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CALIBRATION OF MAMMOGRAPHY IONISATION CHAMBERS

Maria Steenkamp

A research report submitted to the Faculty of Science, University of the Witwatersrand, Johannesburg, in partial fulfilment of the requirements for the degree of Master of Science.

Johannesburg, 2007

DECLARATION

I declare that this research report is my own, unaided work. It is being submitted for the degree of Master of Science in the University of the Witwatersrand, Johannesburg. It has not been submitted before for any degree or examination in any other University.

_____

_____day of _____200_____

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Sincere gratitude to the following people:

University of the Witwatersrand for the opportunity granted to enrol and exploit the expertise and clinical research facilities.

Prof. D.G. van der Merwe for expertise, skills, personal guidance and full support through out this study.

No pupil is greater than his teacher is,
But every pupil, when he has completed his training,
Will be like his teacher **Luke 6 verse 40**

Tshwane University of Technology for granting me lecturer- relief and sabbatical leave to be able to embark on a comprehensive study.

Fellow students for their assistance and moral support.

To my **husband** and **sons** for their belief in my work and tolerance during times when our lives were turned upside down.

My Creator, it is with **GRATITUDE** that I think of and thank **You** for enabling me to complete and achieve this academic goal.

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ABSRACT

The South African national calibration facility is currently not equipped for mammography, dose-measuring equipment. A therapy X-ray machine was used as a calibration unit at the national secondary standards dosimetry laboratory (SSDL) for medium and low energy X-ray, therapy calibrations. It is not necessarily intuitive that the latter calibrations are applicable to diagnostic X-ray beams generated by high frequency generators. The response of measuring equipment calibrated in a therapy X-ray beam, compared to its response in a diagnostic or clinical mammography unit, is unknown.

The aim of the research was to investigate whether there was a measurable difference between the X-ray beam qualities available for low energy diagnostic radiology and radiation therapy, i.e. up to 100 kV. The beam qualities studied included both mammography and conventional diagnostic radiography, i.e. nominally 20 kV to 100 kV. The diagnostic and therapy X-ray tubes under investigation had different target-filter combinations, inherent filtration and theoretically, different X-ray spectra.

Practically, spectrometry of X-ray beams is not possible because of the sophistication of the instrumentation, comprehensive analyses being very time consuming and not practically applicable to the clinical environment (Kharatti and Zarrad, 2003). Furthermore, not all SSDL's or Hospitals have access to spectral analysers. Clinical beam quality is instead specified in terms of both the tube peak voltage and the half-value layer (HVL), the thickness of material that will reduce the maximum output of the X-ray beam to 50%.

The goal was to compare measured HVLl's to the ones recommended by the International Electro technical Commission (IEC-61267, 2005) for available mammography beam qualities. The method was validated using attenuation curves. The attenuation curves were then used to derive the suitability of the X-ray spectra for calibration of mammography ionisation chambers (Waggener and Blough, 1999). One of the low energy therapy units was found to be suitable for introducing a regional calibration service for mammography.

INTRODUCTION

Wilhelm Conrad Roentgen discovered X-rays on 8 November 1895, while experimenting with a discharge tube and noticing a glow on a piece of glass covered with zinc sulphide. Roentgen called the undiscovered radiation X-rays, the "X" being the established symbol for an unknown quality. By the 28 November 1895, he had thoroughly investigated the properties of these rays and had prepared a manuscript describing his experiments (Krane K S, 1988). In recognition of his outstanding contribution to science, Wilhelm Conrad Roentgen was awarded the first Nobel Prize for Physics in 1901.

Today physicians use techniques learned from nuclear physics experiments to perform diagnosis and therapy in the human body. "None other field of science comes to mind in which theory encompasses such a broad spectrum, ranging from the most microscopic to the cosmic, neither is there another field in which direct applications of basic research contain the ultimate limits to be beneficial or detrimental"(Krane K S, 1988).

Soon after the discovery of X-rays and their application in medicine, it was realised that this radiation could be dangerous. A quantitative relationship between the radiation and its physical and biological effects needed to be established. This entailed definition and acceptance of a suitable unit for measuring X-ray energy.

In diagnostic radiology, there has been an increased demand for clinical radiation absorbed dose measurements (dosimetry). Radiation, measurement accuracy, requirements for diagnostic radiology is usually more stringent than those for health physics and less than those for therapy (Sunde, 1992). In diagnostic radiology, different beam qualities have been applied as compared to those for radiation therapy. Quantities such as organ and effective dose are of interest and the uncertainties allowed are much larger in radiology. However, these quantities still refer to radiation received by human tissue and are analogous to absorbed dose and kerma (energy absorbed or transferred per unit mass of material) in a phantom in radiotherapy dosimetry.

Dosimetry requires a clear definition of the incident beam quality. Traditionally, beam quality for X-rays is specified by the peak tube operating voltage of the X-ray machine and the first half-value layer (HVL1). "The first half-value layer, is defined as the thickness of the specified material, which attenuates the air kerma or air kerma rate in the beam to one half of its original value measured without any absorber. The contribution of all scattered radiation, other than any which might be present in the beam concerned, is determined to be excluded" (ICRU report 74, 2005). In practice HVL1, is the thickness of material that will reduce the X-ray beam energy to 50%. Reduction of X-ray beam output occurs because the low energies, having little penetrating power, are easily attenuated in the material through which they pass. Different X-ray beams may under certain circumstances has the same HVL1 but differ in the second half-value layer (HVL2) because of the heterogeneity of the X-ray beam spectrum. The heterogeneity of the X-ray beam spectrum can be expressed in terms of the heterogeneity coefficient (HC), that is the ratio of the HVL1 to the HVL2 (Zoetelief *et al,* 1996).
The HC for a homogenous X-ray beam is unity. A heterogeneous X-ray beam generally has a HC less than 1.00 (Zoetelief *et al,* 1996). The reason for this phenomenon is that the X-ray beam hardens with additional filtration and therefore the HVL2 is always thicker than the HVL1. The beam quality of an X-ray beam is also affected by other parameters related to typical diagnostic and therapeutic X-ray machines e.g. different target/filter combinations, inherent filtration, collimation, scatter, geometric factors, etc.

The ideal would be an international protocol of standard recommendations for dosimetry in Diagnostic Radiology. Standardised quantities and units applicable to diagnostic radiography, as well as the practical aspects of accurate measurement are however still outstanding. The above requirements were recognised by the SSDL of the IAEA/WHO during a meeting of the Scientific Committee, 1988: " Dosimetry measurements of diagnostic radiology, radiation protection and environmental exposures do not require the accuracy of therapy-level measurements, but do require a coherent relationship to the international measurement system" (Freitas and Drexler, 1992). These include the definition and acceptance of various application-specific dosimetric quantities and units, and include the establishment of national diagnostic radiology calibration facilities. Currently there is no official international protocol

addressing dosimetry in diagnostic radiology although the IAEA (IAEA-CN-96/39) and the IEC (IEC- 61267, 2005) are currently addressing this.

Medical Exposures and Population dose

Medical ionising radiation from diagnostic X-ray procedures contributes about 90 % to the population dose from man-made sources (Pernicka *et al*, 2000). Medical exposures are however, not the source of the highest radiation dose to the general population. Radon is the main contributor to doses from environmental radiation.

There is a larger possibility that gamma rays will pass through an individual's body from environmental sources, like soil and building materials and inside the human body, potassium-40 atoms disintegrate emitting beta particles and gamma rays. The latter radiation is grouped under natural radiation, contributing by far the largest percentage to the annual dose of the population (Martin A and Harbison S A, 1986).

Risks associated with Medical Exposure

Experimental evidence of a "safe level of radiation" for prevention of the two most important long-term radiation effects, namely carcinogenesis and mutagenesis is non-existent. It is important that the risks associated with medical examinations are therefore expressed more quantitatively. When diagnostic X-rays are used correctly and doses are controlled carefully, risks are acceptable within the broader context of both the clinical value of the diagnostic information and the risks associated with normal daily living. The detrimental effects of radiation arising from diagnostic radiology are currently estimated to be small in comparison with the potential benefits resulting from the diagnostic information gained (Rainbow *et al,* 1992).

The use of mammography has increased rapidly in many countries. As the female breast seems to be a comparatively radiosensitive organ, optimisation of the radiographic procedure is crucial (Thilander *et al,* 1992). There is a small but significant risk of radiation-induced carcinogenesis associated with the mammography examination and it is important that it is estimated (Carlsson and Dance, 1992).

Limitation of Medical Exposure

Since medical X-ray examinations contribute greatly to the population dose from man-made radiation sources, a requirement exists to optimise the dose, the design and use of X-ray imaging systems (Zoetelief *et al,* 2003). Dose limitation is a responsibility of all participants involved in medical exposures and that entails establishing, using and assessing dose to determine national diagnostic reference levels (DRL's) (ICRP 60, 1991). Measurement of the dosimetric parameters related to the performance of the equipment plays a predominant role in determining the lowest possible dose to either the patient or the staff. (Zoetelief *et al,* 2003).

Radiation Measurements and Instrument Performance

The European Council Directive 97/43/EURATOM (Directive 84/466 Euratom, 1997) on health protection of individuals against the dangers of ionising radiation in relation to medical exposures provides various measures in order to control patient dose from radio-diagnosis including mammography. Dosimeters are important instruments to assess this dose (Witzani J, 2004). An additional objective of dosimetry in medical imaging is the assessment of equipment performance. Dosimetry for medical X-ray imaging includes the need for international standardisation on the practical aspects of instrument calibration and clinical measurement methodology in addition to international recommendations on appropriate radiation quantities and units. (Zoetelief *et al*, 2003).

The role of the ICRU, IAEA and IEC

The present situation in dosimetry for medical imaging indicates clearly the need for international recommendations on appropriate diagnostic radiation quantities and units. The International Commission on Radiation Units and Measurements (ICRU) defines application specific quantities for dosimetry in medical X-ray imaging, new symbols for these units and recommendations for the specification of X-ray beams in terms of both radiation quality and intensity (ICRU, Report 74, 2005). Table 1 shows different radiation quantities recommended for mammography and their application with reference to the patient and radiation worker (Zoetelief *et al*, 2003).

Table 1**: Quantities for Dosimetry in Medical X-ray Imaging**

Radiation Quantities	**Patient related Quantities**	**Worker related Quantities**
Air kerma Free-in-air	Incident air kerma (without backscatter)	Organ dose
Air kerma-area product	Entrance air kerma (including backscatter)	Effective dose

The International Atomic Energy Agency (IAEA) is currently compiling and testing an international code of practice for dosimetry in diagnostic radiology. The main objectives are to achieve and maintain a high level of quality in dosimetry, to improve the implementation of traceable standards at the national level and to ensure the control of dose in medical X-ray imaging, worldwide (Zoetelief *et al*, 2003). The IAEA is emphasizing the practical aspects of establishing calibration facilities at SSDL's in order to disseminate a higher accuracy in clinical diagnostic dosimetry.

Rosser (Rosser, 1998) showed that if only generating tube potential and HVL represented beam quality, there was little consensus in the beam qualities for low- and medium-energy radiotherapy units throughout the UK when comparing the Hospitals with the National Physical Laboratory (NPL) beam qualities.

The International Electrical Commission, IEC, also recommended RQA-M qualities but in most cases, they are not used (IEC-61267, 2005). The standard radiation qualities recommended for mammography calibrations, are listed in table 2.

Table 2**: Characterization of standard RADIATION QUALITIES RQR-M 1 to RQR-M 4.**

Standard Radiation Quality	**X-RAY TUBE VOLTAGE in kV**	**First HALF-VALUE LAYER In mm Aluminium**
RQR-M 1	25	0.28+/- 0.02
RQR-M 2	28	0.31+/- 0.02
RQR-M 3	30	0.33+/- 0.02
RQR-M 4	35	0.36+/- 0.02

The standard radiation conditions for the RQR-M series are relevant to:

- An emitting target of molybdenum (Mo),
- An X-ray tube voltage with a ripple of not more than 4% and
- A total filtration of 0.032mm +/- 0.002mm Mo in the X-ray source assembly

The Role of Secondary Standards Dosimetry Laboratory (SSDL)

In South Africa, the aim is to establish a diagnostic radiology calibration facility and develop national DRL's (SAF/6/010). The SSDL situated at the South African Council for Scientific and Industrial Research (CSIR) National Metrology Laboratory (NML), currently calibrates most of the national measuring equipment used in therapy and radiation protection but no diagnostic radiology service is offered currently.

The objectives of the SSDL are to calibrate field radiation measuring instruments, provide traceability from primary standard dosimetry laboratories (PSDLs) and advise on all aspects of ionising radiation metrology. Services of PSDLs are evolving constantly to keep pace with advances in dosimetry.

A direct consequence of the latter is that each national or regional SSDL must adapt its work to provide the best results under the different circumstances. Currently an instrument is calibrated at a range of appropriate beam qualities available at the CSIR-NML, and the response is determined as a function of the peak voltages and the HVL1 of the laboratory unit. The user then interpolates between these calibration points in the field. These calibrations are

valid for two years and a calibration certificate is issued for the specific measuring equipment. The calibration certificate is an official document used in the clinical environment as an indication of the performance of the field-measuring instrument. Calibrations are directly traceable to primary standards held by Primary Standards Dosimetry Laboratories (PSDLs). Several PSDLs have demonstrated mutual equivalence of their primary standards for most radiation beam qualities employed in radiology (Zoetelief *et al*, 2003). The primary standards for radiation qualities should be developed to match the requirements of the clinical application (Zoetelief *et al*, 2003).

In South Africa, the SSDL has a technical co-operation project with the IAEA to set up diagnostic radiology capabilities. The main objectives of the project are:

- To realize a national measurement standard in the field of diagnostic radiology traceable to the SI units.
- To set up a facility to produce diagnostic radiology X-ray beams in line with IAEA Code of Practice for diagnostic radiology and IEC-61267 in order to disseminate traceability from the abovementioned national measurement standard, thereby enabling the delivery of controlled radiation doses in hospitals.
- To set up a system to disseminate traceability from the abovementioned national measurement standard in line with the ISO/IEC 17025 and the IAEA Code of Practice on dosimetry in diagnostic radiology.

Low- energy therapy X-ray beams

The history of radiotherapy with reference to low energy X-ray beams dates back to the beginning of 1900's. Conventional kilovoltage X-ray therapy machines were used for external beam radiotherapy in the treatment of superficial lesions. Low energy therapy X-ray machines operate at tube voltages up to100 kV. This type of beam produces a very rapidly decreasing depth dose. The skin surface is maximally irradiated but the underlying tissues are spared to an increasing degree with depth. Treatment is limited to about 1-2 mm of tissue and complete absorption of the beam is within 20 mm of soft tissue (Kahn F M, 1992).

A range of beam qualities is normally available from the commercial suppliers of kilovoltage units, and the choice is in principle, arbitrary. Relative dosimetry is then obtained from interpolation of tabulated beam data available (Br.J.Radiol.Suppl 25,1996) or (Rosser K E, 1998), for instance. Little standardisation therefore exists between clinical units in general.

Low-energy diagnostic X-ray beams

Most diagnostic X-ray beams are operated at peak tube voltages ranging from 20-150 kV. Different diagnostic examinations and procedures require the availability of this range of tube voltage. Conventional diagnostic examinations are performed at 50- 150 kV. The image contrast is obtained from the differential X-ray attenuation by bone (high atomic number atoms), air and soft tissue (low atomic number atoms). The amount of beam filtration also plays a minor role in optimising patient dose against image quality. In general, the only restriction in beam quality is the requirement of a minimum of 2.5 mm Al total tube filtration. The low-energy range of diagnostic X-ray beams of 20-40 kV peak tube voltage is exclusively applicable to mammography, an examination that is designed especially for detecting breast pathology. In mammography, the average tissue (glandular) dose is determined from measured kerma (free in air) and various correction and conversion factors. The latter mentioned factors can be obtained from published tables that provide information of the breast entrance skin dose, X-ray tube target/filter combinations employed e.g. Mo/Mo, Mo/Al, Mo/Rh, Rh/Rh, W/Al, etc., beam quality (HVL), breast thickness, compression and the age of the patient (Kharrati and Zarrad, 2003).

OBJECTIVES AND JUSTIFICATION

Nationally measuring equipment (ion chambers and electrometers) is calibrated at the CSIR-NML, using a constant voltage laboratory, therapy X-ray machine. It is not necessarily intuitive that the latter unit would be applicable to clinical diagnostic X-ray beams generated by high frequency generators of different configurations.

This research was undertaken to establish the difference in beam quality between locally-available low-energy diagnostic radiology and radiation therapy X-ray machines with different target-filter combinations and inherent filtrations and theoretically, different X-ray spectra. This investigation attempted to identify whether the recommended mammography beam qualities could then be simulated in order to introduce national mammography ionisation chamber calibrations.

METHODS AND MATERIALS

Four different X-ray machines were investigated, namely two clinical mammography units and two low-energy therapy X-ray machines. The therapy units were operated at constant tube current for all measurements. One of the latter units was the national SSDL unit. The other therapy X-ray machine was a clinical unit. Details of the X-ray units and their parameter settings used in this work are summarised in Table 3, below.

Table 3**: Characteristics of Diagnostic and Therapy X-ray Machines**

MANUFACTURER AND MODEL	**TARGET-FILTER COMBINATION**	**ADDED FILTRATION**	**APPLIED VOLTAGE kV_p**	**TUBE OUTPUT / CURRENT**
Philips Mammo Diagnost UC™	Mo/Mo Mo/Al	0.15 mm Mo 0.15 mm Al	26 36	63 mAs 16 mAs
Bennett Contour M-CTR™	Mo/Mo Mo/Rh	0.15 mm Mo 0.15 mm Rh	36 34	16 mAs 24 mAs
Philips RT 100™ Clinical Therapy X-ray Unit	W/Al	0.15 mm Al 0.30 mm Al 0.40 mm Al 0.55 mm Al 0.78 mm Al 1.25 mm Al 1.75 mm Al	20 30 37 45 55 70 100	10 mA 10 mA 10 mA 10 mA 10 mA 10 mA 8 mA
Pantak Laboratory X-ray Unit	W/Al	2 mm Be	50 60 70 100	20 mA 20 mA 20 mA 20 mA

The experimental research entailed collection of attenuation data through known thicknesses of high purity (99.9%) aluminium filters (attenuators) for all beams. HVL1 and HVL2 were measured and the HC was derived.

EXPERIMENTAL CONDITIONS

The measurements on the mammography units were done with a PTW UNIDOS-E electrometer and a 1.0 cc parallel-plate ionisation chamber M 77334, with a calibration certificate issued by PTW Freiburg Germany [Certificate No. 025019]. This ionisation chamber is rated as being suitable as a diagnostic dosemeter for radiography measurements, including mammography. Its mammography calibration range is from **25 kV to 35 kV**, with HVLs ranging between **0.26 mm Al to 0.65 mm Al**. The calibration factor for the 1 cc flat mammography ionisation chamber:

$$\mathbf{N_a = 1.964\ E\ +07\ Gy/C}$$

The measurements on the therapy units were done with a PTW IQ4 electrometer and a 0.6 cc ionisation chamber M 23333-1608 calibrated at the national SSDL. This ionisation chamber is rated as being suitable as a therapy dosemeter. Its calibration range suitable for the research done is given in table 4.

Table 4**: Calibration Data for IQ4 Dosemeter applicable to Low-Energy Therapy X-ray beams.**

Beam Quality **kV**	**Beam Filtration: Inherent (mm Al) + Additional (mm Perspex + mm Al)**	**HVL1** **mm Al**	**Air-kerma** **Gy / displayed V**	**Air-kerma rate** **mGy /min**	**Calibration Factor** **E+09 Gy / C**
50	0.11 + 4.6 + 0.34	1 mm Al	0.0475 +/- 2%	83	435.90
100	0.11 + 4.6 + 0.74	4 mm Al	0.0460 +/- 2%	109	422.14

HVL MEASUREMENTS

Figure 1 illustrates the schematic arrangements of the apparatus for the HVL-measurements for the therapy X-ray apparatus.

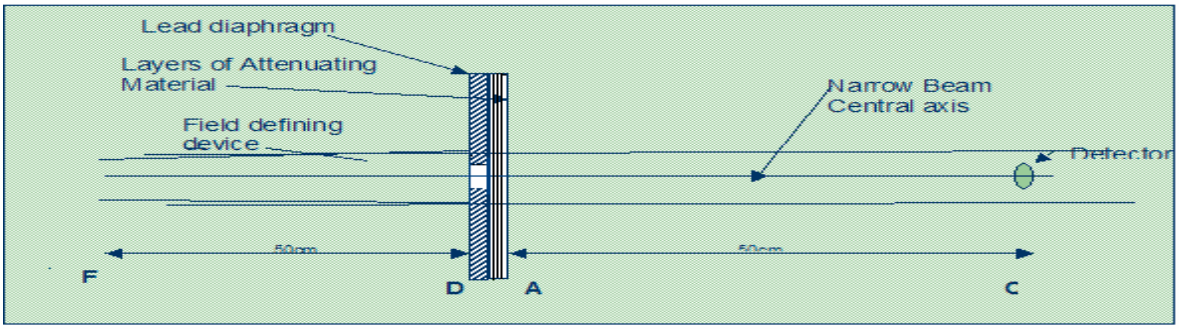


Figure: 1 , Schematic presentation for HVL measurements: therapy X-ray beams.

Owing to the ability to manoeuvre the X-ray tube mounting, the experimental apparatus consisted of a laboratory optical bench with mountings to support the lead diaphragm and aluminium filters. The 0.6 cc ionisation chamber was used for monitoring the X-ray beam attenuation measurements. The narrow beam or “good geometry” condition was the main requirement to produce accurate attenuation measurement data. In practice this requirement was fulfilled by limitation of the X-ray beam divergence with the aid of a 50 cm FSD (film to source distance) applicator that limited the radiation field to a radius of 2.5 cm at the aperture of the diaphragm. The 2.2 cm aperture in the lead diaphragm further limited the radiation field to just beyond the outer dimensions of the sensitive volume of the ionisation chamber to ensure minimal contribution from scattered radiation to the measuring chamber. The alignment of the set-up as well as the centring of the ionisation chamber was done by attaching an X-ray film into a fixed chamber holder. Exposure at the lowest tube load showed a circular pattern that was comparable with both the centre of the diaphragm and ionisation chamber. Aluminium

filters (attenuators) of 99.9% purity according to ISO 2092, were positioned midway between the source and the ionisation chamber for the therapy X-ray beams.
Fixed tube loadings (exposure settings) were used to establish the HVL1 and HVL2.

Figure 2 illustrates the schematic arrangements of the apparatus for the HVL-measurements for the mammography X-ray apparatus

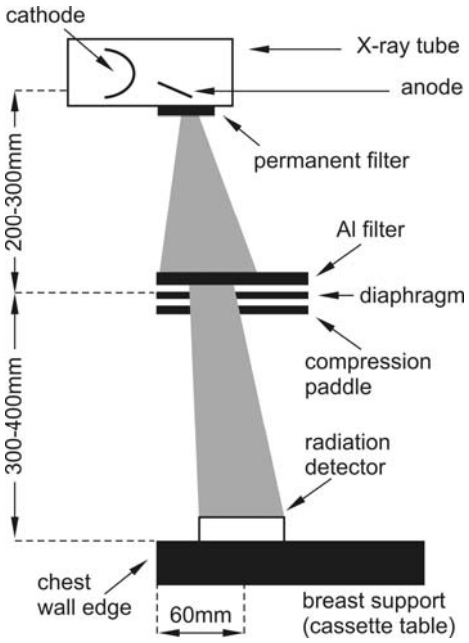


Figure 2: Schematic of the HVL measurements for the mammography Beams [European Protocol, 1996]

In the case of the diagnostic mammography X-ray beams, the 1.0 cc ionisation chamber was placed on top of the breast support with its centre 4.4 cm above the cassette table, 6.0 cm from the chest wall edge and centred laterally. The compression plate was placed between the detector and the X-ray tube. The manual exposure mode was selected. A lead diaphragm 5 mm thick with an aperture small enough to limit the radiation to just beyond the outer dimensions of the ionisation chamber, was used. The lead diaphragm was placed on top of the compression paddle approximately 30.0 cm from the focal spot (European Protocol, 1996).

The latter arrangement satisfied the conditions for good geometry. X-ray films were exposed during the set-up stage to ensure alignment of the beam to the sensitive volume of the ionisation chamber.

The current IAEA recommendation for HVL measurements for mammography requires that the detector be positioned more above the table to avoid backscattered radiation to reach ionisation chamber. The distance from the chest wall is slightly less than used in this work. Measurements would have been different due to different geometrical arrangements but final result the same.

Aluminium filters (attenuators) of 99.9% purity according to ISO 2092 were positioned between the source and the ionisation chamber for the mammography X-ray beams. In this case, a source-chamber distance of at least 60 cm was used owing to the physical restriction imposed by the clinical unit. Fixed tube loadings (exposure settings) were used to establish the HVL1 and HVL2. Two different target-filter combinations per mammography unit were investigated.

GRAPHYCAL ANALYSIS

The computation of the attenuation data was based on the following theory.
The attenuation of a homogeneous photon beam is exponential and given by the formula:

$$K_d = K_O e^{-\mu t} \quad \ldots\ldots\ldots\

RESULTS

The therapeutic and mammography (diagnostic) data obtained were analysed according to the HVL1, HVL2 and HC per beam quality. Since this research attempted to identify whether the IEC-recommended mammography beam qualities could be simulated, the IEC data points (Table 2) are also shown.

Figure 3 shows the HVL1 values obtained for each X-ray machine over the selection of available peak tube voltages

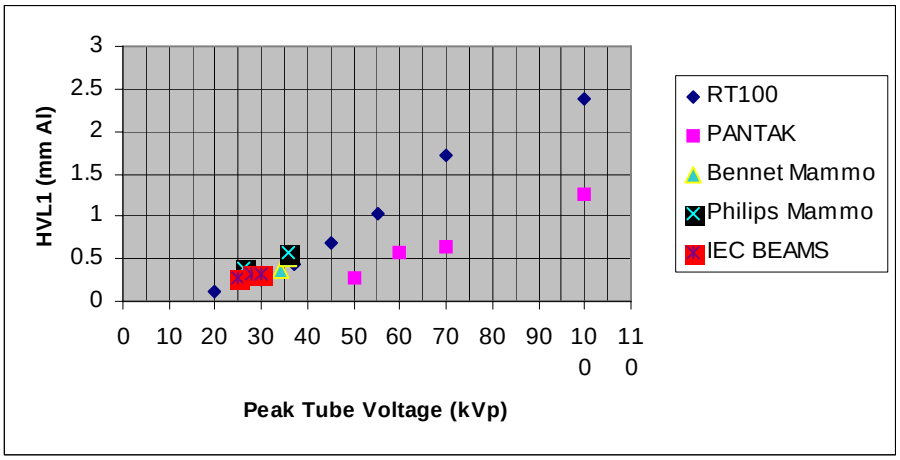


Figure 3**:** Graph of HVL1 versus Peak tube voltage (kVp) for all the units studied, including the IEC calibration points.

The PANTAK, laboratory X-ray machine used at the CSIR NML for calibration of therapy measuring equipment was unstable at beam qualities lower than 50 kVp. The purpose of the Laboratory X-ray machine was originally to provide calibrations for low- and medium-energy therapy dosimetry. The output from this unit at energies below 50 kVp was unstable and it was therefore considered unsuitable for further investigation in this work.

The HVL1 values of the RT100^{TM} were close enough to exercise regression in order to examine the possibility of adapting the unit for mammography calibrations.

Figure 4 is an expanded view of the energy range of interest for this research. Only the clinical X-ray machines and the IEC-recommended beam qualities are indicated. Clearly, the RT100^{TM} required little adaptation to simulate the mammography calibration points.

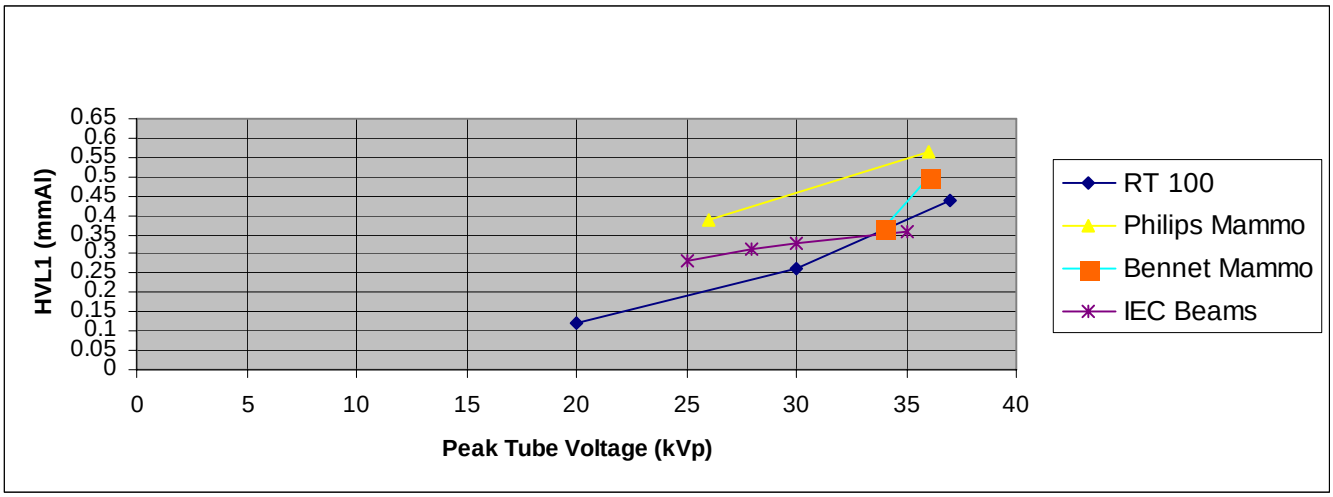


Figure 4: Graph of HVL1 versus Peak Tube Voltage (kVp)

An attempt to add further specification to beam quality was performed in terms of the HVL2 values. Figure 5 shows HVL2 as a function of peak tube voltage over the energy range of interest.

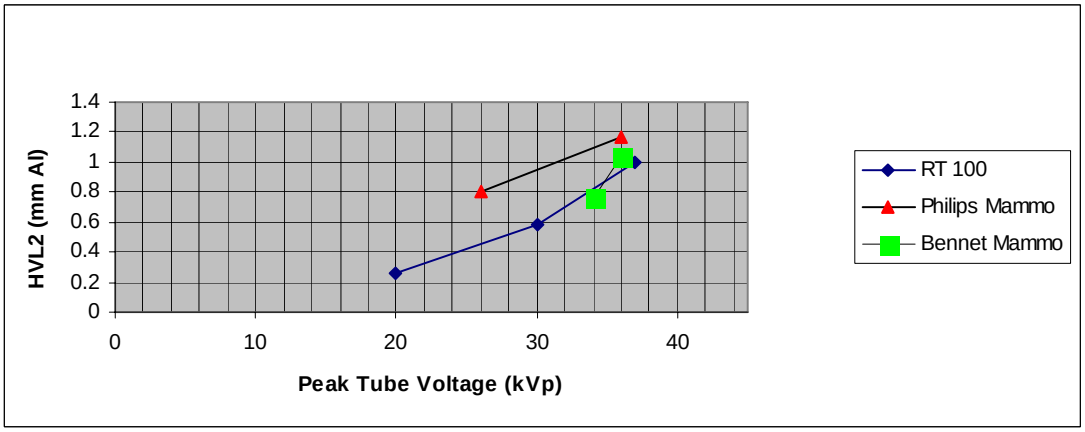


Figure 5: Graph of HVL2 versus Peak Tube Voltage (kVp)

After determination of the HVL1 and HVL2 values for the different X-ray beam spectra, the HC (homogeneity coefficient) was calculated. The mean value and standard deviation of the HC for all the mammography beams was calculated to be 0.48 +/- 0.00 and for the RT100^{TM}, 0.49 +/- 0.03. The homogeneity coefficient for the full X-ray beam energy range was calculated as 0.49 +/- 0.01.

Manipulation, by shifting of the RT100^{TM} raw data was then carried out in order to adjust its HVL1 to match the best fit of the IEC data points together with the clinical mammography units' data points. Figure 6 shows the normalised electrometer readings versus the attenuator (filter) thicknesses per available beam quality relevant to the mammography energy range.

Mathematical functions were used, supplied by ***MATHEMATICA*** soft ware, for determining the best line fit through the data points in order to determine the required increase or decrease in the HVL1 and HVL2 values. This mathematical manipulation is illustrated by figure 6.

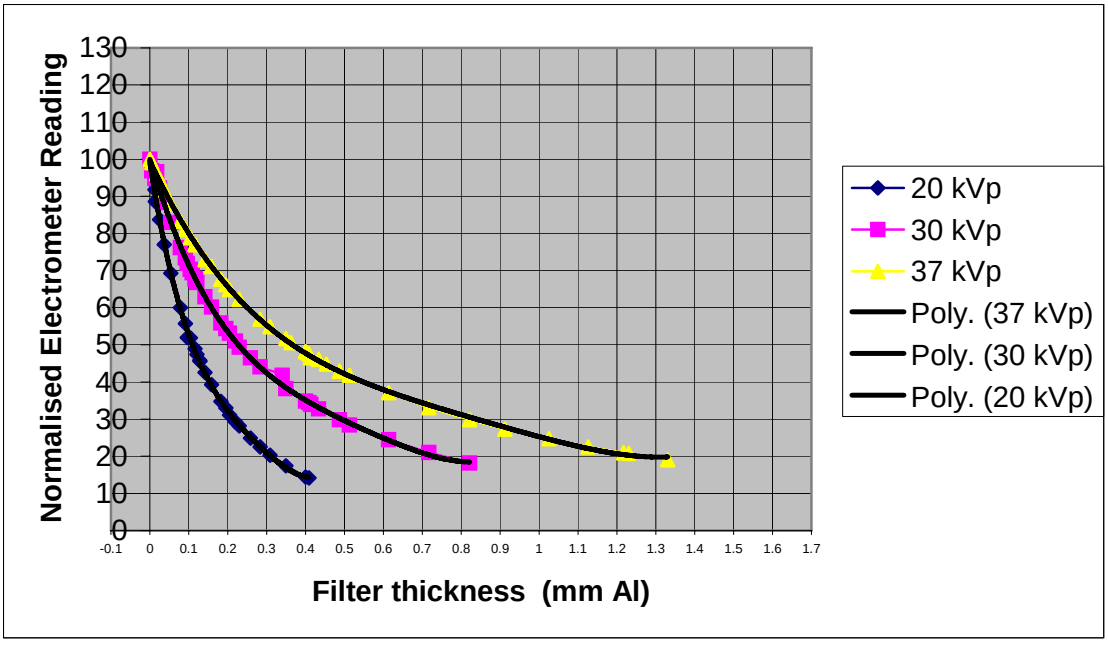


Figure 6: Graph of Normalised Electrometer Readings versus Filter Thickness (mm Al)

The RT100^{TM} data points obtained for 20 kVp, 30 kVp and 37 kVp were fit to a fourth degree polynomial. The fitted polynomial was found to be significant with $R^2 = 0.9997$ at 37 kVp and $R^2 = 0.9990$ at 30 kVp. The filter thickness required to produce the required HVL1 could then be computed. A second fourth degree polynomial function of the form $x = b_0 + b_1y + b_2y^2 + b_3y^3 + b_4y^4$ was also fit to the data and these approximations were found to be significant. The filter thickness was found to be -0.1 mm Al at 37 kVp and +0.1 mm Al at 20 kVp and 30 kVp.

Figure 7 shows the calculated HVL1 data of the RT100^{TM} obtained from the change to the filter thickness. It can be seen that the RT100^{TM} data points would then fit to the mean value of the measured clinical data and the IEC recommended points.

Figure 7: Graph of **corrected** RT100^{TM} HVL1* versus Peak Tube Voltage (kVp)

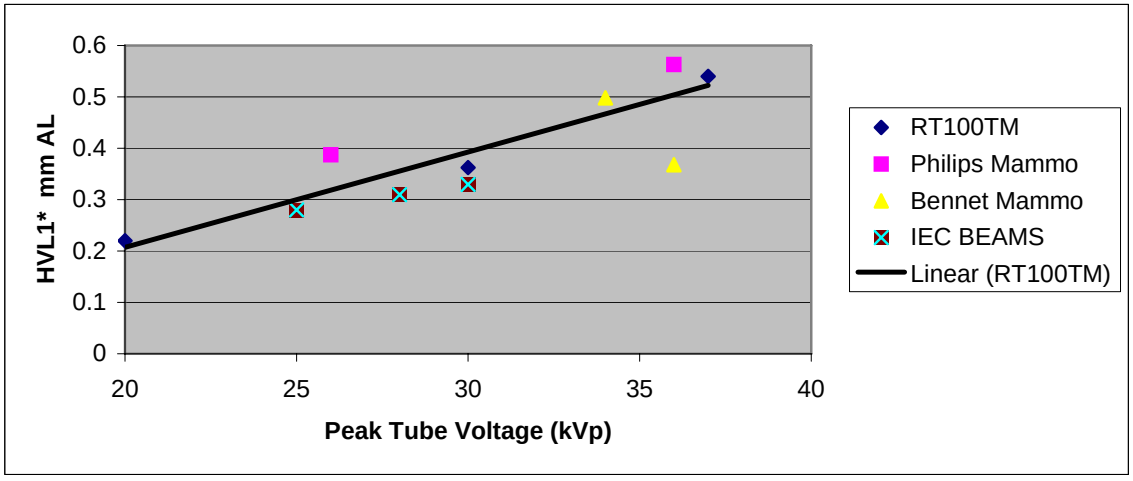


Figure 8 similarly illustrates the impact of this change in filtration to the HVL2 values.

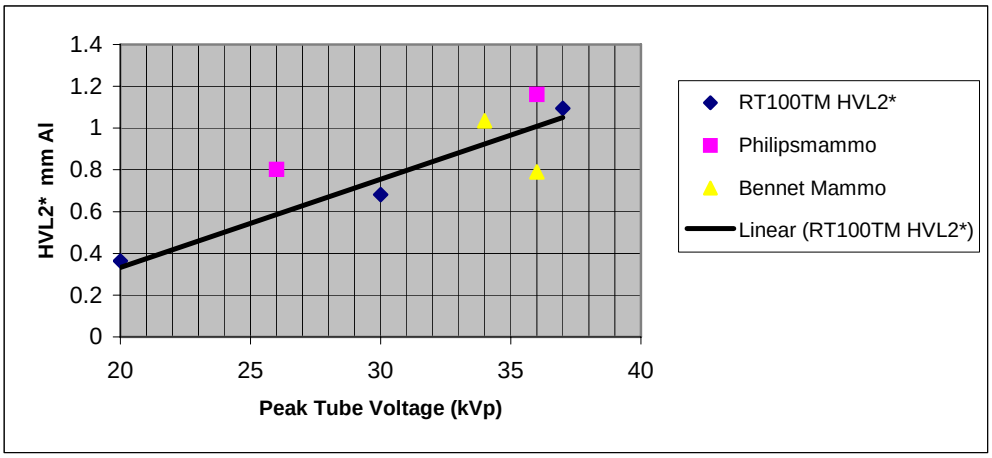


Figure 8: Graph of **corrected** RT 100^{TM} HVL2* versus Peak Tube Voltage (kVp)

Uncertainty of Measurements

The method used in this report for estimating the uncertainty of measurement is approved by the Comite International des Poids et Measures (CIPM) in 1981. According to the Technical Reports Series No. 374 (IAEA, TRS No.374, 1994), the uncertainty associated with a measurement is a parameter that characterizes the dispersion of the values. This parameter is normally an estimated standard deviation. The estimation of an uncertainty might be carried out by some known methods called either Type A or B uncertainty.

Type A is where a series of measurements have been made, the so-called observed values. The best estimate is then given by the arithmetic mean value (average measurement) and the scatter of the observed values around their mean is reflected by the standard deviation.
Type B is the method where there are many sources of measurement uncertainty that cannot be estimated by repeated measurements. Examples of these include not only unknown, although suspected, influences on the measurement process, also pressure and temperature differences, physical data from the literature, correction factors, etc.

According to the CIPM, method type B standard uncertainties must be estimated so that they correspond to standard deviations. Both types A and B are estimated standard deviations therefore, they are combined using statistical rules for combining variances
(the sum of the standard deviations squared).

The overall uncertainty, U (L), was given by:

$$U(L) = k\,\{(\text{the sum of the standard deviations squared})/m\}^{1/2},$$

where k is the confidence level:…… k = 2.0, and
m is the number of beam qualities:…. m = 7 (THERAPY clinical)
m = 4 (THERAPY non-clinical)
m = 4 (DIAGNOSTIC)

The uncertainty of the calibration factor of the **measuring equipment** was **+/- 2%.**
The contributors for the **clinical therapy X-ray machine** were:

- Air pressure……0.3
- Temperature and Humidity……0.3
- Field homogeneity including homogeneity of the added filter..0.5
- Stability of the operating voltage…….. 0.5

The overall uncertainty, U (L), was estimated to be:

$$U(L) = +/- \{[(0.3)^2 + (0.3)^2 + (0.5)^2 + (0.5)^2]/7\}^{1/2}$$

$$= +/- 0.3$$

Under laboratory conditions, the uncertainty contributors for the **laboratory therapy X-ray machine**, were:

- Air pressure……0.3
- Temperature and Humidity……0.3
- Field homogeneity including homogeneity of the added filter..0.5
- Stability of the operating voltage…….. 0.5
- Energy dependence of the chamber……..0.05

The overall uncertainty, U (L), was estimated to be:

$$U(L) = +/- \{[0.3)^2 + (0.3)^2 + (0.5)^2 + (0.5)^2 + (0.05)^2]/4\}^{1/2}$$

$$= +/- 0.4$$

Typical sources of error for the **mammography** HVL measurements are given below (Wagner, et al):

- Effective thickness of attenuators……..+/-0.04
- Monitored output of X-ray source……..…+/-0.07
- Attenuator placement……+/-0.03
- Position in field…….+/-0.05
- Energy dependence of the chamber……..+/-0.03

$$U(L) = +/-\{[(0.04)^2 + (0.07)^2 + (0.03)^2 + (0.05)^2]/4\}^{1/2}$$

$$= +/-0.05$$

DISCUSSION

The use of mammography has increased rapidly in many countries. In diagnostic radiology, there has been an increased demand for more accurate clinical dosimetry. Dosimetry requires a clear definition of the incident beam quality. Currently there are no officially, published international protocols addressing dosimetry in diagnostic radiology.

The low-energy range of diagnostic X-ray beams of 20-40 kV peak tube voltage is exclusively applicable to mammography, an examination that is designed especially for detecting breast pathology. Measurement of the dosimetric parameters related to the performance of the equipment plays a predominant role in determining the lowest possible dose. An additional objective of dosimetry in medical imaging is the assessment of equipment performance.

In conventional diagnostic radiology, similar beam qualities have been applied to those employed in radiation therapy. Traditionally, the combination of peak tube operating voltage of the X-ray machine and the HVL1 has specified X-ray beam quality. The different X-ray machines used in this work intuitively produced quite different beam qualities owing to differences in their different target/filter combinations, pre-set radiographic factors, added filtration, inherent filtration, anode angulations and configuration.

Assisting in the establishment of a national calibration facility for diagnostic radiology dosimetry was the main goal of this study. The research done focussed on mammography energies. The use of mammography has increased rapidly and there is a demand for more accurate clinical dosimetry. Mammography is one of the diagnostic techniques that contribute to the patient dose (Witzani *et al*, 2003) and a significant risk of potential radiation induced carcinogenesis is associated with it (Dance *et al*, Pernicka, 2000). The CSIR-NML currently uses a constant voltage laboratory therapy X-ray machine for low-energy therapy X-ray beam calibrations.

From the research undertaken it is clear that the Pantak laboratory X-ray therapy machine being used by the national CSIR-NML would be completely inappropriate for mammography calibrations without major modifications. The instability of this unit at these low peak tube voltages made the unit very unsuitable for accurate calibrations of low dose measuring equipment. A better correlation was found between the clinical X-ray therapy units and the range of mammography beams investigated.

The RT100^{TM} clinical therapy X-ray machine could be adapted by modifying its added filtration to function as a calibration unit. This option could be exercised at little cost and the beam quality reconfirmed.

For the calibration of dosimeters used exclusively in mammography, only standard radiation qualities produced by a Mo-anode are defined in IEC-61267, 2005. The latter document then further specifies beam quality only by the generating peak tube voltage and the associated HVL1. A more elaborate system of beam quality specification from the International community, which includes at least the HVL2, would further support this work. The results confirm that the radiation beam qualities produced by the RT100^{TM} therapy X-ray machine could be adapted to those recommended by the IEC for mammography. Experimental confirmation of this would support the use of this unit at the SSDL facility. This work could then also be expanded to peak tube voltages up to 100 kVp, to include calibration requirements for other diagnostic modalities.

CONCLUSIONS

- There was a measurable difference between clinical X-ray beam qualities and the national laboratory unit used historically for therapy calibrations purposes.

- The constant voltage X-ray machine at the CSIR NML could not be used for calibration of mammography ionisation chambers.

- An expanded International specification of beam quality would support this work.

- The equivalence in the sensitivity and response of the different ionisation chambers in both therapeutic and diagnostic beams needs further investigation.

- X-ray machines with intuitively quite different beam qualities could be modified to produce equivalent beam quality in terms of the existing IEC specifiers, by manipulating beam filtration accordingly

- The introduction of a national calibration facility for mammography does not require the immediate purchase of a dedicated clinical unit, if the mammography dosimeters to be calibrated do not have significant energy dependence.

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