results from different investigators and provide for the critical analysis of measurement or calculation methods. Uncertainty is defined as the interval about the average value of a series of measurements or calculations which, within a certain level of confidence, is believed to contain the “true” value of a quantity. These are further classified as type A uncertainty (calculated by statistical methods) and type B uncertainty (evaluated by other means).

By use of the Microsoft Excel spreadsheet, uncertainty measurements were calculated and used in the determination of the uncertainty contribution from each of the possible inputs as stated in Table 5.7. The spreadsheet was further used in computing the combined and expanded uncertainty.

CHAPTER FIVE

RESULTS AND DISCUSSION

5.0 Introduction

This chapter presents the results obtained from verification of beam qualities; investigation of the inverse-square law (ISL); determination of the X-ray tube focus positioning; beam profiling; establishment of air kerma reference rates and dose rates for ISO 4037-1 beam qualities; determination of the half value layer (HVL) and establishment of RQR beam qualities for the range 40 kV to 150 kV in accordance with the IEC 61627 standard. The associated uncertainties of measurements are also evaluated.

5.1 Determination and Verification of Beam Qualities

Initial measurements of the half value layer for some narrow beam qualities (N-series) indicated that the measured HVL values were lower by more than 5% of the ISO criteria. In order to overcome these discrepancies, which result from inadequate X-ray beam filtration, additional foils of aluminium were added to the filters. An additional 0.5 mm Al extra foil was added to the permanent filtration device, making the inherent filtration of the tube to be 1 millimetre of beryllium + 3.5 millimetres of aluminium + monitor chamber (specified to be equivalent to 1 millimetre of aluminium by the manufacturer).

To test the effect of this modification, the measured half value layer (HVL) for the inherent filtration at 60 kV (reference) was found to be 2.688 millimetres of aluminium, which differs from the ISO suggested value (2.75 millimetres of aluminium) by -2.3 % (limit ± 5 %). Thus, the inherent filtration was found to be acceptable.

The N40, N60, N80, N100, N120, N150 and N200 ISO Narrow Series beam qualities were verified. Some filters of the wheel were modified, since the criterion was not satisfied in the original form as presented by the manufacturer. These modifications to filters or/and kVp, together with the differences (%) from the ISO values are shown in Table 5.1 below.

Table 5.1: Modifications made in X-ray qualities

Beam	Initial HVL values found			After modification			Modification
Quality	kVp	HVL	% diff	kVp	HVL	% diff	Additional mm Al
N40	40	(1 st) 2.57	-7.30 %	41	(1 st) 2.68	+0.91 %	0.2
N120	120	(2 nd) 1.62	-8.80 %	120	(2 nd) 1.77	-0.88 %	1

The final ISO narrow beam qualities obtained are shown in Table 5.2.

Table 5.2: The final ISO 4037 narrow spectrum series established at KEBS SSDL

ISO 4037		Final added Filtration In millimetres				KEBS	KEBS	ISO	ISO	% diff	% diff
Beam Quality	kVp	Pb	Sn	Cu	Al	1st HVL	2nd HVL	1st HVL	2nd HVL	1st HVL	2nd HVL
N40	40	0.0	0.0	0.2	0.2	0.082	0.086	0.084	0.092	-2.14%	-3.30%
N60	60	0.0	0.0	0.6	0.0	0.240	0.245	0.24	0.26	-0.20%	-3.59%
N80	80	0.0	0.0	2.0	0.0	0.566	0.604	0.58	0.62	-2.41%	-2.58%
N100	100	0.0	0.0	5.0	0.0	1.076	1.137	1.11	1.17	-3.06%	-2.82%
N120	120	0.0	1.0	5.0	1.0	1.676	1.769	1.71	1.77	-1.99%	-0.06%
N150	150	0.0	2.5	0.0	0.0	2.353	2.545	2.36	2.47	-0.30%	3.04%
N200	200	1.0	3.0	2.0	0.0	3.999	3.911	3.99	4.05	0.23%	-3.43%

The verified beam qualities were found to be within the $\pm 5\%$ acceptance threshold and thus further development of reference medical calibration beams was possible.

5.2 Inverse-Square Law and X-ray Tube Focus Positioning

Table 5.3 below shows the results of the determination of the actual position of the tube focus using the inverse square law.

Table 5.3: X-ray focus positioning

SCD	Temp/°C	P/mbar	k_{PT}	Mean Charge (Q)	Corrected Charge Q_{corr}	$\frac{Q_o}{Q}$	$\frac{\sqrt{Q_o}}{\sqrt{Q}}$
70	23.4	846.4	1.2110	279.30 ± 0.00	338.24	0.493	0.702
80	23.4	846.4	1.2110	214.37 ± 0.06	259.60	0.643	0.802
90	23.4	846.4	1.2110	169.80 ± 0.00	205.63	0.812	0.901
100	23.4	846.4	1.2110	137.83 ± 0.29	166.92 (Q_o)	1.000	1.000
115	23.4	846.4	1.2110	104.40 ± 0.00	126.43	1.320	1.149
125	23.4	846.4	1.2110	88.39 ± 0.01	107.04	1.559	1.249
130	23.4	846.4	1.2110	81.64 ± 0.02	98.86	1.688	1.299
150	23.4	846.4	1.2110	61.10 ± 0.01	73.99	2.256	1.502
200	23.4	846.4	1.2110	34.18 ± 0.04	41.40	4.032	2.008

The mis-positioning (x) of the source using the distance indicated by the ruler on the vertical wall was

determined from the slope and intercept of the curve. A plot of $\sqrt{\frac{Q_o}{Q}}$ against SCD is displayed in Fig. 5.1.

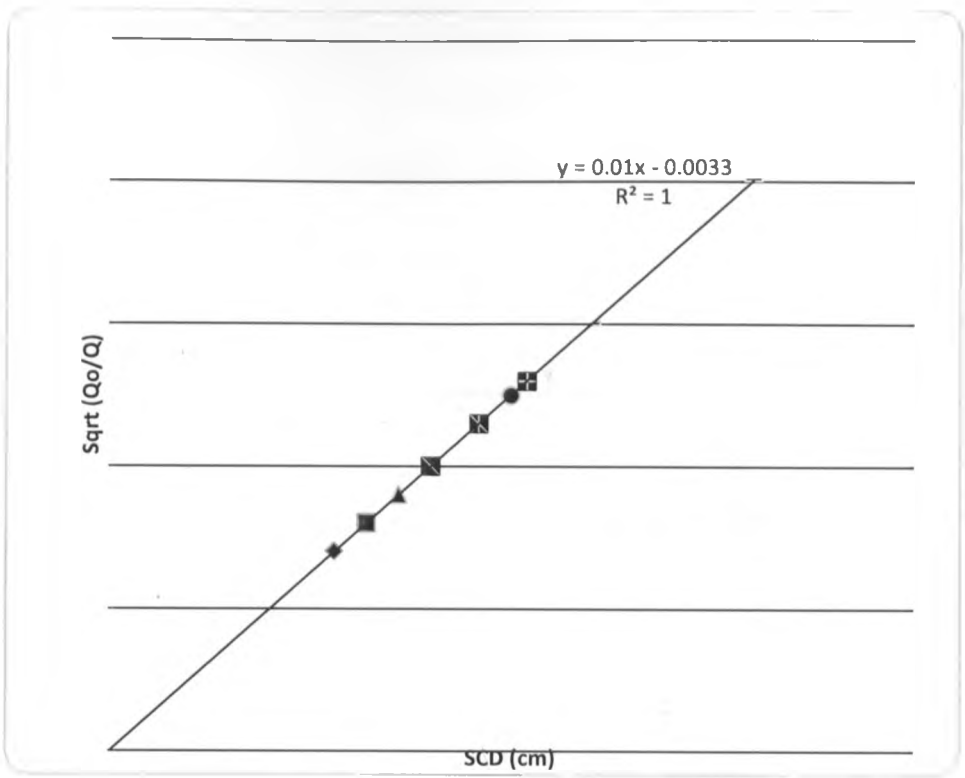


Figure 5.1: Plot of $\sqrt{\frac{Q_0}{Q}}$ against Source to Chamber Distance (SCD).

Using equation 4.3, the mis-positioning (x) of the tube focus using the wall mounted ruler was determined as -0.4 cm. This was taken into account during measurement and in the determination of the uncertainty of measurement in the positioning of the chamber during the experiment. This is, however, quite insignificant (0.2 % of FCD) given that most calibrations are performed at 200 cm distance from focus.

5.3 Beam Profile, Symmetry and Flatness

The beam profile obtained at the focus to chamber distance (FCD) of 100 cm is shown in Figure 5.2. It was deduced that the field size (at 50 %) was 42.7 cm in diameter, the symmetry was -1.2 cm and the 95% flatness region covered a span of 31.8 cm in diameter (i.e: 14.9 cm to the right and

16.9 cm to the left). This measurement is important since it gives the size of the beam at the measurement point and is used to ensure that the whole ionization chamber volume is fully immersed in the beam to ensure total irradiation and thus complete ionization of the air inside it.

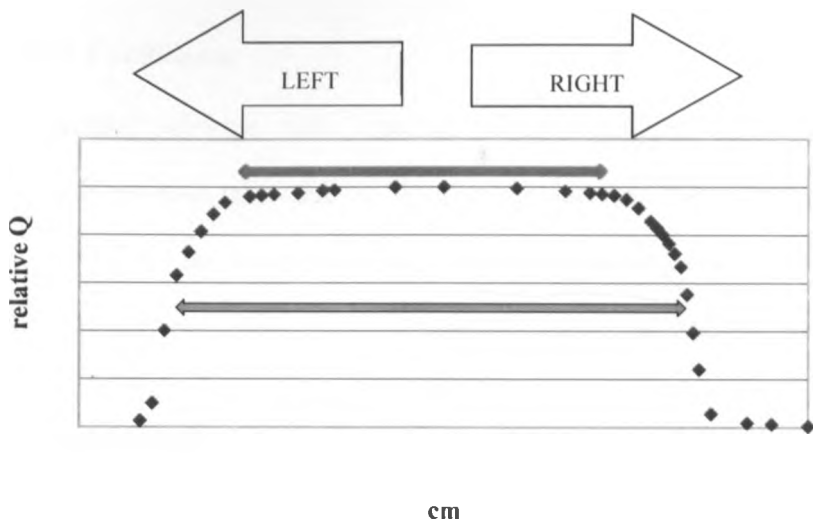


Figure 5.2: X-ray Beam Profile at SCD of 100 cm and aperture diameter of 5 cm.

5.4 Air Kerma Reference and Dose Rates for ISO Beam Qualities

In order for radiation protection calibrations at the SSDL to be feasible, the air kerma rates obtained for various narrow beam qualities (see Tables 4.9 – 4.15 in appendix II) was converted to standard dose rates using the conversion coefficients in Table 4.6 (IAEA, 1999). These coefficients were used to convert air kerma rate (k_{air}) to ambient dose rate ($H^*(10)$).

These conversion coefficients are obtained by use of computation methods relating air Kerma to $H_p(0.07, \alpha)$, $H_p(10, \alpha)$, and $H^*(10)$ in an ICRU slab phantom for tungsten anode X-ray spectra for tube potentials from 40 to 140 kV. These methods allow an appreciable estimation of conversion coefficients for the narrow X-ray spectra indispensable to calibrate the personnel dosimeters in

terms of the personal dose equivalent (Ankerhold *et al.*, 1999). This scenario is vital in converting dosimetric quantities (air kerma) to operational quantities (dose and dose rate).

5.4.1 Conversion Coefficients

Protection of personnel working with ionizing radiation relies on careful and accurate measurement of ambient dose rates and the dose accumulated as a result of their work. The protection of the public and the environment depends on evaluation of radiation and radioactive materials in the environment. Both these scenarios require the use of equipment whose metrological characteristics have been verified and ascertained through calibration. Because of diversity in exposures in both routine and accident conditions, internationally accepted measurement conventions are required for assessment of irradiation of individuals and for monitoring of the environment. Specialized quantities and a substantial collection of reference data are needed for correlation of individual exposures and the associated risk (Wall, 2004).

ICRP Publication 74 provides an extensive and authoritative set of data linking the operational quantities defined by ICRU with the dosimetric and protection quantities defined by ICRP. The operational quantities provide a satisfactory basis for most of the measurements for radiation protection against external radiations. In those cases where it is not so, the data given in the publication provides a basis for designing special measurement programmes, properly interpreting their results and relating them to the protection quantities (ICRU, 2005).

Table 5.4: Conversion coefficients for the ISO Narrow Beam Qualities (Source: ICRP, 1996)

Radiation Quality	Mean Energy, E (keV)	H'(0.07)/K _a Sv.Gy ⁻¹	H*(10)/K _a Sv.Gy ⁻¹
N10	8	0.91	-
N15	12	0.96	-
N20	16	1.00	-
N25	20	1.03	0.51
N30	24	1.10	0.81
N40	33	1.25	1.18
N60	48	1.48	1.59
N80	65	1.60	1.73
N100	83	1.60	1.71
N120	100	1.55	1.64
N150	118	1.50	1.58
N200	164	1.39	1.46
N250	208	1.34	1.39
N300	250	1.31	1.35

The results obtained were as shown in Tables 5.5 (a-g). The correlation with the monitor chamber (MC) readings is shown in the corresponding Figure(s) 5.4. It can be deduced from the monitor chamber calibration factor that the beams were successfully verified and that the X-ray output was stable during the course of the measurements (see Figure 5.3).

Table 5.5: Analysis of beam quality characteristics for the ISO 4037 narrow series

(a) N40 Beam Quality

Conversion Coefficient, $h = 1.18 \text{ SvGy}^{-1}$

	K _{air}	K _{air}	H*(10)	H*(10)	MonCh	MC calib.factor
mA	μGy/min	μGy/h	μSv/h	mSv/h	nC/10s	(μGy/min)/(nC/10s)
2.5	34.148	2048.878	2417.677	2.42	0.877	38.923
5.0	67.240	4034.419	4760.615	4.76	1.727	38.944
10.0	127.144	7628.633	9001.787	9.00	3.265	38.939
15.0	179.044	10742.624	12676.296	12.68	4.590	39.005
20.0	223.510	13410.625	15824.537	15.82	5.717	39.093

(b) N60 Beam QualityConversion Coefficient, $h = 1.59 \text{ SvGy}^{-1}$

	Kair	Kair	H*(10)	H*(10)	MonCh	MC (calib.factor)
mA	μGy/min	μGy/h	μSv/h	mAs/h	nC/10s	(μGy/min)/(nC/10s)
2.5	65.760	3945.619	6273.535	6.27	1.822	36.097
5.0	121.033	7261.991	11546.565	11.55	3.379	35.816
10.0	237.130	14227.806	22622.212	22.62	6.643	35.698
15.0	342.111	20526.658	32637.386	32.64	9.599	35.640
20.0	439.123	26347.367	41892.313	41.89	12.340	35.584

(c) N80 Beam QualityConversion Coefficient, $h = 1.73 \text{ SvGy}^{-1}$

	Kair	Kair	H*(10)	H*(10)	MonCh	MC calib.factor
mA	μGy/min	μGy/h	μSv/h	mAs/h	nC/10s	(μGy/min)/(nC/10s)
2.5	34.176	2050.579	3547.502	3.55	0.744	45.907
5.0	67.771	4066.261	7034.632	7.03	1.508	44.946
10.0	130.782	7846.898	13575.134	13.58	2.952	44.298
15.0	189.506	11370.359	19670.722	19.67	4.298	44.090
20.0	243.515	14610.896	25276.851	25.28	5.551	43.867

(d) N100 Beam QualityConversion Coefficient, $h = 1.71 \text{ SvGy}^{-1}$

	Kair	Kair	H*(10)	H*(10)	MonCh	MC calib.factor
mA	μGy/min	μGy/h	μSv/h	mAs/h	nC/10s	(μGy/min)/(nC/10s)
2.5	17.080	1024.822	1752.445	1.75	0.287	53.457
5.0	34.508	2070.480	3540.521	3.54	0.646	53.380
10.0	66.893	4013.575	6863.213	6.86	1.264	52.910
15.0	98.257	5895.449	10081.219	10.08	1.868	52.603
20.0	127.072	7624.339	13190.106	13.19	2.429	52.620

(e) N120 Beam QualityConversion Coefficient, $h = 1.64 \text{ SvGy}^{-1}$

	Kair	Kair	H*(10)	H*(10)	MonCh	MC calib.factor
mA	μGy/min	μGy/h	μSv/h	mAs/h	nC/10s	(μGy/min)/(nC/10s)
2.5	17.048	1022.860	1677.491	1.68	0.287	52.446
5.0	34.437	2066.212	3388.587	3.39	0.645	53.380
10.0	66.755	4005.301	6568.694	6.57	1.262	52.910
15.0	98.055	5883.296	9648.605	9.65	1.864	52.603
20.0	126.768	7606.088	12473.984	12.47	2.423	52.320

(f) N150 Beam QualityConversion Coefficient, $h = 1.58 \text{ SvGy}^{-1}$

	Kair	Kair	H*(10)	H*(10)	MonCh	MC calib.factor
mA	μGy/min	μGy/h	μSv/h	mAs/h	nC/10s	(μGy/min)/(nC/10s)
2.5	148.04	8882.32	14034.06	14.03	1.791	82.677
5.0	296.55	17792.80	28112.63	28.11	3.613	82.086
10.0	583.07	34984.01	55274.74	55.27	7.128	81.798
15.0	860.58	51634.84	81583.04	81.58	10.762	79.962
20.0	1126.07	67564.31	106751.61	106.75	13.811	81.532

(g) N200 Beam QualityConversion Coefficient, $h = 1.46 \text{ SvGy}^{-1}$

	Kair	Kair	H*(10)	H*(10)	MonCh	MC calib.factor
mA	μGy/min	μGy/h	μSv/h	mAs/h	nC/10s	(μGy/min)/(nC/10s)
2.5	52.392	3143.521	4589.540	4.59	0.963	54.384
5.0	104.523	6271.380	9156.215	9.16	1.928	54.200
10.0	205.698	12341.903	18019.179	18.02	3.814	53.931
15.0	304.015	18240.880	26631.685	26.63	5.665	53.670

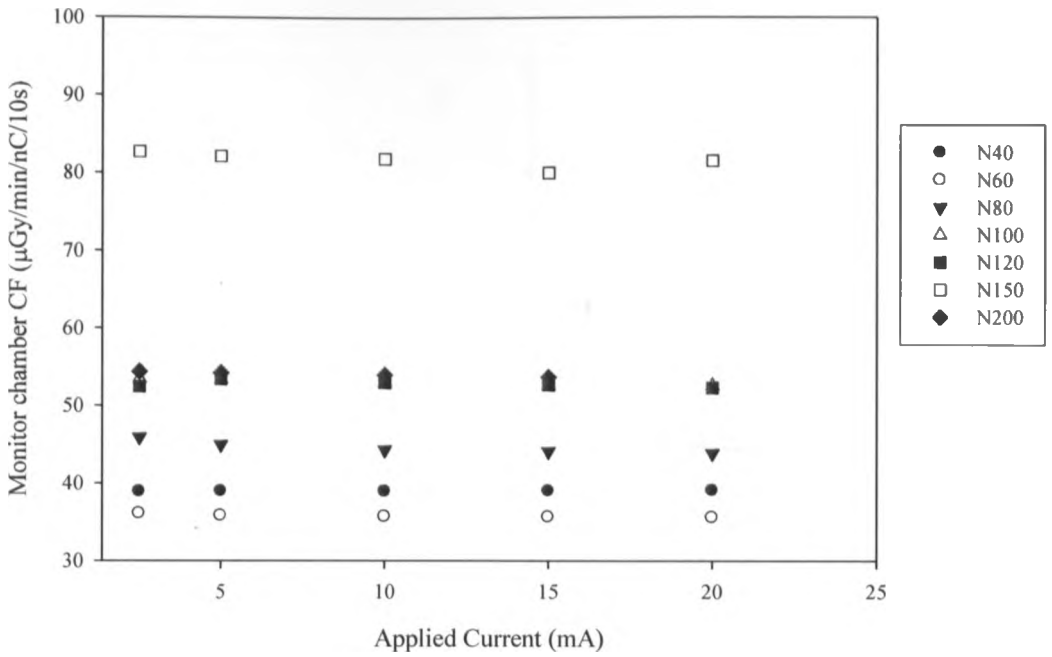


Figure 5.3: Monitor chamber (PTW UNIDOS E) stability during air kerma measurements for ISO 4037 qualities.

5.4.2 Correlations Between the Air Kerma Rates (K_{air}) and Monitor Chamber Readings

Figure 5.4 shows how the air kerma rates and monitor chamber readings vary with current. It can be observed that tube current is proportional to ambient dose rates. This is further confirmed by the increase in the monitor chamber readings that checks the tube output. The results confirm that the system is properly configured with regard to the added filtration and can be relied on in the calibration of radiation protection detectors and diagnostic X-ray equipment.

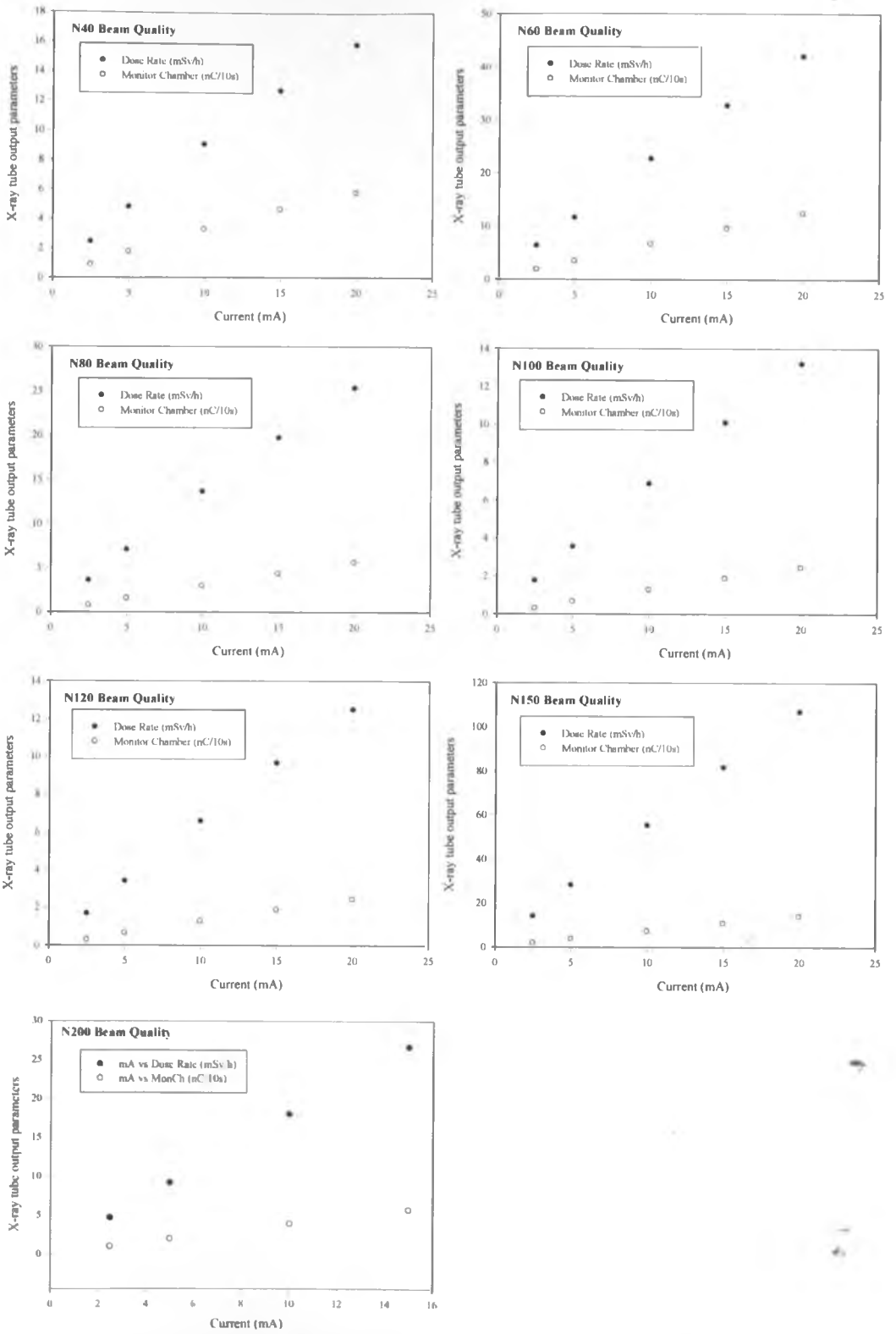


Figure 5.4 : Correlations between air kerma rates and monitor chamber for N40 to N200 qualities.

Table 5.6: Reference RQR beams established at KEBS SSDL

Beam	Voltage	Filter	KEBS	KEBS	KEBS	KEBS	ISO	ISO	ISO	IEC 61627	% diff	% diff	% diff	% diff
Quality	kV	No.	1 ST HVL	2 ND HVL	h	Filtration	1 st HVL	HVL	h	Filtration	1 st HVL	2 nd HVL	h	Filtration
RQR 2	40	1	1.420	1.653	0.86	2.50	1.42	1.75	0.81	2.49	-0.03%	-4.52%	4.80%	0.40%
RQR 3	50	2	1.720	2.236	0.77	2.45	1.78	2.34	0.76	2.46	-3.36%	-4.45%	1.14%	-0.41%
RQR 4	60	3	2.172	2.976	0.73	2.70	2.19	2.96	0.74	2.68	-0.82%	0.54%	-0.35%	0.75%
RQR 5	70	4	2.671	3.767	0.71	3.10	2.58	3.63	0.71	2.83	3.54%	3.76%	-0.22%	4.54%
RQR 6	80	5	2.877	4.029	0.71	3.00	3.01	4.36	0.69	2.99	-4.43%	-4.59%	3.42%	0.33%
RQR 7	90	6	3.417	5.012	0.68	3.30	3.48	5.12	0.68	3.18	-1.80%	-2.11%	0.32%	3.77%
RQR 8	100	7	3.848	5.798	0.66	3.35	3.97	5.84	0.68	3.36	-3.07%	-0.72%	-2.37%	-0.30%
RQR 9	120	8	4.943	7.052	0.70	3.75	5.00	7.35	0.68	3.73	-1.14%	-4.05%	3.04%	0.54%
RQR10	150	9	6.570	9.129	0.72	4.40	6.57	9.13	0.72	4.38	0.00%	-0.01%	0.01%	0.46%

It can be observed in Table 5.6 that the diagnostic reference beam qualities were successfully established at the secondary standards dosimetry laboratory at Kenya Bureau of Standards. These qualities represent the clinical X-ray beams incident on patients during various radiofluoroscopic techniques and can now be used to calibrate hospital systems through transfer standards (multimeters) that measure the peak voltages (kV_p), current (mA), Dose (Gray) and time (seconds) parameters from clinical systems used in diagnostic radiology. Other equipment that can now be calibrated includes ionization chambers and electrometers. The beams were well within the tolerance of $\pm 5\%$ in the requirements set by the IEC 61627 standard. Thus they can actively be used to be transferred through calibration of clinical systems and provide the unbroken traceability chain that was previously lacking. This new capability will also help in standardizing approaches to patient dose assessments for various diagnostic procedures by providing a known reference point and hence provide a true picture of the performance of the technology in Kenya's healthcare system.

The calibration capabilities established in Kenya compare very well with similar systems established elsewhere in the world based on the requirements of the ISO 4037 and IEC 61627 standards. The first half value layer for reference diagnostic beam qualities established in a few countries are sampled and the results shown below (Table 5.7).

Table 5.7: Comparison of first half value layer results from selected countries

COUNTRY	1 ST HALF VALUE LAYER									REFERENCE
	RQR 2	RQR 3	RQR 4	RQR 5	RQR 6	RQR 7	RQR 8	RQR 9	RQR 10	
ISO	1.42	1.78	2.19	2.58	3.01	3.48	3.97	5.00	6.57	ISO 4037
GREECE	0.99	1.49	1.98	2.49	2.88	3.31	3.72	4.72	5.71	www.eeae.gr
BRAZIL	1.43	1.77	2.22	2.52	3.00	3.52	3.90	5.04	6.71	(Franciscatto and Potiens, 2009)
KENYA	1.42	1.72	2.17	2.67	2.87	3.42	3.85	4.94	6.57	This work

The capability and capacity of the SSDL at KEBS has been extended to include calibrations in the medical diagnostic realm. Patient dose optimization can now be undertaken with certainty since quality control checks, as part of overall quality assurance programmes, shall now be undertaken using equipment whose metrological parameters are known.

5.5 Evaluation of Uncertainties

Uncertainty refers to the estimated amount or percentage by which an observed or calculated value may differ from the true value. In this work, an effort was made to quantify the contributions of various parameters and influence factors that impacted on the final results. Table 5.8 below represents the summary of the parameters identified and the overall uncertainty determined for this project.

Table 5.8: Parameters to characterize the uncertainty in the Xradin A4 ionization chamber calibration procedure in the RQR reference radiation beam establishment

Source of uncertainty	Type	Distribution	Divider	Uncertainty (%)
Thermometer Resolution	B	Rectangular	3	0.020
Thermometer Calibration	B	Rectangular	2	0.051
Barometer Resolution	B	Rectangular	3	0.004
Barometer Calibration	B	Rectangular	2.53	0.023
Distance Resolution	B	Rectangular	3	0.115
Distance (Positioning)	B	Rectangular	3	0.577
Chronometer Resolution	B	Rectangular	3	0.002
Reference Ionization Chamber Resolution	B	Rectangular	3	0.001
Reference Ionization Chamber Calibration	B	Rectangular	2	0.450
Reference Ionization Chamber Standard Deviation	A	Normal	10	0.025
Monitor Chamber Resolution	B	Rectangular	3	0.006
Monitor Chamber Standard Deviation	A	Normal	10	0.047
Ionization Chamber Xradin A4 Resolution	B	Rectangular	3	0.021
Ionization Chamber Xradin A4 Standard Deviation	A	Normal	10	0.092
Combined Uncertainty		Normal		0.751
Expanded Uncertainty		Normal (k = 2)		1.503

The results in Table 5.8 show that the main source of uncertainty was the positioning procedure of the Xradin A4 ionization chamber. This uncertainty could be reduced by acquiring an accurate computer controlled positioning set-up.

CHAPTER SIX

CONCLUSIONS AND RECOMMENDATIONS

6.0 Conclusions

This thesis has been devoted to the beam parameter analysis aimed at the development of reference radiation beam qualities (RQR) that facilitate the accurate calibration of diagnostic X-ray equipment and radiation protection detectors as well as describing the methodology to be used in establishing a reference X-ray laboratory for such purposes. The ultimate aim is to increase the scope of the SSDL services at KEBS in the area of calibration traceability of radiation protection and diagnostic X-ray systems.

In this study we have established narrow beam qualities using a Hopewell Design X-ray system based on ISO 4037 criteria; and used them to determine the necessary filtration needed for appropriate RQR compliant with IEC 61267 standard. The beam characteristics were analyzed through the measurement of beam parameters namely; the inherent tube filtration, beam uniformity, radiation field size, and uniformity and flatness measure. The first half-value layers and the homogeneity coefficients were measured for the RQR2, RQR3, RQR4, RQR5, RQR6, RQR7, RQR8, RQR9 and RQR10 IEC beam qualities. The required additional filtration was chosen and adjusted to comply with the IEC 61627 standard criteria.

The main conclusions that arise from the presented work are:

- a) The Air Kerma rate (K_{air}) references values for the ISO 4037 narrow series N40 to N200 radiation protection qualities were measured and verified. The first and second

half value layers (HVLs) on the installed Hopewell X-ray system were determined, modifications (Table 5.1) made and results compared with the ISO criteria. They were found to be in agreement within the $\pm 5\%$ allowable tolerance (Table AII.1-7 in appendix II).

- b) Reference Radiation Beam Qualities (RQR) for the X-ray range 40kV to 150 kV used in diagnostic radiology were developed and established to comply with the criteria in IEC 61267 standard, at the Secondary Standards Dosimetry Laboratory (SSDL) at Kenya Bureau of Standards (KEBS). The values of the RQR were found to be within the $\pm 5\%$ allowable limits (Table 5.1).
- c) A method for the establishment of a radiation protection and diagnostic level calibration laboratory was documented, illustrated and confirmed. This will enable the standardization of approaches in future establishments of similar laboratories.

The international standardization of the radiation beams allows X-ray equipment to be calibrated and type tested at different laboratories under the same conditions and irradiation characteristics. This is done in an effort geared towards exploiting the full benefits of ionizing radiation while keeping doses to exposed individuals as low as reasonably achievable.

To establish and implement reference radiation beam (RQR), it was necessary to measure the first and the second HVL, after having added the calculated filtration in the tests specified the IEC standard. The values obtained were then compared to the values of reference established by this standard. The RQR beam of interest were considered to have been established and implemented since the half value layer (HVL), homogeneity coefficient and total aluminium filtration values were found to be within the tolerances permitted by the standard.

New X-ray radiation beams suitable for calibration of diagnostic and radiation protection instruments are now available at the Kenya Bureau of Standards Secondary Standards Dosimetry Laboratory. The beams are based on two series of X-ray beams, RQR and the ISO narrow described in the IEC standard 61267 (2005). The radiation qualities RQR2 to RQR10 correspond to the beams emerging from the X-ray tube assembly and incident on patients. The beams are calibration alternatives to the ISO narrow qualities that until now had been used for calibration of diagnostic instruments.

Following the criteria defined in the IEC 61267 and ISO 4037-1 standards resulted in HVL and filtration deviating from the values stated by IEC and ISO. This difference became less than 5 % for all RQR and ISO narrow beam qualities after modifications to filters were made.

The feasibility of the introduction of the IEC RQR reference radiations in the Hopewell systems X-ray equipment of the SSDL at KEBS was explored and confirmed in this work, although filters with mixed purity levels (99.9% and 99.99%) lower than the one recommended by the IEC standard (99.999%) were used.

The Hopewell System X-ray system at the KEBS' SSDL delivers exposure as designed over its operating range of 40 kV to 200kV and 2.5 mA to 15 mA. In this work, the exposure rate increased as the current and voltage are increased. The X-ray beam exposure was found to be uniform within 5% across a wide range of operating voltages.

6.1 Recommendation for Further Work

During the course of the research reported in this thesis, mixed lower purity (99.9 % and 99.99 %) aluminium filters were used. It could be worthwhile to continue this work so as to identify any metrology implications for using commercial filters that have purity levels a little bit lower than the one required by the standard. This was not possible in this study due to financial and equipment constraints. It is also necessary to perform a spectral analysis and the effect of the scatter components (probably using the Monte Carlo code) for each beam quality in order to completely characterise the beams. Further work is also suggested for the evaluation of the performance of the clinical X-ray systems downstream using transfer standards calibrated in the reference beams.

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APPENDICES

Appendix I

PTB Calibration Certificate Ref: 6.25-31/09K Dated 2009-07-22

Physikalisch-Technische Bundesanstalt

Braunschweig und Berlin



Kalibrierschein

Calibration Certificate

Gegenstand: Ionisation chamber with display unit
Object

Hersteller: PTW Freiburg
Manufacturer

Typ: Chamber: TW32002 S/N 00349
Type Unit: UNIDOS T10002 S/N 20707

Kennnummer: see above
Serial No

Auftraggeber: Kenya Bureau of Standards KEBS
Applicant Radiation Dosimetry Lab.
Kapiti Road, Off Road Mombasa
PO Box 00200 54974
Nairobi, Kenya

Anzahl der Seiten: 5
Number of pages

Geschäftszeichen: 6.25-31/09K
Reference No.

Kalibrierzeichen: Chamber 5897 Unit 5898
Calibration mark

Datum der Kalibrierung: 2009-07-22
Date of calibration

Im Auftrag: Braunschweig, 2009-07-24
On behalf of PTB

Bearbeiter:
Examiner

Siegel
Seal




Dr. L. Büermann


D. Jahns

Kalibrierscheine ohne Unterschrift und Siegel haben keine Gültigkeit. Dieser Kalibrierschein darf nur unverändert weiterverbreitet werden. Auszüge bedürfen der Genehmigung der Physikalisch-Technischen Bundesanstalt.
Calibration Certificates without signature and seal are not valid. This Calibration Certificate may not be reproduced other than in full. Extracts may be taken only with the permission of the Physikalisch-Technische Bundesanstalt.

1. General information

1.1 Scope of the calibration

Calibration of the ionisation chamber in terms of air kerma.

1.2 W - value

The reference value of the air kerma as obtained by the primary standard measurement is based on $(W/e)_{\text{air}} = (33,97 \pm 0,05) \text{ V}$.

1.3 Conditions prevailing during the calibration (see also 2.1)

1.3.1 Radiation

Gamma radiation from sources of the PTB.
X-radiation produced with constant potential generators.

1.3.2 Climatic conditions

temperature: 23,5°C to 24,1°C
air pressure: 997,6 hPa to 998,6 hPa
rel. humidity: around 50%

1.3.3 Geometrical arrangement

1.3.3.1 Direction of radiation incidence

The line on the chamber stem facing the radiation source.

1.3.3.2 Reference point of the ionisation chamber

Geometrical centre of the chamber.

1.3.3.3 Point of test

The reference point of the chamber was positioned in the central beam at a distance a (see 2.1) from the focal spot.

1.3.4 Leakage current

The effect of leakage currents was eliminated by appropriate corrections.

2. Results of the calibration

The calibration factor is the ratio of the conventional true value of the quantity to be measured to the indication of the instrument to be tested. The value of the air kerma, K_a , to be measured in units of Grays (Gy) is obtained from the reading, M :

$$K_a = N_K \cdot M \cdot k_Q \cdot k_p$$

N_K calibration factor in terms of air kerma,
reference conditions $T=20^\circ\text{C}$, $p=1013,25$ hPa.

k_Q correction factor for the radiation quality.

k_p correction factor for the density of air,
reference conditions $T=20^\circ\text{C}$, $p=1013,25$ hPa.

2.1 Calibration factor for the reference radiation quality and correction factors k_Q for other radiation qualities

2.1.1 Ionisation chamber TM32002 S/N 00349

- | | |
|---|--|
| Q radiation quality | d diameter of the radiation field at the point of test (50% isodose) |
| b additional filtration | k_Q correction factor for the radiation quality |
| s_1 first half value layer | K_a air kerma rate |
| a distance between source and point of test | U rel. Uncertainty of $N_K \cdot k_Q$ according to the 'Guide to the Expression of Uncertainty in Measurement' (ISO, 1995) as derived from the standard uncertainty by applying a coverage factor $k = 2$ |
| N_K calibration factor in terms of air kerma K_a for reference radiation quality S-Cs ⁺ ; potential of the high voltage electrode: ~0V potential of the collector electrode: +400V | |

$N_K = 2,524 \cdot 10^4$ Gy/C chamber on its own

$N'_K = 1,002$ chamber with UNIDOS T10001 S/N 20707, high dose rate range

$N''_K = 1,000$ chamber with UNIDOS T10001 S/N 20707, low dose rate range

Q*	b* in mm	s ₁ in		a in cm	d in cm	K _a in mGy/min	k _Q	U in %
		mm Al	mm Cu					
N-40	4,0 Al + 0,21 Cu	2,68	0,09	150	22,5	0,10	1,092	0,77
N-60	4,0 Al + 0,6 Cu	5,91	0,24	150	22,5	0,16	1,013	0,77
N-80	4,0 Al + 2,0 Cu	9,97	0,58	150	22,5	0,16	0,999	0,77
N-100	4,0 Al + 5,0 Cu	13,03	1,10	150	22,5	0,17	0,991	0,77
N-120	4,0 AL + 5,0 CU + 1,0 Sn	15,04	1,68	150	22,5	0,17	0,985	0,77
N-150	4,0 Al + 2,5 SN	16,58	2,33	150	22,5	0,16	0,979	0,77
N-250	4,0 Al + 2,0 Sn + 3,0 Pb	21,37	5,15	150	22,5	0,16	0,986	0,77
Nuklid S-Cs*	Energie in keV 662			150	38,8	0,36	1	1,0

- * Inherent filtration: 7 mm Be
- * denomination of radiation qualities according to ISO 4037 part 3, characterisation see ISO 4037 part 1

Die Physikalisch-Technische Bundesanstalt (PTB) in Braunschweig und Berlin ist das nationale Metrologieinstitut und die technische Oberbehörde der Bundesrepublik Deutschland für das Messwesen und Teile der Sicherheitstechnik. Die PTB gehört zum Dienstbereich des Bundesministeriums für Wirtschaft und Technologie. Sie erfüllt die Anforderungen an Kalibrier- und Prüflaboratorien auf der Grundlage der DIN EN ISO/IEC 17025.

Zentrale Aufgabe der PTB ist es, die gesetzlichen Einheiten in Übereinstimmung mit dem Internationalen Einheitensystem (SI) darzustellen, zu bewahren und – insbesondere im Rahmen des gesetzlichen und industriellen Messwesens – weiterzugeben. Die PTB steht damit an oberster Stelle der metrologischen Hierarchie in Deutschland. Kalibrierscheine der PTB dokumentieren die Rückführung des Kalibriergegenstandes auf nationale Normale.


Dieser Ergebnisbericht ist in Übereinstimmung mit den Kalibrier- und Messmöglichkeiten (CMCs), wie sie im Anhang C des gegenseitigen Abkommens (MRA) des Internationalen Komitees für Maße und Gewichte enthalten sind. Im Rahmen des MRA wird die Gültigkeit der Ergebnisberichte von allen teilnehmenden Instituten für die im Anhang C spezifizierten Messgrößen, Messbereiche und Messunsicherheiten gegenseitig anerkannt (nähere Informationen unter <http://www.bipm.org>).



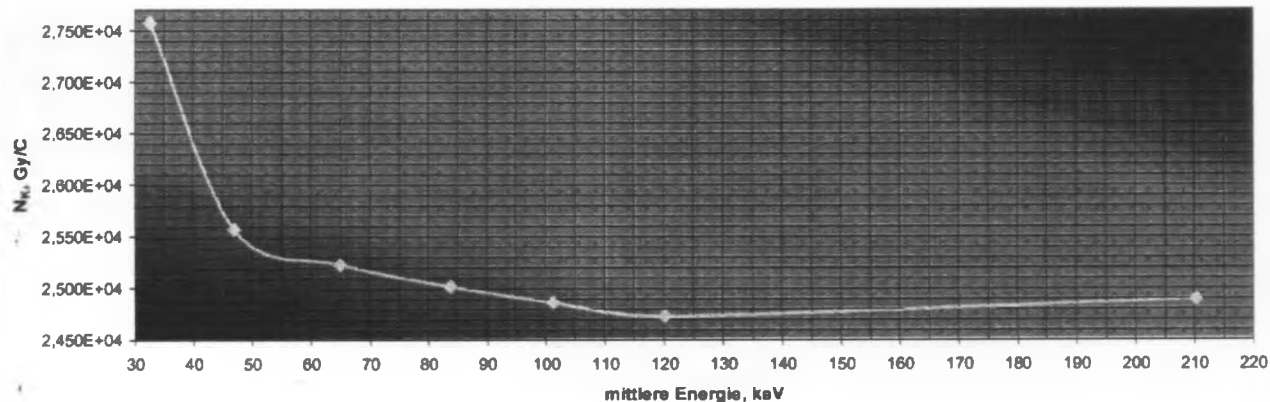
The Physikalisch-Technische Bundesanstalt (PTB) in Braunschweig and Berlin is Germany's National Metrology Institute and the supreme technical authority in the Federal Republic of Germany for metrology and certain sectors of safety engineering. The PTB comes under the auspices of the Federal Ministry of Economics and Technology. It meets the requirements for calibration and testing laboratories as defined in EN ISO/IEC 17025.

The central task of the PTB is to realize and maintain the legal units in compliance with the International System of Units (SI) and to disseminate them - in particular within the framework of legal and industrial metrology. The PTB thus is on top of the metrological hierarchy in Germany. The calibration certificates issued by the PTB document that the calibrated object is traceable to national standards.

This certificate is consistent with the Calibration and Measurement Capabilities (CMCs) included in Appendix C of the Mutual Recognition Arrangement (MRA) drawn up by the International Committee for Weights and Measures (CIPM). Under the MRA, all participating institutes recognize the validity of each other's calibration and measurement certificates for the quantities, ranges and measurement uncertainties specified in Appendix C (for details, see <http://www.bipm.org>).

Strahlungsqualität		mittlere Energie Luftkerma [keV]	Abstand zum Messort [cm]	Strahlungsfeld [cm]	Messung der Luftkerma						Temperatur		Druck	
PTB	ISO 4037 IEC 81267				Dosisleistung [mGy/min]	N_k [Gy/C]	HWS [mm Al]	HWS [mm Cu]	Filterung	k_a	von [°C]	bis [°C]	von [hPa]	bis [hPa]
A40	N-40	32,5	150	22,5	0,17	2,757E+04	2,68	0,09	4,0 mm Al + 0,21 mm Cu	1,092	23,5	24,1	997,6	998,6
A60	N-60	46,8	150	22,5	0,16	2,558E+04	5,91	0,24	4,0 mm Al + 0,6 mm Cu	1,013				
A80	N-80	64,9	150	22,5	0,16	2,522E+04	9,97	0,58	4,0 mm Al + 2,0 mm Cu	0,999				
A100	N-100	83,7	150	22,5	0,17	2,500E+04	13,03	1,1	4,0 mm Al + 5,0 mm Cu	0,991				
A120	N-120	101,1	150	22,5	0,17	2,485E+04	15,04	1,68	4,0 mm Al + 5,0 mm Cu + 1,0 mm Sn	0,985				
A150	N-150	120	150	22,5	0,16	2,472E+04	16,58	2,33	4,0 mm Al + 2,5 mm Sn	0,979				
A250	N-250	210,2	150	22,5	0,16	2,488E+04	21,37	5,15	4,0 mm Al + 2,0 mm Sn + 3,0 mm Pb	0,986				
S-Cs		662	150	38,8	0,36	2,524E+04					Datum der Kalibrierung			
											vom : 22.7.2009		bis : 23.7.2009	

TW 32002 SN 0243



A-Qualität

Bereichsfaktoren

Unidos T10002-20707

Messbereich	Referenz Messwert (Quotient)	Mittelwert (Bereichsfaktor)
Low 230 pC	1,000	1,000
High 24 nC	1,001	1,002

Appendix II

ISO 4037 Narrow Beam Qualities Established at KEBS SSDL

Table AII.1: The N40 Beam Quality Characteristics

mA	2.5	2.5	5.0	5.0	10.0	10.0	15.0	15.0	20.0	20.0
θ/°C	25.2	25.2	25.3	25.3	25.4	25.4	25.4	25.4	25.5	25.5
P/hPa	832.5	832.5	832.5	832.5	832.5	832.5	832.5	832.5	832.5	832.5
K_{PT}	1.23871	1.23871	1.23912	1.23912	1.23954	1.23954	1.23954	1.23954	1.23995	1.23995
Chamber	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh
Units	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s
Q1	1.083	0.7078	2.130	1.390	4.022	2.621	5.670	3.696	7.080	4.603
Q2	1.083	0.7077	2.132	1.393	4.030	2.635	5.677	3.703	7.083	4.609
Q3	1.083	0.7080	2.133	1.394	4.033	2.638	5.676	3.707	7.086	4.615
Q4	1.083	0.7086	2.133	1.394	4.034	2.639	5.678	3.703	7.084	4.613
Q5	1.084	0.7092	2.133	1.396	4.033	2.638	5.677	3.707	7.081	4.615
Q_{mean}	1.083	0.7083	2.132	1.393	4.030	2.634	5.676	3.703	7.083	4.611
SD	0.000	0.631	0.001	0.002	0.005	0.008	0.003	0.004	0.002	0.005
%sd	0.04%	0.09%	0.06%	0.16%	0.12%	0.29%	0.06%	0.12%	0.03%	0.11%
Q_{corr}	1.3418	0.8773	2.6421	1.7266	4.9958	3.2652	7.0351	4.5903	8.7823	5.7174
K_{air}(μGy/min)	34.148	38.923	67.240	38.944	127.144	38.939	179.044	39.005	223.510	39.093

Table AII.2: The N60 Beam Quality Characteristics

mA	2.5	2.5	5.0	5.0	10.0	10.0	15.0	15.0	20.0	20.0
θ/°c	26.1	26.1	26.0	26.0	26.0	26.0	26.0	26.0	26.0	26.0
P/hPa	829.5	829.5	829.5	829.5	829.5	829.5	829.5	829.5	829.5	829.5
K_{PT}	1.24694	1.24694	1.24652	1.24652	1.24652	1.24652	1.24652	1.24652	1.24652	1.24652
Chamber	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh
Units	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s
Q1	2.069	1.459	3.814	2.709	7.474	5.325	10.780	7.698	13.840	9.893
Q2	2.072	1.460	3.816	2.711	7.476	5.328	10.790	7.700	13.840	9.900
Q3	2.073	1.461	3.815	2.712	7.474	5.330	10.780	7.704	13.840	9.906
Q4	2.073	1.461	3.815	2.711	7.476	5.331	10.780	7.700	13.840	9.899
Q5	2.074	1.464	3.816	2.712	7.474	5.331	10.790	7.701	13.850	9.901
Q_{mean}	2.072	1.461	3.815	2.711	7.475	5.329	10.784	7.701	13.842	9.900
SD	0.002	0.002	0.001	0.001	0.001	0.003	0.005	0.002	0.004	0.005
%sd	0.09%	0.13%	0.02%	0.05%	0.01%	0.05%	0.05%	0.03%	0.03%	0.05%
Q_{corr}	2.5839	1.8218	4.7557	3.3793	9.3175	6.6427	13.4425	9.5990	17.2543	12.3403
K_{air} μGy/min	65.760	36.097	121.033	35.816	237.130	35.698	342.111	35.640	439.123	35.584

Table AII.3: The N80 Beam Quality Characteristics

mA	2.5	2.5	5.0	5.0	10.0	10.0	15.0	15.0	20.0	20.0
θ/°C	26.0	26.0	26.0	26.0	26.0	26.0	26.0	25.9	25.9	25.9
P/hPa	832.5	832.5	832.5	832.5	832.5	832.5	832.5	832.5	832.5	832.5
K_{PT}	1.24203	1.24203	1.24203	1.24203	1.24203	1.24203	1.24203	1.24161	1.24161	1.24161
Chamber	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh
Units	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s
Q1	1.080	0.597	2.143	1.207	4.135	2.374	5.991	3.458	7.705	4.468
Q2	1.082	0.598	2.144	1.210	4.138	2.375	5.996	3.462	7.706	4.472
Q3	1.082	0.599	2.145	1.212	4.138	2.378	5.995	3.463	7.707	4.472
Q4	1.081	0.601	2.144	1.219	4.138	2.379	5.998	3.463	7.707	4.471
Q5	1.081	0.602	2.144	1.222	4.138	2.379	5.996	3.463	7.707	4.472
Q_{mean}	1.081	0.599	2.144	1.214	4.137	2.377	5.995	3.462	7.706	4.471
SD	0.001	0.002	0.001	0.006	0.001	0.002	0.003	0.002	0.001	0.002
%sd	0.08%	0.35%	0.03%	0.52%	0.03%	0.10%	0.04%	0.06%	0.01%	0.04%
Q_{corr}	1.3429	0.7445	2.6629	1.5078	5.1388	2.9523	7.4462	4.2982	9.5684	5.5513
K_{air} μGy/min	34.176	45.907	67.771	44.946	130.782	44.298	189.506	44.089	243.515	43.867

Table AII.4: The N100 Beam Quality Characteristics

mA	2.5	2.5	5.0	5.0	10.0	10.0	15.0	15.0	20.0	20.0
θ/°C	27.0	27.0	27.1	27.1	27.1	27.1	27.1	27.1	27.2	27.2
P/hPa	833.5	833.5	833.5	833.5	833.5	833.5	833.5	833.5	833.5	833.5
K_{PT}	1.24469	1.24469	1.24510	1.24510	1.24510	1.24510	1.24510	1.24510	1.24551	1.24551
Chamber	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh
Units	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s
Q1	0.539	0.227	1.089	0.518	2.111	1.014	3.101	1.498	4.005	1.949
Q2	0.539	0.229	1.089	0.518	2.112	1.015	3.101	1.499	4.010	1.950
Q3	0.539	0.230	1.089	0.519	2.111	1.016	3.101	1.500	4.010	1.950
Q4	0.540	0.233	1.089	0.520	2.111	1.016	3.100	1.502	4.009	1.951
Q5	0.540	0.235	1.089	0.521	2.110	1.016	3.101	1.502	4.010	1.950
Q_{mean}	0.539	0.231	1.089	0.519	2.111	1.015	3.101	1.500	4.009	1.950
SD	0.000	0.003	0.000	0.001	0.001	0.001	0.000	0.002	0.002	0.001
%sd	0.05%	1.38%	0.00%	0.25%	0.03%	0.09%	0.01%	0.12%	0.05%	0.04%
Q_{corr}	0.6711	0.2873	1.3559	0.6465	2.6284	1.2643	3.8608	1.8679	4.9930	2.4288
K_{air} μGy/min	17.080	59.457	34.508	53.380	66.893	52.910	98.257	52.603	127.072	52.320

Table AII.5: The N120 Beam Quality Characteristics

mA	2.5	2.5	5.0	5.0	10.0	10.0	15.0	15.0	20.0	20.0
θ/°C	27.2	27.2	27.2	27.2	27.2	27.2	27.2	27.2	27.2	27.2
P/hPa	835.5	835.5	835.5	835.5	835.5	835.5	835.5	835.5	835.5	835.5
K_{PT}	1.24253	1.24253	1.24253	1.24253	1.24253	1.24253	1.24253	1.24253	1.24253	1.24253
Chamber	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh
Units	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s
Q1	0.539	0.227	1.089	0.518	2.111	1.014	3.101	1.498	4.005	1.949
Q2	0.539	0.229	1.089	0.518	2.112	1.015	3.101	1.499	4.010	1.950
Q3	0.539	0.230	1.089	0.519	2.111	1.016	3.101	1.500	4.010	1.950
Q4	0.539	0.233	1.089	0.520	2.111	1.016	3.101	1.502	4.009	1.951
Q5	0.540	0.235	1.089	0.521	2.110	1.016	3.100	1.502	4.010	1.950
Q_{mean}	0.539	0.231	1.089	0.519	2.111	1.015	3.101	1.500	4.009	1.950
SD	0.000	0.003	0.000	0.001	0.001	0.001	0.000	0.002	0.002	0.001
%sd	0.04%	1.38%	0.00%	0.25%	0.03%	0.09%	0.01%	0.12%	0.05%	0.04%
Q_{corr}	0.6698	0.2868	1.3531	0.6451	2.6230	1.2617	3.8528	1.8640	4.9811	2.4229
K_{air} μGy/min	17.048	59.446	34.437	53.380	66.755	52.910	98.055	52.603	126.768	52.320

Table AII.6: The N150 Beam Quality Characteristics

mA	2.5	2.5	5.0	5.0	10.0	10.0	15.0	15.0	20.0	20.0
θ/°C	22.3	22.3	22.3	22.3	22.3	22.3	22.3	22.3	22.2	22.2
P/hPa	830.5	830.5	830.5	830.5	830.5	830.5	830.5	830.5	830.5	830.5
K_{PT}	1.22962	1.22962	1.22962	1.22962	1.22962	1.22962	1.22962	1.22962	1.22920	1.22920
Chamber	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh
Units	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s
Q1	4.726	1.453	9.475	2.936	18.630	5.796	27.500	8.752	35.980	11.220
Q2	4.731	1.455	9.477	2.937	18.630	5.796	27.500	8.753	36.000	11.240
Q3	4.732	1.456	9.478	2.938	18.630	5.797	27.500	8.753	36.000	11.240
Q4	4.732	1.458	9.475	2.939	18.630	5.798	27.500	8.753	36.000	11.240
Q5	4.732	1.459	9.476	2.940	18.640	5.798	27.500	8.752	36.000	11.240
Q_{mean}	4.731	1.456	9.476	2.938	18.632	5.797	27.500	8.753	35.996	11.236
SD	0.003	0.002	0.001	0.002	0.004	0.001	0.000	0.001	0.009	0.009
%sd	0.06%	0.16%	0.01%	0.05%	0.02%	0.02%	0.00%	0.01%	0.02%	0.08%
Q_{corr}	5.8168	1.7906	11.6521	3.6126	22.9103	7.1281	33.8146	10.7624	44.2464	13.8113
K_{air} μGy/min	148.039	82.677	296.547	82.086	583.067	81.798	860.581	79.962	1126.072	81.532

Table AII.7: The N200 Beam Quality Characteristics*

mA	2.5	2.5	5.0	5.0	10.0	10.0	15.0	15.0
θ	22.1	22.1	22.1	22.1	22.0	22.0	22.0	22.0
P	830.5	830.5	830.5	830.5	830.5	830.5	830.5	830.5
kpt	1.22879	1.22879	1.22879	1.22879	1.22837	1.22837	1.22837	1.22837
Chamber	LS01	MonCh	LS01	MonCh	LS01	MonCh	LS01	MonCh
Units	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s	nC/min	nC/10s
Q1	1.725	0.783	3.444	1.567	6.779	3.103	10.020	4.610
Q2	1.727	0.784	3.444	1.568	6.780	3.105	10.020	4.612
Q3	1.725	0.783	3.444	1.570	6.780	3.105	10.020	4.611
Q4	1.727	0.785	3.443	1.571	6.780	3.106	10.020	4.612
Q5	1.727	0.785	3.444	1.571	6.779	3.106	10.020	4.612
Qmean	1.726	0.784	3.444	1.569	6.780	3.105	10.020	4.611
SD	0.001	0.001	0.000	0.002	0.001	0.001	0.000	0.001
%sd	0.06%	0.13%	0.01%	0.12%	0.01%	0.04%	0.00%	0.02%
Qcorr	2.1211	0.9634	4.2317	1.9285	8.3279	3.8141	12.3083	5.6645
Kair μGy/min	52.392	54.384	104.523	54.200	205.698	53.931	304.015	53.670

*X-ray system restricted to 15mA when used with a peak voltage of 200kV. This is designed to protect the tube and elongate its life span. The MPI control console cannot therefore permit selection beyond this value.

Appendix III

X-Ray Set Up

Figure A3.1: Pictures of X-ray setup



(a) A picture of the X-ray equipment layout at KEBS



(b) Visual display for temperature monitoring



(c) The PTW UNIDOS and UNIDOS E electrometers for charge display



(d) The MPI control console for settings of kilovoltage (kV) and current (mA)