Chapter 1

GENERAL INTRODUCTION

1.1. Historical Background

Mammography is used in two main categories: screening and diagnosis. Screening involves examination of asymptomatic women with the aim of detecting breast lesions at an early stage before the lesion becomes palpable. Diagnostic mammography on the other hand is done on women who through physical findings or the symptoms, may show that they are considered to already have breast cancer. The use of X-ray mammography in the screening of asymptomatic women has become very common in many parts of the world, in view of the fact that breast cancer has been found to be one of the most common malignant diseases of women. Breast cancer is reported to be the second highest cause of cancer deaths in women today (after lung cancer), with an estimation of 400,600 deaths from breast cancer for 2001 (GLOBOCAN, 2000). In 2001, World Health Organization (WHO) predicted more than 1.2 million new breast cancer cases (globally) while in United States, both the American Cancer Society and the National Cancer Institute estimated approximately 192,200 new cases of invasive breast cancer cases for 2001. Male breast cancer has also been reported but this accounts for less than 1% of the total (American Cancer Society, 2001; National Cancer Institute, 1999). The reported death rates from breast cancer declined significantly between 1992 and 1996. Statistics on mammography indicate that the incidence of breast cancers per 100,000 women increased by approximately 4% during the 1980s and levelled off to about 100 cases per 100,000 women in the 1990s. Currently over 900,000 new breast cancer cases are registered annually worldwide (Siemens, 2004). Early diagnosis of breast cancer plays a leading role in improving the patient's prognosis. This is accomplished globally by mass screening of healthy women using mammography X-ray techniques. The objective of the screening exercise is to detect breast cancer at an early stage to reduce breast cancer mortality. Clinical studies have proven that with early detection and proper patient management the mortality rate is reduced

(Siemens, 2004). In the screening programs, it is essential that low radiation dose protocols be used to reduce cancer induction by radiation to an absolute minimum, and achieve quality health: the ultimate aim of medical services.

1.2. Mammography Technique

Mammography is the most sensitive technique currently available for the early detection of breast lesions and therefore is the method of choice (Fahrig and Yaffe, 1994a; Simonetti et al 1998). Mammography is a radiographic technique that uses X-rays produced from a molybdenum, rhodium or tungsten or combination of any two of them as anode or target materials with an exit window of beryllium or glass. Some older machines have glass as exit window. The mammography X-ray beam is generated when high-velocity electrons from the cathode collide with the target or anode material. Only one percent of the energy of the streams of electrons is transformed directly into X-ray energy producing bremsstrahlung and characteristic X-rays. The remaining ninety-nine percent of the energy of the electrons is converted into heat and dissipated away. To avoid excessive heating of one point of the anode material surface and cause irreparable damage to the anode material, most modern mammography machines have rotating anodes. The X-ray tube is mounted on tube housing which provides mechanical support and also serves as a container to store oil used to cool the anode during operation. The mammography machine is designed such that the generated X-ray beam is directed through a window by a primary beam restrictor and then passed through a filter[1]. The filter materials commonly used in mammography are molybdenum, rhodium and aluminium. The filtered beam is collimated to desired dimensions for patient imaging.

[1] A material inserted in the X-ray beam to alter the quality of the beam by removing unwanted energy which does not contribute to image information but increases patient dose.

Xeromammography and screen-film mammography are the two main mammography imaging techniques most commonly used for the detection and diagnosis of breast cancer.

1.2.1. Xeromammography

Most xeromammography units employ ceiling-mounted X-ray tubes with tungsten and aluminium as anode or target and filter materials respectively. Xeromammography techniques are usually operated at peak kilovoltage (kVp) ranging between 40 and 55 kVp and a tube current of 300 milliampere (mA). The image receptor is a thin sheet of photoconducting amorphous[1] selenium. Dedicated or specialised mammography X-ray machines can also be used in xeromammography with aluminium as filter material instead of molybdenum filtration which is employed in screen-film mammography. In xeromammography, the X-ray images of the breast are recorded on a uniformly charged selenium plate held in a light-proof cassette. Upon X-ray exposure, the incident X-ray beam selectively dissipates the charge on the selenium plate to form a latent image which is made visible by dusting finely divided thermoplastic powder onto the plate. To form a permanent copy of the image, the visible image is electrostatically transferred to a plastic-coated paper and followed with thermal bonding.

1.2.2 Screen-film mammography

Screen-film mammography employs a specialised or dedicated mammography X-ray unit which consists of a molybdenum target and molybdenum filter usually of thickness 30 μm. In screen-film mammography, the X-ray image of the breast is recorded on a film cassette. The

[1] An amorphous material is vitreous, glass-like, with the atoms in random positions.

mammography cassette consists of a single fluorescent[1] high-detail intensifying screen in close proximity with a single-emulsion[2] film. The formation of the X-ray image on the film involves two steps: (i) the X-ray beam emerging from the patient is transformed into a pattern of visible light by the cassette intensifying screen; (ii) the visible image is captured permanently on the film. Upon chemical development of the film with a film processor, the emulsion (silver iodobromide crystals) that contains the latent X-ray image centres is converted into specks of metallic silver.

For many years, direct exposure with medical X-ray film or industrial film was used for mammography until 1972 when the first screen-film combination was designed for the technique. Films used in screen-film combinations have higher film contrast and require significantly lower radiation exposure than the direct exposure films (approximately 50 to 100 times less) (Barnes and Frey, 1991; Rothenberg and Haus, 1995). Cassettes and screen-film combinations used in conventional mammography are designed such that the performance of the technique is enhanced. The major components of dedicated mammography X-ray machines are: appropriate beam quality, breast compression device, phototiming or automatic exposure control, focal spot-to film distance, grids and magnification.

1.2.2.1. Appropriate beam quality

The mammography X-ray tube is usually operated at kilovoltage of 25 to 32 kVp. With a molybdenum target, the mammography X-ray beam consists of bremsstrahlung and a significant component of molybdenum characteristic X-ray photons. The molybdenum

[1] A fluorescent substance emits a pinpoint of visible light whenever an incident x-ray photon undergoes a photoelectric or Compton collision with one of its atoms.
[2] Emulsion is a suspension of microscopic silver iodobromide crystals in a gelatin.

target produces characteristic X-ray peaks that occur between 17.5 and 19.6 keV. Though not too penetrating, the characteristic lines provide high subject contrast in tissue resulting in enhanced detection of calcifications and image information. To keep exposure times to a minimum, mammographic X-ray tube currents are usually high, at least 100 mA on a large focal spot and 80 mA on a small focal spot. Both mA values are accurate to ± 10% (Lee et al., 1995).

1.2.2.2. Breast compression device

The breast compression device usually a stiff polymethyl methacrylate positioned parallel to the film surface. Good breast compression is an important factor in screen-film mammography. A number of benefits are associated with good breast compression (NCRP, 1986; Barnes and Frey, 1991) namely:

i. Reduction of scattered radiation which reduces significantly the subject contrast. This in turn increases subject contrast leading to improved detection of calcification.
ii. Immobilization of the breast reduces blurring caused by patient motion.
iii. Reduction in radiation dose.
iv. Reduction of geometric unsharpness by locating breast tissue closer to the image receptor i.e. reduction of object-film distance.
v. Production of more uniformly thick breast which leads to even the penetration of the X-ray beam and a lower difference in radiographic density of the breast area under examination.
vi. Spreading the breast tissue so that suspicious lesions can be more easily detected.

1.2.2.3. Phototiming

Phototiming, also known as automatic exposure control (AEC), is designed to automatically provide the radiation exposure needed to produce a mammogram with an acceptable and consistent optical density. Most dedicated mammographic X-ray units have automatic exposure control systems. The radiation exposure is normally controlled by a radiation detector located after the image receptor. Exposure is terminated when the radiation dose received by the detector reaches a pre-determined level which corresponds to the desired optical density. For many mammographic units, the position of the radiation detector can be varied between two or more predetermined positions to facilitate the exposure of breast of differing size and density. To prevent gross over exposure of the breast, in case of automatic exposure control system failure, a guard timer is fitted.

1.2.2.4. Focal spot-to-film distance

The distance between the focal spot[1] and the X-ray film plays an important role in the sharpness of the image formed. One of the first molybdenum target X-ray units had a nominal focal spot diameter of 0.6 mm and focal spot-to-film distance of only 35 cm. It has been observed that mammography units with larger focal spot sizes and short focal spot-to-film distances have excessive geometric unsharpness[2]. Some mammographic units have been designed to produce two focal spots usually 0.3 mm for normal mammography and 0.1 mm for magnification mammography. To minimize geometric unsharpness or blurring, the focal spot size and object-to-image receptor distance should be minimized,

[1] Focal spot is the region on the x-ray tube anode material's surface from where the X-ray beams are generated.
[2] Geometric blurring is the lateral spreading of the image of a structural boundary; that is the distance over which the optical density changes between the structure of interest and its surrounding takes place (Barnes and Frey, 1991).

while maximizing the focal spot-to-object distance. Modern mammography X-ray units have been designed to operate at focal spot-to-film distances of 50 cm or more.

1.2.2.5. Grid

To further reduce scattered radiation and improve image contrast, dedicated mammographic X-ray units include grids. Grids are capable of absorbing about 50 percent of the X-rays exiting from the breast. To offset the reduction in X-rays caused by the grids, an increase in the mAs to about 2 to 2.5 times the value for non-grid techniques is required. The increase in mAs increases patient radiation dose. Grids are of two types: stationary and moving grids. Stationary grids are ultra-high-strip-density grids having extremely fine grid lines of about 80 grid lines/cm (NCRP, 1986). Moving grids are thinner than conventional Bucky[1] grids and have carbon fibre interspace material. Most modern mammographic X-ray units are designed with moving grids to blur the grid lines thereby reducing its presence in the image. A moving grid is essential to good image quality despite the associated increase patient dose. The increase in patient radiation dose associated with the use of grids may be offset by: (i) using higher tube voltage settings, (ii) increase in X-ray beam filtration and (iii) using higher speed recording systems (NCRP, 1986).

1.2.2.6. Magnification

Screen-film mammography utilizes high speed screen-film systems which allow the use of smaller focal spots with corresponding improvement in image sharpness. Magnification is

[1] Bucky is a film cassette holder, which contains a moving grid.

the enlargement of suspicious areas of a mammogram[1] with the goal of better visualization of fine tissue structures and the provision of more diagnostic information. Magnification mammography is another advantage of screen-film mammography.

1.2.2.7. Film processor

Film processor is an integral part of screen-film mammography. The film processing is the stage of the mammographic technique whereby the visible image captured permanently on the X-ray film is developed to produce an image of the breast area under examination. The film processor unit consists of developer, a fixer and water. The developer, a reducing agent or a donor of electrons donates electrons to the silver cations to reduce it to neutral silver atoms; the fixer, usually sodium or ammonium thiosulphate, removes undeveloped silver iodobromide crystals still present on the film by binding tightly with the remaining silver ions to form a water-soluble complex; the processed film is finally washed in water and dried (Wolbarst, 1993). It is important that the film processor is kept in good condition for good and quality mammograms.

The capacity of a mammographic image to convey clinically useful information is measured by three especially important parameters namely; contrast, resolution, and noise level. Since the production of high-contrast, high-resolution and low noise images with the lowest radiation dose possible is the goal of mammography, there have been several investigations carried out for the optimisation and improvement of mammography techniques (Fahrig and Yaffe, 1994a, 1994b; Lado et al., 1997; Bhat et al., 1998a, 1998b; Anastasio et al., 1998; and Kallergi et al., 1999). Dose and image quality in mammography studies conducted by Young

[1] A mammogram is the mammographic X-ray image of the breast developed on an x-ray film.

et al. (1996) concluded that for all breast thicknesses, highest contrast and therefore overall image quality is achieved using conventional mammography techniques. Review of mammography techniques and evaluation of real cost and benefit ratios by Simonetti et al. in 1998 also concluded that conventional mammography remains the most sensitive tool for breast cancer diagnosis. An X-ray beam from a screen-film mammography unit is used in this study.

1.3. Interactions of X-rays with matter

The interaction of X-rays or photons with matter results in a number of possible interaction mechanisms such as: photoelectric absorption, coherent or Rayleigh scattering, incoherent or Compton scattering, pair production and photonuclear interactions. Photoelectric absorption, Compton scattering and pair production are the most important interactions in radiation dose measurements as they lead to partial or complete transfer of photon energy to electron energy which consequently impart energy into matter. Being an elastic scattering interaction, Rayleigh scattering involves the redirection of the photon through a small angle with no energy loss. Photonuclear interactions become significant when photon energy is in excess of a few MeV (Attix, 1986). The kinematics and the interaction probabilities or interaction coefficients involve and depend largely on the energy of the incident photon and the physical properties of the target material such as atomic number, Z, and density.

1.4. Mammography Dosimetry

There is a small but non-negligible risk of radiation-induced carcinogenesis associated with an X-ray examination of the female breast (NCRP, 1986). In fact, in the mid 1970s it was postulated in the U.S. that the National Cancer Institute's screening mammography program

induced more cancers than were found (Bailer, 1977). Although more careful analysis of the radiation dose and associated risks estimates has negated this hypothesis, mammography radiation doses are of concern and are routinely monitored. The magnitude of the absorbed radiation dose to the breast from mammography X-ray beams forms an important part of the quality control of the mammographic examination since it gives an indication of the performance of the mammographic imaging system as well as an estimated risk to the patient. Breast cancer almost always arises in the glandular tissue of the breast. As a result, the mean or average radiation absorbed dose of the glandular tissue is the preferred measure of the radiation risk associated with mammography (NCRP, 1986; Rosenstein et al., 1985). The mean glandular (MGD) dose is the quantity also recommended by International Commission on Radiological Protection (ICRP, 1991) and is used by many national protocols, such as the European Protocol (CEC, 1996).

Mammographic dosimetry is primarily to assess the risk of radiation-induced carcinogenesis in mammographic examinations. Breast dose assessment has therefore been recommended to be included in every mammographic quality assurance programme by some national protocols and institutions such as the European Protocol, using the mean glandular dose (MGD) as the risk assessment parameter since the glandular tissue is the most vulnerable of the breast tissues (European Protocol (CEC, 1996), the British Institute of Physical Sciences in Medicine (IPSM, 1989), and the International Commission on Radiological Protection (ICRP, 1991)). Measurements of the mean glandular dose from mammography have been carried out by a number of investigators using a variety of mammographic techniques.

An accurate knowledge of the output of an X-ray tube is essential in medical diagnostics for ensuring accurate dose levels estimation and for the provision of a useful check on the diagnostic adequacy of the technique. Increased use of mammography has brought into focus the necessity for radiation dose reduction. The application of X-rays and ionizing radiations for diagnostic radiology requires that the procedure is justified and optimized and that the exposure to the patient is kept as low as possible, without compromising image information.

1.5. Introductory Description to Studies

This thesis describes a system that will provide instantaneous radiation information from patients undergoing mammography examination. Standard conventional techniques for the measurement of MGD which involves measurement of tube loading (TL) and entrance surface air kerma (ESAK) for dose calculations were employed to calculate radiation dose from mammography X-ray beams. Other methodologies also presented include: (1) direct measurement of X-ray spectra using a germanium detector and computation of mean glandular dose from the measured spectra; (2) computation of MGD from theoretically generated X-ray spectrum using a Monte Carlo simulation code for electron and photon transport, PENELOPE.

This work also focused on exploring the suitability of diamond as a detection system for mammography dosimetric application. Both single crystal diamond produced under high-pressure high-temperature (HPHT) conditions and polycrystalline diamond produced by chemical vapour deposition (CVD) were used in the study. The tissue equivalence, chemical inertness, basic material and electrical properties as well as the radiation sensitivity of diamond are important parameters in this study. Investigation into the mechanisms necessary to enhance the absorption capabilities of the diamond films was undertaken. A suitable

diamond probe has been constructed for measuring integrated dose from mammography X-ray beams.

1.5.1. Overview of Direct Measurement of Mean Glandular Dose

The standard method of estimating the mean glandular dose on patients undergoing mammography X-ray examinations is based on incident air kerma or entrance surface air kerma (ESAK) measurements without backscatter and the conversion to glandular dose using appropriate conversion factors depending on the type of phantom used (IPSM, 1994). The air kerma value may be determined either for patients or for a standard breast phantom (Stanton *et al*., 1984). Polymethyl methacrylate[1] (PMMA) also known as acrylite, Perspex, Plexiglas and Lucite are normally used as breast substitute phantoms. The incident air kerma is obtained from the product of (1) the exposure time current product (mAs) i.e. tube loading for correct exposure (recommended optical density) of the PMMA phantom and (2) the output (air kerma per mAs) of the X-ray machine with the phantom removed. The mean glandular dose is then obtained either by the relationship

$$D = Kpg.$$

K is the incident air kerma without backscatter at the specified half value layer; g is the ESAK to MGD conversion factor and the p-factor converts air kerma for the PMMA breast substitute phantom to that for the model breast (Dance et al., 1999). A direct radiation dose assessment can also be made with thermoluminescent dosemeters (TLDs) placed on the breast to

[1] PMMA is normally used as a breast substitute phantom. in view of its noble properties such as: (i) high light transmittance with a refractive index of 1.49; (ii) non conducting with high electrical resistance; (iii) unaffected by prolonged exposure to moisture; (iv) can easily be heat-molded without loss of optical clarity; (v) high strength-to-weight ration; (vi) can easily be sawed, drilled, milled and engraved (Boedeker, 2004; MatWeb, 2004). PMMA is also cheap and readily available.

determine the dose (K_s) to the entrance surface of the breast (this will include all backscattered radiation). The MGD has also been calculated through the use of energy fluence or x-ray energy spectra distribution (Boag *et al*., 1976; Shrivastava, 1981; Skubic and Fatouros, 1986; Calicchia *et al*. (1996) or computed by using Monte Carlo techniques (Dance, 1980 and Rosenstein *et al*., 1985). In Chapter 2 of the thesis, the results of the MGD calculations are presented. Part of the MGD measurements has been published and the article presented in Chapter 2 (Assiamah et al., 2004).

1.5.2. Overview of Direct X-Ray Spectra Measurement

An accurate knowledge of the output of an X-ray tube is essential in many areas of study. It forms the basis of almost all image quality simulation. Direct measurement of X-ray spectra is usually performed with a high-purity germanium detector with the signal output being processed by conventional electronics and a multichannel analyzer (MCA). Notable among these works include, Fewell and Shuping (1977), Birch and Marshall (1979), Fewell et al. (1981). Others are Aoki and Koyama (1989), Laitano et al. (1991), and Marshall et al. (1996). The relatively complex nature of X-ray spectral measurement from diagnostic X-ray machines makes direct measurement of X-ray spectra a difficult task. High photon fluxes produced from diagnostic X-ray machines and the consequent probability of pulse pileup occurring, hence distorting the detector output are the main difficulties in the measurement of X-ray spectra from a mammography X-ray machine. Additional problems are:- (i) the relatively low energy of X-rays generated by a mammography machine. These photons can be significantly attenuated by small amounts of any medium including air. (ii) The unavailability of the option to reduce the photon flux emitted from the X-ray tube by using a low current or fluoroscopic

mode on a typical mammography machine compounds the complex nature of direct X-ray measurement from an X-ray machine.

Several solutions have been found to address the problem of pile-up effects. They include, the use of an appropriate gating signal to reduce pulse pileup, the use of an air-free path between the X-ray machine and detector and the use of multi-parameter system (MPS) instead of a standard multi-channel analyzer (MCA). Application of these novel solutions by Wilkinson et al. (2001) to address the high flux production and associated pulse pileup indicate that using MCA or MPS with an air-free path in conjunction with a gating signal is an effective technique to solve the problem. However, an air-free path is difficult to implement and so is unsuitable in a clinical setting where a simple, convenient method is desirable. Wilkinson et al. have suggested the use of an MCA with a gating technique. They however, noted that neither of the techniques generates an absolute measurement of the photon flux since the pileup rejection process rejects some detected events.

In this thesis, the method of data acquisition, adoption and optimization of existing procedures for correcting or stripping spectral data with the aim of placing the incident photon in its primary and correct energy bin has been carried out. The corrected photon flux has been used to calculate mean glandular dose employing appropriate conversion factors. Mass attenuation coefficient data and mass energy-absorption coefficient data for certain materials were employed in various stages of the spectrum corrections. The attenuation coefficient data were obtained from the published results of Hubbell and Seltzer (1996). The mass attenuation and mass energy-absorption coefficient was fitted to an extension of a fifth degree polynomial

equation (Tucker et al., 1991) in order to derive the parameters needed to generate the absorption energy "spectra". This was then used in computing the air kerma *K* (E) and consequently, the MGD. In Chapter 2 the outcome of the spectral measurement and consequent computation of mean glandular dose is given. The extension to the Tucker et al. equation is found necessary, as the published data do not cover the full energy spectrum of interest. In Chapter 3 is presented the steps used to generate the absorption energy "spectra" from existing published data. The effect of ambient temperature, pressure and humidity on the calculated values of photons and air kerma were also considered and the methods involved are presented in Chapter 4 of the thesis.

1.5.3. Overview of Theoretical Computations of X-Ray Spectra

The difficulties associated with direct measurement of X-ray spectra led many researchers to develop mathematical models the use of which enables the calculation of a spectrum for a defined peak kilovoltage and X-ray target/filter combinations to generate X-ray spectra required. The earliest spectra calculations are those of Kulenkampff in 1922 and Kramers in 1923. There were two shortcomings in the Kramers model. The first was the assumption that the original differential energy intensity was constant. The second drawback was that the absorption of the X-rays in the target was not accounted for. Kramers' model has been corrected and modified by later researchers; Unsworth and Greening, (1970); Sundararaman et al., (1973); Soole (1977); and Birch and Marshall (1979). There has been tremendous development and advancement in the generation of X-ray spectra by mathematical models and calculations. Computer simulation has become a convenient and frequently used tool in the study of X-ray mammography for the calculation of radiation doses to the breast (Wu et al.,

1991); for optimizing techniques in mammography (Fahrig and Yaffe, 1994a, 1994b; Sandborg et al., 1994); and for evaluating detector performance (Chen, 1980; Maidment et al., 1993). Computer-generated spectra in mammography enables system designers to predict patient dose more accurately and, hence, aid in the development of better hardware and software systems to reduce patient dose. In recent times, several models have been proposed; they include, the Tucker et al., (1991a, 1991b); Boone and Seibert, (1997); Boone et al. (1997); and Blough et al., (1998) models. The most popular of these models for mammography has been that of Tucker et al. Their calculations attempt to model the physical processes such as self-absorption within the target, and the use of a semi-empirical approach in fitting a parameterized function to previously measured spectra. In their comparison of mammography spectral measurements with spectra produced using several different mathematical models, Wilkinson et al., (2001), have indicated that accurate results can be produced by all mathematical models, but only if the user attempts to match the calculated half value layer (HVL) of the modelled spectrum with the physically measured HVL. The modelled spectra may otherwise be in error and can lead to an underestimation of the dose calculation by up to 20%.

In this thesis a computer code system PENELOPE (PENetration and Energy Loss of Positron and Electrons) which simulates coupled electron-photon transport has been studied and used to compute radiation dose from mammography X-ray beam. The simulation algorithm is based on a scattering model that combines numerical databases with analytical cross section models for different interaction mechanisms (Salvat et al., 2001). A model was developed that

simulates the experimental setup and conditions for evaluating mammography X-ray beam radiation dose. Presented in Chapter 6 is the simulation work using the PENELOPE MC code.

1.5.4. Overview of Dosimetry with Diamond

Diamond is a unique element with attractive physical properties such as wide band gap making it an excellent insulator, wide transmission band, optically transparent, high charge carrier mobility, high breakdown voltage, high thermal conductivity and low thermal expansion coefficient, small dielectric constant, excellent radiation hardness, physically hard and chemically inert. The near tissue equivalence and chemical inertness of diamond have been identified to be important in both medical physics and health physics as a radiation monitoring tool (Nam et al., 1987). Thus, as a radiation dosimeter, it does not require large correction factors to convert the response to a true deposited radiation dose. This is because the energy response or energy absorption of diamond as a dose-sensing element per unit exposure will be similar to that of tissue.

Diamond also has low noise contribution arising from leakage currents due to the large band gap. Further, good sensitivity compared to ionization chambers has been found (Bruzzi et al., 2000) and the insensitivity to radiation damage makes diamond a material more suitable than most solid-state detectors currently being used for on-line dosimetric applications (Plankskoy, 1980; Rustgi, 1995).

The applications of synthetic diamond crystal as radiation sensing elements have been concentrated mainly on its response to charge particles. These applications have been both in nuclear and medical physics. Some of the studies include: (Burgemeister, 1981) rate ionization

chamber radiation detectors in biological environments (Keddy et al., 1987), as a thermoluminescent detector (Nam, 1989), as a sensor for measuring low dose-rates (Grobbelaar et al., 1991), as a near tissue-equivalent probe in electron radiation therapy (van der Merwe, 1994), as detectors in the heavy-ion dosimetry for tumor-therapy (Berdermann et al., 1998). There are also extensive reports on the application of synthetic diamond as radiation detectors in high energy physics studies (Meier, 1999; Adam et al., 2000; Adam et al., 2002; Berdermann et al., 2001; Shu et al., 2001; Bergonzo et al., 2003) due to the radiation hardness of diamond which results from the high energy (80 eV) needed to remove a carbon atom from the diamond lattice and the high thermal conductivity which is about five times higher than for copper.

However, its use for both low energy X-ray measurements and dose determination has not been fully investigated. Diamond has an advantage because of the lower energy required to produce a charge pair, and the fact that diamond can be used in smaller sizes than gas-filled ionization chambers, also permitting a better spatial resolution for measurements of radiation fields with steep dose gradient. The project aim was to research and utilize such key characteristics of diamond. Diamond has been used in a conduction mode whereby electrodes are connected to it and the resulting current from the interaction of the ionizing radiation with the diamond was detected. A new method for calculating total dose to tissue has been established from the studies. The studies on diamond as a radiation detector for mammography beam dosimetry are present Chapter 5. The general conclusions from each of the techniques used in this study is consolidated and reported in Chapter 7.

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