Optimization of Automatic Exposure Control of a Photon Counting Mammography System

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Cancer is today one of the most common causes of death in Sweden and the developed world. The number of incidences of breast cancer, which is the most common cancer form among women, is increasing. Early detection is a key to increasing the survival rate and therefore mammography plays a crucial part in fight against breast cancer. The demands placed on the mammography systems are high; a certain image quality has to be achieved while exposing the breast to a radiation dose as low as reasonably possible. Other features, such as the image acquisition time, are also of interest. Choosing the optimal exposure settings for an examination, so that all objectives are considered, is a demanding task handled by the Automatic Exposure Control, AEC. In this thesis, the AEC of the multi-slit, photon counting Philips MicroDose mammography system has been optimized. A new cost function, which the AEC uses to determine which X-ray tube voltage is appropriate to use at a certain breast thickness, has been introduced. This cost function also considers, among other parameters, the effect on the X-ray tube lifetime which should be minimized. Once the new cost function was introduced, it was investigated which set of tube voltages that is the optimal one to use for patient examinations. It was investigated whether any improvements can be done in terms of decrease in population mean dose, image acquisition time, tube load or increase in tube lifetime. Finally, the effects of weighing counted photons of different energies in an optimal manner to increase the dose efficiency was investigated. It was found that small improvements can be done to the MicroDose system by introducing a new cost function and by changing the set of voltages used to operate the X-ray tube. However, considerable improvements can be done by applying energy weighting.

Key words: Mammography, Automatic Exposure Control, Optimization, Filament lifetime, Photon counting, Energy Weighting.
Sammanfattning


Key words: Mammografi, Automatisk Exponeringskontroll, Optimering, Filamentlivstid, Fotonräkning, Energiviktning.
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# Abbreviations

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<th>Abbreviation</th>
<th>Description</th>
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<tr>
<td>AEC</td>
<td>Automatic Exposure Control</td>
</tr>
<tr>
<td>AGD</td>
<td>Average Glandular Dose</td>
</tr>
<tr>
<td>ESAK</td>
<td>Entrance Surface Air Kerma</td>
</tr>
<tr>
<td>EW</td>
<td>Energy Weighting</td>
</tr>
<tr>
<td>FFDM</td>
<td>Full Field Digital Mammography</td>
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<tr>
<td>FOV</td>
<td>Field Of View</td>
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<tr>
<td>MTF</td>
<td>Modulation Transfer Function</td>
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<tr>
<td>OTF</td>
<td>Optical Transfer Function</td>
</tr>
<tr>
<td>PSF</td>
<td>Point Spread Function</td>
</tr>
<tr>
<td>SNR</td>
<td>Signal to Noise Ratio</td>
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<tr>
<td>SDNR</td>
<td>Signal Difference to Noise Ratio</td>
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Chapter 1

Introduction

1.1 Background

Cancer is today one of the most common causes of death in Sweden and the western world. 30% of all cancer incidences among women are breast cancer, making it the most common form. Since 1958 all incidences of cancer in Sweden have been registered and the number of cases of breast cancer per 100,000 inhabitants has, like for most cancer forms, been increasing since. The mortality of breast cancer, however, has been decreasing, and in 2011 the 10-year survival rate was just above 80% [1]. In 2011, 8382 women and 45 men were diagnosed with breast cancer and for 1405 women and 15 men the outcome was fatal [1].

Early detection is a key to decreasing the mortality rate of breast cancer. The diagnosis can be established through three steps; a clinical examination where a doctor examines the breast using palpation, mammography and/or ultrasound examination and biopsy. Women in Sweden at an age between 40 and 74 years are recommended and offered mammography screenings every 18th to 24th month. The screening programme is estimated to decrease the mortality rate of breast cancer by 16 – 25% and mammography is thus an important key to further increase the survival rate of breast cancer [2].

However, exposing tissue to ionizing radiation is associated with a risk of inducing cancer which places high demands on the mammography systems; a certain image quality has to be achieved by exposing the patient to a radiation dose as low as reasonably possible. In order to achieve this it is important to strictly control the exposure settings for a
mammography examination. Different energy spectra of the X-ray photons give different dose efficiencies. Energies that give dose efficiencies close to the maximal value should be used, but dose efficiency can be traded for improvements in other features (e.g. a shorter image acquisition time). The exposure settings are controlled by an *Automatic Exposure Control*, AEC, who’s task is to choose the best breast specific exposure settings so that all objectives are taken into consideration.

The Philips MicroDose mammography system is a scanning multi-slit full field digital mammography system that uses a photon counting detector. The use of collimators, a scanning geometry and a photon counting detector makes it possible to offer a very high image quality obtained at a patient dose lower than any other full field digital mammography system on the market today [3]. The MicroDose mammography system uses an AEC, that for each specific breast chooses between five X-ray tube voltages so that patient dose, image acquisition time and X-ray tube load are minimized while the required image quality is obtained.

### 1.2 Purpose

The purpose of this thesis is to optimize the automatic exposure control for the Philips MicroDose mammography system. By making the AEC also take the effect on the X-ray tube lifetime into account when choosing the exposure settings, it will be investigated whether any improvements can be done in terms of decreased patient dose, decreased image acquisition time, decreased tube load or increased lifetime of the X-ray tube. It will also be investigated which is the optimal set of voltages to operate the X-ray tube, and if improvements can be done if the set of used voltages was changed. The MicroDose system uses five different voltages, but it will be examined what can be gained by using more, and what the loss would be of using fewer voltages.

Since the MicroDose system is photon counting with more than one energy bin, it presents the opportunity of weighing the counted photons in an optimal manner depending on their energies, so that the dose efficiency can be increased. Finally, the effect of energy weighting will be studied.
1.3 Scope

In this thesis, an AEC taking imaging time, patient dose, tube load, tube lifetime and image quality into account will be studied. Based on this, two optimizations will be performed. One to find the optimal set of voltages to use for patient examinations, and one to choose the best voltage in the set to use at each breast thickness. However, a more extensive optimization could be done by taking more parameters into account. The slit width and the number of detector lines could also be incorporated to the model and be chosen optimally.
Chapter 2

Conceptual Background

2.1 Basic X-ray physics

In order to understand the physical mechanism behind mammography it is first important to acquire some basic knowledge about X-rays physics. X-ray photons are electromagnetic quanta having energies between 100 eV and 100 keV. The wavelengths are thus shorter than for visible light and ultraviolet rays but longer than for gamma rays.

2.1.1 Generation of X-rays

X-rays can be generated in an X-ray tube or in a particle accelerator. In an X-ray tube electrons are released from a heated cathode and are then accelerated by a voltage to collide with an anode. Around 99 % of the energy of the colliding electrons will be released as heat but the rest will give rise to X-rays as the electrons are completely or partly slowed down when interacting with the nuclei of the anode. This radiation is referred to as Bremsstrahlung. The maximum energy of the released photons is equal to the energy of the incoming electrons, corresponding to the situation where the electrons are completely slowed down in an electrostatic interaction with an atom of the anode. Lower energy photons are produced when the electrons are only partly slowed down in the electrostatic interactions. Some of the incoming electrons will also generate X-rays in a different manner when they knock out an inner shell electron of an atom of the anode. The vacancy created can then be filled by an outer shell electron that then releases the energy difference through generation of a high energy photon. The X-ray
spectra generated by an X-ray tube will have the appearance showed in Figure 2.1, the continuous spectrum corresponding to Bremsstrahlung and the peaks corresponding to the characteristic emission lines of the anode material.

![Figure 2.1: X-ray spectrum generated by an X-ray tube operated at 29 kV, with a tungsten target and 0.4 mm aluminium filter.](image)

### 2.1.2 X-ray interaction with matter

Due to the high energy of the X-ray photons they can penetrate through many materials, but they will interact with the matter in three main ways described below.

- **X-ray photons entering a material can interact through Photoelectric absorption.** In this type of interaction the photon releases all of its energy to an inner shell electron of the material whereupon the electron is knocked out from its orbit. The electron will have energy corresponding that of the absorbed photon minus the binding energy at its original position.

- **Through Compton scattering** the incoming photon releases some of its energy to an electron which is knocked out from its orbit, and the photon will be left with lower energy moving at a different angle. Compton scattering is an incoherent type of scattering and the relative probability of its occurrence increases with the energy of the X-rays [4].

- **X-ray photons can interact with matter coherently and not release any of its energy,** a process referred to as **Rayleigh scattering.** The oscillating electromagnetic field of an incoming X-ray photon will affect electrons in the material. This will
cause the electrons to oscillate at the same frequency and thereby emit X-ray photons of the same energy as the incoming ones and released in nearly the same direction.

[5]. When a beam of X-rays pass through a material some of the photons will be lost due to any of the interactions described. The loss of photons from the beam is called attenuation and the higher the density of the matter the X-rays are traversing, the higher is the attenuation. This is used for acquiring an image of a part of the body, a breast in the case of mammography. Different structures in the body will give different attenuations which will appear as different contrasts in the image. The attenuation for a monochromatic beam is described by

$$N(x) = N_0 e^{-\mu x},$$  \hspace{1cm} (2.1)

where $N(x)$ is the number of photons in the beam after traversing a material of thickness $x$. $N_0$ is the number of photons in the beam initially and $\mu$ is the energy dependent linear attenuation coefficient, usually given in units of cm$^{-1}$. A higher attenuation coefficient results in less photons passing the material.

### 2.2 Mammography

Mammography is an examination where X-rays are used to image the breast and potentially detect cancers. However, exposing tissue to ionizing radiation is associated with a risk of inducing cancer which places high demands on the mammography systems; a certain image quality has to be achieved by exposing the patient to a radiation dose as low as reasonably possible. The mammography techniques available today use either screen-film, computed radiography plates or digital image receptors, each category covering several different technical solutions.

#### 2.2.1 Screen Film Mammography

At the beginning of the mammography era, film was used to detect the X-ray photons passing through the breast. Film is still used, even tough it is very rare in developed
countries. Today screen film is used together with an intensifying phosphor screen to increase the quantum efficiency. Although screen-film mammography is an inexpensive technique that yields high resolution images it has drawbacks such as low contrast and risk of under- or overexposure if the exposure time is not strictly controlled. This in addition to the practical consideration of handling the film, e.g. storage, and the lack of possibilities for image post-processing.

2.2.2 Computed Radiography

In computed radiography, CR, imaging plates made of photostimulable phosphor, are used to capture an image of the breast. X-ray photons exiting from the breast will excite electrons in the imaging plate, and the excited charge will be captured in the plate. The plate is then scanned with a laser, causing the excited electrons to return to their original state while emitting blue light in the process. The amount of blue light is proportional to the energy of the incoming X-ray photons. The emitted light is recorded by a detector and is stored digitally.

2.2.3 Digital Mammography

In digital mammography, X-ray radiation that exit the breast is detected by a digital image receptor. In general, radiation is converted into to charge in the detector and the electrical signal is then measured and the digitized information is stored on a computer. This allows for image-processing before the image is displayed. The way of actually detecting photons in digital mammography might vary however.

Direct and indirect detection of photons

When the photons enter the detector they have to be converted into charge in order to be measured. This can be done in an indirect manner, where the incoming X-ray photons are first converted into visible light photons by a scintillator. The emitted visible light can then be detected by photodiodes that convert it into charge. The scintillator might also be connected to a charge couple device, CCD, where the visible light photons produce several electron-hole pairs. The electron-hole pairs are then separated by an
applied voltage and the flow of charges gives an electrical signal that can be measured by readout electronics connected to the detector.

With direct detection X-ray photons are converted directly into charge without the intermediate step of conversion into visible light. By removing a conversion step the noise degrading the image can be decreased. Direct detection of X-ray photons requires high absorption in the detector material and it is obtained by using a material of high atomic number or making the detecting element very thick [4]. The incoming X-ray photons interact with the detecting material and multiple electron-hole pairs are created. The amount of charge created is proportional to the energy of the photon. The electron-hole pairs are, as described previously, separated by an applied voltage and the electrical signal is measured.

**Interpreting the detected signal**

There are two ways of measuring the signal in a digital mammography system. Either, all the charge created by incoming photons can be summed in each pixel, and a total incoming energy value for that pixel is then recorded. This kind of detection devices are called energy integrating. Screen film mammography systems are also energy integrating, since each pixel value is given by the integrated energy incident on that pixel. Another method for measuring the signal is through photon counting. This is done by counting the number of electrical pulses that are above a certain threshold. The main advantage of photon counting techniques is the noise reduction. Noise below the threshold level will not be counted leading to an increased $SNR$. This means that the photon counting detectors can operate very close to the quantum limit [4]. Another advantage is the fact that all counted photons are equally weighted. For an energy integrating system the lower energy photons are weighted less even though they contain more contrast information. As the photon pulses are processed individually with the photon counting technique, it is also possible to add more electronic thresholds to each detector channel and thereby measure the photon energies. This spectral information can be used to enhance the image in different ways explained later.
2.3 Imaging Physics

When acquiring an image of an object its quality has to be evaluated to see whether the information provided by the system is adequate. There are several factors that affect the overall image quality, such as contrast, signal-to-noise ratio and spatial resolution.

2.3.1 Photon Quantum Noise

When measuring the number of incoming photons during a specified time interval for a given X-ray intensity, separate measurements will yield varying results. The reason is that the emission of photons from the tube anode is a Poisson distributed statistical process which gives random fluctuations in the produced beam. Combined with the Binomial distributed process for transmission, the result, which is the number of photons absorbed in the detector, is also Poisson distributed. The mean value of the signal, $\bar{N}$, and the standard deviation, $\sqrt{\bar{N}}$, of the detected signal are obtained if the measurement is made many times. $\sqrt{\bar{N}}$ represents the noise if photon quantum noise is the only noise present in the system. Such a system is called a quantum limited, but in general many other noise sources can affect the image such as thermal noise and dark signal noise [6].

2.3.2 SNR and SDNR

The signal-to-noise ratio, $SNR$, is defined as the mean signal value divided by the root-mean-square of the noise affecting the image. For an ideal imaging system, where the only noise present is the photon quantum noise, this would yield

$$SNR = \frac{\bar{N}}{\sqrt{\bar{N}}} = \sqrt{\bar{N}};$$

(2.2)

but in any imaging system affected by other noise sources the $SNR$ is always less than $\sqrt{\bar{N}}$. The $SNR$ is of great importance to an imaging system but even more useful is the signal-difference-to-noise ratio, $SDNR$. The $SDNR$ is the difference in signal between an object and the background divided by the root-mean-square of the noise, as illustrated in Figure 2.2. The contrast, $C$, is given by the difference in intensity between
the object and the background, divided by the intensity of the background. Thus, the \( SDNR \) can be calculated as \( SNR \cdot C \).

![Diagram of SDNR](image)

**Figure 2.2:** Figure illustrating the concept of SDNR.

The \( SDNR \) depends on the object thickness and on the energy of the X-ray photons. For different thicknesses, maximal \( SDNR \) is obtained at different photon energies, which is shown by a simple model and illustrated in Figure 2.3 [7].

![Graph showing SDNR variation](image)

**Figure 2.3:** Figure showing how SDNR vary with photon energy at different object thicknesses. Reprinted from [7].

### 2.3.3 Resolution

The quality of an image is often measured as the spatial resolution of the image. With resolution it is meant how closely two equally bright point objects can be placed and
still be resolved as two separate objects in the image. To begin with we will consider how objects are imaged in an imaging system.

An infinitely small point object will be smeared out when imaged and therefore the image will be an intensity distribution described by the first order Bessel function. The reason why the point object is imaged as a smeared out point is aberrations of the optics but also diffraction of the light by the optics. An ideal imaging system with no aberrations is called diffraction-limited and even this kind of system will image an infinitely small point object as an intensity distribution, due to diffraction of the light. Diffraction is proportional to the wavelength, so for an X-ray imaging system the effects of diffraction are negligible. Instead, there are other factors affecting the resolution, such as the fact that the focal point on the anode, where the X-rays are produced, is a spread out point rather than a point source. When imaging a point object the intensity distribution in the image plane is referred to as the point spread function, or the \( \text{psf} \).

Two equally bright point objects that are closely spaced might get overlapping point spread functions in the image plane and if the overlap is severe they will not be resolved as separate objects. According to the Rayleigh criterion there has to be a 26% intensity dip between the peaks of the point spread functions for the objects to be classified as resolved. This limit is chosen from the approximate minimum intensity decrease that the human eye can detect [6]. The resolution is a good measure of the image quality but the information provided is not enough. Contrast and noise are also of great importance for the overall image quality.

### 2.3.4 OTF and MTF

When considering an imaging system, let \( I_O(x) \) denote the intensity of the object and \( I_B(x) \) denote the image intensity. In order to fully evaluate the image quality it is important to know how an object is imaged. In the previous section we saw that a point object of intensity \( I \) is imaged as the point spread function, thus we have \( I_O(x) = I \cdot \delta(x) \) and \( I_B(x) = I \cdot \text{psf}(x) \). An object with continuous intensity distribution can be described as a sum of infinitely many closely spaced point objects positioned at \( x' \) and with varying intensities. The image intensity at a point \( x \) will then be a sum of intensity contributions from the different point spread functions,
\[ I_B(x) = \int_{-\infty}^{+\infty} I_O(x') \cdot psf(x - x')dx', \quad (2.3) \]

which is the convolution of the object intensity distribution and the point spread function. In order to sum intensities in this manner it has to be assumed that the illumination is incoherent. The image intensity function can thus be determined for an object if the \( psf \) is known. If a Fourier transformation is performed on the convolution expression it simplifies to a multiplication, giving \( \hat{I}_B(\omega) = \hat{I}_O(\omega) \cdot \hat{psf} \) as a simple relation between the object and image function. The Fourier transformed functions can be expressed as \( \hat{I}_O(\omega) = |\hat{I}_O(\omega)| \cdot e^{i\text{arg}[\hat{I}_O(\omega)]} \) and respectively for \( \hat{I}_B(\omega) \) and \( \hat{psf} \) which gives the following relation [6]

\[ |\hat{I}_B(\omega)| = |\hat{I}_O(\omega)| \cdot |\hat{psf}(\omega)|, \quad (2.4) \]

where \( \omega \) is the spatial frequency with unit \( m^{-1} \). \( \hat{psf} \) is called the Optical Transfer Function, \( OTF \), and it describes the relation between the the Fourier transforms of the object intensity function and the image intensity function. \( |\hat{psf}(\omega)| \) is referred to as the Modulation Transfer Function, \( MTF \), and describes the relation between the modulations of each spatial frequency in the object and the image. For an ideal imaging system the \( MTF \) would be unity, meaning that all spatial frequencies are imaged without any reduction of modulation or amplitude.

Usually, the \( MTF \) decreases from one at a spatial frequency of zero, to zero at a limiting spatial frequency. Frequencies above the limit cannot be imaged and this gives us information similar to the Rayleigh resolution criterion which specifies how closely two objects can be placed and still be resolved as separate objects in the image. But the \( MTF \) contains even more useful information regarding the image quality, namely how well each spatial frequency is imaged. This means that if the \( psf \) of an imaging system is known, the quality of the image obtained can be measured effectively.
2.3.5 Other Metrics

Noise Power Spectrum

Similar to the way that the MTF describes the degree of modulation for each spatial frequency, the Noise Power Spectrum, $NPS$, describes the noise at each spatial frequency, i.e. the variations in how a frequency is imaged over a number of measurements.

Noise Equivalent Quanta

The Noise Equivalent Quanta, $NEQ$, is the minimum number of Poisson distributed X-ray photons that would produce a certain $SNR$. A minimum number of photons would be required if the detector was ideal [8]. The $NEQ$ is thus given by

$$NEQ = SNR_{non-ideal}^2.$$  \hspace{1cm} (2.5)

Detective Quantum Efficiency

The Detective Quantum Efficiency, $DQE$, is defined as the $NEQ$ divided by the actual number of incoming photons that created the image [8]. In a sense it compares the imaging system to an ideal one to determine how efficiently the system detects incoming photons. The $DQE$ is given by

$$DQE = \frac{NEQ}{N_{inc}} = \frac{SNR_{non-ideal}^2}{SNR_{inc}^2}.$$  \hspace{1cm} (2.6)

Dose Efficiency

The Dose Efficiency, $DE$, is defined as the square of the $SDNR$ divided by the average glandular dose, $AGD$, i.e. the imaged quality divided by the dose used. The $DE$ should be close to maximized but can however be traded for other improved features in the system.
Chapter 3

The Philips MicroDose Mammography System

The Philips MicroDose mammography system is a scanning multi-slit photon counting full field digital mammography (FFDM) system. The use of collimators, a scanning technology and a photon counting detector makes it possible to considerably improve the dose efficiency, acquiring high quality images using a much lower dose than other FFDM systems.

3.1 General Description

Figure 3.1 shows the general set up of the MicroDose system. X-rays are produced in an X-ray tube with a tungsten anode. After the X-rays exit the tube the beam is hardened by a 0.4 mm Al filter. This narrows the energy span of the beam and low energy photons, that would not be useful for the image production and that would only increase the patient dose, are removed. The hardened X-ray beam continues on a foreclosed path to the pre-collimator. The purpose of the pre-collimator is to transform the X-ray beam into several smaller beams that matches the post-collimator and the detector. Then the beam passes through the compression plate, whose purpose is to fixate the patient in order to improve image quality, and then enters the patient. Between the patient and the detector is the post-collimator that efficiently removes radiation scattered by
the tissue towards the detector. The scatter rejection of the post-collimator greatly improves the image quality but a drawback of the multi-slit geometry is the fact that only a very small percentage of the produced X-rays pass through the collimators and are used for image acquisition, and this requires a very high tube load. The detector is placed perpendicular to the chest wall, defined as the y-direction. Both collimators and the detector are mounted rigidly together and perform a scanning motion parallel to the chest wall, defined as the x-direction, to obtain a 2D image [4].

![Diagram of Philips MicroDose mammography system](image)

**Figure 3.1:** Illustration of the general set up of the Philips MicroDose mammography system. Reprinted from [4].

### 3.2 The X-ray Tube

The X-ray tube has a tungsten anode. For this kind of system a tungsten target is favourable since tungsten has no characteristic lines in the energy span typically used in mammography, which makes is easy to alter the mean energy of the produced X-ray spectrum. Tungsten is also advantageous due to its high efficiency for photon generation, the efficiency is proportional to the atomic number. Furthermore, it has a high tolerance to heat, which is of great importance in a scanning multi-slit system where the tube load is high [4]. In order to avoid melting of the tungsten anode, the focal point where the electrons hit the target, cannot be too small. For the tube used in the MicroDose system the focal point is 0.3 mm and this will affect the resolution. The voltages on which the
tube is operated are 26, 29, 32, 35 and 38 kV, which gives the same corresponding maximum photon energies in keV.

### 3.3 The Detector

The detector consists of multiple silicon detector lines aligned in the y-direction and to the incoming X-ray beams from the post-collimator. In practise it is not possible to produce a single silicon detector as long as the field of view (FOV) in the y-direction and therefore some shorter silicon detectors are combined to form a complete detector line as long as the FOV. Since silicon has a low atomic number of 14 and thus a small linear attenuation coefficient, the detectors have to be quite deep in the direction of the incoming X-rays. In the y-direction each detector line is segmented into several photon counting channels with readout electronics. During the scan all detector lines sweep the whole FOV and a readout, that registers the number of detected photons in each channel, is made every 50 μm in the scan direction. In the image reconstruction all the images acquired from each line are merged into a single image. This results in a warranty of no dead pixels, i.e. pixels in the image where no information exists. Even though one read out channel is not working, information regarding this pixel will be obtained by channels in the other detector lines [9].

![Figure 3.2: Illustration of the silicon strip-detector lines used in the MicroDose. Reprinted from [9].](image)

When the X-ray photons enter the silicon detectors they produce electrical signals that are proportional to their energy. Only signals above a certain threshold are registered in order to remove noise. Then, information regarding the number of photons detected
in each channel is stored for image reconstruction. When the photons interact through Compton-scattering in the detector, the produced signal might be lower than the threshold and will then not be counted. The photon from the Compton-scattering process will continue its journey and it might in worst case be photo absorbed and give rise to a signal above the threshold. This pulse will contain "false" information both regarding the photon energy and the object attenuation in the pixel where the photon was detected. Compton scattering is therefore something that should be avoided, and since the probability increases with higher photon energies, a too high tube voltage cannot be used [4]. Further, the efficiency of the detector decreases with increasing photon energies. The photon counting technique effectively removes noise and therefore photon quantum noise is the only noise affecting the physical image quality.

![Figure 3.3: Illustration of the photon counting technique. Only photons that give rise to an electrical signal above the threshold are counted, which effectively removes noise. It is possible to add more thresholds and thereby measure the photon energies. Reprinted from [9].](image)

### 3.3.1 Energy weighting

The photon counting technique, unique for the MicroDose system, presents the possibility of weighting the photons in an optimal manner to achieve the best possible dose efficiency and image quality. It would be possible to apply increasing weights to photons of decreasing energies, since those carry more information. This improves the achieved $SDNR$ compared to energy integrating systems where the signal is proportional to the photon energy and a higher weight therefore is applied to higher energy photons.
In the more recent MicroDose systems two thresholds are used. One for separating photon counts from noise and another one dividing the photon counts into a high energy and a low energy group. The images from the low energy and the high energy photons are added with equal weight to provide the image. However, studies have been made to examine how much the image quality could be improved by applying a higher weight to the low energy photon image. The energy weighted pixel values are

\[ p_{ew} = w \cdot p_{low} + p_{high} \]

or

\[ p_{ew} = w \cdot p_{sum} - (w - 1) p_{high}, \]  

(3.1)

where \( p_{low} \) is the pixel value in the low energy image, \( p_{high} \) is the pixel value in the high energy image and \( p_{sum} \) is the summed pixel value. The value of the weight, \( w \), for maximum improvement of image quality will depend on the object material composition and thickness, the energies of the X-ray spectrum used, the energy threshold level of the detector and non-idealities of the system. Through measurements and simulations, optimal weight factors for different object thicknesses and X-ray tube voltages, have been extracted. For each case, the percentual improvement of the \( SDNR \) have been determined as compared to case of equal weighting of low and high energy photons, i.e. \( w = 1 \). Optimal weight factors usually have values between 1.5 and 2 and give \( SDNR \) improvements of up to 6 %. Greater improvement can be achieved for higher voltages. The improvement in dose efficiency can be used to increase the image quality or alternatively to decrease the dose \([10]\).

### 3.4 Scan time and mAs

When the detector is scanned over the field of view there are several parameters referred to time that are of interest for the performance. During the scan the detector counts photons, but every 50 \( \mu \)m a readout is made to register the number of photons counted. During the readout the detector is inactive and do not count any photons. Each such readout takes 28 \( \mu \)s, and this is referred to as the blind time. During the scan a total of
4965 readouts are made which means that the detector is not active for photon counting and has a total blind time of 0.1986 s.

The time between the readouts is determined from the target $SDNR$ that is set for each breast thickness. It can be calculated what number of photons is needed in order to reach the target $SDNR$ at a certain tube voltage. The photon rate, the number of photons exiting the object depending on the tube voltage and the object thickness, has been measured and by using this the time needed in order to register the correct number of photons can be determined. The photon rate is proportional the tube output and to the tube current.

The total time of the scan is then the time needed between readouts multiplied with the number of readouts plus the total blind time. In addition, there is an acceleration and retardation time at the beginning and end of the image acquisition. The time referred to as the exposure time, on the other hand, is the time it takes for the whole detector to pass a certain point. Since the detector width is approximately 50 mm and the scan length is approximately 248 mm, the exposure time is close to one fifth of the total scan time. Even though the total scan time is relatively long, it is the exposure time and not the total scan time that determines the risk of image blurring due to patient movement. The time with risk of image blurring for this type of scanning system is somewhat larger than that of a full-field detector system.

Another commonly used metric is the current time product, measured in units of mAs. The current time product is the product between tube current, which is fixed for each tube voltage, the time between readouts and the factor

$$K = \frac{1}{0.05} \cdot 18 \cdot 0.1 \cdot \frac{660}{532}.$$  \hspace{1cm} (3.2)

This factor transforms the time between readouts to the effective time between readouts. This is done to account for that the radiation passes through multiple slits in the pre-collimator and the effective time is therefore higher. For the factor $K$, 18 is the number of slits, 0.1 is the slit width in mm, 0.05 is the distance between readouts in mm and 660/532 scales the slit width to the image plane.
Chapter 3. The Philips MicroDose Mammography System

3.5 Automatic Exposure Control

When making a mammogram, parameters like the average energy of the X-rays, determined by tube voltage, and the time of the scan need to be determined so that the patient dose is minimized while a certain image quality is still obtained. These parameters depend on the breast thickness which is measured mechanically when the breast is compressed, but also on the density, measured as the glandular fraction in percent. The five possible energies used by the MicroDose system give different scan times and doses when used for acquiring an image of the required image quality or SDNR. It is then a balance to choose the most suitable tube voltage so that the dose is minimized and the scan time is not too long. A long scan time is associated with image blurring due to movement of the patient and also discomfort of the patient. A maximum scan time of 15 seconds is therefore required and also a minimum scan time of 3 seconds is possible. The required minimum SDNR depends on the breast thickness since a high SDNR can easily be obtained for thinner breasts while it is a challenge to obtain a good SDNR for thicker breasts with acceptable scan time and dose. For the MicroDose system there are two so called dose levels that could be used. Each dose level specifies a target SDNR at each breast thickness. Dose level C100, which is most commonly used, has lower requirements on the SDNR compared to dose level C120.

It is the task of the Automatic Exposure Control, AEC, to choose the most suitable exposure settings so that all objectives are considered. The AEC of the MicroDose system considers that the patient dose, scan time and energy used by the tube should be minimized while the target SDNR is obtained. First, the tube voltage is chosen based on the measured breast thickness. A conventional AEC would then measure the transmission, and in this way determine the density, in a specific region of the image field and set the appropriate exposure time accordingly. The AEC detector that is placed between the patient and the detector could be movable or consist of many separate detectors in order to increase the probability of measuring at the densest part. But there is a risk of underexposure in systems like this, if the densest part happens to not be examined by the AEC detector [11]. In digital mammography with full field detectors, the transmission can also be measured with a pre-exposure, then using the imaging detector as AEC detector. For a scanning system this way of measuring the
transmission would require a pre-scan but as described in the next section there are alternative ways that are better [12].

### 3.5.1 Exposure that matches transmission distribution

Since the density and radiological thickness is different at each point in the image field it would be optimal to change the scan velocity through out the scan depending on the density at each point. This would eliminate the risk of underexposure of dense parts and also minimize the dose and acquisition time by increasing the scan velocity at low density regions. The criterion is that the target $SDNR$ is obtained at each point. In comparison to a constant scan speed system, where the $SDNR$ is reached at the highest density parts but exceeded at other parts, the dose and scan time can be reduced [12].

In the MicroDose system, the imaging detector is also used as the AEC detector. The first line in the scanning detector measures the transmission at each point in the image and this information is used to adjust the speed of the scan. However, the scan speed cannot vary in the y-direction which reduces the problem to a one dimensional one. The scan velocity is adjusted to match the highest density at each line in the y-direction of the image. The acceleration of the detector is limited to $2.5 \text{ m/s}^2$ and the maximum velocity is $0.2 \text{ m/s}$, this in order to avoid acceleration effects in the image and to achieve reasonable regulation of the detector velocity [11].
Chapter 4

Optimizing the exposure settings

4.1 Precalculations

As explained in Chapter 3, the task of the AEC is to choose the most suitable exposure settings so that the dose is minimized while a certain target $SDNR$ is still obtained within an acceptable examination time. The energy used by the X-ray tube should also be kept low to avoid over heating of the tube and to maximize the workflow. It is also of interest to minimize the wear of the X-ray tube and to choose exposure settings so that the tube lifetime is maximized. The lifetime of the tube is not considered by the AEC of the MicroDose system today but will be considered in this thesis.

The exposure parameters that can be controlled are the photon energy, determined by the tube voltage, and the scan time. In the MicroDose system the set of voltages used has been determined through earlier design. The scan times, doses, tube energies and effects on filament lifetime for all possible voltages are computed by the AEC and it is then determined, through a simple optimization, which voltage is the most appropriate to use at each thickness. It is shown how the scan time, dose, energy in the tube and effect on the tube filament lifetime are calculated by the AEC. Then, an optimization with respect to one parameter at a time or to a function of parameters is used to choose the appropriate tube voltages at all thicknesses. This requires an input of the glandular fraction, which can be estimated in advance, and a constant scan velocity is considered here. The possible tube voltages investigated in the following calculations are ranging...
Chapter 4. *Optimizing the exposure settings*

from 26 to 38 kV in steps of 3 kV, i.e. the same voltages currently used by the MicroDose system.

### 4.1.1 Theoretical determination of scan time, energy and dose

When the breast thickness and glandular fraction are determined the scan times needed to obtain the required $SDNR$ can be computed for the five different tube voltages used in the MicroDose system. The number of incoming photons needed, $N$, in order to reach the target $SDNR$ can be determined as

$$SDNR = C \cdot SNR = C \cdot \sqrt{N} \quad \Rightarrow \quad N = \left(\frac{SDNR}{C}\right)^2.$$  \(4.1\)

The contrast, $C$, is interpolated from tabulated values where the contrast has been measured for different thicknesses of Poly(methyl methacrylate), PMMA. The equivalent thickness of PMMA has to be determined for the given breast thickness, glandular fraction and tube voltage. At 32 kV or example, a 30 mm breast corresponds to 29 mm PMMA while a 60 mm breast corresponds to 50 mm PMMA. This because thicker breasts tend to have a lower glandular fraction. Then, the time needed between readouts where the detector is active to count photons, $t_{act}$, for each detector line can be determined if the photon rate, $N_{rate}$ is known,

$$t_{act} = \frac{N}{N_{rate}}.$$  \(4.2\)

The photon rate, given in photons/ms, depends on the tube voltage, breast thickness and glandular fraction. It is important to take into account that some percentage of the readout channels might not work which would increase the active time needed.

To determine the total scan time the number of readouts during the scan, $n$, the time of each readout when the detector is inactive, $t_{inact}$, and also the acceleration and retardation time, $t_{acc}$, at the beginning and end of the scan all need to be known. The acceleration and retardation distances are both 10 mm, where 2 mm lie outside of the FOV and 8 mm overlap with the FOV where detector readouts are made. This means that the detector maintains a constant scan speed on 232 mm of the 248 mm FOV. Since
a readout is made every 50 \( \mu \text{m} \), 4640 readouts are made during this part of the scan and the time \( t_{\text{act}} \) is needed between each readout. The acceleration time is easily determined when the acceleration is assumed constant and the acceleration distance and final speed are known. The total scan time is then calculated as

\[
t_{\text{tot}} = 4640 \cdot t_{\text{act}} + 4965 \cdot t_{\text{inact}} + 2 \cdot t_{\text{acc}}.
\]  

(4.3)

The energy, \( E \), used by the X-ray tube during an examination can be calculated from the tube voltage, \( U \), the tube current, \( I \), and the total scan time as

\[
E = U \cdot I \cdot t_{\text{tot}}.
\]  

(4.4)

The energy used by the X-ray tube should preferably be kept as low as possible since it correlates to workflow. A high energy is associated with a high production of heat, which requires time for cooling before examinations can be continued.

Next, the patient dose can be determined. The average glandular dose, \( AGD \), is measured in the unit Gray, Gy, which is energy absorbed per kilogram tissue. The \( AGD \) can be determined from the entrance surface air kerma, \( ESAK \), also given in Gy. \( ESAK \) is a measure of the kinetic energy per unit mass released by the ionizing radiation in the air right before the breast. The \( ESAK \) is determined from the number of photons detected, \( N_{\text{detec}} \), divided by the number of photons per mGy, \( N_{m\text{Gy}} \). \( N_{m\text{Gy}} \) has been measured for 45 mm PMMA and therefore \( N_{\text{detec}} \) has to be determined at the same thickness. \( N_{\text{detec}} \) is given by the photon rate multiplied with the sum of the active and inactive time. In order to find the \( ESAK \) for the current breast thickness, \( \tau \), a scaling factor needs to be added since the \( ESAK \) decreases with the square of the distance from the X-ray source. The distance between the tube and the object holder is 640 mm which gives

\[
ESAK = \left( \frac{640 - 45}{640 - \tau} \right)^2 \cdot \frac{N_{\text{detec}}}{N_{m\text{Gy}}}.
\]  

(4.5)

There are three conversion factors, \( c \), \( g \) and \( s \), between the \( ESAK \) and the \( AGD \) [13]. The conversion factors depend on the breast thickness and on the half value layer, \( HVL \)
(a measure of the radiation penetrating power). When the factors are interpolated from tabulated values the dose is calculated as

\[ ADG = c \cdot g \cdot s \cdot \text{ESAK}. \]  

(4.6)

In this way it is possible to compute the theoretical scan times, tube energies and doses for different breast compositions using different tube voltages. Figure 4.2 shows how computed scan times for different voltages, for breast thicknesses ranging from 0 to 120 mm with mean glandular fraction, match actual output data from the MicroDose system. In the MicroDose system the tube voltage used is set depending on the breast thickness which is illustrated in Figure 4.1. However, the use of voltages can change over time if the output of photons from the tube has changed or if the number of dead read out channels has changed.

![Figure 4.1: An example of how the voltages are used depending on the breast thickness for the MicroDose system today. Showing the use at two different dose levels.](image)

4.1.2 Population mean values

A thickness distribution has been determined from a group of 7393 patients. Figure 4.3 shows the distribution of the group with a mean thickness of approximately 53.6 mm. By theoretical determination of the scan time, energy and dose for each breast thickness when using the predetermined tube voltages it is possible to calculate the mean scan time, energy and dose for the population obtained with the MicroDose system today. At dose level C100 a mean scan time of approximately 7.16 s and a mean dose of
0.464 mGy is obtained. The mean X-ray tube energy of an examination is calculated to approximately 40.9 kJ. At dose level C120, which requires higher image quality, the mean scan time is 8.70 s, the mean dose 0.733 mGy and the average energy per examination is 55.1 kJ.

4.1.3 Effect on the X-ray filament lifetime

The lifetime of the X-ray tube is limited by the lifetime of the filament, the cathode emitting electrons. The lifetime of the filament, $L_{fil}$, is dependent on the filament
current, $I_{fil}$, used to heat the filament in order to get thermionic emission, in a strongly non-linear manner given by

$$L_{fil} = 4 \cdot 10^{17} \cdot I_{fil}^{-22.94}$$

(4.7) [14]. A low filament current gives a longer filament lifetime. The filament current effects the tube current, $I_{tube}$, i.e. the current flowing between the cathode and the anode which is proportional to the tube output. For each possible voltage in the MicroDose system today, the tube current is set to a certain value between 158 and 180 mA. In order to determine what filament current is needed to obtain the specific tube current, an interpolation is made from from curves describing the relation between tube current and filament current at different voltages. The curves are found in the X-ray tube description from the producer, Varian, and a reproduction of these are shown in Figure B.1 of Appendix B. The relation between filament and tube current depend on the tube voltage and the higher the voltage the smaller is the filament current needed to achieve the same tube current. Using a higher voltage is thus favourable in order to get a long filament lifetime. For every voltage with its fixed tube current the lifetime of the filament can be determined. When using a tube voltage of 26 kV and a tube current of 158 mA, the lifetime of the filament is found to be 125.3 hours, and for tube settings of 38 kV and 180 mA the calculated lifetime is 160.5 hours.

In order to find the average lifetime of a filament used for patient examinations one has to consider the the breast thickness distribution of the patients. The distribution described in Section 4.1.2 will be used, normalized to one. A good measure of the filament lifetime in this case is the number of exposures that could be performed in the lifetime, denoted $n_{exp}$. The inverse of the number of examination that could be performed is calculated as

$$\frac{1}{n_{exp}} = \int_{\tau} \frac{s(\tau)}{L_{fil}(U(\tau))} n(\tau) d\tau,$$

(4.8)

where $s(\tau)$ is the scan time at a certain thickness, $\tau$, $L_{fil}(U)$ is the filament lifetime dependent on the voltage, $U(\tau)$ is the voltage used depending on the thickness and $n(\tau)$ is the thickness distribution. The average lifetime of a filament in the MicroDose system
today, used for patient examinations, is in this manner calculated to 65,010 exposures when dose level C100 is used. At dose level C120 the system uses the higher voltages more, which results in a longer lifetime in hours, but at the same time the scan times are increased. The average lifetime of a filament, at dose level C120 is thus 57,440 exposures. It might be of interest to study the strong dependence of the tube lifetime in hours on the tube current. An increased tube current would yield an increased tube output of photons, resulting in lower scan time but at the same time a shorter tube lifetime in hours and in the number of examinations that could be performed. It has to be considered whether the decrease in scan time is worth the decrease in tube lifetime. In fact, a 10 % increase of tube current yields approximately a 9 % decrease in population mean scan time but a decrease of approximately 18 % in tube lifetime, to 53,020 exposures. Correspondingly, it can be considered if it is worth to increase the population mean scan time in order to achieve an increased tube lifetime. A 10 % decrease of tube current yields approximately a 10 % increase in population mean scan time but an increase of approximately 21 % in tube lifetime, to 78,950 exposures.

### 4.2 Single parameter optimization

The scan times, doses, tube energies and effects on filament lifetime for all possible voltages have been computed and it can then be determined, through a simple optimization, which voltage is the most appropriate to use at each breast thickness. Initially, the most suitable voltage is determined by optimizing for one parameter at a time, either by minimizing the scan time, minimizing the patient dose, minimizing the tube energy or by minimizing the effect on the lifetime of the X-ray tube filament. As a measure of how an examination will effect the filament lifetime, the inverse of the number of such examinations that could be performed is used as a figure of merit. The number of exposures that could be performed is the filament lifetime at the specific voltage divided by the scan time and the inverse of this merit is to be minimized. The lifetime of the filament can be determined after the voltages have been chosen at all thickness.

However, the allowed time for a scan is restricted to be between 3 and 15 seconds. If a voltage would yield a scan time below 3 or above 15 seconds the time would automatically be set to 3 or 15 seconds respectively. The target $SDNR$ would be exceeded in the former case and not reached in the latter. Only voltages that exactly meet the target for the
\( SDNR \) can be chosen. In those cases where no voltage manage to meet the exact target, the one with the smallest deviation from the target \( SDNR \) is chosen. Also, one has to take into account that the chosen voltage has to be able to reach the target \( SDNR \) for the *worst case scenario* if possible. The *worst case scenario* means that a breast of the highest possible glandular fraction is examined, that the actual thickness is 2 mm larger than measured (the required accuracy of the compression paddle is 2 mm) and that 10\% of the readout channels are not working. The chosen voltage has to be able to reach the target \( SDNR \) in below 15 seconds for this extreme case. For example, a voltage of 26 kV would reach the target \( SDNR \) in less than 15 seconds for a breast of thickness 45 mm under normal circumstances but not in the worst case scenario. Therefore a voltage of 26 kV should not be used for this thickness. The extreme case on the other side of the scale would mean that the examined breast has the lowest possible glandular fraction, that the actual thickness is 2 mm less than measured and that all readout channels are working. This would lead to an unusually short scan time and the chosen voltage should not be such that the \( SDNR \) is unnecessarily exceeded for this extreme case.

The population described in Section 4.1.2 is considered to see what would happen if examinations were performed on the entire group as to optimize for one parameter at a time, given only the restrictions mentioned above. Table 4.1 shows the result of the single parameter optimization. It shows the population mean scan time, mean \( AGD \), mean tube energy per examination and the tube filament lifetime in number of exposures, if the tube voltage for each examination was chosen in order to optimize either of the four parameters at a time.

<table>
<thead>
<tr>
<th></th>
<th>Scan time [s]</th>
<th>AGD [mGy]</th>
<th>Tube energy [kJ]</th>
<th>Filament lifetime [numb. of exp.]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Optimized for</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Scan time</td>
<td>5.223</td>
<td>0.5984</td>
<td>35.64</td>
<td>110,320</td>
</tr>
<tr>
<td>AGD</td>
<td>8.816</td>
<td>0.4320</td>
<td>45.18</td>
<td>52,140</td>
</tr>
<tr>
<td>Tube energy</td>
<td>5.223</td>
<td>0.5984</td>
<td>35.64</td>
<td>110,320</td>
</tr>
<tr>
<td>Filament lifetime</td>
<td>5.223</td>
<td>0.5984</td>
<td>35.64</td>
<td>110,320</td>
</tr>
</tbody>
</table>

**Table 4.1:** The table is showing population mean scan time [s], population mean dose [mGy], mean tube energy per examination [kJ] and the filament lifetime [numb. of exp.] when the tube voltages are chosen so that one of the parameters is optimized at a time.

When optimizing for either scan time, tube energy and tube lifetime the system tends to choose among the highest possible voltages, while when optimizing with respect to
the dose the lowest voltages are chosen. This is illustrated in Figure 4.4.

Figure 4.4: Figures (a) and (b) illustrate what tube voltages are used for different breast thicknesses when optimizing with respect to one parameter at a time. Choosing voltages in order to minimize scan time, tube energy or to maximize the filament lifetime give the same result.

4.3 Optimizing a function of parameters

In reality it is not feasible to chose the tube voltage by optimizing for one single parameter at a time since it would lead to unacceptable values of the the remaining parameters. Instead, a function combining all parameters is considered and minimized at each breast thickness. A suitable cost function to minimize will depend on the thickness, $\tau$, and voltage, $U$, and will have the following form

$$C(U, \tau) = \gamma_d \cdot \ln \left( \frac{\text{AGD}}{\text{AGD}_{\text{min}}} \right) + \gamma_t \cdot \ln \left( \frac{t}{t_{\text{min}}} \right) + \gamma_e \cdot \ln \left( \frac{E}{E_{\text{min}}} \right) + \gamma_l \cdot \ln \left( \frac{l}{l_{\text{min}}} \right) + \gamma_{\text{SDNR}} \cdot \left| \ln \left( \frac{\text{SDNR}}{\text{SDNR}_{\text{target}}} \right) \right|,$$

(4.9)

where $\gamma_d$ to $\gamma_{\text{SDNR}}$ are constants determining how strongly each parameter weighs and $\text{AGD}_{\text{min}}$, $t_{\text{min}}$, $E_{\text{min}}$ and $l_{\text{min}}$ are constants corresponding to the minimum dose, scan time, tube energy and minimum effect on the filament lifetime respectively. For this cost function each term has a lowest value of zero, corresponding to the case were the value of a parameter equals the minimum value possible. In the calculations, the effect that an examination would have on the filament lifetime is measured as the inverse of the
number of such examinations that could be performed. Thus, \( l \) is calculated as the scan
time over the lifetime in hours at the specific voltage. The number of such exposures
that could be performed should be maximized, but the inverse should be minimized.
The actual lifetime of the filament can only be determined after the voltages have been
chosen for all thicknesses. What is novel with this cost function compared to the one
already incorporated in the MicroDose system is that it takes into account the lifetime
of the filament.

In the cost function a term considering the deviation from the target \( SDNR \) is incor-
porated. The constant, \( \gamma_{SDNR} \), in this term is set to 5000 so that a voltage not meeting
the exact \( SDNR \) target is not chosen if other voltages would meet it. If none of the
possible voltages can meet the exact target the one with the smallest deviation from the
target will be chosen. Similar to the single parameter optimization, the chosen voltage
has to work for the extreme cases as well. I.e. if either of the extreme cases need a scan
time below 3 or above 15 seconds there will be a deviation from the target \( SDNR \) which
will lead to a very high value of the cost function. Such a voltage will not be chosen if
there is another voltage meeting target even for extreme cases.

In order to perform the optimization, the values of scan time, dose, tube energy and
effect on filament lifetime are calculated for each breast thickness and each possible
tube voltage. The values are stored in a matrix of dimensions \([n \times m \times q]\), where \( n \)
is the number of parameters in the cost function which in this case equals five, \( m \) is
the number of possible voltages and \( q \) is the number of thicknesses. Then the value of
\( C(U, \tau) \) is calculated at each voltage and thickness diminishing the matrix dimensions
to \([1 \times m \times q]\). The voltages minimizing \( C(U, \tau) \) at each thickness are found and stored.
For the voltages chosen through the optimization, the population mean scan time, dose
and tube energy are determined and the filament lifetime is found through integration
as described in Section 4.1.3.

### 4.3.1 Assigning the cost function constants

An important step of the optimization is to choose the constants \( \gamma_d \) to \( \gamma_l \) so that ac-
ceptable values of all parameters are obtained. Since optimizing for scan time, tube
energy and filament lifetime lead to similar results, and changing one of the correspond-
ing constants will lead to changes in all three of the parameters, the constants \( \gamma_t, \gamma_e \)
and $\gamma_l$ are set to equal. An approach is then to set $\gamma_d = x \cdot \gamma_{t,e,l}$ and investigate how all four parameters change with increasing values of $x$. One must also consider which parameters are the most important to optimize, e.g., a low dose and a short scan time of greater importance than a long filament lifetime. Due to the restriction of keeping the dose very low one might expect that a relatively high value of $x$ is appropriate to use.

Figures 4.5(a) to 4.5(d) show how population mean scan time, dose, tube energy and filament lifetime change when choosing voltages by minimizing $C(U, \tau)$ with different relations between $\gamma_d$ and $\gamma_{t,e,l}$. The dashed lines represents the limiting values displayed in Table 4.1 found through a single parameter optimization. For low values of $x$ population mean scan time, tube energy and filament lifetime are almost optimal while the population mean dose is suboptimal. When the value of $x$ increases the dose reaches 0.4320 mGy which is the optimal value obtained when optimizing only with respect to dose. Scan time and tube energy increase more slowly but for very high values of $x$ they reach their maximal values as illustrated in Table 4.2.

![Figures 4.5](image-url)  

**Figure 4.5:** Figures (a) to (d) illustrate how population mean scan time, dose, tube energy and filament lifetime change with different relations between the constants, $\gamma_d$ and $\gamma_{t,e,l}$, of the cost function $C(U, \tau)$ in Eq. 4.9.
Chapter 4. *Optimizing the exposure settings*

Figure 4.6 shows how population mean scan time, mean tube energy per examination and filament lifetime vary with population mean dose for different values of $\gamma_d$ ranging between 1 and 6.

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Figure 4.6: The curves are showing how population mean scan time, mean tube energy per exposure and filament lifetime vary with population mean dose for different values of $\gamma_d$ ranging between 1 and 6. Voltages [26, 29, 32, 35, 38] are used. Stars show parameter values of the MicroDose system today, for comparison.

Table 4.2: The table is showing population mean scan time [s], population mean dose [mGy], population mean tube energy per exposure [kJ] and the filament lifetime [numb. of exp.] when the tube voltages are chosen by optimizing with respect to one parameter at a time or with respect to cost functions with different constants.

<table>
<thead>
<tr>
<th>Optimized for</th>
<th>Scan time [s]</th>
<th>AGD [mGy]</th>
<th>Tube energy [kJ]</th>
<th>Filament lifetime [numb. of exp.]</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\gamma_d = 4.7$</td>
<td>6.853</td>
<td>0.4734</td>
<td>39.96</td>
<td>69,340</td>
</tr>
<tr>
<td>$\gamma_d = 5.5$</td>
<td>7.420</td>
<td>0.531</td>
<td>41.57</td>
<td>62,460</td>
</tr>
<tr>
<td>$\gamma_d = 30$</td>
<td>8.816</td>
<td>0.4320</td>
<td>45.18</td>
<td>52,140</td>
</tr>
</tbody>
</table>

If a restriction is set that the mean dose cannot be more than 10 % greater than the minimum possible value of 0.4320 mGy then $\gamma_d$ should be 4.7 times $\gamma_{t,e,l}$. This gives a mean scan time that is approximately 31 % higher than the minimum possible value,
a 12 % higher mean tube energy and a 37 % lower filament lifetime as compared to optimal values. The values are displayed in Table 4.2. A restriction that the mean dose cannot exceed the minimum value with more than 5 % leads to a $\gamma_d$ of 5.5 times $\gamma_{t,e,l}$. For these optimization constants the mean scan time will be 42 % higher, the mean tube energy 17 % higher and filament lifetime 43 % lower than optimal values.

### 4.3.2 Thickness dependent constants

In order to improve the optimization it is possible to make the constant $\gamma_d$ of the cost function thickness dependent and change its value for thicker and smaller breasts. Thick breast are exposed to a relatively high dose and the scan time is still as long as up to 9 s. For these breasts it might be better to ease on the constraint of keeping the dose low so that the scan time can be decreased. For thinner breasts that are already exposed to a very low dose it might also be a good idea to accept a higher dose in order to obtain a shorter scan time. Figures 4.7(a) to 4.7(d) illustrate how mean scan time, dose, tube energy and filament lifetime change with different relations between the constants, $\gamma_d$ and $\gamma_{t,e,l}$, for three groups of patients, with breast thickness between 0 and 30 mm, 31 and 70 mm and 71 and 110 mm. From the figures it is possible to determine which relationship between $\gamma_d$ and $\gamma_{t,e,l}$ that is appropriate for each group of thicknesses. Figure 4.7(a) suggests that there is no point in choosing a smaller $x$ than 3.5 for breasts thicker than 70 mm since it will lead to no decrease in scan time. The same figure suggests that an $x$ smaller than 3.5 is best for breasts thinner than 30 mm but this will also lead to a dramatic increase in dose according to Figure 4.7(b).

Figure 4.8 illustrates what improvements can be done by decreasing or increasing $\gamma_d$ and in that way change the priority of obtaining a low dose. From the figure it is seen that it would be possible to decrease the long scan time for breasts thicker than 70 mm by decreasing the value of $\gamma_d$. The figure also suggests that scan times for breasts of thickness between 16 and 22 mm could be decreased if allowing an increase in the already low dose for these breasts. If $\gamma_d$ is set to 3.5 for thicknesses below 23 mm and above 70 mm and to 4.7 for all other thicknesses scan time cuts can be made at an acceptable cost of dose, which is illustrated in Figure 4.9. It is also possible to make reductions in dose by increasing the value of $\gamma_d$, however, it would signify large increases in scan time.
Chapter 4. Optimizing the exposure settings

(a) Mean scan time

(b) Mean AGD

(c) Mean tube energy

(d) Filament lifetime

Figure 4.7: Figures (a) to (d) show how mean scan time, mean AGD, mean tube energy per examination and filament lifetime changes for different relations between $\gamma_d$ and $\gamma_{t,e,l}$ for three groups of patients, with breast thickness between 0 and 30 mm, 31 and 70 mm and 71 and 110 mm.

The thickness dependent $\gamma_d$ described above, with lower priority for a low dose for thicknesses below 23 and above 70 mm, will provide an increased population mean dose. However, the increase in dose are for patients with breast thickness below 23 mm who are already exposed to a very low dose and for patients with breast thickness above 70 mm for whom it is more important to decrease the long scan time. For the thickness dependent $\gamma_d$ varying between 3.5 and 4.7 the population mean scan time is 6.727 s, the mean dose is 0.4828 mGy, the mean tube energy per examination is 39.58 kJ and the filament lifetime is 71,350 examinations.

Although, thickness dependent values of the constants of the cost function is a good way of taking the different objectives at different thicknesses into consideration, it requires a more proper investigation and will not be used further in this report.
Figure 4.8: Middle curves show scan time and dose for each breast thickness when $\gamma_d$ is set to 4.7. Dotted curves correspond to $\gamma_d = 3.5$ which means a lower priority on obtaining a low dose. Dashed curves correspond to $\gamma_d = 5.5$ which means a higher priority on obtaining a low dose. Tube voltages of 26, 29, 32, 35 and 38 kV are used.

Figure 4.9: Dashed curve shows scan time and dose if a thickness dependent $\gamma_d$ is used, compared to scan time and dose with a constant $\gamma_d$ of 4.7 illustrated by the solid curve. The thickness dependent $\gamma_d$ has a value of 4.7 for thicknesses between 23 and 70 mm and of 3.5 for other thicknesses.

4.3.3 Effect of output changes on filament lifetime

It is important that the optimization function provide stable and reasonable results when the tube current and thereby the tube output is increased or decreased, since it is possible
that such changes will be made. When the tube current is increased the scan time is shortened and the filament lifetime is decreased. If $\gamma_d$ of the optimization function is set to 4.7 and $\gamma_t$, $\gamma_e$ and $\gamma_l$ to 1, a 10% increase of current, and thereby output, would lead to an expected decrease in population mean scan time and tube lifetime but also to an increase in the mean dose. This means that with the current cost function, dose can be sacrificed in order to increase the filament lifetime. The conclusion that the filament lifetime has too much influence in the cost function can be drawn. If the dose was to be kept constant when the tube current was increased by 10% the filament lifetime would be 57,120 examinations compared to 61,200 examinations which would be obtained if the dose was increased.

Figure 4.10 illustrates how population mean dose, scan time and filament lifetime changes for different tube currents and values of $\gamma_l$ between zero and one. The cases with the same tube currents as today, 10% higher and 10% lower tube currents are studied. A $\gamma_l$ of zero would mean that the lifetime has no influence when choosing the optimal voltages and that is not what we want. A $\gamma_l$ of 0.5 will, however, give the lifetime some influence but give more stable results for the population mean dose when making output changes. With $\gamma_l$ of 0.5, $\gamma_d$ of 4.7 and the other constants at previous values the population mean scan time will be 7.59 s, mean dose 0.448 mGy, mean tube energy 42.07 kJ and the filament lifetime 60,720 examinations.

The special case where the output of photons is increased but the tube current is kept constant, could be obtained by using a better X-ray tube or a better detector or by increasing the slit width. A 10% increase in output would in such a case give a dose of 0.4733 mGy, a scan time of 6.278 s, a tube energy of 36.49 kJ and a filament lifetime of 75,680 exposures, if $\gamma_d$ was set to 4.7 and $\gamma_{t,e,l}$ to 1.
Figure 4.10: The figure illustrates how population mean scan time, dose and filament lifetime changes with different values of $\gamma_l$ for three different tube outputs, which are regulated by the tube current. $n=1$ corresponds to the same tube currents as today, $n=1.1$ corresponds to 10 % higher and $n=0.9$ to 10 % lower tube currents.
Chapter 5

Optimizing the set of voltages

In all calculations so far tube voltages between 26 and 38 kV in steps of 3 kV have been used, since those are the voltages currently used by the MicroDose system. But once the new cost function has been determined, it is possible to compare the results with the ones obtained when using another set of five voltages. The range of voltages of interest is between 25 and 40 kV, i.e. we want to find an optimal set of five voltages between 25 and 40 kV to use. The cost function that was determined in Chapter 4 and that is used in the following simulations has $\gamma_d = 4.7$, $\gamma_t = 1$, $\gamma_E = 1$, $\gamma_l = 0.5$ and $\gamma_{SDNR} = 5000$.

5.1 Contrast measurements

Values of contrasts for different voltages and thicknesses of PMMA are needed in the calculations to determine the exposure settings. Data exists for thicknesses between 0 and 100 mm PMMA and for all integer voltages between 26 and 38 kV, but since voltages of 25 and 40 kV are to be studied as well, new contrast measurements will be done.

Some contrast measurements will be done at 25 kV for smaller thicknesses since this voltage would predominantly be used in this region but also at one larger thickness. Correspondingly some measurements will be done at 40 kV for larger thicknesses and one measurement at a smaller thickness. A few measurements are also done in the region where data already exists as a verification and to determine whether there is a factor
difference between the old and new data. In total 24 measurements are done at PMMA thicknesses between 20 and 80 mm.

For thicknesses 0, 10, 90 and 100 mm, values of the contrasts at 25 and 40 kV are extrapolated from existing data at the same thickness and other voltages. In this way the new data will match the already existing data. Then a cubic interpolation is made to assign contrast values to all other thicknesses for 25 and 40 kV. Figure 5.1 illustrates where measurements are done and where values are extrapolated and interpolated. The result is shown in Figure 5.2. When new and old data was fused together a dip in contrast values was obtained for low voltages and large thicknesses. This will not effect the simulations since such low voltages are not used for large thicknesses.

<table>
<thead>
<tr>
<th>[mm]</th>
<th>25</th>
<th>26</th>
<th>29</th>
<th>32</th>
<th>35</th>
<th>38</th>
<th>40</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>e.p.</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>e.p.</td>
<td></td>
</tr>
<tr>
<td>10</td>
<td>e.p.</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>e.p.</td>
<td></td>
</tr>
<tr>
<td>20</td>
<td>**</td>
<td></td>
<td>*</td>
<td></td>
<td></td>
<td>*</td>
<td></td>
</tr>
<tr>
<td>30</td>
<td>**</td>
<td></td>
<td></td>
<td>i.p.</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>40</td>
<td>**</td>
<td></td>
<td>*</td>
<td>**</td>
<td>i.p.</td>
<td></td>
<td>i.p.</td>
</tr>
<tr>
<td>50</td>
<td>i.p.</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>**</td>
<td></td>
</tr>
<tr>
<td>60</td>
<td>i.p.</td>
<td></td>
<td>**</td>
<td>*</td>
<td></td>
<td>**</td>
<td></td>
</tr>
<tr>
<td>70</td>
<td>i.p.</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>**</td>
<td></td>
</tr>
<tr>
<td>80</td>
<td>*</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>**</td>
<td></td>
</tr>
<tr>
<td>90</td>
<td>e.p.</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>e.p.</td>
<td></td>
</tr>
<tr>
<td>100</td>
<td>e.p.</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>e.p.</td>
<td></td>
</tr>
</tbody>
</table>

*Figure 5.1: The table illustrates how contrast data is measured and extended to 25 and 40 kV.*

At 39 kV, no contrast measurements have been done. To acquire data for this voltage as well, values are extrapolated for every tenth millimetre of thickness between 0 and
Figure 5.2: The figure illustrates how contrast vary with PMMA thickness and tube voltage.

100 mm, from existing values at other voltages. Thereafter, an interpolation is made for all other thicknesses.

5.2 Introducing an optimization function

At each breast thickness an optimization, to find the optimal voltage to use, is done by minimizing the cost function. In this optimization one of five feasible voltages is chosen as the most appropriate one at the given thickness. In order to determine which set of five voltages that is favourable to use for the entire population a new optimization function need to be introduced,

$$ y(U, \gamma) = \frac{\int C(U, t)n(t)dt}{\int n(t)dt} $$

(5.1)

The optimal set of voltages is found by minimizing Equation 5.1, which has five voltages as free parameters. $C(U, \tau)$ is the cost function given by Equation 4.9 with predetermined values of all $\gamma$, and $n(\tau)$ is the thickness distribution. The constraints on the free parameters is that they are all integer voltages between and including 25 and 40 kV. The following optimization problem is formulated,
Chapter 5. Optimizing the set of voltages

\[
\begin{align*}
\text{minimize} \quad & y(U, \gamma) \\
\text{subject to} \quad & U_i \in [25, 40] \quad \forall i \\
& U_i \in \mathbb{Z} \quad \forall i \\
& i \in [1, 5].
\end{align*}
\] (5.2)

In the optimization function given by Equation 5.1 a thickness distribution is part of the integration. The distribution described in Section 4.1.2 might be used in this place.

5.2.1 Fixating the outer voltages

Fixating the lowest and highest voltage to specific values, and only have three voltages as free parameters, will facilitate the solving of the problem but also ensure that good voltage options exist for very small and large breast thicknesses, that will not affect the value of \( y(U, \gamma) \) considerably since they are rare. As a lowest voltage 25 or 26 kV should be used, and as a highest voltage 38 or 40 kV should be used. To investigate which voltage is the most suitable as lowest and highest, Figure 5.3 is produced.

![Figure 5.3](image)

**Figure 5.3:** Figure 5.3(a) illustrates how scan time and dose varies with breast thickness at 25 and 26 kV, for thicknesses up to 25 mm. Figure 5.3(b) illustrates how scan time and dose varies with breast thickness at 38 and 40 kV, for thicknesses above 80 mm.

For breast thicknesses below 25 mm the dose reduction that could be obtained by using 25 kV instead of 26 kV is small. And since the dose is already very low for these thicknesses, further reduction is not a high priority. Furthermore, the range of thickness
at which the target $SDNR$ is unnecessarily exceeded is not much higher at 26 kV than for 25 kV. For thicknesses over 80 mm the reduction in scan time by using 40 kV instead of 38 kV is small compared to the dose increase. The decrease in scan time could potentially be used to obtain a better image quality instead, but since the current target image quality is always met below 15 seconds, this is not a high priority. Also, the efficiency of the detector decreases with increased voltage. The investigation leads to the conclusion that 26 and 38 kV are adequate to use since the gain of choosing 25 and 40 kV does not weigh up for the losses.

5.3 Performing the optimization

Once the minimization problem has been formulated, the MATLAB script in Appendix A.3 is used to find a set of optimal voltages for the given cost function. When the lowest and highest voltage are fixated to 26 and 38 kV respectively the optimal set of voltages to use at dose level C100 is found to be $[26, 28, 31, 34, 38]$. These voltages used with the chosen cost function will give a population mean scan time of 8.17 s, mean dose of 0.433 mGy, mean tube energy of 43.3 kJ and an average lifetime of 56,550 exposures. The optimal point yields a very low dose but the other parameters obtain values that are worse than those of today. This result indicates that the dose might have been too highly weighted in the cost function when $\gamma_d$ was set to 4.7.

It is investigated if another, more suitable, optimum can be obtained with another value of $\gamma_d$ in the cost function. For a range of values of $\gamma_d$ between 3 and 5 the optimal set of voltages and corresponding parameter values are found and the result is displayed in Table 5.1 and Figure 5.4.

If it is acceptable to keep the dose at approximately the same level as today a $\gamma_d$ of 3.5 should be used in the cost function. The optimal set of voltages to use would then be $[26, 28, 32, 34, 38]$, and which would result in a small decrease in population mean scan time, mean tube energy per examination and an increase in the average lifetime of the tube filament. Compared to the mean values of the MicroDose system today the average scan time could be decreased by 1.4 %, the average energy per examination could be decreased by 1.2 % and the average filament lifetime could be increased by 3.1 % if the dose is kept at the same level. If also accepting that the scan time is kept at the
same level as today, the tube output could be decreased. A decrease in tube current, and thereby tube output, of 1.4 % would keep the scan time at approximately the same value as today but give an increase in filament lifetime of 5.9 %.

<table>
<thead>
<tr>
<th>$\gamma_d$</th>
<th>Optimal voltages</th>
<th>Mean scan time [s]</th>
<th>Mean AGD [mGy]</th>
<th>Mean tube energy [kJ]</th>
<th>Filament lifetime [exp.]</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.0</td>
<td>26, 28, 33, 35, 38</td>
<td>6.453</td>
<td>0.4918</td>
<td>38.73</td>
<td>76,350</td>
</tr>
<tr>
<td>3.25</td>
<td>26, 28, 33, 34, 38</td>
<td>6.815</td>
<td>0.4740</td>
<td>39.74</td>
<td>70,710</td>
</tr>
<tr>
<td>3.5</td>
<td>26, 28, 32, 34, 38</td>
<td>7.065</td>
<td>0.4635</td>
<td>40.43</td>
<td>67,040</td>
</tr>
<tr>
<td>3.75</td>
<td>26, 28, 32, 34, 38</td>
<td>7.302</td>
<td>0.4554</td>
<td>41.05</td>
<td>64,330</td>
</tr>
<tr>
<td>4.0</td>
<td>26, 28, 32, 34, 38</td>
<td>7.512</td>
<td>0.4496</td>
<td>41.57</td>
<td>62,360</td>
</tr>
<tr>
<td>4.25</td>
<td>26, 28, 32, 34, 38</td>
<td>7.706</td>
<td>0.4442</td>
<td>42.08</td>
<td>60,520</td>
</tr>
<tr>
<td>4.5</td>
<td>26, 28, 32, 34, 38</td>
<td>7.979</td>
<td>0.4378</td>
<td>42.77</td>
<td>58,330</td>
</tr>
<tr>
<td>4.75</td>
<td>26, 28, 31, 34, 38</td>
<td>8.174</td>
<td>0.4327</td>
<td>43.34</td>
<td>56,550</td>
</tr>
<tr>
<td>5.0</td>
<td>26, 28, 31, 34, 38</td>
<td>8.431</td>
<td>0.4269</td>
<td>44.04</td>
<td>54,580</td>
</tr>
</tbody>
</table>

Table 5.1: For nine values of $\gamma_d$ ranging between 3 and 5 in steps of 0.25 the optimal set of five voltages and corresponding parameter mean values are determined at dose level C100.

At dose level C120 other voltages are optimal to use. With the cost function with $\gamma_d$
of 3.5 the optimal set of five voltages at this dose level is found to be [28, 31, 33, 35, 38]. For this result all voltages are freely chosen and 28 is chosen as the lowest voltage. This would yield a population mean dose of 0.698 mGy which is 4.7 % lower than at dose level C120 today. The population mean scan time would, however, be 9.40 s which is an increase of 8.1 % compared to today. The average tube energy per examination would be increased with 3.0 % to a value of 55.5 kJ and the average filament lifetime would be decreased with 9.2 % to 52,140 exposures. These numbers are displayed in Table B.1 of Appendix B. If the dose was to be kept at the same level as today, i.e. at approximately 0.733 mGy a $\gamma_d$ of 2.8 would have to be used together with voltages [28, 32, 33, 35, 38]. This would give a 0.3 % decrease in scan time, 0.5 % decrease in tube energy and a 1.8 % increase in filament lifetime. The improvements that could be done are modest at both dose levels, but even smaller at dose level C120.

### 5.4 Using fewer or more tube voltage values

The MicroDose system uses five different voltages for patient examinations, from 26 to 38 kV in steps of 3 kV. With this set of voltages good image quality can be achieved at reasonable scan times and patient doses for the entire population. An advantage is also that the calibration time for the system is kept short. However, it is of interest to investigate how much could be gained, in terms of minimized scan time and patient dose, by using more than five voltages, and perhaps even all integer voltages between 26 and 38 kV or between 25 and 40 kV. Correspondingly, it is of interest to investigate what the loss of using fewer voltages would be. A longer calibration time is acceptable if would greatly benefit the patients. And the calibration time could be cut by using fewer voltages if it would not signify sacrifices for the patients.

In an attempt to answer the above questions, simulations where the system was allowed to choose different number of voltages between 26 and 38 kV, were performed. The system was allowed to choose voltages freely up to a set of five. Thereafter, for sets of up to eight voltages, the lowest and highest voltage was fixated to 26 and 38 kV since those were voltages always chosen by the system. Finally, the system was allowed to use all thirteen integer voltages between 26 and 38 and all sixteen voltages between 25 and 40. The result, at dose level C100, is illustrated in Table 5.2 and Figure 5.5. Corresponding results for simulations at dose level C120 are shown in Table B.1 and
Figure B.4 of Appendix B. At dose level C120 however, the lowest voltage was not fixated to 26 kV but was allowed to be chosen freely. Figure B.3 of Appendix B illustrate how the chosen voltages at dose level C100 are used.

**Optimal voltages**

<table>
<thead>
<tr>
<th>Optimal voltages</th>
<th>Mean scan time [s]</th>
<th>Mean AGD [mGy]</th>
<th>Mean tube energy [kJ]</th>
<th>Filament lifetime [exp.]</th>
</tr>
</thead>
<tbody>
<tr>
<td>37</td>
<td>5.537</td>
<td>0.5903</td>
<td>36.87</td>
<td>100,040</td>
</tr>
<tr>
<td>26, 38</td>
<td>5.614</td>
<td>0.5802</td>
<td>36.46</td>
<td>99,360</td>
</tr>
<tr>
<td>26, 33, 38</td>
<td>6.599</td>
<td>0.4860</td>
<td>39.40</td>
<td>72,780</td>
</tr>
<tr>
<td>26, 28, 33, 38</td>
<td>7.121</td>
<td>0.4650</td>
<td>40.53</td>
<td>66,880</td>
</tr>
<tr>
<td>26, 28, 32, 34, 38</td>
<td>7.065</td>
<td>0.4635</td>
<td>40.43</td>
<td>67,040</td>
</tr>
<tr>
<td>26, 28, 31, 33, 34, 38</td>
<td>7.063</td>
<td>0.4630</td>
<td>40.42</td>
<td>67,080</td>
</tr>
<tr>
<td>26, 28, 31, 33, 34, 35, 38</td>
<td>7.086</td>
<td>0.4615</td>
<td>40.50</td>
<td>66,740</td>
</tr>
<tr>
<td>26, 27, 28, 31, 33, 34, 35, 38</td>
<td>7.074</td>
<td>0.4617</td>
<td>40.47</td>
<td>66,830</td>
</tr>
<tr>
<td>26, 38</td>
<td>7.086</td>
<td>0.4613</td>
<td>40.50</td>
<td>66,730</td>
</tr>
<tr>
<td>25:40</td>
<td>7.039</td>
<td>0.4639</td>
<td>40.31</td>
<td>67,510</td>
</tr>
</tbody>
</table>

Table 5.2: Table illustrates how all parameter values change, at dose level C100, with different number of voltages. Each set of voltages is the optimal one found through optimization. However, for sets of more than five voltages the two outer voltages are fixated.

Figure 5.5: The figure shows how population mean scan time, dose and filament lifetime vary when different numbers of optimal voltages are used at dose level C100.
Chapter 6

Energy Weighting

As explained in Section 3.3.1 the photon counting technique used by the MicroDose system presents the possibility of using energy weighting in order to maximize the dose efficiency and in that manner either improve the image quality or decrease the patient dose. A study, including a large number of experimental measurements and simulations, has been made and is presented in [10]. Maximal $SDNR$ improvements that can be obtained with energy weighting for different object thicknesses and voltages have been determined. These have been used to investigate how energy weighting will affect the previous simulations of this report.

6.1 SDNR improvements with energy weighting

Simulation results made by Johan Berglund, illustrating $SDNR$ improvements obtained with different energy weighting factors at different PMMA thicknesses and voltages are used. One of these plots is shown in Figure 6.1.

From that data, the maximum $SDNR$ improvement at PMMA thicknesses between 20 and 70 mm and at voltages between 26 and 38 kV are collected. From the data, values at thicknesses below 20 and above 70 mm and for 25 and 40 kV are then extrapolated. A table containing $SDNR$ improvements at PMMA thicknesses between 0 and 100 mm and voltages between 25 and 40 kV is made and used in the MATLAB script to simulate how energy weighting will affect examination parameters.
6.2 Effect of energy weighting

In order to simulate how energy weighting will affect the imaging process and the image it has to be incorporated into the MATLAB script. Simulations were done at dose level C100. When energy weighting is used the obtained $SDNR$ will be higher, typically with the factor $ew$ between 1.02 and 1.06, meaning an increase of 2 to 6 percent compared to when energy weighting is not used. The $SDNR$ is thus given by

$$SDNR = C \cdot SNR \cdot ew,$$

(6.1)

where the $SNR$ can be replaced with $\sqrt{N}$, since photon quantum noise is the only noise affecting the image. In the MATLAB script the the number of photons needed to obtain the target $SDNR$ is calculated and when energy weighting is implemented it is given by

$$N = \sqrt{\frac{SDNR}{C \cdot ew}}.$$  

(6.2)

Once implemented the effects of energy weighting can be simulated. To begin with, there are three cases where it is of interest to investigate how energy weighting will
effect the mean values of scan time, dose, tube energy and the filament lifetime. The first case is when the voltages [26,29,32,35,38] are used as is optimal with the new cost function. Secondly, when the same voltages are used in the fixed manner as they are in the MicroDose system today, which is illustrated in Figure 4.1. At last, when the optimal set of voltages, [26,28,32,34,38], found in Section 5.3 is used in the optimal manner given the cost function. These results are illustrated in Table 6.1.

<table>
<thead>
<tr>
<th>Voltages</th>
<th>Mode</th>
<th>Mean scan time [s]</th>
<th>Mean AGD [mGy]</th>
<th>Mean tube energy [kJ]</th>
<th>Filament lifetime [exp.]</th>
</tr>
</thead>
<tbody>
<tr>
<td>26,29,32,35,38</td>
<td>no EW</td>
<td>6.771</td>
<td>0.4760</td>
<td>39.76</td>
<td>70,200</td>
</tr>
<tr>
<td></td>
<td>EW</td>
<td>5.851</td>
<td>0.4617</td>
<td>35.59</td>
<td>84,270</td>
</tr>
<tr>
<td>26,29,32,35,38*</td>
<td>no EW</td>
<td>7.164</td>
<td>0.4637</td>
<td>40.90</td>
<td>65,010</td>
</tr>
<tr>
<td></td>
<td>EW</td>
<td>6.697</td>
<td>0.4329</td>
<td>38.21</td>
<td>69,530</td>
</tr>
<tr>
<td>26,28,32,34,38</td>
<td>no EW</td>
<td>7.065</td>
<td>0.4635</td>
<td>40.43</td>
<td>67,040</td>
</tr>
<tr>
<td></td>
<td>EW</td>
<td>5.889</td>
<td>0.4632</td>
<td>35.61</td>
<td>84,510</td>
</tr>
</tbody>
</table>

Table 6.1: The effects of energy weighting on the population mean value of scan time, dose, tube energy per examination and filament lifetime are displayed for three different cases. When today’s voltages are used with the new cost function, when today’s voltages are used in the same manner as today (marked with *) as illustrated in Figure 4.1, or when the new cost function is used together with the optimal set of voltages found in Section 5.3.

6.3 Optimal voltages when using EW

It is possible that another value of $\gamma_d$ in the cost function and another set of voltages is optimal when using energy weighting. For values of $\gamma_d$ ranging from 3 to 5 in steps of 0.25 the optimal set of voltages and corresponding parameter values are found and are displayed in Table 6.2. If the dose is to be kept at approximately the same level as today a $\gamma_d$ of 3.5 is appropriate. This would lead to a significant decrease in mean scan time of 17.5 %, in tube energy of 12.8 % and increase of 29.1 % in the filament lifetime. If the scan time is to be kept at approximately the same value as today a $\gamma_d$ of 5.0 would be appropriate. It would signify a reduction in dose of 9.9 %, a reduction in tube energy of 4.1 % and an increase in filament lifetime of 0.6 %. If comparing the result to the case when energy weighting is not used, in Table 5.1, it is seen that slightly higher voltages are preferred.
Table 6.2: For nine values of $\gamma_d$ ranging between 3 and 5 in steps of 0.25 the optimal set of five voltages and corresponding parameter mean values are determined when energy weighting is used.

The nine optimal points of Table 6.2 are plotted in Figure 6.2 together with curves showing parameter values obtained with different values of $\gamma_d$ and voltages [26,29,32,35,38].

Figure 6.2: The curves shows parameter mean values using voltages [26, 29, 32, 35, 38] for different values of $\gamma_d$ when energy weighting is used. Dots show parameter mean values obtained at the optimal set of five voltages for different values of $\gamma_d$ ranging between 3 and 5 in steps of 0.25. Stars show parameter mean values that would be obtained if energy weighting was applied today, with no changes in use of voltages. Plus signs show parameter mean values for the MicroDose system today, without energy weighting.
Chapter 7

Summary and Conclusion

7.1 Optimizing the exposure settings

In Chapter 4 a new cost function was introduced. The cost function included five parameters: dose, scan time, tube energy, effect on the filament lifetime and image quality measured as $SDNR$. The constants of the cost function were determined so that acceptable values of all parameters were obtained at different tube outputs.

The chosen cost function highly prioritized a low dose. Figure 7.1 shows how scan time, tube energy and filament lifetime values change with dose for different values of the cost function constant $\gamma_d$. $\gamma_t$ and $\gamma_e$ were set to 1 while $\gamma_l$ was set to 0.5. Voltages [26, 29, 32, 35, 38] were used. It is seen that very small improvements can be done by introducing a new cost function and using the same set of voltages as the MicroDose system today.

7.2 Optimizing the set of voltages

Before performing an optimization to find the optimal set of voltages, it was discussed which voltages are appropriate to use as lowest and highest, in Section 5.2.1. The conclusion drawn was that 26 and 38 kV, i.e. the same ones as used by the MicroDose system today, are the best. Using 25 kV instead of 26 kV would not lead to significant reduction in dose for small breast thicknesses and using 40 kV instead of 38 kV would
Figure 7.1: The curves show how population mean scan time, mean tube energy per exposure and filament lifetime vary with population mean dose for different values of $\gamma_d$ ranging between 1 and 6. $\gamma_t$ and $\gamma_e$ are set to 1 and $\gamma_l$ to 0.5. Voltages [26, 29, 32, 35, 38] kV are used. Stars show parameter values of the MicroDose system today, for comparison.

yield a too high increase in dose compared to the reduction in scan time for large breast thicknesses.

At the end of Chapter 4 the cost function constants were set to $\gamma_d = 4.7$, $\gamma_t = 1$, $\gamma_e = 1$ and $\gamma_l = 0.5$. When the optimization was performed for this cost function, the optimal set was found to be [26, 28, 31, 34, 38] kV. This gave a population mean dose of 0.433 mGy, which is 6.6% lower than the value of today, while all other parameters obtained worse values than today. It was decided that the dose was too strongly prioritized and a new value of the cost function constant $\gamma_d$ was determined so that the optimal set voltages would yield a population mean dose approximately at the same value as today. This result is displayed in Table 7.2, the optimal set of voltages to use is then [26, 28, 32, 34, 38] kV. If the dose is kept at the same level as today, small improvements can be done for the rest of the parameters. If the tube current, and thereby the tube output, is decreased so that also the scan time is kept at the same value as today, the improvement in filament lifetime is greater, which is also illustrated in Table 7.2. Other optimal points at dose level C100 are also displayed in Table 5.1. For the rest of the simulations, however, the cost function with a value of $\gamma_d$ of 3.5 was used.
Chapter 7. Summary and Conclusion

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<table>
<thead>
<tr>
<th>Constants</th>
<th>Mode</th>
<th>Mean scan time</th>
<th>Mean AGD</th>
<th>Mean tube energy</th>
<th>Filament lifetime</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\gamma_d = 3.5$, $\gamma_t = 1$, $\gamma_e = 1$, $\gamma_l = 0.5$</td>
<td>26, 28, 32, 34, 38, 38</td>
<td>1.4% 0% 1.2% 3.1%</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 7.1: Percentual improvement, compared to today’s values, in each parameter as a certain cost function is used with the corresponding optimal set of voltages. The results at two different tube outputs, n, are shown. Results are for dose level C100.

At dose level C120, even smaller improvements can be achieved. If the dose is to be kept at approximately the same level as today, i.e. at 0.733 mGy at this dose level, a $\gamma_d$ of 2.8 should be used together with the optimal voltages [28, 32, 33, 35, 38] kV. This would lead to a 0.3 % decrease in population mean scan time, a 0.5 % decrease in mean tube energy per examination and the number of examinations that could be performed in a lifetime of the tube filament would be increased with 1.8 %.

In Section 5.4 it was investigated what could be gained by using more than five voltages and what the loss would be of using fewer. The result, at dose level C100, is displayed in Table 5.2 and Figure 5.5. For dose level C120, corresponding results are displayed in Table B.1 and Figure B.4. It was found that very small improvements could be done by using more voltages and even by using all integer voltages between 26 and 38 kV or between 25 and 40 kV. Using only four voltages would not lead to significant degradation in any parameter value and could perhaps be considered.

7.3 Energy Weighting

In Chapter 6 the effects of energy weighting were examined. Table 6.1 shows the result if energy weighting was applied to the MicroDose system today, when using voltages [26, 29, 32, 35, 38] kV in the same manner as today. It would give approximately a 7 % reduction in scan time, dose and tube energy, corresponding to the square of the energy weighting factor, $ew$. The filament lifetime would be increased by 7 %. The effects of energy weighting can however be used to reach different objectives. If the X-ray tube output was decreased by 7 % at the same time as energy weighting was used, the scan time would remain unchanged but the filament lifetime would be increased by 22 % to
a value of 79,220 exposures. Dose and energy would both be reduced by 7 %. These results are shown in Table 7.2.

<table>
<thead>
<tr>
<th>Mode</th>
<th>Mean scan time</th>
<th>Mean AGD</th>
<th>Mean tube energy</th>
<th>Filament lifetime</th>
</tr>
</thead>
<tbody>
<tr>
<td>26, 29, 32, 35, 38</td>
<td>7%</td>
<td>7%</td>
<td>7%</td>
<td>7%</td>
</tr>
<tr>
<td>n=1</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>26, 29, 32, 35, 38</td>
<td>0%</td>
<td>7%</td>
<td>7%</td>
<td>22%</td>
</tr>
<tr>
<td>n=0.93</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 7.2: Percentual improvement, compared to today’s values, if energy weighting was applied today.

Decreasing the tube output in order to achieve a longer lifetime instead of keeping the scan time low can be applied for the other cases displayed in Table 6.1 as well. In the case when using voltages [26,29,32,35,38] with the new cost function, the tube output could be decreased by 13 %. At this point the dose would have approximately the same value as today, the scan time would be 7% lower, the energy 13 % lower and the filament lifetime would be increased by 72 % to a value of 111,600 examinations. In the third case, when using the optimal voltages with the new cost function, the output could be decreased by 17 %. At this point the the mean scan time would have approximately the same value as today, the mean dose would be 1.6 % lower, the tube energy 13 % lower and the filament lifetime would be increased by 92 % to a value of 124,800 exposures. The result for the last case is displayed in Table 7.3.

<table>
<thead>
<tr>
<th>Constants</th>
<th>Mode</th>
<th>Mean scan time</th>
<th>Mean AGD</th>
<th>Mean tube energy</th>
<th>Filament lifetime</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\gamma_d = 3.5$, $\gamma_t = 1$,</td>
<td>26, 28, 32, 34, 38</td>
<td>18%</td>
<td>0%</td>
<td>13%</td>
<td>30%</td>
</tr>
<tr>
<td>n=1</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>$\gamma_e = 1$, $\gamma_l = 0.5$</td>
<td>26, 28, 32, 34, 38</td>
<td>0%</td>
<td>1.6%</td>
<td>13%</td>
<td>92%</td>
</tr>
<tr>
<td>n=0.83</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 7.3: Percentual improvement, compared to today’s values, if energy weighting was applied with the new cost function and optimal set of voltages.

Another objective could be obtaining a better spatial resolution. This could be achieved by reducing the slit width of the collimators and the cost would be an increased scan time. But such considerations were not within the scope of this thesis.
7.4 Conclusion

The Philips MicroDose mammography system is today operated on tube voltages 26, 29, 32, 35 and 38 kV and the exposure settings are determined by an AEC that takes into account the dose, scan time, tube load and image quality. Potential improvements such as decreased scan time and tube load and increased filament lifetime are of great interest. The population mean dose obtained with the MicroDose system is very low compared to the other mammography systems on the market today, but further reduction would also be of interest.

By introducing a new cost function, that also takes the filament lifetime into account, and adjusting the set of used voltages to an optimal one, very small improvements could be achieved. If the dose is kept at the same level as today, the scan time would be reduced by 1.4%, the tube energy by 1.2% and the filament lifetime could be increased with 3.1% at dose level C100. If the new cost function and optimal set of voltages were applied at the same time as the tube current was decreased, the dose and scan time could be kept at the same values as today while the tube energy would be decreased with 1.2% and the filament lifetime would be increased with 5.9%. At dose level C120, even smaller improvements could be done. The conclusion is that the system design for the MicroDose today is good with only small possible improvements. Nevertheless, relations between important parameters have been assessed, and knowledge of the complex optimization problem has been expanded. This may be of great use for future development and reoptimization if constrains and priorities were to be changed.

Great improvements can be done in all AEC parameters by using energy weighting. However, it has to be determined which parameter is the most important to optimize. Is it better to achieve large improvement in one of the parameters or to achieve smaller improvements in all parameters at the same time?
Appendix A

Matlab Code

A.1 find_time_dose_energy_lifetime.m

function [ total_scantime AGD tube_energy filament_affect target_SDNR SDNR ] = ...
   find_time_dose_energy_lifetime( constraint , gland_to_PMMA_table , ...
   contrast_table , photon_rate_table , filament_current_table , ...
% Constants .................................................................

n = n_in;
dead_time = 28*10^(-6);  
total_dead_time = 4965*dead_time;  
t = thickness_in;  
mA = n*find_mA(kV_in);  
photons_per_mGy = find_photons_per_mGy(kV_in);  
HVL = find_HVL(kV_in);

% Determine time between readouts.................................
gf = constraint(3,t+1);  
target_SDNR = constraint(4,t+1)/constraint(6,t+1);  
equiv_PMMA = find_equiv_PMMA(gland_to_PMMA_table, gf, kV_in,t);  
contrast = find_contrast(contrast_table, kV_in, equiv_PMMA);  
N = (target_SDNR/contrast)^2;  
lost_channel_factor = 1.0395;  
photon_rate = n*find_photon_rate(photon_rate_table, kV_in, equiv_PMMA);  
time = lost_channel_factor*N/(photon_rate*18*10^3);

% Time at two worst case scenarios...............................%

% 1: max glandularity, measure 2 mm too thin, 10% dead channels  
% 2: min glandularity, measure 2 mm too thick, 0% dead channels  
gf_max = constraint(2,t+3);  
equiv_PMMA_max = find_equiv_PMMA(gland_to_PMMA_table, gf_max, kV_in,t+2);  
contrast_max = find_contrast(contrast_table, kV_in, equiv_PMMA_max);  
N_max = (target_SDNR/contrast_max)^2;  
lost_channel_factor_max = 1.11;  
photon_rate_max = n*find_photon_rate(photon_rate_table, kV_in, equiv_PMMA_max);  
time_max = lost_channel_factor_max*N_max/(photon_rate_max*18*10^3);

if t<2
  t=2;
end

gf_min = constraint(1,t-1);  
equiv_PMMA_min = find_equiv_PMMA(gland_to_PMMA_table, gf_min, kV_in,t-2);  
contrast_min = find_contrast(contrast_table, kV_in, equiv_PMMA_min);  
N_min = (target_SDNR/contrast_min)^2;  
lost_channel_factor_min = 1;  
photon_rate_min = n*find_photon_rate(photon_rate_table, kV_in, equiv_PMMA_min);  
time_min = lost_channel_factor_min*N_min/(photon_rate_min*18*10^3);

% Determine mAs.........................................................

K = 1/0.05*18*0.1*660/532;  
mAs = mA*K*(time+dead_time);
% Determine total scan times.................................
active_time = 4640*time;
acc_time = 0.01/(0.5*((50e-6)/(time+dead_time)));
total_scantime = active_time+total_dead_time+2*acc_time;

acc_time_max = 0.01/(0.5*((50e-6)/(time_max+dead_time)));
total_scantime_max = 4640*time_max+total_dead_time+2*acc_time_max;

acc_time_min = 0.01/(0.5*((50e-6)/(time_min+dead_time)));
total_scantime_min = 4640*time_min+total_dead_time+2*acc_time_min;

% Determine deviation from target SDNR if times not in [3,15]s..............
if total_scantime_min<3
time_3s = (3-5765*dead_time)/5440;
SDNR = contrast_min*sqrt(time_3s*photon_rate_min*18*1e3/...  
lost_channel_factor_min);
if total_scantime<3
time = time_3s;
total_scantime = 3;
end
elseif total_scantime_max>15

time_15s = (15-5765*dead_time)/5440;
SDNR = contrast_max*sqrt(time_15s*photon_rate_max*18*1e3/...  
lost_channel_factor_max);
if total_scantime>15
time = time_15s;
total_scantime = 15;
end
else
SDNR = target_SDNR;
end

% Determine dose........................................
c = Dance_c_breast(t,HVL);
g = Dance_g_breast(t,HVL);
s = Dance_new_s_breast(t);
N_detec = n*find_photon_rate(photon_rate_table,kV_in,46)*...  
18*(time+dead_time)*1e3;
ESAK = (595/(640-t))^2*N_detec/photons_per_mGy;
AGD = EASK*c*g*s;

% Determine tube energy..................................
tube_energy = kV_in*mA*total_scantime*1e-3;

% Determine affect on filament lifetime......................
filament_current = find_filament_current(filament_current_table,kV_in,mA);
filament_lifetime = 4*1e17*filament_current^(-22.94);
filament_affect = total_scantime/(filament_lifetime*60*60);
A.2 cost_function_population.m

% % Author: Deborah Merzan % % Created: 2013-12-30 % % Edited: 2014-03-17 % % Purpose: For all thicknesses, determine which voltage in the set kV, % will minimize the cost function. Compute population mean % scan time [s], population mean dose [mGy], mean energy per % per examination [kJ], filament lifetime [h]. % % Plot: Scan times at all possible voltages and chosen scan times % versus breast thickness in mm. % % Parameters: None % % Return: None

clear all, close all, clc

% Directories and folders to search
table_dir = '..\..\AEC_files\';
table_dir2 = '..\Excel Tables\';
addpath('C:\Users\310148753\Documents\Debbi, Examensarbete\Compute_Scan_Time\m-files_debbi\Functions');
addpath('C:\Users\310148753\Documents\Debbi, Examensarbete\Compute_Scan_Time\m-files_debbi\Dance Functions');

% Reading data to tables...................................................
gland_to_PMMA_table = mexreadArraymultid([table_dir ...
'GlandularityConversion.Array3D']);
%Value: PMMAEquivalent (mm)
%1. gf: Index N corresponds to (N+1)*10 %.  
%2. kV: Index N corresponds to N + 24 kV.  
%3. mm: Index N corresponds to N-1 mm. 

[contrast_num contrast_txt contrast_raw] = xlsread([table_dir2 ...
'Contrasts_new.xls']);
contrast_table = contrast_num(2:end,2:end);
%% Value: contrast
%% 1. kV 25 to 40
%% 2. PMMA thickness from 0 to 100 mm

constraint = mexreadArraymultid([table_dir ... 'C100.Array2D']);
%% 1. Values: 1 = min gf (%), 2 = max gf (%), 3 = mean gf (%),
%% 4 = CNR target, 5 = max exptime (s)
%% 2. mm: Index N corresponds to N-1 mm.

thickness_distr = mexreadArraymultid([table_dir ... 'ThicknessDistr.Array2D']);
%% 1. breast thickness in mm, from 0 to 110
%% 2. number of patients with certain thickness, total of 7393 size(photon_rate_table)

[photon_num photon_txt photon_raw] = xlsread([table_dir2 ... 'Low_Photon_Rate2.xls']);
photon_rate_table = photon_num(2:end,1:end);
%% Value: number of incoming photons/ms for one detector line
%% 1. kV (26,29,32,35,38)
%% 2. PMMA thickness from 0 to 100 mm

[filament_num filament_txt filament_raw] = xlsread([table_dir2 ... 'Tube vs Filament current.xls']);
filament_current_table = str2double(filament_raw(6:12,2:10));
%% Value: filament current [A]
%% 1. kV (26,29,32,35,38)
%% 2. Tube current [mA]

tic

%% Constants.................................................................

kV = [26 29 32 35 38];
thickness = [0:110];
n = 1;
dead_time = 28*10^(-6);
total_dead_time = 4965*dead_time;
numb_of_patients = sum(thickness_distr);
%% Cost function constants
t0=3;%% minimum scan time
d0=0.08;%% minimum dose
e0=10;%% minimum tube energy
l0=3e-006;%% minimum lifetime affect

%% Determine scan time, dose, energy and filament affect for all thicknesses
%% and kVs and save in values_matrix().
for i = 1:length(thickness)
for k = 1:length(kV)
    th = thickness(i);
    % Thickness dependent gamma_d
    % gamma_d_dep = find_gamma_d(th);
    [total_scantime AGD tube_energy filament_affect target_SDNR SDNR] = ...
        find_time_dose_energy_lifetime(constraint, gland_to_PMMA_table,...
            contrast_table, photon_rate_table, filament_current_table, kV(k),... 
            th, n);
    values_matrix(1,k,i) = total_scantime;
    values_matrix(2,k,i) = AGD;
    values_matrix(3,k,i) = tube_energy;
    values_matrix(4,k,i) = filament_affect;
    % Calculate function value at each kV and thickness..............
    t = total_scantime; 
    d = AGD;
    e = tube_energy;
    l = filament_affect;
    % Cost function
    function_value = 3.5* log (d/d0 )+1* log (t/t0 )+1* log (e/e0 )+ ... 
        0.5* log (l/l0 )+5000* abs (log ( SDNR / target_SDNR ));
    function_value_matrix(1,k,i) = function_value;
end
end

% Find kV which minimizes the cost function
% Save time, dose, energy and affect at the optimal kVs
for i = 1:length(thickness)
    [min2_value optimal_index] = min(function_value_matrix(1,:,i));
    index2_matrix(1,i) = optimal_index;% holds index of all optimal kV
    opt_matrix2(1,i) = values_matrix(1, optimal_index, i);% time
    opt_matrix2(2,i) = values_matrix(2, optimal_index, i);% dose
    opt_matrix2(3,i) = values_matrix(3, optimal_index, i);% tube energy
    opt_matrix2(4,i) = values_matrix(4, optimal_index, i);% filament affect
    opt_matrix2(5,i) = function_value_matrix(1, optimal_index, i);
    mA = n*find_mA(kV(optimal_index));
    fil_lifetime_vector(1,i) = find_filament_lifetime ...
        (filament_current_table, kV(optimal_index), mA);
end

all_times = 0;
all_doses = 0;
all_tubeenergies = 0;
average_lifetime = 0;
function_values = 0;

% Add all scan times and doses to determine population mean
for i = 1:length(thickness_distr)
    all_times = all_times+opt_matrix2(1,i)*thickness_distr(i);
end
all_doses = all_doses + opt_matrix2(2,i)*thickness_distr(i);
all_tubeenergies = all_tubeenergies+opt_matrix2(3,i)*thickness_distr(i);
function_values = function_values+opt_matrix2(5,i)*thickness_distr(i);
end

mean_time = all_times/numb_of_patients
mean_dose = all_doses/numb_of_patients
mean_tubeenergy = all_tubeenergies/numb_of_patients
mean_function_value = function_values/numb_of_patients;

% Integrate to calculate filament lifetime in number of examinations
f=(opt_matrix2(1,:)/(fil_lifetime_vector*3600))*thickness_distr;
average_lifetime=1/(f/sum(thickness_distr))

% Calculate value of optimization function, y(kV, gamma)
y = opt_matrix2(5,:)*thickness_distr/sum(thickness_distr)

toc

% Plot scan times vs breast thickness for all kVs and plot chosen scan times
figure(1)
for i = 1:length(kV)
    plot(thickness,squeeze(values_matrix(1,i,:)), ':')
    hold all
end
hold all
plot(thickness, opt_matrix2(1,:), 'k')
xlim([0 110])
ylim([0 20])
xlabel('Breast thickness [mm]', 'FontSize',12)
ylabel('Scan time [s]', 'FontSize',12)
title('C100', 'FontSize',12)
hleg1 = legend(num2str(kV(1)), ' kV',num2str(kV(2)), ' kV',...
    num2str(kV(3)), ' kV',num2str(kV(4)), ' kV',num2str(kV(5)), ' kV',...
    'Used kV');

A.3 optimal_voltages.m

% % Author: Deborah Merzan % % Created: 2014-02-27 %
% Edited: 2014-03-18
%
% Purpose: Determine which set of five voltages in an interval, here
% 26 to 38 kV, minimizes the optimization function. The
% optimization function is the integral of the cost function
% times the population thickness distribution, divided by the
% integral over the thickness distribution.
%
% Parameters: None
%
% Return: None

clear all, close all, clc

% Directories and folders to search
table_dir = '..\..\AEC_files\';
table_dir2 = '..\Excel Tables\';
addpath('C:\Users\310148753\Documents\Debbi, Examensarbete\Compute_Scan_Time\m-files_debbi\Functions');
addpath('C:\Users\310148753\Documents\Debbi, Examensarbete\Compute_Scan_Time\m-files_debbi\Dance Functions');

% Reading data to tables..................................................
% Same as in Appendix A.2.

tic

thickness = [0:110];
best_value = intmax;
% Cost function constants
t0 = 3; % minimum scan time
d0 = 0.08; % minimum dose
e0 = 10; % minimum tube energy
l0 = 3e-006; % minimum lifetime affect

for i = 26
    for j = i+1:35
        for k = j+1:36
            for l = k+1:37
                for m = 38
                    index = [i,j,k,l,m];
                    all_scantimes = 0;
                    all_doses = 0;
                    all_energies = 0;
                    for n = 1:length(thickness)
                        for o = 1:5
                            kV = index(o);
                            th = thickness(n);
Appendix A. MATLAB Code

```matlab
[totalscantime, AGD, tube_energy, filament_effect,...
target_SDNR, SDNR] = ...
find_time_dose_energy_lifetime(constRAINT,...
gland_to_PMMA_table, contrast_table, ...)
photon_rate_table, filament_current_table,...
kV, th, 1);

values_matrix(1, o, n) = totalscantime;
values_matrix(2, o, n) = AGD;
values_matrix(3, o, n) = tube_energy;
values_matrix(4, o, n) = filament_effect;

cost_function_value = 5*log(AGD/d0) + ...
1*log(totalscantime/t0) + ...
1*log(tube_energy/e0) + ...
0.5*log(filament_effect/l0) + ...
5000*abs(log(SDNR/target_SDNR));

cost_functions(o, n) = cost_function_value;
end

[min_value, min_index] = min(cost_functions(:, n));
optimal_cost_functions(1, n) = cost_functions(min_index, n);

all_scantimes = all_scantimes*...
values_matrix(1, min_index, n)*thickness_distr(n);
all_doses = all_doses*...
values_matrix(2, min_index, n)*thickness_distr(n);
all_energies = all_energies*...
values_matrix(3, min_index, n)*thickness_distr(n);

mA = 1*find_mA(index(min_index));
fil_lifetime_vector(1, n) = find_filament_lifetime(...
filament_current_table, index(min_index), mA);
scantime_vector(1, n) = values_matrix(1, min_index, n);
end

mean_scantime = all_scantimes/sum(thickness_distr);
mean_dose = all_doses/sum(thickness_distr);
mean_energy = all_energies/sum(thickness_distr);

y = optimal_cost_functions*thickness_distr/sum(thickness_distr);
if y<best_value
best_value = y;
optimal_kV = [i, j, k, l, m];
opt_mean_scantime = mean_scantime;
opt_mean_dose = mean_dose;
opt_mean_energy = mean_energy;
f = thickness_distr.*scantime_vector';
average_lifetime = fil_lifetime_vector*f/sum(f);
end
```
```matlab
end
end
end
end
end
toc
```
Appendix B

Plots and Tables

B.1 Background

![Figure B.1](image)

**Figure B.1:** Curves show how tube current vary with filament current at different voltages.
B.2 Optimal voltages

(a) One voltage

(b) Two voltages

(c) Three voltages

(d) Four voltages

(e) Five voltages

(f) Six voltages

**Figure B.2**
Figure B.3: In Figures B.2(a) to B.3(d) different numbers of voltages are used. Each set of voltages is the optimal one at dose level C100, found through the optimization described in Chapter 5. The figures show how scan time vary with thickness for all feasible voltages and illustrate which voltage is used at each thickness.
<table>
<thead>
<tr>
<th>Optimal voltages</th>
<th>Mean scan time [s]</th>
<th>Mean AGD [mGy]</th>
<th>Mean tube energy [kJ]</th>
<th>Filament lifetime [exp.]</th>
</tr>
</thead>
<tbody>
<tr>
<td>38</td>
<td>7.387</td>
<td>0.8524</td>
<td>50.53</td>
<td>78,200</td>
</tr>
<tr>
<td>29, 38</td>
<td>8.721</td>
<td>0.7663</td>
<td>53.60</td>
<td>60,360</td>
</tr>
<tr>
<td>28, 33, 38</td>
<td>9.168</td>
<td>0.7184</td>
<td>54.76</td>
<td>54,770</td>
</tr>
<tr>
<td>28, 33, 35, 38</td>
<td>9.371</td>
<td>0.7019</td>
<td>55.36</td>
<td>52,800</td>
</tr>
<tr>
<td>28, 31, 33, 35, 38</td>
<td>9.400</td>
<td>0.6984</td>
<td>55.51</td>
<td>52,140</td>
</tr>
<tr>
<td>28, 31, 33, 34, 35, 38</td>
<td>9.439</td>
<td>0.6962</td>
<td>55.64</td>
<td>51,801</td>
</tr>
<tr>
<td>28, 31, 32, 33, 34, 35, 38</td>
<td>9.449</td>
<td>0.6954</td>
<td>55.68</td>
<td>51,650</td>
</tr>
<tr>
<td>28, 31, 32, 33, 34, 35, 36, 38</td>
<td>9.476</td>
<td>0.6934</td>
<td>55.77</td>
<td>51,410</td>
</tr>
<tr>
<td>26:38</td>
<td>9.493</td>
<td>0.6929</td>
<td>55.80</td>
<td>51,320</td>
</tr>
<tr>
<td>25:40</td>
<td>9.376</td>
<td>0.6983</td>
<td>55.20</td>
<td>52,820</td>
</tr>
</tbody>
</table>

**Table B.1:** Table shows how all parameter values change, at dose level C120, with different number of voltages. Each set of voltages is the optimal one found through optimization. However, for sets of more than five voltages the last voltage is fixated.

**Figure B.4:** The figure shows how population mean scan time, dose and filament lifetime vary when different numbers of optimal voltages are used at dose level C120.
Bibliography


