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

Johannesburg, 2008
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.

_________________________

23rd day of September 2008
Abstract

Aim: Four clinically measurable dose descriptors: the ratio of absorbed doses at depths 1 cm and 2 cm (D₁/D₂), the ratio of absorbed doses at depths 2 cm and 5 cm (D₂/D₅), the tissue phantom ratio of depths 1 cm to 2 cm (TPR₁,₂), and the tissue phantom ratio of depths 2 cm to 5 cm (TPR₂,₅) were investigated in relation to the quality of superficial (low energy) and orthovoltage (medium energy) x-ray beams.

Methods and Materials: D₁/D₂ and TPR₁,₂ were measured on a Gulmay D3300 unit. D₂/D₅ and TPR₂,₅ were measured on a Gulmay D3300 unit and a Pantak Therapax DXT 300 unit. Different field sizes, half-value layers (HVLs), and distances from the source were investigated in a 30 cm × 30 cm × 30 cm water phantom and a 20 cm × 20 cm × 10 cm solid acrylic phantom. A PTW M30001 0.6 cc cylindrical ionization chamber and a T10008 electrometer system were utilized for all measurements.

Results: D₁/D₂ reflected the changes expected in the penetration of superficial x-rays due to beam hardening and D₂/D₅ was found to vary appreciably with field size, source-to-surface distance (SSD) and HVL. The most practical conditions for the measurement of TPR₁,₂ as a potential beam quality specifier could not be established in this work and TPR₂,₅ varied with field size and HVL, irrespective of the distance from the source and the energy.

Conclusions: D₁/D₂ and D₂/D₅ could be a practical quality index in field sizes of at least 11.28 cm diameter defined at an SSD of 50 cm. Measurements at different HVLs and source-to-chamber distances (SCDs) are needed to establish the most practical measurement conditions of TPR₁,₂. TPR₂,₅ measurements were more accurate than D₂/D₅, and should be investigated further as the beam quality index for orthovoltage x-ray beams.
This work is dedicated to my mother, Mrs. Agness K. Mdziniso, and my daughter, Letsie Melokuhle Mdziniso. I give thanks to God for their spiritual and financial upkeep.
Acknowledgements

This research report is the result of collaboration with the Division of Medical Physics at Johannesburg Hospital. Prof. Debbie van der Merwe is thanked for her continuous guidance and good advice.

Finally, I thank my instructors, Mr. Sibusiso Jozela of the Johannesburg Hospital and Ms. Zakithi Msimang of the CSIR National Metrology Laboratory in Pretoria. Profound thanks is given for their assistance with transport between Johannesburg Hospital and Pretoria.

Everyone else who assisted the aspiring academic physicist in becoming familiar with the operation of all the equipment involved, is also acknowledged.
CONTENTS

DECLARATION i
ABSTRACT ii
ACKNOWLEDGEMENTS iv

1 INTRODUCTION

1.1 General Introduction 1
1.2 Theory of X-Rays 1
   1.2.1 Nature and properties of x-rays 1
   1.2.2 X-ray production 3
   1.2.3 Clinical x-ray beams 6
   1.2.4 X-ray machines for radiotherapy 6
   1.2.5 X-ray energy spectra and the effect of added filtration 8
   1.2.6 Kilovoltage x-ray beam radiotherapy 8
1.3 Kilovoltage X-ray Beam Dosimetry 9
   1.3.1 Categories of kilovoltage x-ray dosimetry 10
   1.3.2 Factors that affect the dosimetry process 11
   1.3.3 Current status of kilovoltage x-ray beam dosimetry 12
   1.3.4 Mass-energy absorption coefficients 16
   1.3.5 Water kerma backscatter factors 17
   1.3.6 Central-axis depth dose data 18
   1.3.7 Beam quality specification 20
   1.3.8 HVL as the beam quality index 21
1.4 Research Objective 23

2 MATERIALS AND METHODS 26
2.1 HVL Determination 29
2.2 Measurement of Absorbed-Dose Ratios 32
2.3 Measurement of Tissue-Phantom Ratios 33
2.4 Correlation of Measured Ratios with Conversion Factors 33

3 RESULTS AND DISCUSSION 35

3.1 Alternative Approaches to Kilovoltage X-Ray Beam Dosimetry 36

4 CONCLUSIONS 48

REFERENCES 49

Glossary 54

APPENDIX A. DOSIMETRIC CONVERSION FACTORS 56
LIST OF FIGURES

Figure 1.1 Atomic model for the production of bremsstrahlung radiation 4

Figure 1.2 Typical x-ray spectrum produced by a tungsten target 5

Figure 1.3 Schematic representation of an x-ray tube 7

Figure 1.4 Wide variation obtained by Coffey et al (2001) between tube potential and HVL for superficial x-ray beams 14

Figure 1.5 Wide variation obtained by Coffey et al (2001) between tube potential and HVL for orthovoltage x-ray beams 15

Figure 1.6 Variation obtained by Aukett et al (1996) between mean mass energy absorption coefficient ratios of water to air and HVL for superficial x-rays 15

Figure 1.7 Comparison by Rosser (1999) of the variation between mean mass energy absorption coefficient ratios of water to air and HVL for orthovoltage x-rays, obtained by using different dosimetry protocols 16

Figure 1.8 Relationship obtained by Eissa et al (2005) between HVL and the water kerma backscatter factors (BSF), for a 10 cm field diameter at 100 cm SSD 23
Figure 2.1 Schematic diagram of applicator design and geometrical alignment testing for the Gulmay D3300 unit

Figure 2.2 Example of a typical radiographic check used to verify the alignment of the x-ray source, the diaphragm and the detector prior to the HVL measurements

Figure 2.3 Illustration of the geometric setup that was used for the HVL measurements

Figure 3.1 Variation of $D_1/D_2$ with field size for two clinical beam qualities produced by different added filtrations at 95 kV, at an SSD of 50 cm

Figure 3.2 Variation of $D_2/D_3$ with field diameter for different HVL beams at an SSD of 50 cm

Figure 3.3 Variation of $D_2/D_5$ with SSD for different field diameters and HVLs

Figure 3.4 Variation of $D_2/D_3$ with published BSF at an SSD of 50 cm for medium-energy x-ray beams produced by different added filtrations and field sizes

Figure 3.5 Variation of the ratios of absorbed doses with the published ratios of PDDs for superficial and orthovoltage x-ray beam qualities
Figure 3.6 Variation of $D_2/D_5$ with the ratio of mean mass energy absorption coefficients of water to air averaged over the photon spectrum at 2 cm in water, at an SSD of 50 cm for a 10 cm × 10 cm field size 43

Figure 3.7 Variation of $D_2/D_5$ with free in-air mean mass energy absorption coefficient ratios of water to air, for three SSDs and field diameters 44

Figure 3.8 Variation of TPR$_{2.5}$ with field diameter for orthovoltage x-ray beam qualities 45

Figure 3.9 Variation of TPR$_{2.5}$ with HVL (mm Cu) for different field diameters 45

Figure 3.10 Variation of TPR$_{1.2}$ with field diameter, for the two clinical beam qualities produced by different added filtrations at 95 kV 47
LIST OF TABLES

**Table 2.1** Results of HVL measurements for the beams produced on the clinical unit 31

**Table 2.2** Values of HVL and machine settings for the beams produced on the laboratory unit 32

**Table A.1** Percentage Depth Dose (PDD) as a function of field diameter and HVL (mm Al) at depths 1 cm and 2 cm and 30 cm SSD, for superficial x-rays 57

**Table A.2** PDD as a function of field diameter and HVL (mm Cu) at depths 2 cm and 5 cm and 50 cm SSD, for orthovoltage x-rays 57

**Table A.3** BSF as a function of HVL (mm Al) and field diameter at 30 cm SSD, for superficial x-rays 58

**Table A.4** BSF as a function of field size and HVL (mm Cu) at 50 cm SSD, for orthovoltage x-rays 58

**Table A.5** Ratios of mean mass energy absorption coefficients of water to air \([\mu_{\text{en}}/\rho]_{w,\text{air}}\) averaged over the photon spectrum at 2 cm depth in water for a 10 cm × 10 cm field size at 50 cm SSD 58
Table A.6 Free in-air $(\mu_{en}/\rho)_{w,\text{air}}$ as a function of HVL, for superficial and orthovoltage x-rays
1 INTRODUCTION

1.1 General Introduction

X-rays were discovered by Wilhelm Conrad Roentgen, a German physicist, on 8 November 1895. Roentgen was studying the properties of cathode rays, and had placed a cathode ray tube in a box in a dark room to block light from leaking. To his amazement, a sheet of paper coated with a barium platinocyanide and placed four feet away, glowed when the tube was switched on in the closed box. An unknown kind of ray was being emitted from the tube and caused the paper to glow. Roentgen called these x-rays, the “x” denoting the unknown. He tried to stop the rays by using different substances to cast shadows of solid objects. When his hand was held between the tube and a barium platinocyanide screen, the bones of his hand were visible on the screen (Dove 2003).

Since then, x-rays have transformed and brought about new technologies in medicine. It became possible to visualize the inside of the body without performing invasive surgery. X-rays became an important diagnostic and therapeutic tool for doctors and dentists the world over. Nowadays, x-rays are used widely in medical and industrial applications (Chang et al. 2002). They are used in diagnostic radiology for the diagnosis and treatment of disease. They are also used in radiation oncology (radiotherapy) for the localization and treatment of disease (Podgorsak 2005).

1.2 Theory of X-Rays

1.2.1 Nature and properties of x-rays

X-radiation is a form of energy in transit. X-rays are electromagnetic radiation similar to light (Dove 2003). They can be considered as small packets of energy capable of
ionizing any atom or molecule, hence the term ionizing radiation. The deposition of energy by x-rays however, may not cause ionization in all cases. Instead, they may excite the molecules of the medium with which they interact. X-rays have the following important properties:

1. They have mass \(E = mc^2\),
2. They are electrically neutral,
3. They travel in a vacuum at the speed of light,
4. They travel in straight lines, but diverge from a central focus,
5. They have enough energy to ionize and cause biological damage, and
6. They can cause certain substances to fluoresce and affect photographic film.

The most abundant molecule in the human body is water. X-rays may ionize or excite water molecules, blocking their normal function. Once absorbed into the body, they interact with water molecules in tissue to produce free radicals that are able to diffuse far enough to reach and damage the nuclear DNA (Hall 2000). Free radicals induce single strand breaks (repairable) and/or double strand breaks (irrepairable) in DNA. This in turn results in cell death, delayed cell division or abnormal cell growth. Deterministic effects (such as reddening of the skin) may ensue if cell death and delayed cell division accompany an exposure to x-rays. Stochastic effects (such as cancer) may ensue if an irradiated cell is modified rather than killed.

Considered as electromagnetic waves, the energy and frequency of x-rays is related through the relation:

\[ E = h\nu, \quad (1.1) \]
where: E is the energy [J], h is Planck’s constant \(6.63 \times 10^{-34}\) Js, and \(\nu\) is the frequency [Hz]. X-rays are also characterized as particles and in this way, they have a velocity \(v\), mass \(m\) and momentum \(p\) given by:

\[
p = mv = mc = E / c = h / \lambda,
\]

where \(c\) is the speed of light \(3 \times 10^8\) ms\(^{-1}\) and \(\lambda\) is the wavelength [m]. These particles are called photons and are delivered in packets called quanta. If the particle energy is greater than about 2 – 3 eV, then the photons are capable of ionizing atoms. To break a chemical bond, one requires energies in the order of 2 – 10 eV. This can be delivered by electromagnetic waves in the ultraviolet region or above (Dove 2003).

1.2.2 X-ray production

X-rays are produced extranuclearly by high-energy electrons incident on a metal target made from materials such as tungsten or molybdenum. The electrons interact with the nuclei of the metal target to form bremsstrahlung radiation. The energy spectrum of x-rays due to bremsstrahlung depends on the energy levels of the atomic electrons and on their velocity. When an electron passes near a positively charged nucleus, it is attracted to the nucleus and deflected away from its original trajectory. The electron may or may not lose energy. If it does not lose energy, the process is called elastic scattering, and no x-ray photon is produced. If it loses energy, the process is termed inelastic scattering, and an x-ray photon is created. The radiation thus produced is called bremsstrahlung radiation (Dove 2003). Figure 1.1 is the atomic model for the production of bremsstrahlung radiation.
Figure 1.1 Deflection of a high-energy electron by the nucleus. $E_1$ is the initial (before the collision) energy, yet $E_2$ is the final (after the collision) energy of the electron. Shown in the figure is the production of bremsstrahlung radiation when $E_1 > E_2$ (Dove 2003).

The probability of the electron losing energy increases as the atomic number of the target increases (Podgorsak 2005). The electron may interact with many nuclei, thus the energies of the x-ray photons generated by this process are distributed over a wide range of values (Dove 2003). Figure 1.2 shows a typical x-ray spectrum produced in a tungsten target. In this figure, white radiation refers to bremsstrahlung radiation. Bremsstrahlung radiation constitutes a major fraction of x-rays emerging from an x-ray tube.

The high-energy electrons may interact with the orbital electrons of the metal target to form characteristic radiation (Dove 2003). Characteristic x-rays are produced by electron-electron interactions between the incident electron and the atomic electrons in the target material. The incident electron ejects an inner shell electron, such as a K, L or M shell electron, from the target atom leaving the target atom ionized. This causes electrons to plummet from higher levels in order to occupy the lower levels. The energy difference between the higher levels and lower levels is emitted as
characteristic radiation or characteristic line spectra (Ejere 2006).

![Diagram of X-ray intensity in the tube, showing characteristic radiation and white radiation.](image)

Figure 1.2 Typical x-ray spectrum produced by a tungsten target (Dove 2003).

Characteristic radiation is characteristic of the target atoms and the shells between which atomic transitions took place. With higher atomic number targets, the characteristic radiation emitted may be of high enough energy to be considered as part of the x-ray spectrum emerging from the x-ray tube. The threshold energy that the primary incident electron must possess to eject an orbital electron is known as the binding energy (Bulz 2001). It is specific to the shell and to the target material of the electron to be ejected. Hence the incident electron must have kinetic energy greater than the binding energy of the inner shell electron. Part of this energy is used to overcome the inner shell’s binding energy and the remainder is used as kinetic energy for the ejected electron (Maniquis 2006). The ejected electron may induce secondary ionizations leading to large numbers of characteristic x-rays at a few discrete energies denoted α_i, β_i (i = 1, 2) in figure 1.2.
1.2.3 Clinical x-ray beams

Clinical x-ray beams typically range in energy from 10 kV to 50 MV. They are produced when electrons with kinetic energies between 10 keV and 50 MeV respectively, are decelerated in metallic targets. A typical spectrum of a clinical x-ray beam consists of line spectra that are characteristic of the target material. These line spectra are superimposed onto a continuous bremsstrahlung spectrum. The bremsstrahlung spectrum originates in the x-ray target. The characteristic line spectra, on the other hand, originate in the target and in any attenuators placed in the beam (Podgorsak 2005).

Clinical x-ray beams for external beam radiotherapy fall into three categories, depending on the kinetic energies of the electrons involved in their production. These are: superficial x-rays, orthovoltage x-rays and megavoltage x-rays. Of interest in this research are the superficial and orthovoltage x-rays, commonly known as kilovoltage x-rays. Kilovoltage x-ray beams have been employed in radiotherapy for many years, and have regained popularity in recent years (Li et al. 1997). Superficial x-rays are produced by electrons with kinetic energies between 40 keV and 100 keV, and are used to irradiate surface lesions. Orthovoltage x-rays are produced by electrons with kinetic energies between 100 keV and 500 keV (Podgorsak 2005).

1.2.4 X-ray machines for radiotherapy

Superficial and orthovoltage x-rays used in radiotherapy are produced by means of x-ray machines. The main components of such machines are: an x-ray tube, a ceiling or floor mount for the x-ray tube, a target cooling system, a control console, and an x-ray power generator. This section shall focus on the x-ray tube.
An x-ray tube is contained within an evacuated glass envelope. A conventional x-ray tube comprises a metal filament (cathode) that emits electrons when resistively heated to over 1000 °C (thermionic emission). The anode is the target. It emits x-rays when bombarded by the accelerated electrons produced by thermionic emission from the cathode. The intensity of the x-ray is proportional to the electron current and the square of the accelerating voltage (Chang et al. 2002).

Figure 1.3 is a schematic diagram of the x-ray tube. The cathode is made of a tungsten filament and is heated to produce electrons. When a high voltage is applied between the anode (+) and cathode (-), the electrons move towards the anode. The anode is made of copper with a tungsten target where the electrons hit the anode. The electrons are focused onto the tungsten target by means of a focusing cup, which is made of molybdenum. The tube has a metal housing and is surrounded with oil which serves to insulate and prevent sparking within the various electrical components (Podgorsak 2005).

Figure 1.3 Schematic representation of an x-ray tube (Dove 2003).
X-rays are produced when a stream of fast moving electrons collide with the tungsten anode. The kinetic energy of the electrons is converted into x-radiation and heat. More than 99 % of the electron energy is converted into heat. About 1 ~ 5 % of the electron’s kinetic energy is converted into x-rays, which are divided into two groups: characteristic x-rays and bremsstrahlung x-rays (Podgorsak 2005).

1.2.5 X-ray energy spectra and the effect of added filtration

The energy spectrum of x-rays produced by an x-ray machine exhibit a continuous distribution of energies, with the bremsstrahlung photons superimposed by characteristic radiation of discrete energies. Added filtration, positioned externally to the x-ray tube, modifies the energy spectrum by primarily affecting the low-energy part of the spectrum. It relatively enriches the beam with higher-energy photons by absorbing the lower-energy components of the spectrum. As the filtration is increased, the transmitted beam becomes harder and therefore achieves greater penetrating power. The total intensity of the beam on the other hand, decreases with increasing filtration.

It is conventional practice to describe the quality of an x-ray beam by its penetrating ability. However, an ideal way to characterize the quality of an x-ray beam is to specify its spectral distribution. Spectral distributions are difficult to measure and this has led to a crude but simpler beam quality index, the HVL.

1.2.6 Kilovoltage x-ray beam radiotherapy

Superficial therapy refers to radiation therapy obtained with x-ray tube potentials between 40 kV and 100 kV. Superficial therapy is used clinically to irradiate surface lesions. Added filtration of up to 6 mm Al is used to remove very low energy photons and harden the beam. The irradiated area is defined by an applicator cone attached to
the head of the tube housing. Irradiation is performed at short SSDs and the lesion depth must be less than a few millimeters. Hence, this range of kilovoltage x-rays is used when surface or shallow lesions are treated. Tissue at more than a moderate depth is therefore spared when treating surface lesions in this way (Arsenault et al. 2005).

Orthovoltage therapy refers to radiation therapy obtained with x-ray tube potentials between 100 kV and 500 kV. This deeper radiotherapy tool requires beam currents of up to 20 mA. Added filtration, equivalent to HVL values of 1 to 4 mm Cu is utilized to eliminate lower energy photons and harden the beam. Cone applicators or movable diaphragms are again used to define these beams. The applicators are made of metal and have a clear plastic end to aid in viewing the target region. SSDs of approximately 50 cm are chosen. The depth dose in this range is dependent on kV, HVL, SSD and field size. Maximum dose occurs close to the skin, with 90% of the dose being delivered within a tissue depth of 2 cm (Arsenault et al. 2005).

Kilovoltage radiotherapy units continue to be used for superficial therapy. Simpler design, unique range of application, and their traditional type of technology set them apart from higher energy devices (Arsenault et al. 2005)

1.3 Kilovoltage X-ray Beam Dosimetry

In order to evaluate the risk/benefit ratio of exposing a patient to ionizing radiation, it is necessary to measure the radiation dose to which the subject is exposed. Medical kilovoltage x-ray beam dosimetry deals with the determination of the radiation dose deposited at points of interest in a given medium. The ultimate goal is to determine accurately the three-dimensional dose distribution planned and delivered to the patient. This is very important not only for the safety of the patient but is also crucial
for the technical advancement of radiotherapy. A number of kilovoltage x-ray beam dosimetry protocols have recently been published (Eissa et al. 2005). These protocols provide guidance and have helped maintain a high level of accuracy and consistency in clinical reference dosimetry. However, the challenge remains in performing a complete characterization of the three-dimensional dose distribution delivered to a patient (Chen et al. 2007).

1.3.1 Categories of kilovoltage x-ray dosimetry

Part of the external beam radiotherapy process that involves radiation dosimetry is as follows:

1. **Basic Beam Dosimetry**
   - 1.1 Beam quality
   - 1.2 Absolute dose output or reference dosimetry
   - 1.3 Relative dose distribution or relative dosimetry
   - 1.4 Dosimetry parameters for computational beam modeling
   - 1.5 Verification of dose-computation algorithms

2. **Patient Dosimetry**
   - 2.1 Dose distribution planning
   - 2.2 In-phantom dose verification
   - 2.3 In vivo dose verification of treatment delivery

This includes basic radiation beam characterization in a homogeneous water phantom, beam modeling for dose computation, patient-specific dose planning, radiation delivery monitoring and verification (Chen et al. 2007).
1.3.2 **Factors affecting the dosimetry process**

The determination of absorbed dose using a dosimetry protocol is not a straightforward task. In order to use a dosimetry protocol, one first needs to determine the quality of the clinical beam (Chen et al. 2007). Radiation quality influences radiation dose through mechanisms by which photons of different energies interact with tissue. Five major factors that affect an accurate determination of absorbed dose in beam characterization and patient dosimetry include:

1. Variation of the primary fluence spectrum in the irradiated volume,
2. Variation of the angular distribution of primary ionizing particles in the irradiated volume,
3. Variation of the effective dose rate in the irradiated volume,
4. Existence of high spatial dose gradients in the irradiated volume, and
5. Existence of tissue heterogeneities in patient dosimetry.

In photon beams, the photon fluence decreases continuously with depth in water along the central axis. The fluence spectrum also varies with depth and lateral distance from the central axis, albeit at a slower rate. The spectral variation with depth is primarily due to photon scattering in the medium. As the depth increases, the proportion of scattered photons increases. This, in turn, results in a shift of the energy spectrum towards lower photon energies. The fluence of the secondary electrons generated by the photons also decreases with depth. This occurs at nearly the same rate as the photon fluence, for depths greater than the maximum buildup (Chen et al. 2007).

The angular distribution of the primary ionizing particles varies from point to point in the irradiated volume. Along the central axis, the angular distribution changes from
nearly mono-directional, along the beam’s forward direction at the surface, to an increasingly broader distribution with increasing depth. Near the beam edge, a non-symmetric angular distribution results from the loss of side scatter equilibrium. High spatial dose gradients exist in photon beams at the depth of maximum dose and in the penumbra. The existence of such dose gradients requires detectors that have a high spatial resolution for the accurate determination of the dose in those regions (Chen et al. 2007).

Tissue heterogeneity is always present in patients. The variation in atomic composition across a spatial region in a patient creates abrupt changes in the primary and secondary particle fluence. This creates regions (near the interfaces) in which charged particle equilibrium does not exist and, performing or verifying dosimetry in such regions a complex one (Chen et al. 2007).

1.3.3 **Current status of kilovoltage x-ray beam dosimetry**

The dosimetry of kilovoltage x-rays has always been based on the concept of treating the dosimeter as an exposure meter. An air ionization chamber is calibrated in terms of exposure (or air kerma) at the quality of interest. Its in-phantom reading is interpreted as measuring exposure (or air kerma) at the chamber centre. For orthovoltage x-rays for example, the dose to water at a depth \( z \) according to Aukett et al (1996) is given by:

\[
D_{w,z} = M_N k_{ch} \left[ (\bar{\mu}_{en} / \rho)_{w,air} \right]_{z,\phi},
\]  

where:

- \( M \) is the electrometer reading corrected to the same ambient conditions as the chamber calibration factor,
- \( N_k \) is the air kerma calibration factor of the instrument for standard ambient conditions.
conditions and for the radiation quality of the incident beam, $k_{ch}$ is a correction factor that takes account of various effects, including the modification of the primary spectrum by water, and the attenuation and scattering by water in the cavity, and

\[ \left( \frac{\mu_{en}}{\rho} \right)_{w,\text{air}} \, z, \phi \]

is the ratio of the mean mass energy-absorption coefficients of water (w) to air averaged over the photon spectrum at depth $z$ in water for field size $\phi$.

For superficial x-rays on the other hand, measurements are made free in air to obtain air kerma, $K_{air}$. The air kerma is converted to water kerma in air, $(K_{\text{water}})_{air}$, through the ratio of the mean mass energy-absorption coefficients of water to air, $(\mu_{en}/\rho)_{w,\text{air}}$. The backscatter factor $(B_w)$ is then used to convert $(K_{\text{water}})_{air}$ to water kerma at the surface of a water phantom. The dose at the surface ($z = 0$) of a water phantom is thus given by:

\[
D_{w,z=0} = MN_K B_w P_{stem,\text{air}} \left( \frac{\mu_{en}}{\rho} \right)_{w,\text{air}} \text{air},
\]

where $P_{stem,\text{air}}$ is the chamber stem correction factor which accounts for the change in photon scatter from the chamber stem between the calibration and measurement conditions (Coffey et al. 2001).

Several parameters required for the determination of absorbed dose depend on the accurate specification of the quality of the x-ray beam. Such parameters include the $(\mu_{en}/\rho)_{w,\text{air}}$ and $B_w$ that appear in equations (1.3) and (1.4). According to Coffey et al (2001), the specification of a kilovoltage x-ray beam requires detailed knowledge of the photon fluence spectrum at the point of interest. Andreo et al (2001) suggested the use of more than one beam quality parameter to characterize a kilovoltage x-ray spectrum for dosimetry. In one approach, HVL is used as the beam quality index for
kilovoltage x-ray radiotherapy (Coffey et al. 2001). Another approach however, uses a combination of kV_p and HVL for beam quality specification (Nahum 1994).

Andreo et al (2001), Coffey et al (2001), and Rosser (1998) showed that using only kV_p or HVL is insufficient to specify the quality of a kilovoltage x-ray beam. For example, commonly used clinical beams may have a wide range of HVLs corresponding to the same kV_p. The exposure (or air-kerma) to dose-to-water conversion and other related chamber correction factors may vary significantly for these beams (Coffey et al. 2001).

Figures 1.4 and 1.5 show the wide range in kilovoltage x-ray beams currently used in routine radiotherapy practice. Emphasized in these figures is that a wide range of HVLs may correspond to the same kV_p. Likewise Figures 1.6 and 1.7 reflect the uncertainty in determining the ratios of mean mass energy-absorption coefficients based on HVL, from using different dosimetry protocols. Hence, there is large uncertainty in the specification of a kilovoltage x-ray beam using HVL only.

![Figure 1.4 Tube potential as a function of HVL for low-energy x-rays as reported by North American clinics (Coffey et al. 2001).](image-url)
Figure 1.5 Relation between tube potential and reported HVL values for medium-energy beams as reported by North American clinics (Coffey et al. 2001).

Figure 1.6 Mean mass energy-absorption coefficient ratios of water to air as a function of HVL, corresponding to the in-air (primary) spectra. The circles and filled triangles correspond to 113 different spectra (Aukett et al. 1996).
1.3.4 Mass-energy absorption coefficients

An x-ray photon incident on a medium produces secondary charged particles. The relative biological effectiveness (RBE) of photons increases with decreasing photon energy (Beatty et al. 1999). As the energy of the photons decrease, the energy of the secondary charged particles decreases. This corresponds to an increase of the linear energy transfer (LET).

Most of the kinetic energy of the secondary charged particles set in motion by the photons is deposited in the medium. This occurs through inelastic collisions (ionization and excitation) with atomic electrons of the medium. The energy deposition usually occurs along the tracks of the charged particles, which can extend
away from the interaction site. The mass energy absorption coefficient describes the fraction of the kinetic energy of secondary charged particles that is deposited in the medium per unit mass (Podgorsak 2005).

Li et al (1997) showed that the ratios of mean mass-energy absorption coefficients for water to air, \((\bar{\mu}_{en} / \rho)_{w,air}\), depend on the beam size and depth in a phantom. For the in-phantom method of kilovoltage x-ray beam dosimetry therefore, \([(\bar{\mu}_{en} / \rho)_{w,air}]_z\) will be depth, SSD and field size dependent (Ma and Seuntjens 1999).

1.3.5 Water kerma backscatter factors

When x-rays impinge on a scattering medium, the photons are absorbed or scattered by different interaction processes. Any point on the surface of the medium receives unattenuated primary radiation plus scattered radiation. The dose at any point of an exposed phantom can be calculated by estimating the primary radiation reaching the point of interest and the corresponding contribution of scattered radiation. Because of the complexity of such calculations however, an empirical approach to this problem has been practiced using experimental measurements. These measurements involve for example, the concepts of surface backscatter, tissue-air ratios and scatter-air ratios (Bek-Uzarov et al. 1999).

The quantity that characterizes the contribution of backscattered radiation to the surface dose (or kerma) is called the backscatter factor, denoted BSF or \(B_w\). According to its definition, \(B_w\) depends on the way the radiation beam is generated, filtered and measured, on the type and thickness of underlying scatter material, and the size of the beam cross section at the scatter surface. There are a few slightly differing definitions of the BSF. \(B_w\) was defined in a 1973 report of the ICRU (Grosswendt 1984) as the ratio of the exposure at the surface of a water phantom to
that part of the exposure which is due to primary photons. In a later report of the ICRU (1976), the BSF was defined as the ratio of the absorbed dose rate at the intersection of the beam axis with the phantom surface to the absorbed dose at the same point in space. The BSF was redefined by Grosswendt (1984) as the ratio of the kerma rate to a specified material at a point on the surface of a phantom to the kerma rate to the same material at the same point in space in the absence of the phantom (Bek-Uzarov et al. 1999).

For x-rays of up to 100 kV, \((\bar{\rho}_{en}/\rho)_{w,air}\) and \(B_w\) are used in the conversion of air kerma, measured free in air, to water kerma on the surface of a water phantom. For clinical radiotherapy, similar conversion factors are needed for the determination of absorbed dose to biological tissues on the surface of a human body (Ma and Seuntjens 1999).

1.3.6 Central-axis depth dose data

The PDD was introduced to characterize the central axis dose distribution for SSD techniques of treatment. The dose at depth is normalized with respect to the dose at a reference depth for a fixed SSD. Mathematically, PDD is defined as the quotient, expressed as a percentage, of the absorbed dose at any depth \(d\) in a phantom to the absorbed dose at a fixed reference depth \(d_0\), along the beam central axis.

\[
PDD = \frac{D_d}{D_{d_0}} \times 100
\]

(1.5)

For low- and medium-energy x-rays, the reference depth is at the surface of a phantom, that is \(d_0 = 0\).
Factors that affect the central axis depth dose distribution include the beam quality or energy, depth, field size and shape, SSD and the beam collimation. The PDD increases with beam energy beyond $d_0$. Higher-energy beams have greater penetrating power, higher HVL and deliver a higher PDD. The beam quality affects the PDD by virtue of the average attenuation coefficient $\bar{\mu}$ (Khan 2003). As $\bar{\mu}$ decreases, the beam penetrating power increases. This results in a higher PDD at any given depth beyond the build-up region. For kilovoltage x-rays, the dose builds up on or near the surface.

A point distal from the focal spot of an x-ray target will receive the radiation as if it was emitted by a point source. The exposure or dose rate in air from the source varies inversely as the square of the distance from the source. This inverse square variation results in an increase in the PDD as the SSD increases. Only the dose rate at a point decreases with increase in SSD. This effect is more pronounced at smaller distances from the source than at larger distances. This means that the PDD decreases rapidly nearer the source.

The PDD at depth $d$ and SSD $f_1$ is related to the PDD at depth $d$ and SSD $f_2$ for a field size $\phi$ through the relation:

$$\frac{PDD(d, \phi, f_1)}{PDD(d, \phi, f_2)} = \left( \frac{f_1 + d_0}{f_2 + d_0} \right)^2 \times \left( \frac{f_2 + d}{f_1 + d} \right)^2,$$

(1.6)

where the right hand side of equation (1.6) is referred to the Mayneord F factor. That is:

$$F = \left( \frac{f_1 + d_0}{f_2 + d_0} \right)^2 \times \left( \frac{f_2 + d}{f_1 + d} \right)^2$$

(1.7)
For kilovoltage x-ray beams ($\leq 500$ kV), the PDD at depth $d$ and SSD $f$ for field size $\phi$ can also be calculated from:

$$PDD(d, \phi, f) = TAR(d, \phi_d) \times \frac{1}{\sqrt{BSF(\phi)}} \times \left(\frac{f + d_0}{f + d}\right)^2 \times 100,$$

(1.8)

where $\phi_d$ is the field size at depth $d$ and TAR is the tissue-air ratio.

1.3.7 **Beam quality specification**

Various parameters are used as x-ray beam quality indices. These include the photon spectrum, the first HVL, the nominal accelerating potential (NAP), and the beam penetration into tissue-equivalent media. A complete x-ray spectrum is very difficult to measure even though it gives the most rigorous description of x-ray beam quality (Podgorsak 2005).

The first HVL is practical for kilovoltage beam quality description however, HVL is not practical to use in the megavoltage energy range because the attenuation coefficient is a slowly varying function of beam energy (Podgorsak 2005).

For kilovoltage x-ray beams, the nominal accelerating potential is equivalent to the tube potential. This determines the proportion of high energy photons in an x-ray beam. Superficial patient dose can be reduced by adding extra filtration to the x-ray beam. This also increases the beam’s penetration into tissue and influences the x-ray spectrum.

In kilovoltage x-ray beam radiotherapy, the absorbed dose at the prescription point in a patient should be known with an overall uncertainty of 3.5 % (Ab and Antti 1999). This requirement arises because of the narrow dose margin between the dose needed
for tumor control and the dose causing complications in normal healthy tissues. The size of this margin depends on the radio-sensitivity of tissues and on the quality of the radiation beam (Ab and Antti 1999).

The most accurate method to characterize the quality of any radiation beam is to measure its spectral distribution. One of the main problems in clinical radiation dosimetry is the precise determination of the distribution of absorbed dose in the regions of interest of an irradiated human body (Grosswendt 1984). The specification of beam quality in terms of the HVL is limited because it reflects little about the energy distribution of the photons present in the beam (Andreo et al. 1997).

1.3.8 HVL as the beam quality index

X-rays are attenuated in matter such that their intensity is reduced exponentially as the thickness of the absorber increases. This exponential reduction in intensity, besides the absorber thickness, also depends on the energy of the x-rays and the atomic number and density of the absorber material. The thickness of the absorber required to reduce the intensity of the original beam by one half is known as the half-value layer, HVL (Beh et al. 2004). Andreo et al (1997) and Coffey et al (2001) define HVL as the thickness of a specified attenuator that reduces the air-kerma rate in a narrow beam to one half its original value. This is used clinically to specify the quality of an x-ray beam, and is expressed in mm Al for low energy x-rays (40 kV ≤ kV_p ≤ 100 kV) and in mm Cu for medium energy x-rays (100 kV < kV_p ≤ 500 kV).

The quality of a kilovoltage x-ray beam depends on many factors other than the tube potential. Such factors include the target angle and material, window material and thickness, monitor chamber material and thickness, added filtration material and thickness, and the source-chamber-distance (air attenuation and scattering may
change the beam quality significantly for low-energy photons). The uncertainties in measured HVLs can vary significantly due to the differences in the experimental setup, the measurement procedures and the dosimeters used (Andreo et al. 1997; Jozela 2007).

A number of variables in measurement conditions affect the x-ray spectrum and hence influence the selection of the appropriate values of \((\bar{\mu}_n / \rho)_{w,air}\) and \(B_w\). Two variables common to absorbed dose measurements are the field size (excluding the influence of the actual applicator) and the depth within the water phantom. In general, \((\bar{\mu}_n / \rho)_{w,air}\) decreases with an increase in field size and this decrease is more pronounced for harder beam qualities. As the field size increases, a proportion of lower energy photons is enhanced by backscatter from the field periphery, hence the spectrum becomes slightly “softer”. This results in a lowering of the mean energy, the HVL and a subsequent lowering in \((\bar{\mu}_n / \rho)_{w,air}\). Since monoenergetic values of \((\bar{\mu}_n / \rho)_{w,air}\) increase between 100 kV and 500 kV, a spectrum primarily limited to photon energies within this range will actually become softer with an increase in depth. The opposite effect occurs with the increase in added filtration.

Eissa et al (2005) showed that different BSF values could be obtained for the same HVL and attributed this to the use of different applied kV and added filtration. Figure 1.8 shows the relationship obtained by Eissa et al (2005) between the HVL and the BSF for a 10 cm field diameter at 100 cm SSD. It was concluded that HVL was not a reliable index for the BSF.
1.4 Research Objective

Absorbed dose to water is the end point of clinical dosimetry measurements for radiotherapy. To determine absorbed dose to water it is essential that the quality of the x-ray beam is known accurately (Rosser 1998). Most of the dosimetry protocols established for kilovoltage x-rays recommend that absorbed dose measurements are made with reference to air kerma calibrations. This requires the use of the HVL and the generating tube potential (kV$_p$) for beam quality specification. The determination of HVL however, is subject to errors resulting from the measurement geometry. An invasive measurement of kV$_p$ is also seldom confirmed in the clinical environment. Hence there is a need for an alternative quality index for kilovoltage x-rays. Andreo et al (2001) suggested that a new quality index for kilovoltage x-rays based on the quantity absorbed dose to water should be adopted for future Codes of Practice.
Rosser (1998) investigated three parameters: HVL, mean energy at 2 cm depth in water, and \( D_2/D_5 \) as a function of \( (\bar{u}_{en} / \rho)_{w,air} \). The conclusion reached was that HVL alone did not uniquely define the quality of an x-ray beam, and that the mean energy at 2 cm depth in water was an impractical quality index for routine use. It was suggested that the preferred quality index for future Codes of Practice be \( D_2/D_5 \). Rosser (1998) however, measured this quality index at an SSD of 100 cm and a 10 cm \( \times \) 10 cm field at the phantom surface. These conditions are not typical in the clinical situation. Further work is therefore required to determine a more practical SSD and field size for end users in radiotherapy centers.

Jozela (2007) compared and analyzed two clinically measurable quantities: HVL and \( D_2/D_5 \), and showed that a relationship may be established between HVL and \( D_2/D_5 \). The use of \( D_2/D_5 \) as a tool to verify the beam quality was also found to simplify quality control in the clinical environment.

Eissa et al (2005) investigated three ratios of absorbed dose: \( D_{2.5}/D_5 \), \( D_2/D_5 \) and \( D_5/D_{10} \) in relation to the BSF. Measurements were carried out with different applied kV at an SSD of 100 cm for a 10 cm field diameter. There was an explicit correspondence between \( B_w \) and \( D_{2.5}/D_5 \), and therefore \( D_{2.5}/D_5 \) was suggested to be a more convenient medium energy quality index than the HVL. A wider range was obtained for \( D_5/D_{10} \) when compared to \( D_2/D_5 \), and a unique value of the BSF would therefore be obtained from \( D_5/D_{10} \) (Eissa et al. 2005). This study concluded that the use of in-water dose ratios, as an alternative beam quality index for medium energy x-rays, would potentially give a more unique definition of the beam quality in water than the HVL. Depths beyond 5 cm however, are not practical in the clinical situation when using kilovoltage x-ray therapy. This is due to very weak signals at these depths e.g. the typical PDD in a 100 kVp, 3 mm Al HVL beam is of the order of 10 % at
depth 10 cm and 30 cm SSD. Hence, the ratio $D_5/D_{10}$ was not considered in this research.

This research aimed at furthering the investigation of the ratios of absorbed dose as a function of:

1. SSD,
2. $kV_p$ (and/or beam hardening), and
3. field size (defined at the phantom surface).

Tissue-phantom ratios were investigated for various SCDs. Depths of 1 cm and 2 cm were used for superficial x-ray beams, and depths of 2 cm and 5 cm for orthovoltage x-ray beams. Investigations of the ratios $D_1/D_2$, $D_2/D_5$, $TPR_{1,2}$ and $TPR_{2,5}$ were carried out. TPR was thought to be more useful clinically because of the lower uncertainty involved in its measurement given that it is easy to measure, it theoretically does not depend on the SCD, and there is reduced positional uncertainty in its measurement.
2 MATERIALS AND METHODS

Experiments for this research were carried out on two different units: a Gulmay D3300 unit at the Johannesburg Hospital and a Pantak Therapax DXT 300 unit at the National Secondary Standards Dosimetry Laboratory (NSSDL).

The above kilovoltage x-ray therapy units have the following characteristic features:

1. The Gulmay D3300: This unit is situated at the Johannesburg Hospital and was used for some experiments in this research. It is a combination superficial and orthovoltage x-ray unit capable of generating tube potentials in the range of 40 – 300 kV. It consists of a floor mounted tubestand, a bipolar oil-cooled metal ceramic x-ray tube, a high stability generator (Gulmay CP320) and a software-based user interface and controller (Gulmay TP1). The system is fully user-configurable across its range of operation with regard to kVp, tube current (mA) and external tube filtration, making it possible to obtain a wide range of beam qualities for clinical use (Evans et al. 2001).

The dose monitoring system comprises an unsealed parallel plate ionization chamber and electrometer. Compensation for the monitor chamber response, owing to ambient temperature and pressure conditions, is achieved automatically via software corrections. Temperature data are obtained from a copper-constantan (type T) thermocouple sensor in close proximity to the monitor chamber. Ambient pressure data are derived from a pressure transducer (Evans et al. 2001). Figure 2.1 is a schematic representation of applicator design and geometrical alignment testing for the Gulmay D3300 x-ray unit.
Dose output calibration of the unit is achieved by determining the absorbed dose delivered to a reference applicator defined for the beam energy in use. Software corrections are then stored, which set the output in units of absorbed dose per monitor unit (MU). When any other applicator is used with the particular beam energy, the monitor chamber signal is corrected in the software for the effect of change in the magnitude of radiation backscattered into the monitor chamber. This correction results in the same displayed dose rate for any beam, irrespective of the applicator selected (Evans et al. 2001).

The system also includes a dose rate interlock, which monitors the steady-state beam dose rate against the stored nominal value (obtained at calibration), and terminates the beam for values outside ± 3 % of the nominal value. This interlock feature has the
advantage of detecting significant changes in beam dose rate owing to, for example, target material deposition on the tube exit window, any high tension cable/generator fault or any failure in the monitor chamber. To guard against primary dose monitor failure, an independent back-up timer is automatically set by the software system to terminate treatment at a time limit that is 5% greater than that calculated on the basis of the stored nominal dose rate and the number of MUs set (Evans et al. 2001).

2. The Pantak Therapax DXT 300: This unit is located at the NSSDL and was also used in this research. The principal components of this kilovoltage x-ray therapy unit include a computerized control console (CPU), a high voltage control system and generator, a cooling system (oil to water or oil to air), a Comet x-ray tube, a dosimetry system and a set of applicators. The unit is provided with either a pedestal wall mount or a ceiling-mounted support system (Gerig et al. 1994).

The x-ray tube has a power rating of 300 kW, and is configured to operate at kVp between 40 kV and 300 kV. When operated, the control system ensures that the tube never operates above 3 kW and that the tube current never exceeds 30 mA. The comet x-ray tube is of metal-ceramic design and has an anode angle of 30° and a 5 mm thick beryllium window (Gerig et al. 1994).

The Therapax DXT 300 was the first commercially available unit with its own internal dosimetry system. This consists of a PTW Diamentor-M3 250 cm³ pancake chamber, operating at 500 V bias. In clinical mode the unit can be operated to terminate treatment by dose in terms of monitor units (MU) with timer backup, or to terminate by time with a second independent timer as backup in a similar fashion to traditional orthovoltage units. The ion chamber is located between the slot provided to accommodate the added filtration and the housing slot for the beam applicator (Gerig et al. 1994).
The system provides a mechanism for the user to configure up to eight combinations of kV, mA, and beam hardening filter. These combinations are encoded in software and provide the mechanism for auto-selection of kV and mA once a beam filter is inserted (Gerig et al. 1994).

The internal ion chamber is unsealed. The Therapax unit provides a mechanism through the control console software to adjust the monitor system gain in order to compensate for variations in the mass of air in the ion chamber resulting from changes in temperature and pressure (Gerig et al. 1994).

2.1 HVL Determination

A PTW M30001 0.6 cc cylindrical ionization chamber and T10008 electrometer system, calibrated in terms of air kerma, were used to measure the HVL on the hospital unit at three nominal kVp’s. The energy response of the ionization chamber was within ± 1.0 % over the range of kVp’s considered. A lead diaphragm thick enough to attenuate the primary beam to 0.1 % was used to define the beam diameter to about 4 cm at the position of the detector. The attenuating materials utilized were a set of aluminium foils at 95 kV and thin sheets of copper at 180 kV and 300 kV. An optical bench was used to align the detector, attenuators and the lead diaphragm. The detector was placed 62 cm away from the attenuating material and the diaphragm (to minimize scatter), according to the Code of Practice given by Coffey et al (2001). A radiographic check of the alignment of the source, the diaphragm and the detector was performed using a series of exposures on film. Figure 2.2 shows an example of a typical radiographic alignment check.

A monitor chamber was used to correct for variations of the air kerma rate that might arise from the added filtration during the HVL determination. Figure 2.3 shows the geometric setup that was used for this experiment.
Figure 2.2 Radiograph of the source, diaphragm and detector for the alignment verification check.

Figure 2.3 Experimental setup for the HVL determination (Coffey et al. 2001).
If $M_0 \ [nC]$ was the detector reading when there is no attenuator in the beam, and $M(x)$ the detector reading when there was an attenuator of thickness $x \ [mm]$ in the beam, then:

$$M(x) = M_0 e^{-\mu x},$$

(2.1)

where $\mu$ is the effective linear attenuation coefficient of the attenuator material at that particular beam quality. Therefore,

$$\ln \frac{M}{M_0} = -\mu x$$

(2.2)

The HVL was determined at 95 kV$_p$ for added filtrations of 2.5 mm Al and a combination of 0.1 mm Cu and 1.00 mm Al. It was also determined at 180 kV$_p$ for a combination of 0.55 mm Cu and 1.0 mm Al added filtration in the beam. Table 2.1 shows the results of this experiment for the hospital unit.

Table 2.1 Results of HVL measurements for the beams produced on the hospital unit. Given also are the percentage errors incurred in determining the HVL.

<table>
<thead>
<tr>
<th>kV$_p$ (kV)</th>
<th>Added Filtration</th>
<th>HVL ± 1.0 %</th>
</tr>
</thead>
<tbody>
<tr>
<td>95</td>
<td>2.5 mm Al</td>
<td>3.25 mm Al</td>
</tr>
<tr>
<td></td>
<td>0.1 mm Cu + 1.00 mm Al</td>
<td>4.04 mm Al</td>
</tr>
<tr>
<td>180</td>
<td>0.55mm Cu + 1.0 mm Al</td>
<td>1.06 mm Cu</td>
</tr>
</tbody>
</table>

HVLs were not measured for the beams produced on the NSSDL unit. They were provided to the user together with the machine settings, with a stated overall
uncertainty of ± 0.5%. Table 2.2 shows the given HVLs and machine setting parameters for the NSSDL unit.

Table 2.2 Values of HVL and machine settings for the beams produced on the laboratory unit.

<table>
<thead>
<tr>
<th>kV$_p$</th>
<th>mA</th>
<th>Filters</th>
<th>HVL (mm Al)</th>
<th>HVL (mm Cu)</th>
</tr>
</thead>
<tbody>
<tr>
<td>100</td>
<td>20</td>
<td>3.4 mm Al</td>
<td>4.0</td>
<td>0.15</td>
</tr>
<tr>
<td>105</td>
<td>20</td>
<td>0.1 mm Cu + 1 mm Al</td>
<td>5.0</td>
<td>0.20</td>
</tr>
<tr>
<td>135</td>
<td>20</td>
<td>0.27 mmCu + 1 mm Al</td>
<td>8.8</td>
<td>0.50</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Inherent filtration + 3.4 mm Al + 4.6 mm perspex</td>
<td></td>
<td></td>
</tr>
<tr>
<td>180</td>
<td>15</td>
<td>0.42 mm Cu + 1 mm Al</td>
<td>12.3</td>
<td>1.0</td>
</tr>
<tr>
<td>220</td>
<td>15</td>
<td>1.2 mm Cu + 1 mm Al</td>
<td>16.1</td>
<td>2.0</td>
</tr>
<tr>
<td>280</td>
<td>12</td>
<td>1.4 mm Sn + 0.25 mm Cu + 1 mm Al</td>
<td>20.0</td>
<td>4.0</td>
</tr>
</tbody>
</table>

2.2 Measurement of Absorbed-Dose Ratios

D$_1$/D$_2$ was measured in a 30 cm × 30 cm × 30 cm water phantom at an SSD of 50 cm on the hospital unit. It was measured for the two clinical beam qualities produced by different added filtrations at 95 kV. Five different field sizes were employed at each hospital beam quality. An open ended circular applicator defined a field diameter of 4.29 cm at 50 cm from the source. Closed end applicators were further utilized to define square fields with dimensions 4 cm × 4 cm, 8 cm × 8 cm, 10 cm × 10 cm and 15 cm × 15 cm at 50 cm from the source. These were converted to the equivalent field diameters using the ‘equivalent circle’ concept recommended by Aird et al.
Pressure and temperature were monitored during all measurements. The internal timer of the electrometer was used such that the ionization current over 30 seconds was measured at each datum point.

Likewise, $D_2/D_5$ was measured in a 30 cm $\times$ 30 cm $\times$ 30 cm water phantom and a 20 cm $\times$ 20 cm $\times$ 10 cm solid acrylic phantom. This quantity was measured on both the hospital and NSSDL units for the orthovoltage x-ray beam qualities. Different field sizes at SSDs of 50 cm, 60 cm, 70 cm, 80 cm, 90 cm and 100 cm were measured. The field sizes were confirmed radiographically.

2.3 Measurement of Tissue-Phantom Ratios

$\text{TPR}_{1,2}$ and $\text{TPR}_{2,5}$ were also measured in the water and solid phantoms for the superficial and orthovoltage beams, respectively. These were also measured for different field sizes and HVLs. $\text{TPR}_{1,2}$ was measured at SCDs of 60 cm, 70 cm and 80 cm and $\text{TPR}_{2,5}$ on the other hand, was measured at SCDs of 60 cm, 70 cm, 80 cm, 90 cm and 100 cm.

2.4 Correlation of Measured Ratios with Conversion Factors

The experimental $D_1/D_2$ and $D_2/D_5$ were correlated to the published BSF and PDD data given by Aird et al (1996). Values of $(\mu_{en}/\rho)_{w,\text{air}}$ averaged over the photon spectrum at 2 cm depth in water for a 10 cm $\times$ 10 cm field size at an SSD of 50 cm, were interpolated from the data given by Ma and Seuntjens (1999). The aim was to study the relationship to $D_2/D_5$ in order to establish if there was agreement with recent publications. Free in-air values of $(\mu_{en}/\rho)_{w,\text{air}}$ were also calculated for the orthovoltage x-ray beams using the free an-air data given by Coffey et al (2001). PDDs at depths 1 cm, 2 cm and 5 cm were interpolated from the central-axis depth dose data given by Aird et al (1996). These were then converted to PDDs at an SSD.
50 cm for the superficial x-rays using equation 1.6. The aim was to investigate the agreement between published dose ratios and the measured $D_1/D_2$ and $D_2/D_5$, respectively.
3 RESULTS AND DISCUSSION

Current dosimetry protocols for kilovoltage x-ray radiotherapy are based on the air kerma response of a dosimeter. These protocols utilize HVL and/or kVp for beam quality specification. HVL and kVp are then used to define conversion factors such as: $(\overline{\mu}_{en}/\rho)_{w,air}$ which converts air kerma to water kerma, and $B_w$ (or BSF) to convert to the dose at the surface of the water phantom. Recent investigations however, show that using only the HVL and kVp is insufficient to specify the quality of kilovoltage x-ray beams (Andreo et al. 2001). Beams of a particular HVL for example, can be produced by either light filtration of a higher kVp beam or heavy filtration of lower kVp beam. Similarly, clinical beams were found to have a wide range of HVLs corresponding to the same kVp. A wide range of $(\overline{\mu}_{en}/\rho)_{w,air}$ and $B_w$ corresponds to the same HVL and kVp.

HVL is a quantity that describes the primary photon fluence free in-air. It does not cater for changes in the photon fluence at depth due to the thickness of the overlying phantom material. Also, HVL does not take into account changes in the photon fluence at the surface of a phantom due to the thickness of the underlying scatter material. Moreover, HVL does not take into account changes in the photon fluence due to the size of the beam cross section both in air and at depth. The effects of scatter are bypassed with the concept of HVL. However, in the clinical situation both the primary fluence and scatter contribute to the dose deposited at a point in a phantom. Thus, HVL is not a representative quality index for dose measurements in the clinical situation.
3.1 Alternative Approaches to Kilovoltage X-Ray Beam Dosimetry

Clearly, there is a need for an alternative quality index in kilovoltage x-rays. The most suitable quantity could be one directly related to the absorbed dose to water, which is the end point of clinical dosimetry measurements for radiotherapy. The Code of Practice given by Andreo et al (2001) for high-energy photon beams for example, employs the tissue-phantom ratio of depths of 20 cm to 10 cm (TPR$_{20,10}$) measured in water for a field size of 10 cm × 10 cm at an SCD of 100 cm. The protocol by Almond et al (1999) for clinical reference dosimetry of high-energy photon beams on the other hand, employs the percentage depth dose of depth 10 cm measured in water for a field size of 10 cm × 10 cm at an SSD of 100 cm. Recently published investigations showed that an analogous quantity for kilovoltage x-rays could probably be the ratio of doses at different depths in a phantom. Recommended quantities: $D_{2}/D_{5}$, $D_{2.5}/D_{5}$ and $D_{5}/D_{10}$ have been investigated as functions of HVL, $(\bar{\mu}_{en}/\rho)_{w,air}$ and $B_{w}$ (Eissa et al. 2005; Jozela 2007; Rosser 1998). These quantities correlated well. All published measurements with the exception of Jozela’s (2007) however, were taken at an SSD of 100 cm for a 10 cm × 10 cm field. These measurement conditions are not typical of the clinical situation. Further work was therefore required to suggest a more practical measurement condition for end users in radiotherapy centres and calibration laboratories.

The present work focused on depths less than or equal to 5 cm, typical of the clinical prescription. $D_{1}/D_{2}$ and TPR$_{1,2}$ were measured for superficial x-ray beams, while $D_{2}/D_{5}$ and TPR$_{2,5}$ were measured for orthovoltage x-ray beams, for a range of field sizes and distances from the source.

Figure 3.1 shows how $D_{1}/D_{2}$ varied with field diameter at an SSD of 50 cm for the two clinical beam qualities at 95 kV with the different added filtrations. Also shown
is tabulated $D_1/D_2$ data from Aird et al (1996) for the equivalent HVLs and SSDs. The standard deviations in the values of $D_1/D_2$ were too small to be indicated as error bars in the figure. The $D_1/D_2$ measurements reflected the increase in the penetration of low-energy x-ray beams due to beam hardening. $D_1/D_2$ decreased rapidly with an increase in field diameter for field diameters less than 11.28 cm at an SSD of 50 cm. For field diameters greater than 11.28 cm, $D_1/D_2$ decreased more slowly. This implies that investigations of $D_1/D_2$ to characterize superficial x-ray beams at depth should be made for equivalent field diameters of at least 11.28 cm (10 cm × 10 cm) at an SSD of 50 cm. Changes in $D_1/D_2$ with field diameter result mainly from in-phantom scatter at these field sizes and minimal collimator scatter influences the measurements. $D_1/D_2$ decreased with an increase in HVL as expected however, there was inadequate data to relate $D_1/D_2$ to different HVLs and SSDs.

Figure 3.1 Variation of $D_1/D_2$ with field diameter for superficial x-ray beam qualities at an SSD of 50 cm. BJR refers to the ratios published by Aird et al (1996).
D₁/D₂ values were found to deviate significantly from the PDD₁,₂ values published by Aird et al (1996) for both the 3.25 mm Al and 4.04 mm Al HVL beams, at an SSD of 50 cm and for all field sizes investigated. The percentage error between measured and published data could be due to the uncertainty in the HVL measurement, as well as the uncertainty introduced by changing the position of the ionization chamber with depth when using percentage depth dose ratios. Using a 0.6 cc cylindrical ionization chamber as close as 1 cm to the phantom surface probably also introduced a significant perturbation effect, which was not taken into account. Differences in HVL corresponding to the same tube potential as shown in figure 1.4 could also cause differences between the measured and published dose ratios. These measurements should therefore, be verified and confirmed with a low energy parallel plate x-ray ionization chamber with a uniform energy response.

Figure 3.2 shows how D₂/D₅ varied with field diameter for different HVL beams at an SSD of 50 cm. Again, the D₂/D₅ data from Aird et al (1996) for the equivalent HVL beams and SSD is also plotted. The ratio of doses at depths 2 cm and 5 cm (D₂/D₅) proved to be a practical quality index for medium-energy x-ray beams, irrespective of the unit used to generate the beam. D₂/D₅ varied significantly with SSD, field size and HVL when measured for the orthovoltage x-ray beams generated between 105 kV and 300 kV. For measurements taken at an SSD of 50 cm, D₂/D₅ was found to decrease rapidly with an increase in field size for equivalent field diameters less than 11.28 cm (10 cm × 10 cm), and then more slowly for field diameters greater than 11.28 cm. For clinical dosimetry measurements at an SSD of 50 cm therefore, D₂/D₅ should again be measured for field diameters of at least 11.28 cm (i.e. 10 cm × 10 cm). At these field sizes, the in-phantom scatter should overshadow the air and applicator scatter contributions to the measurements.
Figure 3.2 Variation of D₂/D₅ with field diameter for different HVL beams at an SSD of 50 cm. Standard deviations in D₂/D₅ were too small to be shown as error bars to each datum point. BJR refers to the ratios published by Aird et al (1996).

D₂/D₅ also varied with SSD and HVL, decreasing with an increase in each of these variables. Values of D₂/D₅ at SSDs f₁ and f₂ are theoretically related by the Mayneord factor F as given in equations 1.6 and 1.7. This relationship was found to be accurately obeyed at SSDs of 50 cm and 100 cm. For measurements made at other SSDs (that is, 60 cm, 70 cm, 80 cm and 90 cm) however, a percentage error of up to ±2.0% was incurred in the validation of this relationship per field size and HVL. This was within the positional uncertainty of the ionization chamber each time the measurement depth was changed per SSD and field size. Figure 3.3 shows how D₂/D₅ varied with SSD for different field diameters and HVLs. Also shown is the BJR data.
(extracted from Aird et al. 1996) for the equivalent field diameters and HVL. The maximum deviation from the BJR data was within 1.0%.

![Graph showing the variation of D2/D5 with SSD for different HVLs (mm Cu). Standard deviations in D2/D5 were too small to be indicated as error bars to each datum point. A radiographic check was used to verify that the field diameters were 5 cm, 6 cm, 7 cm, 8 cm and 9 cm at SSDs of 60 cm, 70 cm, 80 cm, 90 cm and 100 cm, respectively.](image)

It was possible to derive an equation relating D2/D5 to HVL per SSD and field size. For a field diameter of 11.28 cm at an SSD of 50 cm for example, D2/D5 was found to vary with HVL according to the following equation.

\[
D_2/D_5 = 1.4259HVL^{-0.0417} \quad \text{for HVL (mm Cu) } \in [0.50, 4.00] \quad (3.1)
\]
This was in agreement with the results of Jozela (2007) who showed that if $D_2/D_5$ was given for a particular HVL (SSD and field size constant), it was possible to predict the value of $D_2/D_5$ at another HVL using a relationship of the above nature. A relationship like this would clearly simplify beam quality specification for orthovoltage x-ray beams.

When correlated with BSF data from Aird et al (1996), it was also possible to predict changes in BSF using $D_2/D_5$ for an SSD of 50 cm. This was in agreement with the results of Eissa et al (2005), who showed that it would be convenient to obtain a unique value of the BSF using $D_2/D_5$ rather than HVL. Measurements of $D_2/D_5$ were also found to be within $\pm 1\%$ agreement with the published central axis depth dose data of Aird et al (1996), for different HVLs and field sizes. Figure 3.4 shows how $D_2/D_5$ varied with BSF for the orthovoltage x-ray qualities at an SSD of 50 cm. Figure 3.5 shows how $D_2/D_5$ varied with the ratio of the published PDDs at depths.

![Figure 3.4 Variation of $D_2/D_5$ with published water kerma BSF at an SSD of 50 cm for orthovoltage x-ray beams produced by different added filtrations and field sizes.](image)

41
Figure 3.5 Variation of the ratios of absorbed doses with the published ratios of PDDs for superficial and orthovoltage x-ray qualities. The left lower dotted curve shows the variation of PDD\(_{1,2}\) with \(D_{1}/D_{2}\), and all other curves show the variation of PDD\(_{2,5}\) with \(D_{2}/D_{5}\). As seen in the figure, orthovoltage measurements were in good agreement with published data.

2 cm and 5 cm (PDD\(_{2,5}\)) respectively, for selected beam qualities at an SSD of 50 cm.

When \(D_{2}/D_{5}\) was correlated with the free in-air data of Coffey et al (2001), it was found to increase with a decrease in \((\overline{\mu}_{en}/\rho)_{w,air}\) for different SSDs and field sizes. When correlated with the ratios of mean mass energy absorption coefficients given by Ma and Seuntjens (1999) averaged over the photon spectrum at 2 cm depth in water, \(D_{2}/D_{5}\) was found to decrease with an increase in \((\overline{\mu}_{en}/\rho)_{w,air}\). This was in agreement with the results of Jozela (2007) and Rosser (1998), both of whom showed that \(D_{2}/D_{5}\) could be a well-situated index of \((\overline{\mu}_{en}/\rho)_{w,air}\) for clinical, orthovoltage x-ray
beam dosimetry. Figure 3.6 shows how $D_2/D_5$ varied with $(\bar{\mu}_{en}/\rho)_{w,air}$ averaged over the photon spectrum at 2 cm depth in water. Figures 3.7 shows how $D_2/D_5$ varied with the free in-air $(\bar{\mu}_{en}/\rho)_{air}$ for three different equivalent field diameters.

![Graph showing variation of $D_2/D_5$ with the ratio of mean mass energy coefficients of water to air averaged over the photon spectrum at 2 cm depth in water. These data were derived for orthovoltage x-ray beams generated at different kV_p and with different HVLs from Ma and Seuntjens (1999) for a 10 cm × 10 cm field size at an SSD of 50 cm.]

Measurements of $D_2/D_5$ on the two different units were indistinguishable to within ±1.0 %. $D_2/D_5$ could therefore potentially characterize orthovoltage x-ray beams irrespective of the therapy unit used to generate the user’s beam.
Free in-air Mean Mass Energy Absorption Coefficient Ratios of Water to Air

SSD = 60 cm, Ø = 5 cm
SSD = 80 cm, Ø = 7 cm
SSD = 100 cm, Ø = 9 cm

Figure 3.7 Variation of $D_2/D_5$ with calculated free in-air mean mass energy absorption coefficient ratios of water to air $\left[\left(\overline{\mu}_{en}/\rho\right)_w\right]_{air}$ for three SSDs and field diameters. Only the HVL was varied to extract $\left[\left(\overline{\mu}_{en}/\rho\right)_w\right]_{air}$ from the free-in-air data given by Coffey et al (2001), irrespective of $\phi$.

Measurements of the tissue-phantom ratio of depths 2 cm to 5 cm (TPR$_{2,5}$) confirmed its potential to accurately characterize orthovoltage x-ray beams at depth. TPR$_{2,5}$ was found to vary primarily with field size and HVL, irrespective of the SCD. TPR$_{2,5}$ varied significantly with field size for field diameters of at least 5 cm, decreasing with an increase in field size. Figures 3.8 and 3.9 show how TPR$_{2,5}$ varied with field size and HVL (mm Cu) for selected field diameters ($\phi$), respectively. There are no error bars to each datum point due to the very small standard deviations in the measurements.
Figure 3.8 Variation of $\text{TPR}_{2,5}$ with the field diameter ($\phi$) for orthovoltage x-ray beam qualities.

Figure 3.9 Variation of $\text{TPR}_{2,5}$ with HVL (mm Cu) for different field diameters ($\phi$).
TPR_{2.5} was found to be a more practical and reliable quantity to measure in that it varied appreciably with field size and HVL, irrespective of SCD. This was confirmed by graphical solutions to different field sizes and HVLs. The value of TPR_{2.5} was therefore predictable for several field sizes and HVLs. The following equations were found to approximate the variation of TPR_{2.5} with HVL for different equivalent field diameters Ø:

\[
\begin{align*}
Ø= 5\text{cm}: & \quad \text{TPR}_{2.5} = 1.558\text{HVL}^{-0.0571}; \quad \text{HVL (mm Cu)} \in [0.50, 4.00] \\
Ø= 7\text{cm}: & \quad \text{TPR}_{2.5} = 1.450\text{HVL}^{-0.0562}; \quad \text{HVL (mm Cu)} \in [0.50, 4.00] \\
Ø= 9\text{cm}: & \quad \text{TPR}_{2.5} = 1.408\text{HVL}^{-0.0536}; \quad \text{HVL (mm Cu)} \in [0.50, 4.00]
\end{align*}
\]

Changing the SCD did not alter values of TPR_{2.5} for the same field diameters and HVLs. Furthermore, the measurements of TPR_{2.5} on the two different units for the same HVLs and field diameters were in agreement to within ± 1.0 %. Thus TPR_{2.5} has the potential to accurately characterize orthovoltage x-rays irrespective of the user’s beam.

The most practical conditions for the measurement of the tissue-phantom ratio of depths 1 cm to 2 cm (TPR_{1,2}) as a potential beam quality specifier could not be established. Figure 3.10 shows the variation of TPR_{1,2} with field size for the superficial x-ray qualities. There are no error bars to each datum point due to the very small standard deviations in the readings. It is clear that more HVLs and SCDs should be investigated to come up with meaningful TPR_{1,2} data.
Figure 3.10 Variation of TPR$_{1,2}$ with field diameter at different SCDs for two superficial x-ray beam qualities produced by different added filtrations at 95 kV.
4 CONCLUSIONS

1. This work showed that $D_1/D_2$ could be a practical quality index for superficial x-ray beams when measured for equivalent field diameters of at least 11.28 cm at an SSD of 50 cm. This quantity however, needs to be verified and investigated further using a low energy parallel plate chamber for instance, to minimize perturbation effects.

2. More HVLs and SCDs should be investigated in order to establish the most practical conditions for the measurement of $TPR_{1,2}$ as a potential beam quality specifier for superficial x-rays. No definitive correlation could be obtained from this work.

3. $D_2/D_5$ was found to be a convenient beam quality index for orthovoltage x-ray beams when measured for equivalent field diameters of at least 11.28 cm (i.e. 10 cm × 10 cm) at an SSD of 50 cm.

4. $TPR_{2.5}$ was found to be more accurate than $D_2/D_5$ and this work showed that it should be investigated further as a beam quality index for orthovoltage x-ray beams. Two different units correlated well in this study. The TPR data varied appreciably with field size and HVL, giving rise to a broad range of data, which could result in a unique qualifier of orthovoltage x-ray beam quality.
REFERENCES


Dove, EL 2003, Physics of medical imaging, Biomedical Engineering-University of Iowa.


Ejere, D 2006, Optimization of cone-beam CT image quality for image guided radiotherapy, Amsterdam: University of Amsterdam.


ICRU, 1976, *Determination of absorbed dose in a patient irradiated by beams of x or gamma rays in radiotherapy procedures*.


Glossary

A number of quantities and units have been defined for describing the radiation beam. Radiation units that might come into play in kilovoltage x-ray beam dosimetry, and therefore should be drawn to the reader’s attention include: Exposure, Air Kerma, Absorbed-Dose, Equivalent Dose, Effective Dose, Energy Fluence, Particle Fluence, Photon Fluence, and Fluence Spectrum.

**Exposure:** This is a measure of the quantity of ionization produced in air, by x- or gamma radiation per unit mass. The SI unit is the Coulomb per kilogram [C/kg]. The older unit is the Roentgen [R].

**Air Kerma:** This quantity is often used as an alternative to radiation exposure, particularly in diagnostic radiology. Kerma is an acronym for Kinetic Energy Released per unit Mass. The SI unit is the Gray [Gy]. An air kerma of 1 Gy represents the transfer of 1 Joule of energy from the radiation beam to air (per kilogram of air). An advantage of using air kerma is that it translates roughly to tissue (and hence) skin absorbed dose.

**Absorbed-Dose:** This is a measure of the amount of energy imparted to matter by ionizing radiation per unit mass of irradiated material. The SI unit is the Gray [Gy]. The older unit is the rad [radiation absorbed dose].

**Equivalent Dose:** This quantity takes into account the biological damage caused by different types of radiation. The SI unit is the Sievert [Sv]. The older unit is the rem [roentgen equivalent man]. The sub-unit, millisievert [mSv], is used more often because of the large size of the Sievert.
For exposure to x-rays, Equivalent Dose and Absorbed-Dose have the same numerical value.

**Effective Dose:** This quantity was introduced for radiation protection purposes. It correlates well with the overall harmful effects caused by exposure to the various types and distribution of ionizing radiation. The SI unit is the Sievert [Sv].

**Photon Fluence and Energy Fluence:** are usually used to describe photon beams and may also be used in describing charged particle beams.

- The **particle fluence** $\Phi$ is defined as the quotient $dN/dA$, where $dN$ is the number of particles incident on a sphere of cross-sectional area $dA$:

  $$\Phi = \frac{dN}{dA} \text{ [m}^2\text{]}$$

- The **energy fluence** $\Psi$ is defined as the quotient of $dE$ by $dA$, where $dE$ is the radiant energy incident on a sphere of cross-sectional area $dA$:

  $$\Psi = \frac{dE}{dA} \text{ [Jm}^2\text{]}$$

The concepts of particle fluence spectrum and energy fluence spectrum replace the particle fluence and energy, respectively, in polyenergetic radiation beams.


APPENDIX A. DOSIMETRIC CONVERSION FACTORS

Certain conversion factors are required to determine the dose at a point in media. Of interest in this research were the water kerma based backscatter factors $B_w$, the percentage depth doses PDDs, and the ratios of mass energy-absorption coefficients of water to air $(\mu_{en}/\rho)_{water, air}$. Such factors exhibit beam quality and medium material dependence. $(\mu_{en}/\rho)_{water, air}$ is used to convert air kerma in air to water kerma in a small mass of medium in air, yet $B_w$ converts water kerma in air to the dose at the surface of water phantom. PDD at most converts the dose at a central axis point in a medium to the dose at the depth of maximum dose. Results of this research demonstrated that the dosimetric ratios $D_1/D_2$ and $D_2/D_5$ exhibit field size and beam quality dependence. Also, $TPR_{1, 2}$ and $TPR_{2, 5}$ depend on the field size and beam quality. Such ratios can, therefore, be related to the backscatter factors, PDDs and mass-energy absorption coefficient ratios. In fact, PDD, $B_w$ and $(\mu_{en}/\rho)_{water, air}$ can be intrinsic properties (or components) of these dosimetric ratios.

For the purposes of this research, PDDs were calculated from tables A.1 and A.2 by the Lagrange interpolation technique. Tables A.3 and A.4 give the BSF data that were utilized for the purposes of this research. Table A.5 was used to derive the ratios of mean mass energy absorption coefficients of water to air averaged over the photon spectrum at 2 cm depth in water. Table A.6 gives the $[(\mu_{en}/\rho)_{water, air}]_{air}$ data that were utilized for the purposes of this research.
Table A.1 Percentage Depth Doses as a function of field diameter $\phi$ and HVL (mm Al) at depths 1 cm and 2 cm. These data were taken from Aird et al (1996) for an SSD of 30 cm.

<table>
<thead>
<tr>
<th>HVL (mm Al)</th>
<th>$\phi = 4$</th>
<th>5</th>
<th>6</th>
<th>8</th>
<th>10</th>
<th>15</th>
<th>20</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.00</td>
<td>78.9</td>
<td>80.6</td>
<td>81.8</td>
<td>83.5</td>
<td>84.5</td>
<td>86.6</td>
<td>87.8</td>
</tr>
<tr>
<td>4.00</td>
<td>81.8</td>
<td>83.3</td>
<td>84.4</td>
<td>86.1</td>
<td>87.2</td>
<td>89.0</td>
<td>90.3</td>
</tr>
<tr>
<td>8.00</td>
<td>85.0</td>
<td>86.5</td>
<td>88.0</td>
<td>90.0</td>
<td>91.5</td>
<td>93.5</td>
<td>95.0</td>
</tr>
</tbody>
</table>

Depth 2 cm

<table>
<thead>
<tr>
<th>HVL (mm Al)</th>
<th>0.5</th>
<th>1.0</th>
<th>2.0</th>
<th>3.0</th>
<th>4.0</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.00</td>
<td>60.5</td>
<td>62.9</td>
<td>64.8</td>
<td>67.1</td>
<td>68.8</td>
</tr>
<tr>
<td>4.00</td>
<td>63.8</td>
<td>66.3</td>
<td>68.3</td>
<td>71.1</td>
<td>73.3</td>
</tr>
<tr>
<td>8.00</td>
<td>70.5</td>
<td>73.0</td>
<td>75.0</td>
<td>79.0</td>
<td>82.5</td>
</tr>
</tbody>
</table>

Table A.2 PDD as a function of field diameter $\phi$ and HVL (mm Cu) at depths 2 cm and 5 cm for medium-energy x-rays. These data were taken from Aird et al (1996) for an SSD of 50 cm.

<table>
<thead>
<tr>
<th>HVL (mm Cu)</th>
<th>Depth 2 cm</th>
<th>Depth 5 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>4.51</td>
<td>5.64</td>
<td>6.77</td>
</tr>
<tr>
<td>0.5</td>
<td>74.4</td>
<td>77.1</td>
</tr>
<tr>
<td>1.0</td>
<td>78.5</td>
<td>80.5</td>
</tr>
<tr>
<td>2.0</td>
<td>80.3</td>
<td>81.8</td>
</tr>
<tr>
<td>3.0</td>
<td>80.1</td>
<td>81.9</td>
</tr>
<tr>
<td>4.0</td>
<td>80.9</td>
<td>83.0</td>
</tr>
</tbody>
</table>

Depth 5 cm

<table>
<thead>
<tr>
<th>HVL (mm Cu)</th>
<th>Depth 5 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.5</td>
<td>41.6</td>
</tr>
<tr>
<td>1.0</td>
<td>46.6</td>
</tr>
<tr>
<td>2.0</td>
<td>49.3</td>
</tr>
<tr>
<td>3.0</td>
<td>50.7</td>
</tr>
<tr>
<td>4.0</td>
<td>52.3</td>
</tr>
</tbody>
</table>
Table A.3 Water kerma backscatter factors BSF as a function of HVL (mm Al) and field diameter ($\phi$) at 30 cm SSD. This data was extracted from Aird et al (1996).

<table>
<thead>
<tr>
<th>$\phi$ (cm)</th>
<th>0</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
<th>8</th>
<th>10</th>
<th>15</th>
</tr>
</thead>
<tbody>
<tr>
<td>HVL (mm Al)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1.0</td>
<td>1.0</td>
<td>1.05</td>
<td>1.09</td>
<td>1.11</td>
<td>1.12</td>
<td>1.13</td>
<td>1.14</td>
<td>1.15</td>
<td>1.16</td>
<td>1.17</td>
</tr>
<tr>
<td>2.0</td>
<td>1.0</td>
<td>1.06</td>
<td>1.11</td>
<td>1.14</td>
<td>1.16</td>
<td>1.19</td>
<td>1.20</td>
<td>1.22</td>
<td>1.24</td>
<td>1.27</td>
</tr>
<tr>
<td>3.0</td>
<td>1.0</td>
<td>1.06</td>
<td>1.12</td>
<td>1.16</td>
<td>1.19</td>
<td>1.22</td>
<td>1.24</td>
<td>1.27</td>
<td>1.29</td>
<td>1.33</td>
</tr>
<tr>
<td>4.0</td>
<td>1.0</td>
<td>1.06</td>
<td>1.12</td>
<td>1.17</td>
<td>1.20</td>
<td>1.23</td>
<td>1.26</td>
<td>1.30</td>
<td>1.33</td>
<td>1.38</td>
</tr>
<tr>
<td>8.0</td>
<td>1.0</td>
<td>1.05</td>
<td>1.11</td>
<td>1.16</td>
<td>1.20</td>
<td>1.24</td>
<td>1.27</td>
<td>1.32</td>
<td>1.36</td>
<td>1.43</td>
</tr>
</tbody>
</table>

Table A.4 Backscatter factors for medium-energy x-rays expressed as a function of field size and HVL (mm Cu). These data were extracted from Aird et al (1996) for an SSD of 50 cm.

<table>
<thead>
<tr>
<th>Field size (cm)</th>
<th>4 × 4</th>
<th>5 × 5</th>
<th>6 × 6</th>
<th>7 × 7</th>
<th>8 × 8</th>
<th>10 × 10</th>
<th>12 × 12</th>
<th>15 × 15</th>
<th>20 × 20</th>
</tr>
</thead>
<tbody>
<tr>
<td>Equivalent Diameter (cm)</td>
<td>4.51</td>
<td>5.64</td>
<td>6.77</td>
<td>7.90</td>
<td>9.03</td>
<td>11.28</td>
<td>13.54</td>
<td>16.93</td>
<td>22.57</td>
</tr>
<tr>
<td>HVL (mm Cu)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0.5</td>
<td>1.224</td>
<td>1.263</td>
<td>1.296</td>
<td>1.327</td>
<td>1.352</td>
<td>1.392</td>
<td>1.423</td>
<td>1.462</td>
<td>1.510</td>
</tr>
<tr>
<td>1.0</td>
<td>1.194</td>
<td>1.232</td>
<td>1.267</td>
<td>1.295</td>
<td>1.321</td>
<td>1.368</td>
<td>1.404</td>
<td>1.448</td>
<td>1.503</td>
</tr>
<tr>
<td>2.0</td>
<td>1.151</td>
<td>1.184</td>
<td>1.212</td>
<td>1.236</td>
<td>1.259</td>
<td>1.301</td>
<td>1.335</td>
<td>1.376</td>
<td>1.431</td>
</tr>
<tr>
<td>3.0</td>
<td>1.121</td>
<td>1.145</td>
<td>1.170</td>
<td>1.193</td>
<td>1.214</td>
<td>1.252</td>
<td>1.281</td>
<td>1.317</td>
<td>1.363</td>
</tr>
<tr>
<td>4.0</td>
<td>1.098</td>
<td>1.120</td>
<td>1.140</td>
<td>1.162</td>
<td>1.180</td>
<td>1.213</td>
<td>1.241</td>
<td>1.276</td>
<td>1.318</td>
</tr>
</tbody>
</table>

Table A.5 Ratios of mean mass energy absorption coefficients of water to air averaged over the photon spectrum at 2 cm depth in water. These data were extracted from Coffey et al (2001) for a 10 cm × 10 cm at 50 cm SSD.

<table>
<thead>
<tr>
<th>HVL (mm Cu)</th>
<th>HVL (mm Al)</th>
<th>$[(\bar{\mu}<em>{en}/\rho)</em>{water}]<em>{en}/[\rho]</em>{air}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.1</td>
<td>2.9</td>
<td>1.026</td>
</tr>
<tr>
<td>0.2</td>
<td>4.8</td>
<td>1.032</td>
</tr>
<tr>
<td>0.3</td>
<td>6.3</td>
<td>1.037</td>
</tr>
<tr>
<td>0.4</td>
<td>7.5</td>
<td>1.041</td>
</tr>
<tr>
<td>0.5</td>
<td>8.5</td>
<td>1.046</td>
</tr>
<tr>
<td>0.6</td>
<td>9.3</td>
<td>1.050</td>
</tr>
<tr>
<td>0.8</td>
<td>10.8</td>
<td>1.055</td>
</tr>
<tr>
<td>1.0</td>
<td>12.0</td>
<td>1.060</td>
</tr>
<tr>
<td>1.5</td>
<td>14.2</td>
<td>1.072</td>
</tr>
<tr>
<td>2.0</td>
<td>15.8</td>
<td>1.081</td>
</tr>
<tr>
<td>3.0</td>
<td>17.9</td>
<td>1.094</td>
</tr>
<tr>
<td>4.0</td>
<td>19.3</td>
<td>1.101</td>
</tr>
<tr>
<td>5.0</td>
<td>20.3</td>
<td>1.105</td>
</tr>
</tbody>
</table>
Table A.6 Ratios of mean mass energy absorption coefficients water to air, free in air, to convert air kerma to water as a function of HVL. These data were extracted from Coffey et al (2001).

<table>
<thead>
<tr>
<th>HVL (mm Al)</th>
<th>HVL (mm Cu)</th>
<th>([\left(\frac{\mu_{en}}{\rho}\right)<em>{water}]</em>{air})</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.03</td>
<td></td>
<td>1.047</td>
</tr>
<tr>
<td>0.04</td>
<td></td>
<td>1.047</td>
</tr>
<tr>
<td>0.05</td>
<td></td>
<td>1.046</td>
</tr>
<tr>
<td>0.06</td>
<td></td>
<td>1.046</td>
</tr>
<tr>
<td>0.08</td>
<td></td>
<td>1.044</td>
</tr>
<tr>
<td>0.10</td>
<td></td>
<td>1.044</td>
</tr>
<tr>
<td>0.12</td>
<td></td>
<td>1.043</td>
</tr>
<tr>
<td>0.15</td>
<td></td>
<td>1.041</td>
</tr>
<tr>
<td>0.20</td>
<td></td>
<td>1.039</td>
</tr>
<tr>
<td>0.30</td>
<td></td>
<td>1.035</td>
</tr>
<tr>
<td>0.40</td>
<td></td>
<td>1.031</td>
</tr>
<tr>
<td>0.50</td>
<td></td>
<td>1.028</td>
</tr>
<tr>
<td>0.60</td>
<td></td>
<td>1.026</td>
</tr>
<tr>
<td>0.80</td>
<td></td>
<td>1.022</td>
</tr>
<tr>
<td>1.00</td>
<td></td>
<td>1.020</td>
</tr>
<tr>
<td>1.20</td>
<td></td>
<td>1.018</td>
</tr>
<tr>
<td>1.50</td>
<td></td>
<td>1.017</td>
</tr>
<tr>
<td>2.00</td>
<td></td>
<td>1.018</td>
</tr>
<tr>
<td>3.00</td>
<td></td>
<td>1.021</td>
</tr>
<tr>
<td>4.00</td>
<td></td>
<td>1.025</td>
</tr>
<tr>
<td>5.00</td>
<td></td>
<td>1.029</td>
</tr>
<tr>
<td>6.00</td>
<td></td>
<td>1.034</td>
</tr>
<tr>
<td>8.00</td>
<td></td>
<td>1.045</td>
</tr>
<tr>
<td>0.10</td>
<td></td>
<td>1.020</td>
</tr>
<tr>
<td>0.20</td>
<td></td>
<td>1.028</td>
</tr>
<tr>
<td>0.30</td>
<td></td>
<td>1.035</td>
</tr>
<tr>
<td>0.40</td>
<td></td>
<td>1.043</td>
</tr>
<tr>
<td>0.50</td>
<td></td>
<td>1.050</td>
</tr>
<tr>
<td>0.60</td>
<td></td>
<td>1.056</td>
</tr>
<tr>
<td>0.80</td>
<td></td>
<td>1.068</td>
</tr>
<tr>
<td>1.00</td>
<td></td>
<td>1.076</td>
</tr>
<tr>
<td>1.50</td>
<td></td>
<td>1.085</td>
</tr>
<tr>
<td>2.00</td>
<td></td>
<td>1.089</td>
</tr>
<tr>
<td>3.00</td>
<td></td>
<td>1.100</td>
</tr>
<tr>
<td>4.00</td>
<td></td>
<td>1.106</td>
</tr>
<tr>
<td>5.00</td>
<td></td>
<td>1.109</td>
</tr>
</tbody>
</table>