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ΤΕΧΝΙΚΕΣ ΠΡΟΣΟΜΟΙΩΣΗΣ MONTE CARLO

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BREAST DOSE DISTRIBUTION STUDIES IN MAGNIFICATION MAMMOGRAPHY USING MONTE CARLO SIMULATION

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To my family

To Pagona
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A.1. Introduction

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A.1.1. THE PROBLEM

Breast cancer is nowadays the most common form of cancer for women all over the world. Although the number of fatal cases has been reduced significantly due to the development of technology, it still remains the most usual case of death for women with cancer (American Cancer Society). Medical experts agree that successful treatment of breast cancer is linked to early diagnosis. It is through mammography screening programs that a large percentage of breast cancers can be detected and effectively treated, as mammography is currently the technique with the highest sensitivity available for early detection of breast cancer on asymptomatic women. However, mammography utilizes ionizing radiation, a known carcinogen that has a cumulative effect on the body and this fact, in addition to the fact that the target group of the procedure is every healthy woman has triggered an extensive discussion about possible risks associated with these examinations, specifically the radiation-induced breast cancer, and some risk-benefit analyses have been reported.

Dosimetry has been an established procedure in clinical practice, with the Entrance Surface Dose (ESD) being the first parameter that was associated to the potential risk of mammography. Various measures of dose were proposed as alternatives to ESD. Some authors suggested that mid-breast dose might be a better measure of risk and Boag et al (1976) proposed the use of the total energy imparted to the breast which could be measured from phantom-based measurement of depth dose curves or entrance and exit doses. However, neither of these measures was completely satisfactory. Karlsson et al (1976) suggested that the average dose to the glandular tissues within the breast could be a better measure of risk. Significant fractions of the energy absorbed by the breast are deposited in skin, fat and connective tissue, whereas it is believed that it is the glandular tissue which has the highest risk of radiation-induced carcinogenesis. Average Glandular Dose (AGD) is also the quantity recommended by the ICRP and is used in many national protocols.

Several computational procedures have been proposed in order to calculate the energy imparted in the breast. Most of these procedures are based on Monte Carlo simulation and are used in the calculation of the conversion coefficients from surface dose or air kerma to effective dose.

So, the biological “cost” of mammography can be expressed through the AGD, which is directly related to breast cancer probability (Wu et al 1991, Klein et al
1997, Hammerstein et al 1979, Stanton et al 1984, Shrivastava 1981, Karlsson et al 1976, Wu et al 1994). However, its direct measurement is difficult and air KERMA (Kinetic Energy Released per unit of Mass) to AGD conversion factors are often required (Delis et al 2004a, Dance 1990, Dance et al 2000a, Zoetelief and Jansen 1995). Although studies have been reported towards the direction of dose distribution and calculation of AGD for contact mammography using Monte Carlo simulation (Dance 1990, Dance et al 2000a, Zoetelief and Jansen 1995), only few data have been published for magnification mammography. Law (2005) and and McParland (2000) have studied experimentally the dose in magnification mammography, while Liu et al (1995) calculated the AGD in magnification mammography, using Monte Carlo simulation. In the present study, a Monte Carlo simulation program, dedicated for mammography is utilized in order to study the dose distribution, in terms of AGD, Entrance Surface Dose (ESD) and Percentage Depth Dose (PDD), in magnification mammography. More specifically, the effect of magnification factor and tube voltage in AGD and ESD is considered. Also, the influence of various anode/filter material combinations on the dose delivered to the breast is studied, and air KERMA to AGD conversion factors that can be applied to magnification geometries are also derived.
A.1.2. THESIS ORIGINALITY

The originality of this Master Thesis consists of:

- The use of a wide range of mammographic spectra as input for the simulation studies, including from the most recently introduced ones to the oldest and almost abandoned ones, to study their effect on Average Glandular Dose, Entrance Surface Dose and Percentage Depth Dose, in magnification mammography.
- The extension of a previously developed Monte Carlo simulation code (Delis et al 2004) in order to include calculation of the Average Glandular Dose.
- The adaptation of air Kinetic Energy Released per unit Mass (KERMA) to Average Glandular Breast Dose conversion factors formula in order to include magnification geometries and dependence on the tube voltage.
A.1.3. PUBLICATIONS

This work has been submitted for publication in international journal and parts of it will be presented in international conferences.

Publications in peer reviewed international journals

Publications in international conference proceedings
- M Koutalonis, H Delis, G Spyrou, L Costaridou, G Tzanakos and G Panayiotakis, “Monte Carlo assessment of Average Glandular Dose and Percentage Depth Dose in magnification mammography” ICRP 10, 10th International Symposium in Radiation Physics, September 17-22 2006, Coimbra, Portugal
A.1.4. THESIS LAYOUT

The layout of this thesis is presented as following:

Section A.2 is an introduction to mammography (contact and magnification) and section A.3 includes an overview of the main dosimetric quantities of radiation dosimetry used in this study. Information concerning the history and the basic parameters of Monte Carlo Simulation are found in Section A.4.

The materials and methods section begins with Section B.1, where the geometry of the contact and magnification mammography is presented, followed by a brief description of the simulation of the mammographic procedure in Section B.2. The mammographic spectra, as well as the breast phantom generated for the calculations are presented in Sections B.3 and B.4, with the validation of the AGD calculation being presented in Section B.5. Finally, the proposed air KERMA to Average Glandular Dose conversion factors followed by their validation are presented in Sections B.6 and B.7.

Results of the effects of the magnification factor, the tube voltage and the anode/filter material combination on the AGD, ESD and PDD are presented in Sections C.1, C.2 and C.3, with the corresponding discussion. The last chapter of the Results and Discussion section (Section C.4) contains the results, given in tables, about the conversion factors calculated for the case of magnification mammography.

Finally, some conclusions concerning the results produced during the course of this study are being presented in Section D.1, followed by suggestions of furthermore required for the enrichment of this study (Section D.2).
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A.2 MAMMOGRAPHY

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A.2.1. INTRODUCTION TO MAMMOGRAPHY

The most credible tool for the detection of breast cancer is x-ray mammography. Although the use of x-rays in order to examine the breast was first introduced more than 90 years ago, modern mammography has only existed since 1969, when the first dedicated x-ray machines used just for breast imaging became available. Since then, the technology has advanced a great deal, so that today’s mammogram is very different even from those of the mid-1980s.

Mammography is a specific technique that uses a low-dose x-ray system for the imaging of the breast. It can image early findings associated to breast cancer up to two years before becoming palpable to the patient or the physician (Sickles 1980). Images are processed using either film screen or digital techniques and can be viewed on film at a view box or as soft copy on a digital mammography work station. The resulting images are examined by one or more radiologists, who look for changes or inconsistencies in the breast tissue. A mammographic unit consists of a rectangular box that houses the tube in which x-rays are produced. The unit is dedicated equipment because it is used exclusively for breasts examinations with special accessories that allow only the breast to be exposed to the x-rays. Although current guidelines differ significantly among different organizations, the overall tendency suggests screening mammography for women annually, beginning at age of 40 (American Cancer Society). Studies have shown that annual mammograms lead to early detection of breast cancers, when they are most curable and breast-conservation therapies are available.

Two kinds of mammograms are possible, the screening and the diagnostic one. Screening mammography is the examination of the breast of non symptomatic women, while diagnostic mammography is oriented to women who either have a breast complaint (for example a breast mass or nipple discharge) or have had an abnormality found during screening mammography. During diagnostic mammography, more images are required to carefully study the breast condition. Magnification mammography is an extra step from the routine views and it is only performed when there is something that requires a “closer” look, such as a suspicious abnormality or a differentiation from a previous mammogram (Hermann et al 2002).
During the magnification technique, the breast is placed closer to the x-ray focus, while the image receptor is at a remote distance (Liu et al 1995), (Figure 3). The advantages of magnification mammography compared to the conventional (contact) one are related to higher contrast, reduced image noise and improved spatial resolution, as far as the image quality is concerned (Funke et al 1998, Funke et al 1997, Doi and Imhof 1977, Sickles et al 1977, Kohama et al 2004, Liu et al 1995, Spyrou et al 2002). Moreover, the air gap between the breast and the image receptor reduces the scattered radiation, making the use of an antiscatter grid unnecessary.

On the other hand, magnification mammography has some disadvantages. The dose deposited in the breast is higher due to the decreased focus to skin distance (FSD), combined with the decreased irradiated volume of the breast. Also, in order to reduce the unsharpness caused by the smaller FSD, a smaller focal spot is required, which means that the tube current has to be reduced in order to protect the anode from being destroyed due to the increased thermal loading (Funke et al 1997, Liu et al 1995). This in turns, results in an increase of the exposure time and possible appearance of motion unsharpness in the image due to anatomical motion (Hermann et al 2002, Spyrou et al 2002).

**A.2.2. BENEFITS AND RISKS IN MAMMOGRAPHY**

The profitable role of mammography and the contribution to the early diagnosis and therefore to the cure possibility of breast cancer has already been mentioned. However, mammography, just like every other radiation based technique is not free of any risk. Following, the main benefits and risks are summarized for the mammographic practice.

**Benefits**

- Imaging of the breast improves a physician’s ability to detect small tumors. When cancers are small, the woman has more treatment options and a cure is more likely.
- The use of screening mammography increases the detection of small abnormal tissue growths confined to the milk ducts in the breast, called ductal carcinoma in situ. These tumors cannot harm patients if they are removed at this early stage and mammography is the only proven method to reliably detect these tumors.
Risks

- The effective radiation dose from a mammogram is about 0.3 mSv, which is about the same as the average background radiation a person receives in one month. Although the magnitude may not seem important, the fact that mammography is a non-symptomatic technique makes the dose limitation task obligatory.

- False Positive Mammograms. Five to ten percent of screening mammography results are abnormal and require more testing (additional mammograms, fine needle aspiration, ultrasound or biopsy), while most of the follow-up tests confirm that no cancer was present. It is estimated that a woman who has yearly mammograms between ages 40 and 49 would have about a 30% chance of having a false-positive mammogram at some point in that decade and about 7 to 8 percent chance of having a breast biopsy within the 10-year period. The same estimate of false-positive mammograms is about 25% for women over the age of 50.

Mammographic technique aims towards the optimization and the implementation of the ALARA principle (As Low As Reasonably Achievable). To this direction several studies have been published concerning combined studies for both image quality and radiation dose in mammography (Ng et al 2000, Martin et al 1999, Calicchia et al 1996, Court and Speller 1995, Fahrig and Yaffe 1994, Guibelable et al 1994, Desponds et al 1991, Thilander et al 1989) investigating possible ways of minimizing the risks of the procedure, while keeping the image quality in acceptable diagnostic level.
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A.3 Radiation Dosimetry

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A.3.1.1. ABSORBED DOSE

Absorbed dose is a non-stochastic quantity applicable to both indirectly and directly ionizing radiations. For indirectly ionizing radiations, energy is imparted to matter in a two step process. In the first step the indirectly ionizing radiation transfers energy as kinetic energy to secondary charged particles. In the second step these charged particles transfer some of their kinetic energy to the medium resulting in absorbed dose and lose some of their energy in the form of bremsstrahlung losses.

The absorbed dose is related to the stochastic quantity energy imparted and is defined as the mean energy $E$ imparted by ionizing radiation to matter of mass $m$ in a finite volume $V$ by:

$$D = \frac{dE}{dm}$$

The unit for absorbed dose is Joule per kilogram (Jkg$^{-1}$) and the special name for this unit is gray (Gy). It corresponds to 1 joule of energy deposited in 1 kilogram of water.

Since the maximum dose for the diagnostic energy range is deposited near the surface of the patient, a very commonly used dose value is the Entrance Surface Dose (ESD), referring to the dose deposited to the upper layer of the patient’s skin.

A.3.1.2. AVERAGE GLANDULAR DOSE

Average Glandular Dose (AGD) is defined as the mean dose deposited only in the mammary gland of the breast. As there is a significant risk of radiation induced carcinogenesis associated with x-ray mammography, the determination of AGD forms an important part of the quality control of mammographic imaging systems. Because of the difficulty of estimating AGD directly, the entrance air kerma at the upper surface of the breast is determined and the AGD is calculated by multiplying by appropriate conversion factors.
A.3.1.3. KINETIC ENERGY RELEASED PER UNIT MASS

Kinetic Energy Released per unit Mass (KERMA) is a non-stochastic quantity applicable to indirectly ionizing radiations, such as photons and neutrons. It quantifies the average amount of energy transferred from the indirectly ionizing radiation to directly ionizing radiation (charged particles-electrons) without concerns to what happens after this transfer, and is defined as the energy transferred per unit mass:

$$K = \frac{dE}{dm}$$  \hspace{1cm} (2)

The unit of KERMA is joule per kilogram (Jkg$^{-1}$) and the special name for this unit is gray (Gy), where 1 Gy = 1 Jkg$^{-1}$.

A.3.1.4. PERCENTAGE DEPTH DOSE

Dose distributions inside a phantom or a patient are usually normalized to $D_{max} = 100\%$ at the depth of dose maximum $z_{max}$ and then referred to as the percentage depth dose distributions. The percentage depth dose is thus defined as follows:

$$PDD = 100 \frac{D_Q}{D_P}$$  \hspace{1cm} (3)

where $D_Q$ is the dose at a point Q at depth z and $D_P$ is the dose at point P at $z_{max}$. 
CHAPTER A
INTRODUCTION – THEORETICAL PART

A.4 Monte Carlo Simulation
A.3.1. INTRODUCTION TO MONTE CARLO METHODS

Numerical methods that are known as Monte Carlo methods can be described as statistical simulation methods, where statistical simulation is defined in quite general terms to be any method that utilizes sequences of random numbers to perform the simulation. Monte Carlo methods have been used for centuries, but only in the past several decades has the technique gained the status of a full-fledged numerical method capable of addressing the most complex applications. The name “Monte Carlo” was coined by Metropolis (inspired by Ulam's interest in poker) during the Manhattan Project of World War II, because of the similarity of statistical simulation to games of chance, and because the capital of Monaco was a center for gambling and similar pursuits. Monte Carlo is now used routinely in many diverse fields, from the simulation of complex physical phenomena such as radiation transport in the earth's atmosphere and the simulation of the esoteric subnuclear processes in high energy physics experiments, to the simulation of games of chance. The analogy of Monte Carlo methods to games of chance is a good one, but the “game” is a physical system, and the outcome of the game is not a pot of money or stack of chips (unless simulated) but rather a solution to some problem. The “winner” is the scientist, who judges the value of his results on their intrinsic worth, rather than the extrinsic worth of his holdings.

Statistical simulation methods may be contrasted to conventional numerical discretization methods, which typically are applied to ordinary or partial differential equations that describe some underlying physical or mathematical system. In many applications of Monte Carlo, the physical process is simulated directly, and there is no need to even write down the differential equations that describe the behaviour of the system. The only requirement is that the physical (or mathematical) system be described by probability density functions (pdf's). Once the pdf's are known, the Monte Carlo simulation can proceed by random sampling from the pdf's. Many simulations are then performed (multiple “trials” or “histories”) and the desired result is taken as an average over the number of observations (which may be a single observation or perhaps millions of observations). In many practical applications, one can predict the statistical error (the “variance”) in this average result, and hence an estimate of the number of Monte Carlo trials that are needed to achieve a given error.
Figure 1 illustrates the idea of Monte Carlo, or statistical simulation as applied to an arbitrary physical system. Assuming that the evolution of the physical system can be described by probability density functions (pdf’s), then the Monte Carlo simulation can proceed by sampling from these pdf’s, which necessitates a fast and effective way to generate random numbers uniformly distributed on the interval [0,1]. The outcomes of these random samplings, or trials, must be accumulated or tallied in an appropriate manner to produce the desired result, but the essential characteristic of Monte Carlo is the use of random sampling techniques (and perhaps other algebra to manipulate the outcomes) to arrive at a solution of the physical problem. In contrast, a conventional numerical solution approach would start with the mathematical model of the physical system, discretizing the differential equations and then solving a set of algebraic equations for the unknown state of the system.

It should be kept in mind though that this general description of Monte Carlo methods may not directly apply to some applications. It is natural to think that Monte Carlo methods are used to simulate random, or stochastic, processes, since these can be described by pdf’s. However, this coupling is actually too restrictive because many Monte Carlo applications have no apparent stochastic content, such as the evaluation of a definite integral or the inversion of a system of linear equations. However, in these cases and others, one can pose the desired solution in terms of pdf’s, and while this transformation may seem artificial, this step allows the system to be treated as a stochastic process for the purpose of simulation and hence Monte Carlo methods can
be applied to simulate the system. Therefore, we take a broad view of the definition of Monte Carlo methods and include in the Monte Carlo rubric all methods that involve statistical simulation of some underlying system, whether or not the system represents a real physical process.

Some of the applications of Monte Carlo simulation are in the following fields: nuclear reactors design, quantum chromodynamics, radiation cancer therapy, traffic flow, econometrics, VLSI design and medical imaging.

In the following table 1, some of the Monte Carlo codes that have been implemented are presented together with their applications.

<table>
<thead>
<tr>
<th>Monte Carlo Code</th>
<th>Description</th>
<th>Applications</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ho-Chunk 2004</td>
<td>calculates scattering and polarization in stellar geometries</td>
<td>computing optical/near-IR images where most of the emitted radiation comes from the central star</td>
</tr>
<tr>
<td>MCML by Lihong Wang and Steven Jacques 1992</td>
<td>a steady-state Monte Carlo simulation program for multi-layered turbid media with an infinitely narrow photon beam as the light source</td>
<td>a very flexible method for simulating light propagation in tissue</td>
</tr>
<tr>
<td>MCNP 1994</td>
<td>describes particle transport</td>
<td>criticality safety, oil-well logging, nuclear energy, nuclear safeguards, fusion research, medical technology (mammography)</td>
</tr>
<tr>
<td>FOTELP 2003</td>
<td>photon, electron and positron transport</td>
<td>dosimetry, radiation shielding, radiotherapy, detector and counter efficiency evaluation</td>
</tr>
</tbody>
</table>
**INTRODUCTION-THEORETICAL PART**

<table>
<thead>
<tr>
<th>Code</th>
<th>Description</th>
<th>Application</th>
</tr>
</thead>
<tbody>
<tr>
<td>BEAM 1995</td>
<td>simulate the radiation beams from radiotherapy units including high-energy electron and photon beams, $^{60}$Co beams and ortho-voltage units</td>
<td>radiotherapy</td>
</tr>
<tr>
<td>EGS 1994</td>
<td>general purpose package for the Monte Carlo simulation of the coupled transport of electrons and photons in an arbitrary geometry for particles with energies from a few keV up to several TeV</td>
<td>clinical radiation dosimetry (mammography)</td>
</tr>
<tr>
<td>CASINO 1995</td>
<td>a single scattering Monte Carlo Simulation of electron trajectory in solid specially designed for low-beam interaction</td>
<td>electron beam interactions</td>
</tr>
</tbody>
</table>

*Table 1* Monte Carlo codes utilized and their applications.
A.3.2. FUNDAMENTAL THEORY OF RANDOM NUMBERS

For the implementation of a Monte Carlo simulation process, the first and major part is an infinite sequence of random numbers. A random number is a particular value of a continuous variable uniformly distributed on the unit interval which, together with others of its kind, satisfies certain conditions. A high quality random number sequence is a long stream of numbers with the characteristic that the occurrence of each number in the sequence is unpredictable. Although databases of real random numbers can be found their use is limited due to the large size, which makes the simulation process very slow. Thus, mathematical algorithms for the generation of “pseudorandom numbers” have been introduced. The outputs of pseudorandom numbers generators are not random; they only approximate some of the properties of random numbers. Careful mathematical analysis is required to ensure that the generated numbers are sufficiently “random”.

By far the most popular random number generators in use today are special cases of the following scheme, introduced by D.H. Lehmer (1951). For the implementation of this scheme, four “magic numbers” are selected:

- \( m \), the modulus;
- \( a \), the multiplier;
- \( c \), the increment;
- \( X_0 \), the starting value;

Given a modulus \( m \), a multiplier \( a \) and a starting value – “seed” \( X_0 \) the desired sequence of random numbers \( X_n \) is then obtained by setting:

\[
X_{n+1}=(aX_n+c) \mod m
\]  

(4)

This is called a linear congruential sequence.

Before a random number generator can be regarded as acceptable, its output must pass certain standard randomness tests (Frequency Test, Serial Test, Gap Test, Poker Test, Serial Correlation Test etc), which check very carefully the statistical properties of uniformity and independence. The length of the period of a random number generator must be long enough to avoid repetitions in the sequence of numbers used during the simulation process, as otherwise correlations can be produced. There are simulations however, where the set of independent numbers needed might exceed the repetition period of the generator being used. When the
generator is “well behaved”, even if the sequence of numbers is used more than once, the probability of having more than one particle history starting in the same position of the sequence of random numbers is practically negligible. This means that when the end of the sequence is reached it will be started again during some of the sampling procedures used along the simulation.

The selection of a random value of a specific quantity, from a continuous probability density function is realized in Monte Carlo simulation through sampling methods, which are namely: the inversion method and the rejection method. In the code utilized in this study both methods were used.
CHAPTER B

MATERIALS & METHOD

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B.2. SIMULATION OF THE MAMMOGRAPHIC PROCEDURE
B.3. MAMMOGRAPHIC SPECTRA
B.4. MATHEMATICAL BREAST PHANTOM
B.5. VALIDATION OF THE AGD CALCULATION
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B.7. VALIDATION OF THE INDEPENDENCE OF THE FORMULA FACTORS
B.1. GEOMETRY

The desired magnification degree in magnification mammography is achieved through simple geometric changes with respect to contact mammography, as presented in figure 2. According to these changes, the breast is placed closer to the x-ray tube, while the detector remains at the same distance as in contact geometry. The antiscatter grid is also removed because its presence would cause high deposition of dose in the breast. In magnification geometries the FSD ranges from 24 cm to 56 cm, resulting in magnification factors of 2.1 to about 1.0 (contact mammography), for breast thickness of 4 cm. The focal spot dimensions utilized in the present study were 0.01 mm x 0.01 mm, in order to be compatible with systems available in clinical practice. As already mentioned before, smaller focal spot dimensions are necessary in the case of magnification mammography in order to reduce the unsharpness caused by the smaller FSD.

![Diagram](image)

**Figure 2** Contact (a) and magnification (b) geometry in mammography. In magnification geometry the antiscatter grid is removed and the breast is placed closer to the focal spot.
B.2. SIMULATION OF THE MAMMOGRAPHIC PROCEDURE

The irradiation of the mathematical breast phantom, used in this study, was simulated using a set of earlier developed and validated Monte Carlo software, DOSIS, for deriving the dose delivered to the breast during the mammographic process (Delis et al. 2005a). This software was differentiated from its initial version in order to include magnification geometries and calculation of the AGD.

The flow chart of the proposed simulation model is presented in figure 3. The “history” of each Monte Carlo photon initiates with the assignment of the initial energy which is defined through Monte Carlo sampling from the selected spectral distribution and the initial direction which is defined depending on the selected geometry. When a photon reaches the phantom it can follow a sequence of possible interactions for this energy range (photoelectric absorption, coherent and incoherent scattering). The type of the interaction (based on the atomic cross sections), the energy deposited (photoelectric, Compton) and the scattering angles (Compton, Rayleigh) have been sampled and included in the model. The air KERMA is calculated exactly before the entrance of each photon in the phantom and the AGD at the end of the photon history inside the phantom.

Figure 3 The flow chart of the proposed simulation model.
In order to minimize the statistical fluctuation of the results, the number of x-ray photons in all studies was $10^8$, resulting in execution (computational) time of 16 minutes for each simulation, utilizing a Pentium 4 processor, at 3.0 GHz.

After the assignment of the initial energy and direction, the photon enters the phantom. On the basis of the total cross-sections the free path length of the photon is defined. At the end of this free path the photon is forced to undergo an interaction, based on the sampling of the cross sections and therefore the interaction probability according to the energy that it initially had. The possible interactions for the mammographic energy range are the photoelectric effect, the coherent scatter (Compton) and the incoherent scatter (Rayleigh), with the photoelectric being by far the most dominant effect. The cross sections that were used for the simulation process come from the database of the National Institute of Standards and Technology and from Hubbel \textit{et al} (1975). The history of the photon continues until one of the following happens:

- Photoelectric effect inside the phantom; the photon deposits the total of its energy at the point of the interaction causing radiation dose.
- Escape of the photon towards the detector; the photon is recorded on a 2D matrix, representing an ideal x-ray detector, and contributes to the image formation.
- Escape the phantom towards another direction causing scattered radiation.

\textbf{B.3. MAMMOGRAPHIC SPECTRA}

The mammographic spectra utilized in this study were derived using a previously created and validated program (Delis \textit{et al} 2004). The produced spectra, were corrected for the anode self absorption, and the inherent filtration by beryllium (Be) window of 1 mm, while the heel effect was not taken into consideration. The spectra utilized for the present study were derived for anode materials such as W, Rh and Mo, operating at tube voltages from 24 kVp to 34 kVp to cover the mammographic energy range, and filtered with Al and k-edges (Mo, Rh, Ru, Nb or Pd) filters. The 17 anode/filter material combinations used in this study were: Mo/0.030mmMo, Mo/0.029mmRh, Mo/0.030mmNb, Mo/0.51mmAl,
Rh/0.029mmRh, Rh/0.51mmAl, Rh/0.030mmNb, Rh/0.030mmRu, W/0.50mmAl, W/1.00mmAl, W/2.00mmAl, W/3.00mmAl, W/0.030mmMo, W/0.030mmRh, W/0.050mmNb, W/0.050mmPd+2.00mmAl and W/0.050mmZr.

B.4. MATHEMATICAL BREAST PHANTOM

In order to assure compatibility with similar published studies (Dance 1990, Dance et al 2000a, Zoetelief and Jansen 1995), the breast phantom was simulated by perspex of 4 cm thickness, divided into voxels of 1 mm$^3$. A 5 mm thick skin layer of adipose tissue was assumed at the upper and lower surface of the phantom, being equivalent to 3.94 mm of perspex, while a 3 mm thick perspex compression plate was also used. The Perspex-adipose tissue equivalence was calculated by irradiations of the phantom and the demand equivalent thicknesses to cause the same exposure at the detector plane. The energy absorbed in the skin layers was not included in the calculation of AGD, as the carcinogenic risk for the skin is considered minimal in mammography (Liu et al 1995).

The deposited energy in each irradiated voxel of the phantom was considered and the total dose was calculated by adding all the voxel energies and dividing the sum by the total irradiated mass, considering a homogeneous phantom. The equivalent breast tissue thickness of the utilized perspex phantom was calculated using the data provided by Dance (1990) about the equivalence of perspex and breast tissue thickness. Then, a correction factor for 50% glandularity was calculated for each spectrum used, in order to calculate the AGD from the mean breast dose, for this glandularity. For this calculation, the entrance spectrum and the interaction coefficients of the adipose and glandular breast tissue were utilized for each energy, in order to estimate the interaction probability of each kind of tissue.

B.5. VALIDATION OF THE AGD CALCULATION

In addition to the already published validation results of both the Monte Carlo simulation code and the mammographic spectra utilized in this study (Spyrou et al 1998, Delis et al 2005b, Delis et al 2004), the results concerning the AGD were compared with those of Boone (1999), as an additional validation. Boone presented AGD values in unit mrad per R, which were converted into unit mGy·mGy$^{-1}$ for the
needs of this comparison. A 100% glandular breast of 4 cm thickness was utilized, while the skin thickness was set to 4 mm of adipose tissue. For consistency of the validation results, the utilized spectra were the same as those utilized by Boone et al (1997), Boone and Seibert (1997) and Boone and Chavez (1996), for deriving the original results. The anode/filter material combinations used were the Mo/0.030mmMo, Mo/0.025mmRh, Rh/0.025mmRh and W/0.050mmRh for several values of tube voltage ranging from 20 kVp to 40 kVp.

The AGD results from the present study were found to differ from those of Boone (1999), on average, by 3.13%. This small variation can be attributed to differences in the Monte Carlo simulation code, the utilized interaction coefficients and the breast phantom composition.

B.6. AIR KERMA TO AGD CONVERSION FACTORS

Several authors have referred to conversion factors in order to calculate the AGD (D) in the breast by measuring the incident air KERMA (K) on the surface of the irradiated phantom (Dance 1990, Dance et al 2000a, Zoetelief and Jansen 1995). As g is defined the overall conversion factor, equal to the quotient of AGD to air KERMA:

\[ D = g \cdot K \]  

Published studies (Dance 1990, Dance et al 2000a) have proposed a combination of three independent conversion factors for contact mammography, as described in the equation:

\[ D = (g_{Mo/Mo} \cdot c \cdot s) \cdot K \]  

where \( g_{Mo/Mo} \) is the conversion factor of incident air KERMA to AGD for Mo/0.030mmMo at 28 kVp, \( c \) is a factor that corrects for glandularity different from 50% and \( s \) corrects for any anode/filter material combination, other than the Mo/Mo one.

For the needs of the present study, the calculation of correction factors has been adopted to account for various magnification factors, anode/filter material combinations and tube voltages. The AGD \( (D_m) \) for a magnification factor \( m \) is
associated with the air KERMA \((K_m)\) for the same magnification factor, with the following expression (Koutalonis et al 2006):

\[
D_m = (g_{Mo/Mo} \cdot m \cdot s \cdot v \cdot c) \cdot K_m
\]  

(7)

The \(g_{Mo/Mo}\) is the proposed conversion factor for Mo/0.030mmMo at 28 kVp (Dance 1990, Dance et al 2000a), and was utilized as a “golden standard normalization index” for the rest of this study.

The \(m\) factor corrects for magnification. Magnification factors from 1.0 to 2.1 where considered, with a step of 0.1. For each magnification factor, the \(g\) factor was normalized with respect to the \(g\) factor for magnification 1.0, in order to correct for the magnification considered each time. The calculations were made using the Mo/0.030mmMo combination at 28 kVp.

The \(s\) factor corrects for any anode/filter combination, other than Mo/0.030mmMo, and is independent of the magnification applied (Dance et al 2000a). The \(g\) factor was calculated again for each anode/filter combination and fixed tube voltage of 28 kVp, and divided by the \(g\) factor for Mo/0.030mmMo at 28 kVp. 17 anode/filter combinations were considered for the calculation of \(s\) factor, while the magnification factor was 1.6.

The \(v\) factor corrects for any different applied tube voltage between 24-34 kVp, with respect to 28 kVp. This is also independent of the magnification, and was calculated by dividing the \(g\) factor for any tube voltage of each different anode/filter combination used, by the \(g\) factor for 28 kVp. The magnification factor considered was 1.6.

Finally, \(c\) factor corrects for any glandularity other than 50\%, as proposed by Dance (1990). For the present study, this factor is considered unity, being equivalent to glandularity of 50\%. 


B.7. VALIDATION OF THE INDEPENDENCE OF THE FORMULA FACTORS

The validity of the independence of the factors proposed in formula (7) is performed by comparing the Monte Carlo calculated overall g factor, from the formula:

\[ D_m = g \cdot K_m \]  

with the product of the conversion factors in formula (7). This validation was carried out for a random sample of 100 anode/filter material combinations, considering various magnification factors and tube voltages, rather than the total of 2057 possible combinations, derived from the 11 tube voltages, 11 magnification factors and 17 anode/filter material combinations. The average variation between the two calculations was 0.14% and was attributed to statistical fluctuations.
CHAPTER C

RESULTS AND DISCUSSION

C.1. EFFECT OF MAGNIFICATION FACTOR ON ESD AND AGD………32
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C.4. EFFECT OF MAGNIFICATION FACTOR, ANODE/FILTER
COMBINATION AND TUBE VOLTAGE ON g FACTOR…………………38
In the following paragraphs, the effects of magnification factor, tube voltage and anode/filter material combination on the AGD, ESD and PDD are presented, as well as their contribution to the overall air KERMA to AGD conversion factor.

C.1. EFFECT OF MAGNIFICATION FACTOR ON ESD AND AGD

Low energy x-ray beams used in mammography do not present any skin sparing effect, since the dose maximum occurs on the surface of the skin and there is no buildup region. ESD is considered to be the “golden standard” in clinical mammographic dosimetry and for this reason it was included in this study.

The effect of magnification factor on the ESD is presented in figure 4 for the Mo/0.030mmMo combination at 28 kVp, showing an increase in ESD with magnification. As the irradiated phantom is placed closer to the focal spot, more photons scattered from the collimators reach the surface of the phantom, increasing the surface dose. However, this contribution is considered negligible. The most substantial reason for the increase in ESD is the inverse square law. The decrease of focus to phantom distance by increasing the magnification, results in higher ESD, as demonstrated in figure 4.

![Figure 4](image)

*Figure 4* The influence of magnification factor on ESD for the Mo/0.030mmMo combination at 28 kVp and for 4 cm breast phantom.
In figure 5, the influence of magnification on AGD is presented, for the Mo/0.030mmMo at 28 kVp. In the present study, AGD is defined as the dose delivered only to the irradiated portion of the phantom, rather than the whole, as previously proposed (Liu et al 1995). This was considered appropriate, since AGD is otherwise significantly underestimated, due to the large amount of non-irradiated phantom. According to this assumption, because of the smaller irradiated breast volume and the effect of the inverse square law, the AGD increases with increasing magnification.

**Figure 5** The influence of magnification factor on AGD for the Mo/0.030mmMo combination at 28 kVp and for 4 cm breast phantom.
C.2. EFFECT OF TUBE VOLTAGE ON ESD AND AGD

The effect of tube voltage on the ESD and AGD for the Mo/0.030mmMo combination and for magnification factor 1.6 is presented in figures 6 and 7, respectively. Increase in tube voltage results in “hardening” of the beam, which causes the increase of the mean free path length of the photons, leading to decrease of the ESD, since smaller amount of photons deposit their energy inside the upper layers of the phantom. For example, the increase of the tube voltage from 26 kVp to 28 kVp reduces the ESD by 1.82%, according to this study.

![Graph showing the relationship between tube voltage and ESD](image)

**Figure 6** The influence of tube voltage on ESD for the Mo/0.030mmMo combination at 28 kVp and for 4 cm breast phantom.

As far as the AGD is concerned, for the same number of photons used, it increases with tube voltage because more photons and with higher energies reach the phantom and deposit their energy in it. In figure 7, a comparison between magnification factors 1.0 and 1.6, as well as a comparison of the results obtained from the Monte Carlo simulation and the calculated conversion factors for magnification factor 1.0 is demonstrated. The average variation between results from the Monte Carlo simulation and those from the conversion factors is 0.57%.
Figure 7 The influence of tube voltage on AGD for the Mo/0.030mmMo combination and for magnification factors 1.6 and 1.0. Results from the use of conversion factors are also presented, for magnification factor 1.0.

C.3. EFFECT OF ANODE/FILTER MATERIAL COMBINATION ON ESD, AGD AND PDD

Various combinations of anode/filter materials were used to evaluate their dose characteristics. The results, concerning Mid-Plane Dose (MPD), Percentage Depth Dose (PDD), AGD and ESD are normalized with respect to Mo/Mo(0.030 mm) combination, as it is considered the optimum choice for mammography and are presented in figures 8 and 9. The magnification factor for these calculations was 1.6.

Generally, as the k-absorption edge of the filter material increases, the beam becomes harder. This is due to the fact that high energy photons from the beam are absorbed less. The increase in the Al filter thickness results in a more penetrating beam, since low energy photons are cut off, making the beam “harder”. Finally, anode materials with high k-emission edges contain larger percentage of photons with higher energies and so they produce harder beams. AGD and MPD increase with these three
RESULTS AND DISCUSSION

parameters (filter material k-absorption edge, filter thickness and anode material k-emission edge), as is also confirmed by our results.

Comparing combinations Mo/Mo, Mo/Rh and Mo/Nb, the effect of the filter material’s k-absorption edge on AGD and MPD can be noticed. The use of Rh as filter material results in higher AGD and MPD than using Mo or Nb because Rh has the larger k-absorption edge among these three materials. On the other hand, Nb gives a very “soft” beam, containing a lot of low energy photons. The same effect is obvious for combinations Rh/Rh, Rh/Nb and Rh/Ru. The impact of anode material’s k-emission edge on the same dose characteristics can be noticed by comparing combinations Mo/Rh, Rh/Rh and W/Rh. W has larger k-emission edges than Mo and Rh and so, when used as anode material results in higher AGD and MPD. On the other hand, Mo has the lower k-edges and gives a “soft” beam, compared to the other two materials.

Figure 8 The normalized MPD and AGD with respect to Mo/0.030mmMo for various anode/filter combinations.

The impact of anode material’s k-emission edge on the same dose characteristics can be noticed by comparing combinations Mo/Rh, Rh/Rh and W/Rh. W has larger k-emission edges (outside the mammographic energy range) than Mo and Rh and when used as anode material, results in higher AGD and lower ESD
(figure 9). On the other hand, Mo has the lower k-edges and gives a “soft” beam, compared to the other two anode materials.

Combinations Mo/Nb, Rh/Nb, W/Mo, W/Nb and W/Zr that use filter material with low k-absorption edge, result in lower AGD than Mo/Mo, while all the other combinations result in higher AGD. However, combinations causing lower AGD, simultaneously cause higher ESD than Mo/Mo as shown in figure 9 and so, there should be a compromise before choosing the optimum anode/filter material combination.

![Figure 9](image_url) The relative ESD with respect to the ESD of the Mo/0.030mmMo combination for various anode/filter combinations for 28 kVp and for magnification factor 1.6. The dotted horizontal line passes from the value corresponding to the Mo/0.030mmMo combination.

Finally, by noticing combinations W/0.5Al, W/1.0Al, W/2.0Al and W/3.0Al, the influence of the Al thickness on AGD and PDD is obtained. These four combinations present the increase in the dose characteristics considered, as the Al filter thickness increases. The effect on PDD is presented in figure 10.
C.4. EFFECT OF MAGNIFICATION FACTOR, ANODE/FILTER COMBINATION AND TUBE VOLTAGE ON g FACTOR

Monte Carlo calculations of air KERMA and AGD where utilized in order to derive the $g_{Mo/Mo}$ that is 0.168. This value is considered a constant and is used in subsequent calculations of AGD, based on formula (7).

Results concerning m factor are presented in table 2. Although, as presented in figure 5, AGD increases with the magnification factor, the m correction factor decreases with the magnification, as presented in table 2. The reason for this is that m factor is the quotient of the AGD and air KERMA measured on the surface of the phantom. However, air KERMA increases according to the inverse square law, while AGD increases less rapidly. So, the m correction factor decreases with magnification.
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<table>
<thead>
<tr>
<th>Magnification Factor</th>
<th>m Correction Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.0</td>
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</tr>
<tr>
<td>1.1</td>
<td>0.987</td>
</tr>
<tr>
<td>1.2</td>
<td>0.975</td>
</tr>
<tr>
<td>1.3</td>
<td>0.964</td>
</tr>
<tr>
<td>1.4</td>
<td>0.953</td>
</tr>
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<td>0.942</td>
</tr>
<tr>
<td>1.6</td>
<td>0.931</td>
</tr>
<tr>
<td>1.7</td>
<td>0.921</td>
</tr>
<tr>
<td>1.8</td>
<td>0.911</td>
</tr>
<tr>
<td>1.9</td>
<td>0.901</td>
</tr>
<tr>
<td>2.0</td>
<td>0.891</td>
</tr>
<tr>
<td>2.1</td>
<td>0.882</td>
</tr>
</tbody>
</table>

Table 2: The m correction factor for magnification from 1.0 to 2.1, corresponding to focus to skin distance from 56 cm to 24 cm.

Results concerning s and v factors, for the 17 anode/filter material combinations and the 11 tube voltages studied, are presented in tables 3 and 4. The combined effect of s and v factors is presented in table 5, offering a more practical alternative for direct correction, depending on the selected spectrum.

The effect of the filter material’s k-absorption edge can be noticed in table 3 where conversion factor s is presented. For example, considering Mo as anode material and for a fixed value of tube voltage, Rh as filter material has 25% and 16% larger conversion factors than Nb and Mo, correspondingly, resulting in this way in higher AGD. In the same table, the impact of anode material’s k-emission edge is presented, where W/Rh combination has larger s conversion factor than Mo/Rh and Rh/Rh.

The effect of tube voltage on the combined s and v factor is obvious in table 5. The increase of tube voltage from 24 kVp to 34 kVp leads to about 25% larger combined conversion factor for all combinations, which is translated into a corresponding increase in AGD, for a fixed value of air KERMA.
### RESULTS AND DISCUSSION

<table>
<thead>
<tr>
<th>Combination</th>
<th>s Correction Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo/0.030mmMo</td>
<td>1.000</td>
</tr>
<tr>
<td>Mo/0.029mmRh</td>
<td>1.194</td>
</tr>
<tr>
<td>Mo/0.030mmNb</td>
<td>0.899</td>
</tr>
<tr>
<td>Mo/0.51mmAl</td>
<td>1.224</td>
</tr>
<tr>
<td>Rh/0.029mmRh</td>
<td>1.235</td>
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<tr>
<td>Rh/0.51mmAl</td>
<td>1.268</td>
</tr>
<tr>
<td>Rh/0.030mmNb</td>
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</tr>
<tr>
<td>Rh/0.030mmRu</td>
<td>1.170</td>
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<tr>
<td>W/0.50mmAl</td>
<td>1.285</td>
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<tr>
<td>W/1.00mmAl</td>
<td>1.581</td>
</tr>
<tr>
<td>W/2.00mmAl</td>
<td>1.891</td>
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<tr>
<td>W/3.00mmAl</td>
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<td>W/0.030mmMo</td>
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<tr>
<td>W/0.030mmRh</td>
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<td>W/0.050mmPd+2.00mmAl</td>
<td>1.936</td>
</tr>
<tr>
<td>W/0.050mmZr</td>
<td>0.815</td>
</tr>
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</table>

**Table 3** The s correction factor for the various anode/filter combinations.
<table>
<thead>
<tr>
<th>Combination\kVp</th>
<th>24</th>
<th>25</th>
<th>26</th>
<th>27</th>
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<th>29</th>
<th>30</th>
<th>31</th>
<th>32</th>
<th>33</th>
<th>34</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo/0.51mmAl</td>
<td>0.860</td>
<td>0.902</td>
<td>0.938</td>
<td>0.970</td>
<td>1.000</td>
<td>1.025</td>
<td>1.049</td>
<td>1.071</td>
<td>1.092</td>
<td>1.111</td>
<td>1.129</td>
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<tr>
<td>Mo/0.030mmMo</td>
<td>0.891</td>
<td>0.922</td>
<td>0.952</td>
<td>0.979</td>
<td>1.000</td>
<td>1.028</td>
<td>1.048</td>
<td>1.068</td>
<td>1.088</td>
<td>1.107</td>
<td>1.123</td>
</tr>
<tr>
<td>Mo/0.030mmNb</td>
<td>0.865</td>
<td>0.904</td>
<td>0.940</td>
<td>0.973</td>
<td>1.000</td>
<td>1.034</td>
<td>1.062</td>
<td>1.088</td>
<td>1.114</td>
<td>1.139</td>
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<tr>
<td>Mo/0.029mmRh</td>
<td>0.893</td>
<td>0.929</td>
<td>0.957</td>
<td>0.980</td>
<td>1.000</td>
<td>1.016</td>
<td>1.031</td>
<td>1.045</td>
<td>1.059</td>
<td>1.071</td>
<td>1.083</td>
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<td>Rh/0.51mmAl</td>
<td>0.814</td>
<td>0.867</td>
<td>0.915</td>
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<td>1.000</td>
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<td>Rh/0.030mmNb</td>
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<td>0.897</td>
<td>0.932</td>
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<td>1.000</td>
<td>1.034</td>
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<tr>
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<td>0.935</td>
<td>0.970</td>
<td>1.000</td>
<td>1.028</td>
<td>1.052</td>
<td>1.076</td>
<td>1.098</td>
<td>1.120</td>
<td>1.140</td>
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<tr>
<td>Rh/0.030mmRu</td>
<td>0.862</td>
<td>0.903</td>
<td>0.939</td>
<td>0.971</td>
<td>1.000</td>
<td>1.027</td>
<td>1.051</td>
<td>1.076</td>
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<td>W/0.50mmAl</td>
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<td>1.000</td>
<td>1.041</td>
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<td>1.029</td>
<td>1.059</td>
<td>1.091</td>
<td>1.125</td>
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<tr>
<td>W/0.050mmNb</td>
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<td>0.947</td>
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<td>1.151</td>
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Table 4 The v correction factor for the various anode/filter combinations and for tube voltage between 24-34 kVp.
<table>
<thead>
<tr>
<th>Combination\kVp</th>
<th>24</th>
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<th>30</th>
<th>31</th>
<th>32</th>
<th>33</th>
<th>34</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo/0.51mmAl</td>
<td>1.053</td>
<td>1.104</td>
<td>1.148</td>
<td>1.187</td>
<td>1.224</td>
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<td>1.284</td>
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<td>0.957</td>
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<td>1.210</td>
<td>1.241</td>
<td>1.271</td>
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<td>1.328</td>
<td>1.356</td>
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<td>0.905</td>
<td>0.938</td>
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*Table 5* The combined s and v correction factor for the various anode/filter combinations and for tube voltage between 24-34 kVp.
CHAPTER D

CONCLUSIONS AND FUTURE WORK

D.1. CONCLUSIONS........................................................................................................44
D.2. FUTURE WORK......................................................................................................44
**D.1. CONCLUSIONS**

Conversion factors for estimating AGD from measurements of incident air KERMA on breast surface are essential, since they represent the only way to estimate the AGD in clinical practice. In the present study correction factors adapted to take into account magnification geometries and tube voltage are introduced. The effect of 11 magnification factors, 17 anode/filter material combinations and 11 tube voltages on absolute dose indices for certain magnification factor as well as on conversion factors was investigated.

AGD and PDD were found to increase mainly with the filter material’s k-absorption edge, the Al filter’s thickness and the anode material’s k-emission edge, while AGD increased with the tube voltage. Rh/Nb, W/Zr, W/Nb, W/Mo and Mo/Nb are anode/filter material combinations resulting in lower AGD than the Mo/Mo one, causing simultaneously higher ESD.

**D.2. FUTURE WORK**

Future work will focus on the simulation of breast phantoms with various thicknesses and glandularities. Furthermore, additional parameters affecting image quality, in magnification mammography, should be considered, aiming to optimize the figure of merit in magnification mammography.
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# APPENDIX I

## ABBREVIATIONS

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
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<tr>
<td>AGD</td>
<td>Average Glandular Dose</td>
</tr>
<tr>
<td>Al</td>
<td>Aluminum</td>
</tr>
<tr>
<td>Be</td>
<td>Beryllium</td>
</tr>
<tr>
<td>ESD</td>
<td>Entrance Surface Dose</td>
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<tr>
<td>FSD</td>
<td>Focus-Skin Distance</td>
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<tr>
<td>KERMA</td>
<td>Kinetic Energy Released per unit Mass</td>
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<td>Mo</td>
<td>Molybdenum</td>
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<td>MPD</td>
<td>Mid-Plane Dose</td>
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<td>Nb</td>
<td>Niobium</td>
</tr>
<tr>
<td>Pd</td>
<td>Palladium</td>
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<td>PDD</td>
<td>Percentage Depth Dose</td>
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<tr>
<td>PDF</td>
<td>Probability Density Function</td>
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<tr>
<td>Rh</td>
<td>Rhodium</td>
</tr>
<tr>
<td>Ru</td>
<td>Ruthenium</td>
</tr>
<tr>
<td>W</td>
<td>Tungsten</td>
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<tr>
<td>Zr</td>
<td>Zirconium</td>
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APPENDIX II

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Figure 1 Monte Carlo simulation of physical system.

Figure 2 Contact (a) and magnification (b) geometry in mammography. In magnification geometry the antiscatter grid is removed and the breast is placed closer to the focal spot.

Figure 3 The flow chart of the proposed simulation model.

Figure 4 The influence of magnification factor on ESD for the Mo/0.030mmMo combination at 28 kVp and for 4 cm breast phantom.

Figure 5 The influence of magnification factor on AGD for the Mo/0.030mmMo combination at 28 kVp and for 4 cm breast phantom.

Figure 6 The influence of tube voltage on ESD for the Mo/0.030mmMo combination at 28 kVp and for 4 cm breast phantom.

Figure 7 The influence of tube voltage on AGD for the Mo/0.030mmMo combination and for magnification factors 1.6 and 1.0. Results from the use of conversion factors are also presented, for magnification factor 1.0.

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Table 3 The s correction factor for the various anode/filter combinations.

Table 4 The v correction factor for the various anode/filter combinations and for tube voltage between 24-34 kVp.

Table 5 The combined s and v correction factor for the various anode/filter combinations and for tube voltage between 24-34 kVp.
ABSTRACT

Magnification mammography is a special technique used in cases where breast complaints are noticed by a woman or when an abnormality is found in a screening mammogram. The carcinogenic risk in mammography is related to the dose deposited in the glandular tissue of the breast rather than the adipose, and Average Glandular Dose (AGD) is the quantity taken into consideration during a mammographic exam. Direct measurement of the AGD is not feasible during clinical practice and thus, the incident air Kinetic Energy Released per unit of MAss (KERMA) on the breast surface is used to estimate the glandular dose, with the help of proper conversion factors.

Additional conversion factors adapted for magnification and tube voltage are calculated, using Monte Carlo simulation. The effect of magnification factor, tube voltage and various anode/filter material combinations on AGD, ESD and PDD is also studied.

Results demonstrate that, for fixed glandularity, the estimation of AGD utilizing conversion factors depends on magnification factor, anode/filter combination and tube voltage applied. AGD was found to increase mainly with filter material’s k-absorption edge, filter’s Al thickness, anode material’s k-emission edge and tube voltage. Rh/Nb, W/Zr, W/Nb, W/Mo and Mo/Nb are combinations resulting in lower AGD and higher ESD, compared to the Mo/Mo one.
ΠΕΡΙΛΗΨΗ

Οι μεγεθυντικές λήψεις στη μαστογραφία είναι μια ειδική τεχνική που χρησιμοποιείται σε περιπτώσεις όπου παρατηρούνται ενοχλήσεις στο μαστό μιας γυναίκας ή κάποια πιθανή ανομαλία εμφανίζεται σε μία μαστογραφία. Η πιθανότητα καρκινογένεσης στη μαστογραφία σχετίζεται δίκαια με τη δόση που εναποτίθεται στο μαζικό αδένα του μαστού, σε αντίθεση με τον λιπώδη υπόπυρο, και η μέση δόση στο μαζικό αδένα είναι το μέγεθος το οποίο λαμβάνεται σοβαρά υπόψη κατά τη διάρκεια μιας μαστογραφίας. Η μέση μέτρηση του μεγέθους αυτού δεν είναι δυνατή στην κλινική πράξη και γι’αυτό, η προσπίπτουσα κινητική ενέργεια που εκλέπτεται ανά μονάδα μάζας στην επιφάνεια του μαστού χρησιμοποιείται για τον υπολογισμό της δόσης στο μαζικό υπόστημα, με τη βοήθεια καταλλήλων παραγόντων μετατροπής.

Επίπλεον παράγοντες μετατροπής προσαρμοσμένοι ώστε να περιλαμβάνουν τη μεγέθυνση καθώς και την εξάρτηση από την εφαρμοξόμενη τάση υπολογίστηκαν στην συγκεκριμένη εργασία. Επιπλέον, μελετήθηκε η επίδραση του βαθμού μεγέθυνσης, της εφαρμοξόμενης τάσης και του συνδυασμού υλικών ανόδου/φίλτρου στη μέση δόση στο μαζικό υπόστημα, στη δόση εισόδου και στην επί τοις εκατό δόση βάθους.

Τα αποτελέσματα δείχνουν ότι για συγκεκριμένη περιεκτικότητα σε μαζικό αδένα, η υπολογιζόμενη δόση κάνοντας χρήση παραγόντων μετατροπής εξαρτάται από το βαθμό μεγέθυνσης, την εφαρμοξόμενη τάση και το συνδυασμό υλικών ανόδου/φίλτρου. Η μέση δόση βρέθηκε να αυξάνεται με την K-αιχμή απορρόφησης του υλικού της ανόδου, με το πάχος του φίλτρου Αλουμινίου, με την K-αιχμή εκπομπής του υλικού της ανόδου και την τάση. Τέλος, συνδυασμοί όπως οι Rh/Nb, W/Zr, W/Nb, W/Mo και Mo/Nb προκαλούν μικρότερη δόση στο μαζικό υπόστημα από τον Mo/Mo και υψηλότερη επιφανειακή δόση.