To my grandmother, Elena Carolina Thomas de Letchford
July 28, 1919 - August 8, 2004
ACKNOWLEDGMENTS

There are many people who assisted me with this long endeavor, and I wish to express my gratitude to them.

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Mammography, a successful screening mechanism, still has room for improvement, particularly in false negative exams. Image quality improvements could be made by reducing the presence of scattered radiation through the image processing technique evaluated here. The study aim was to develop a point spread function (PSF), then use a spatially-variant Wiener filter to remove the scatter PSF from the image. Selection of the PSF used for scatter correction was performed on a pixel-by-pixel basis to account for apparent tissue thickness over each pixel.

The PSF was developed from phantom measurements using Hologic™ Selenia® mammography units and dependent on target/filter combination, tube potential, tissue thickness, and grid presence. The parameters used to describe the PSF, scatter fraction (SF) and mean radial extent (MRE) were obtained with physical measurements and verified with Monte Carlo simulations. The SF’s measured ranged from 0.05-0.17 with grid and from 0.25-0.52 without grid. The MRE’s ranged from 3.2-15 mm with grid and from 19-47 mm without grid. Differences in SF between physical and Monte Carlo measurements were -0.04 to +0.07. Monte Carlo MRE and physical data agreed with goodness-of-fit values of 0.96-1.00 with grid and 0.65-0.86 without grid.
The image noise decreased after scatter correction approximately 60-70\%. Correlation of CNR with MRE was statistically significant before scatter correction \((r = -0.67 \text{ with grid and } -0.89 \text{ without grid})\), but not significant after scatter correction \((r = -0.24 \text{ with grid and } -0.37 \text{ without grid})\). Spatial resolution appeared to have improved, as assessed qualitatively. This method surpassed simpler forms of processing including simple deconvolution and histogram manipulation.

Fifty clinical images representative of parameters common to screening mammography were collected from Mo-target Selenia® units and scatter-corrected. Three radiologists assessed their preference for the original or scatter-corrected images in eight categories related to image quality and anatomical features visibility. Scatter-corrected images were significantly preferred in all categories, with \(p\) ranging from 0.01 to nearly zero. These results indicate the potential to improve clinical image quality using scatter correction. Possibilities for future research include clinical imaging without a grid and applications to magnification imaging and digital breast tomosynthesis.
CHAPTER 1
INTRODUCTION AND BACKGROUND MATERIAL

Introduction

The goal of this study is to develop an image processing algorithm to correct for
the effects of scattered radiation in mammograms. The first step in this study is to
characterize the scattered radiation using physical measurements acquired on clinical
mammography systems. The background information in the following sections therefore
details the relevant aspects of mammography system design and the basic physics of
radiation scatter.

Once the radiation scatter is investigated, the data will be used to develop
mathematical functions describing the amount and distribution of the scatter as
functions of imaging technique and breast thickness. An image processing program can
then be developed to remove the scatter from mammography images. For this reason, a
basic introduction to image processing is included in the following sections.

The goal of removing the scatter from the images is to improve the image quality
parameters that are degraded by scatter. Thus, an explanation of image quality is also
included in this chapter.

The Practice of Mammography

Roughly 12% of women in the U.S. will be diagnosed with breast cancer in their
lifetimes, which accounts for one-third of the total cancer incidence in women. One
quarter of the women who are diagnosed with breast cancer will die from it; on average,
each of these deaths results in eleven years of lost life.¹ Mammography permits early
detection of many of these cancers, which in turn leads to earlier intervention and
increases the likelihood that the patient can be treated successfully with less toxic
therapy. Overall mortality from breast cancer is also reduced by mammography screening.

In the last century, the practice of mammography has developed many features which distinguish it from general radiography. Although it is still fundamentally "radiography of the breast", mammography is a distinct specialty with highly-specialized equipment and imaging techniques. This degree of specialization is necessary because radiographing breasts, which are essentially all soft tissue, is technologically difficult. Contrast among tissues in a radiograph is determined by differences in the attenuation coefficients of those tissues. The attenuation coefficients of the tissues of interest in mammography (adipose tissue, glandular tissue, and carcinoma) are all very similar. Carcinoma and glandular tissue are particularly close, which means that it is even harder to detect cancer in dense breasts than in fatty breasts. Microcalcifications, which are often indicative of cancer, do have a markedly different attenuation coefficient than the surrounding tissues but are still difficult to detect due to their size. In fact, their sizes are so small that spatial resolution requirements in mammography are driven in large part by the need to detect these tiny specks. (The American College of Radiology requires mammography units to be able to image specks with diameters of 0.32 mm as a minimum standard of image quality.)

These unique imaging demands mean that excellent low-contrast detectability and high spatial resolution are vital to performing high quality mammography. However, the fact that mammography is used as a screening examination for millions of women per year means that the radiation dose needs to be kept as low as possible, because breast tissue is very sensitive to the carcinogenic effects of radiation. Unfortunately,
these goals are conflicting, since contrast detectability improves as radiation dose increases (due to the effects of noise, which are discussed in more detail later in this chapter).

### Motivation for This Study

Mammography is a successful screening mechanism that has saved many lives; however, there is still room for improvement. The usefulness of a diagnostic test is described by Bayes' theorem, which states that the probability of having a disease given a positive result on a diagnostic test, \( P(D^+|R^+) \), can be calculated:

\[
P(D^+|R^+) = \frac{P(R^+|D^+), P(D^+)}{P(R^+)}
\]

where \( P(D^+) \) is the probability of a person having the disease, \( P(R^+) \) is the probability of obtaining a positive test result, and \( P(R^+|D^+) \) is the probability of testing positive given that the disease is present. When the test is positive and disease is present, the test result is called a "true positive"; the probability of a true positive is equal to \( P(R^+|D^+) \). When the test is negative \( (R^-) \) and the disease is not present \( (D^-) \), the result is a true negative, equal to \( P(R^-|D^-) \). When a test is positive but the disease is not actually present, the result is a false positive, equal to \( P(R^+|D^-) \) or \( 1 - P(R^-|D^-) \). When the test is negative but the disease is actually present, the result is a false negative, equal to \( P(R^-|D^+) \) or \( 1 - P(R^+|D^+) \).

Perhaps the biggest cause for concern in mammography is the rate of false negatives, which average around 20\%,\(^7\) indicating that a large number of breast cancers are not detected by mammography. False negatives are of particular concern in women with dense (highly glandular) breast tissue, because glandular tissue has an attenuation coefficient very close to that of carcinoma\(^4\). Another concern is the rate of
false positives, which was reported by Brewer et al. to be around 11%.\textsuperscript{8} A false positive result causes the patient to undergo the unnecessary pain, stress, and expense of additional diagnostic procedures. The major cause of false positives is a lack of specificity in mammography, because it is not always possible to determine whether a lesion is benign or malignant from the mammography images.

An improvement in image quality would allow small or low-contrast lesions to be detected more easily, thus potentially reducing the false negative rate, allowing a greater number of cancers to be detected earlier, and improving treatment outcomes. The false positive rate could also improve if lesions were seen more clearly.

Another motivation for this study is the possibility of reducing radiation dose to the breast. If other methods of scatter correction are available, it might be possible to remove the grid, or at least reduce its impact on dose, potentially resulting in a sizeable dose reduction. A complete removal of the grid would result in a dose reduction equal to the Bucky factor, which is typically a factor of 2-3. Over the millions of mammography exams performed each year, such a reduction in dose would predict a noticeable reduction in the number of future cancers predicted to be caused by mammograms worldwide. The BEIR VII report indicates that the excess relative risk per sievert (Sv) for breast cancer is about 2 for women over age 35 at the time of exposure, and is well-described by a linear function\textsuperscript{1}. Thus, a reduction in breast dose by a factor of two would be expected to reduce the risk of breast cancer caused by mammography by a factor of two. The excess absolute risk reported in the BEIR VII report is 6.7 per 10,000 persons per year per Sv (with 95% confidence limits of 4.9 and 8.7). Over the approximately 40 million mammography procedures performed per year\textsuperscript{9}, and assuming
four exposures per procedure with an average glandular dose of 2 mGy per exposure, that dose reduction amounts to a savings of approximately 54 cancer cases per year (with 95% confidence limits of 39 and 70 cancer cases per year).

**Mammography Equipment Design**

The unique imaging demands of mammography mean that specialized equipment is required to perform mammography examinations. Mammography equipment differs from general radiography equipment in everything from the general geometry and mechanics to the x-ray spectrum and detectors used.

**Geometry and Mechanics**

The mammography machine is basically a C-arm configuration, with the x-ray tube at the top and the detector at the bottom (Figure 1-1). The entire C-arm can be moved up and down to allow for a range of patient heights, or rotated to allow for different views, but the tube and detector do not move in relation to each other, because the source-to-image distance (SID) is typically fixed, typically somewhere between 60-70 cm. This SID provides a compromise between factors that encourage a long SID and factors that are optimized by a short SID. For example, to reduce focal spot blur and to avoid beam non-uniformity due to the heel effect, a long SID would be best. But to keep exposure times short enough to reduce patient motion and tube heating (a concern with non-tungsten targets), a short SID is required. The best compromise turns out to be around 60-70 cm. The Hologic™ Selenia® (Hologic™, Inc., Bedford, MA) system used in this study has an SID of 66 cm.

The mammography unit has a carbon fiber breast support surface that is typically about 2 cm above the detector. The space between the breast support and the detector is referred to as the air gap, and is where the grid (to be discussed later in this chapter)
is located (Figure 1-1). In screening mammography, the breast is positioned directly on the breast support surface, with the grid in place between the breast support surface and the detector. An alternate geometry is used for the diagnostic technique of magnification imaging, in which the images are acquired with the breast positioned on a platform suspended about 20-30 cm above the detector. Magnification images do not use a grid, because the large air gap between the breast and the detector acts as a scatter rejection mechanism.

Compression is always used in mammography, so there is always a compression paddle in the beam. The compression paddle is made of clear plastic, typically with a thickness of about 2 mm. There is a range of sizes and styles available for different applications, but the most common paddles in use (and the only ones considered in this study) are the large and small full-field screening paddles, which match the size of the large and small fields of view used for large and small breasts.

The mammography generator is not significantly different from that of a general radiographic unit and is not discussed here, other than to mention that high-frequency generators are standard in mammography because they minimize voltage ripple. Ripple is undesirable because the resulting fluctuations in the tube potential can increase dose and exposure time and alter the image contrast. This problem is particularly pronounced in mammography due to the low photon energies used; a relatively small amount of ripple alters the tube potential by a significant percentage.

**The x-ray tube**

Like any other x-ray machine, photons are produced in an x-ray tube, and the photons used to make the image pass through a tube port when exiting the tube. (The remaining photons are shielded by the tube housing.) Unlike a traditional x-ray tube, the
tube port of the mammography tube is made of beryllium, which is necessary to avoid too much attenuation of the low-energy spectrum. The tube also differs by the use of specialized target and filter materials, which increase contrast and reduce dose, and the use of a much smaller focal spot than that of a traditional x-ray tube, which increases spatial resolution.

All mammography tubes are positioned so that the central axis of the beam passes through the tissue at the chest wall, as opposed to the geometry used in general radiography in which the central axis is at the center of the image (Figure 1-2). This configuration allows all the breast tissue at the chest wall to be imaged without an unused portion of the x-ray beam extending into the patient’s torso. The “other half” of the beam is simply removed by collimation. The anode is on the anterior (opposite the chest wall) side of the tube, which allows the transmitted intensity of the x rays to be more uniform throughout the image because the heel effect causes a drop-off in intensity in the anterior direction. Thus, the areas of the breast which are usually thicker and therefore more attenuating (chest wall) receive more radiation than the areas which are usually less attenuating (toward the nipple). This arrangement was more important in the days of screen-film mammography because the dynamic range of the film was very limited, but it is still important in digital imaging as a dose-saving measure and to keep noise levels more constant throughout the field of view.

One important consequence of this geometry is the impact on the effective focal spot size. The nominal focal spot size in mammography is 0.3 mm for the large focal spot and 0.1 mm for the small focal spot, but that number can be misleading. The effective focal spot size is often quoted as being a single number (at a location defined
by the manufacturer, usually at or near the center of the x-ray field), but it actually changes size depending on the position in the field. Specifically, its projection gets larger approaching the cathode end of the tube and smaller approaching the anode end. Thus, the spatial resolution in the image is the worst at the chest wall and gradually improves towards the nipple.

**The antiscatter grid**

For contact (i.e., non-magnification) mammography, the radiation exiting the breast passes through a grid before reaching the detector. The grid is simply a device designed to reject scattered radiation, the effects of which are discussed in more detail later in this chapter. The grid does not remove all of the scattered radiation, but it does remove enough to improve image contrast by up to 40%. However, there is a price to pay for that improvement, because the grid also blocks about 25-30% of the primary radiation that is needed to form the image. Thus, more radiation must be used to compensate, which raises the dose to the patient. This increase in dose is called the Bucky factor, and it is typically about 2-3, which means that more than twice as much radiation must be used with a grid than without.

As in general radiography, the traditional (linear) grid in mammography is made of lead strips interspaced with a material such as carbon fiber or air (Figure 1-3). The grid ratio (ratio of the height of the grid to the width of the interspaces) is usually about 5:1, which is less than the grid ratio used in general radiography. A higher grid ratio provides better scatter rejection but increases the dose penalty. Grid frequencies are usually about 30-50 lines/cm, and the grid moves during the exposure to blur the grid lines so they do not appear on the image.
The Hologic™ Selenia® unit used in this study uses a cellular or crosshatched, rather than a linear, grid (Figure 1-3). This grid is made of copper rather than lead, with air interspaces, and a grid ratio of 4:1. The grid frequency is 23 lines/cm. Despite the lower grid ratio and frequency, the grid rejects scatter in 2 dimensions rather than only 1 dimension, and research has shown that a cellular grid improves contrast by about 5-9% over common linear grids, depending on the techniques used. The Bucky factor is also around 5% less than that of typical linear grids used in mammography.

The Mammography X-Ray Spectra

Of the many important differences between general radiography and mammography, the x-ray spectrum is critical. The applied x-ray tube voltage must be much lower in mammography in order to visualize the small contrast differences among the different soft tissues and between soft tissue and microcalcifications (Figure 1-4). Mammography typically uses x-ray spectra ranging from about 25 kVp to 35 kVp, as opposed to general radiography, in which beam energies typically range from about 50 kVp to 120 kVp. Although breasts are nearly entirely soft tissue, the large variability in compressed breast thickness—from less than 2 cm to more than 8 cm—is the main reason there is a range of tube potentials available. Breast glandularity also plays a role in kVp selection, because dense breasts attenuate more radiation than fatty breasts of similar thickness. Beam energies at the lower end of the spectral range provide better contrast, but it is not possible to penetrate thick or dense breasts at those energies without resorting to unreasonably long exposure times (leading to increased motion) and high absorbed doses.

Computer modeling studies have shown that the ideal spectrum to balance signal-to-noise ratio and dose in mammography would consist of a monoenergetic
beam of 15-25 keV, depending on breast thickness. The best way to approximate this ideal is to use characteristic x rays to make the images rather than bremsstrahlung x rays. However, medical x-ray tubes are not able to emit only characteristic x rays, because they produce a composite spectrum comprised of characteristic and bremsstrahlung x rays. Filters are used to shape the undesirable parts of the bremsstrahlung spectrum. In mammography, this shaping occurs both by absorption of low-energy x rays, mostly due to the thickness of the filter, and by absorption of higher-energy x rays due to the K-edge of the filter material. The K-edge is the energy at which an incident photon is able to overcome the binding energy of a K-shell electron and remove it via the photoelectric effect. Above this energy, transmission through a filter drops off markedly. The filters used in the Hologic™ Selenia® systems have the following K-edges: molybdenum, 20.0 keV; rhodium, 23.2 keV; silver, 25.5 keV.

The combination of the target material in the x-ray tube (embedded in the anode) and the material of the filter used to shape the spectrum is referred to as the “target/filter combination.” Current digital mammography units have at least two filter choices, and some have multiple targets, as well. Traditionally (i.e., in screen-film mammography), the target used in the x-ray tube is molybdenum (Mo), with Mo or rhodium (Rh) filtration, depending on breast thickness. A Rh target with a Rh or aluminum (Al) filter is also sometimes used, for thicker breasts. The use of these target/filter combinations is still extremely common, even in digital mammography. However, the advent of digital mammography has allowed the use of other target/filter combinations, as well—even ones that do not produce useful characteristic x rays. For example, tungsten (W) targets paired with Rh or silver (Ag) filters are becoming increasingly common, because digital
detectors have different absorption characteristics from intensifying screens. Digital acquisition allows the displayed image to undergo image processing to improve the image appearance. (Image processing will be discussed in more detail later in this chapter.) Thus, dose and image quality are not necessarily linked in the same way they are in screen-film systems. Several researchers have claimed that tungsten targets produce image quality equivalent to Mo targets, with lower dose.\textsuperscript{13-15} The spectra of the target/filter combinations present in the mammography units used in this study are shown in Figure 1-5. For the example of these 30 kVp spectra, the average beam energies are: 16.7 keV for Mo/Mo, 17.6 keV for Mo/Rh, 19.3 keV for W/Rh, and 20.3 keV for W/Ag.

**The Mammography Detector**

The x-ray absorber used in the Hologic\textsuperscript{TM} Selenia\textsuperscript{®} detector is a thin sheet of amorphous selenium (a-Se), shown schematically in Figure 1-6. The a-Se is sandwiched between two electrodes: the one at the top (i.e., toward the x-ray tube) is continuous, and the one on the bottom is broken up into a large matrix of dels (short for “detector elements”). In order to clarify the description of this type of detector, in which the signal does not involve light conversion, the term “direct digital detector” is used in this discussion. Other detector types are not included in this discussion, as they are not relevant to this project.

When photons hit the a-Se, they remove electrons from some of the selenium atoms via the photoelectric effect. The electrodes generate an electric field that attracts the freed electrons toward the dels, where they are collected (Figure 1-7). Each del acts as a capacitor and stores the charge until it is queried to release the charge (i.e., in the "reading" process). A thin film transistor (TFT) is attached to each del and acts as a
switch to release the charge when it is time to be read. Each TFT is connected to a control line and a data line, which are oriented perpendicular to each other; the control line signals the switch to close, allowing data to pass through the data line (Figure 1-8).

Each of the dels corresponds to a pixel in the final image. The Hologic™ Selenia® system has a maximum image size of 3420 x 4096 pixels, which means there are 14,008,320 dels in the detector. Spatial resolution is determined by the size of the dels (which are 70 µm in the Hologic™ Selenia® detector), and if there is a problem with a del, there is a problem with that image pixel. "Dead pixels" are common artifacts in digital images, and they are caused by non-functioning dels. Another common artifact is a defective line, which results from a problem with the connection for a column, causing a column of dead pixels in the image. The dead column might span just a few pixels at the edge of the detector, or it might extend the entire length of the detector, depending on where the connection is broken, potentially requiring detector replacement.

**Image Display and Processing**

Once the electrical signal from the dels is read out, the next step in the image display process is to associate the amplitude of the signal from each pixel with a number. This process is highly dependent on the manufacturer and calibration, so it is not valid to assume that a pixel value of “600” from one unit indicates the same level of signal as a pixel value of “600” from another unit. The distribution of pixel values throughout the image is called the histogram, which is displayed as a graph with pixel value on the abscissa and frequency of that pixel value on the ordinate. An example of a typical histogram is shown in Figure 1-9.
Each of these pixel values is then associated with a gray level in the image by use of a look-up table (LUT), which provides the display program with a means of mapping pixel values into gray scale values. The LUT is selected automatically by the image processing software and is often dependent on the image view (e.g., CC, MLO, magnification, implant). Some manufacturers provide customizable look-up tables to suit radiologist preference. Windowing and leveling the image pixel values further adjust the look-up table, causing adjustment of the displayed image in contrast and brightness.

The mammography image viewed by the radiologist does not display the distribution of pixel values exactly as they emerge from the detector; rather, the image has undergone considerable processing to improve the image quality. Image processing of digital mammograms happens on multiple levels. The first step is the flat-field correction, which is applied to all images and normally cannot be disabled. Further processing, often specific to the image view, may then be applied to accentuate certain features of the image or regions of the histogram.

The flat-field correction is essentially a map that is generated by radiographing a uniform object (usually a slab of acrylic). The map is acquired at a variety of tube potentials, with all target/filter combinations available on the unit, and in both the contact and magnification modes. The values from each detector pixel are normalized to create the correction maps, and the appropriate correction map is then applied to every clinical image simply by dividing each pixel value by the corresponding number on the map. The flat-field correction has a few purposes: to equalize gain across the detector, to disguise minor detector artifacts, and to correct for the heel effect. Gain equalization is needed because the detector is subdivided into detector blocks for read-out purposes.
Each block of dels has a gain associated with it; electronic gains are never exactly the same among detector blocks because of small variations in the electronics. If no correction were applied, the detector blocks would be clearly visible in the image because the overall brightness of each block would be slightly different. The flat-field correction is also used to "correct" dead pixels and other defects. A dead pixel is replaced in the image by the average of the surrounding pixels. A defective line can also be replaced by the average of the neighboring lines, although a large number of defective pixels (either in a line or sprinkled throughout the image) requires detector replacement. This practice of disguising dead pixels is controversial because of the potential to miss microcalcifications. The radiologist is usually unaware of the number or location of dead pixels in the image and so has no idea whether there might be information missing from the image. The manufacturer, rather than the radiologist, typically makes the decision whether the number of dead pixels is sufficient to merit detector replacement. This problem is unique to digital imaging, in contrast to imaging with film, an analog detector.

Other defects, such as spots in the detector that are lighter or darker than the rest, can also be "mapped out" by applying corrections to those pixels. Even the heel effect, which is caused by the x-ray tube rather than the detector, is mapped out by the flat-field correction.

Other forms of image processing generally include algorithms to enhance the visibility of the nipple and skin line and to optimize window/level settings for the specific view. Additional forms of image processing are likely also used, and details about the algorithms are proprietary to the manufacturer. Many of the algorithms are concerned
with displaying the data in a reasonable dynamic range. Some examples (which may or may not be used by any particular manufacturer) include.\textsuperscript{16}

- \textbf{Histogram-Based Intensity Windowing.} The window/level settings are determined based on an analysis of the breast glandularity, which can be estimated by comparing the sizes of the “humps” in the histogram (Figure 1-9). Thus, a fatty breast will be displayed different from a glandular breast.

- \textbf{Mixture-Modal Intensity Windowing.} The image is segmented to determine the areas comprising the background, fatty tissue, glandular tissue, and muscle. Each of these regions is displayed using different window-level settings, combined into one image. This process improves the contrast of all types of tissue in the breast, without emphasizing one to the detriment of the others.

- \textbf{Unsharp Masking.} The program creates a low-pass filtered (blurred) version of the image, multiplies it by a weighting factor, and subtracts it from the original image. This process has the effect of reducing contrast in large structures to limit dynamic range while enhancing the visibility of details such as calcifications and mass borders.

- \textbf{Peripheral Equalization.} This algorithm flattens intensities across the image, so that the thinner peripheral tissue can be displayed with the same window/level settings as the thicker central tissue. This algorithm is also a means of dynamic range compression, but it can actually improve image contrast because the part of the dynamic range that was used to represent thickness of the breast tissue can now be reallocated to the entire breast.

Although the details of the algorithms each manufacturer uses are generally not available, all manufacturers manipulate the images to some extent before they are viewed. Thus, the apparent image quality is always the result of the detector’s capabilities combined with the effects of image processing.

\textbf{Image Quality}

A major goal of this study is to assess the impact of scatter correction on image quality. The apparent quality of an image is determined by the combined effects of noise, contrast, and spatial resolution, which are discussed in detail in the following sections.
Image Noise, Signal-to-Noise Ratio, and the Noise Power Spectrum

Noise is a random statistical fluctuation in a signal. In the case of digital imaging, the signal of interest is the array of pixel values in the image. The fundamentals of image noise are described below. In medical imaging, however, knowledge of the image noise by itself is rarely as important as understanding the effects of noise on image quality. The sections on signal-to-noise ratio (SNR) and the noise power spectrum (NPS) describe noise measurements that are useful for imaging applications.

Fundamentals of image noise

Assume there exists an ideal, completely noiseless imaging system. If the detector was covered with a perfectly uniform slab of some object, then exposed to a uniform x-ray beam, the resulting pixel values would be exactly the same over the entire area of the detector. In fact, if the detector response and thickness and attenuation properties of the object were known, the exact pixel values could be calculated without ever acquiring an image. However, this procedure is never possible in reality because noise has the effect of adding or subtracting random values from the “expected” signal. The mean of all the pixel values under a uniform attenuator should still be nearly equal to the expected pixel value, but the individual pixel values vary. The degree of variation can be described by the standard deviation of the signal, which is essentially a measure of the noise characteristics of the imaging system. The greater the standard deviation in the signal, the further the system is from the ideal, noiseless system. Noise is a problem because it can obscure small or low contrast objects in an image, so it is preferable to have a system as close to the ideal as possible.

Image noise has several causes. The first of these is photon (or quantum) noise. In the above example, it was assumed that the system had an x-ray beam that was
uniform in time and space. However, such a thing does not exist. It is a basic fact of physics that there is some variation in the number of photons incident on a particular point per unit time. Even if the timer, current, and voltage of an x-ray tube had perfect reproducibility, so that the field was uniform in time and space, and a radiation detector had perfect efficiency, the detector would still record a slightly different signal each time a measurement was made. It should be noted that this phenomenon is not the result of an imperfect production of x rays from an x-ray tube; it is also seen in the case of natural radioactivity and in other areas of the electromagnetic spectrum, such as with visible light photons.

The second cause of image noise is due to a non-ideal detector, because some random variation occurs during the processing of the signal. Thus, even if several detectors detected exactly the same number of photons, it is likely that they would all produce slightly different signals. In the case of direct digital detectors such as those used in the Hologic™ Selenia® system, there are several causes of detector noise. The first is the conversion of x-ray energy into an electronic charge in the amorphous selenium. There is a statistical variation in the number, energy, and location of electrons produced during this process. It is also true that these electrons are not collected with 100% efficiency, so there is some variation in signal caused by incomplete signal collection. Once the signal is collected, there is additional noise associated with the electronics involved in relaying and processing the signal.

Detector noise is strongly related to system design and thus is not something that is generally affected by imaging techniques. Quantum noise, on the other hand, may be
assumed to be Gaussian and directly related to the number of photons detected by the
equation:
\[ \sigma = \sqrt{N}, \]  
(1-2)
where \( \sigma \) is the standard deviation of the signal, and \( N \) is the number of photons. This
equation states that the absolute fluctuation in the signal per pixel increases as the
square root of the total signal.

**Signal-to-noise ratio**

The total amount of noise is not really the quantity of interest; rather, it is the ratio
of the total signal to the noise (the signal-to-noise ratio, or SNR) that determines the
image quality. Neither the signal nor the noise separately is sufficient, as should be
evident from knowledge of basic statistics. For example, if the average signal is 10, but
the standard deviation of that signal is 100 (SNR = 0.1), then it is not possible to
determine whether the “true” signal is significantly different from zero. However, if the
signal is 10 and the standard deviation is 1 (SNR = 10), then there is a 95% confidence
that the “true” signal is between 8 and 12 (i.e., a range which firmly excludes zero). In
order to identify an object with 100% accuracy, the SNR needs to be higher than 5. This
requirement is known as Rose's criterion.4

The SNR is the ratio of \( N / \sigma \), and by substitution of Equation 1-2, it is apparent
that for a Gaussian distribution,
\[ SNR = \sqrt{N}. \]  
(1-3)
Thus, although the absolute noise increases with the number of photons, the SNR
increases as well. Image quality therefore improves as the number of detected photons
increases. In terms of x-ray imaging, the most common ways of improving the SNR
include increasing the tube current-time product (mAs), in order to generate more photons from the x-ray tube, and increasing the tube potential, which both produces more photons and increases object penetration. Using a filter with a higher K-edge can also improve SNR by increasing the average beam energy (Figure 1-5), which improves object penetration. This effect on SNR is the major reason why many mammography units have multiple target and filter choices. Raising the tube potential increases penetration as well as the number of photons, but most of the photons above the K-edge are attenuated, so the effect on the average beam energy is relatively small, and increasing the K-edge has a much larger effect on the average energy. The trade-off to the increase in penetration is a decrease in contrast, which is discussed in more detail later in this chapter.

**Noise power spectrum**

While it is true that the SNR is a useful measurement from the standpoint of an observer viewing the image, for research purposes it is often necessary to have more detailed information about the image noise, i.e., a description of the noise structure. If the only contribution to image noise were from quantum effects, and if the beam were perfectly uniform across the detector, the noise would have a truly random distribution across the image. In reality, however, the noise at any two points in the image is generally statistically related. This relationship is referred to as spatial structure. In the case of digital mammography, noise structure results from factors including beam non-uniformity; variations in the detector materials (such as non-uniformities in the structure of the amorphous selenium); electronic noise affecting charge collection, read-out, and amplification; and computation "errors" occurring during processing.
The autocorrelation function, $c(\Delta x, \Delta y)$, describes the statistical relationship between two points in an image as a function of distance:

$$c(\Delta x, \Delta y) = \lim_{X,Y \to \infty} \frac{1}{2X} \frac{1}{2Y} \int_{-X}^{X} \int_{-Y}^{Y} \Delta D(x, y) \Delta D(x + \Delta x, y + \Delta y) dx dy,$$

where $\Delta D$ is the difference between the signal at the point of interest and the mean.\(^{17}\) There are two conditions for this equation to be accurate: the image noise must be stationary and ergodic. "Stationary" means that the autocorrelation function depends only on the distance between two points, but not on the absolute locations of those points within the image.\(^{18}\) "Ergodic" means that, as long as enough photons have been recorded to provide a reasonable statistical sample, the noise structure does not vary from one image to the next.\(^{19}\)

It is often more useful to describe the noise structure as a function of frequency rather than spatial distance. The noise power spectrum (NPS), also called the Wiener spectrum, is defined to be the Fourier transform of the autocorrelation function:

$$W(u, v) = \lim_{X,Y \to \infty} \frac{1}{2X} \frac{1}{2Y} \left| \int_{-X}^{X} \int_{-Y}^{Y} \Delta D(x, y) e^{-2\pi i (ux + vy)} dx dy \right|^2.$$

Equation 1-5 is obviously two-dimensional. Often, however, the NPS is isotropic, and for comparison purposes it is simpler to use a one-dimensional NPS. The one-dimensional NPS is simply a cross-section through the two-dimensional NPS, starting from zero frequency to the maximum frequency of interest\(^{17}\) (Figure 1-10).

**Contrast and Contrast-to-Noise Ratio**

The second major component of image quality is image contrast, which may be defined as the difference in signal between one object and another in an image. In the context of digital imaging, it can be described as the difference in monitor brightness
between two objects, as seen by the radiologist. The final image contrast depends on subject contrast, detector properties, image processing, and display characteristics.

There are numerous types of contrast that ultimately contribute to the image contrast. The most basic of these is object contrast: two identical objects have no associated image contrast. In the case of x-ray imaging, object contrast is determined by differences in any of the physical characteristics of tissues which affect attenuation. Examples include thickness, density, and atomic number. Object contrast is a fundamental limitation in mammography, because there is very little difference among the physical characteristics of different types of breast tissue. Adipose and glandular tissues have very similar densities and molecular compositions; glandular tissue and carcinoma are even more similar, as shown in Figure 1-4.\(^4\) Even calcifications, which have a markedly different density and composition from breast tissue, suffer from poor object contrast because they are so small that their projection onto a del suffers from in-plane partial volume effect.

Subject contrast is the difference in x-ray intensities transmitted by different objects. It is affected by both the object contrast and the beam energy used, because the attenuation coefficients of the objects being imaged generally decrease with photon energy. The need to maximize subject contrast is the primary reason why the required mammography beam energies are so low compared to those in other subdisciplines of diagnostic radiology, because the lower the beam energy, the more the attenuation coefficients of the different breast tissues differ (Figure 1-4). The 25 kVp–35 kVp energy range typically used in mammography is a compromise between maximizing subject contrast and retaining adequate signal to keep exposure times and the absorbed dose
to the breast reasonably low. Filtration also influences the average beam energy by removing low-energy photons and selectively allowing higher-energy photons to pass, depending on the relationship between the photon energy and the filter K-edge. Using less filtration (with the cost of an increase in dose) or a filter with a lower K-edge (at the cost of both an increase in dose and exposure time) will also improve subject contrast.

The fidelity with which the subject contrast is transmitted to the final image depends next on the detector. For instance, if a detector is not equally sensitive to all photon energies in the clinical range, the subject contrast is not faithfully captured. Once the signal is read out from the detector, image processing may affect image contrast by accentuating or suppressing different areas of the histogram. The display settings, such as the LUT and display window and level used, and the abilities of the computer monitor and graphics card also determine contrast visibility.

Like noise, however, contrast is not really the property of interest; instead, the contrast-to-noise ratio (CNR) is a much better indicator of whether an object is visible, and for much the same reason explained previously. If the image background has an average pixel value of 10 and an imaged object has an average pixel value of 15, the absolute contrast of that imaged object is 5. However, the visibility of a difference of 5 pixel values depends on the noise characteristics of the image. If the standard deviation of the image background is 1, then the CNR = 5, and the object should be clearly visible in the image. However, if the standard deviation of the background is 100, then the CNR = 0.05 and it is less visible. The absolute visibility of an object depends both on the CNR and on the display window and level.
The CNR of an imaged object can be improved either by increasing the image contrast or by decreasing the amount of noise. The choice of which option to pursue in a clinical situation depends on the primary reason that the CNR is poor. For example, consider the result of using an inappropriate tube potential. Use of an inappropriately high tube potential causes poor contrast, and therefore a poor CNR. The solution to the poor CNR is obviously to lower the tube potential. However, use of an inappropriately low tube potential at constant tube current and time also results in a poor CNR because poor penetration of the breast results in excessive noise. In this case, the solution is to raise the tube potential in order to lower the noise.

**Spatial Resolution and the Modulation Transfer Function**

The third major component of image quality is spatial resolution, which is an indication of how well a system is able to image small objects. Another way to consider spatial resolution is that it is related to the amount of blur the imaging system adds to the image, because a system with poor spatial resolution blurs an image more than a system with good spatial resolution. In a digital mammography system, the causes of blurring can be broken into analog and digital components. The analog causes of blur include the finite focal spot size and the collection properties of the detector. (Radiation scatter, which is also an analog process, is typically excluded from discussions of system spatial resolution, as it is related specifically to the object being imaged rather than the imaging system.) Digital causes of blur include the finite pixel size and image processing effects.

In digital mammography, the system spatial resolution is primarily limited by the pixel size. This effect is described by the Nyquist theorem, which states that the maximum spatial frequency ($f_N$) that can be resolved is:
\[ f_N = \frac{1}{2x}, \quad (1-6) \]

where \( x \) is the sampling distance. In the case of a digital detector, the sampling distance is equal to the pixel size. The common example used to illustrate the Nyquist theorem is the case of a line pair phantom. Thus for a line pair phantom composed of alternating black and white lines, the Nyquist theorem states the fairly obvious conclusion that two pixels are needed to represent one line pair, since one must be black and one must be white. Other effects can certainly degrade the spatial resolution further, but it can never be better than that specified by the Nyquist theorem.

There are many ways to assess the spatial resolution of an imaging system. One of the most intuitive is the point spread function (PSF), which describes the system response to an impulse (Figure 1-11). No imaging system is capable of producing a perfect response to an infinitesimal impulse; there is always some blurring from the effects described above. The width of the PSF, typically measured at one-half or one-tenth of the maximum value, is a common measure of spatial resolution. However, the PSF is not a very practical way to measure the spatial resolution of a digital system, first, because it is very difficult to produce anything resembling an impulse from an x-ray tube, and second, because the response to an impulse is likely to be small in relation to the pixel size. Thus, the most likely result from the application of an impulse to a digital imaging system is just to receive the signal from a single pixel, which does not provide much useful information about the shape of the spread function (Figure 1-11). It should be noted that the PSF can still be used to describe phenomena which produce signal spread on a scale much larger than the pixel size; in fact, the PSF is used in this research to describe the spread caused by radiation scatter. However, the theoretical
definition of the system spatial resolution as the PSF in response to a true infinitesimal
impulse does not work well with digital systems.

A more practical way to measure the spatial resolution of a digital system is with
the modulation transfer function (MTF). Unlike the PSF, which is a measure of blur in
the spatial domain, the MTF describes spatial resolution in the frequency domain. The
MTF is often measured using a line-pair phantom, which has lead lines and spaces at a
range of spatial frequencies (Figure 1-12). The MTF is a graph of the modulation of a
signal (i.e., the fraction of the contrast that is transferred) as a function of frequency.
When the line pairs are large in relation to the spatial resolution, the system is able to
record the signal modulation between the bars accurately (Figure 1-13), and the
modulation approaches one (or 100%). In fact, the MTF is normalized to be equal to
one at zero frequency. As the line pairs decrease in width as frequency increases,
however, blur from lines starts to overlap with the signals of their neighboring spaces,
reducing the modulation. Eventually, the system is unable to distinguish the lines and
the spaces at all, and the modulation is reduced to zero.

Besides being fairly easy to measure, the MTF is a very convenient way to
compare different imaging systems (or components of imaging systems). The lower the
frequency at which the modulation falls to zero, the worse the system spatial resolution.
However, being able to compare the response across a frequency range of interest is
often more important than knowing the frequency at which the modulation falls to zero.
For example, no digital detector comes close to the limiting spatial resolution of screen-
film cassettes, which can exceed resolutions of 20 lp/mm. However, at frequencies
less than the limiting Nyquist frequency, some digital detectors (including the Hologic™
Selenia® system used in this study) have MTF’s higher than that of screen-film. Thus, screen-film systems excel at displaying very small objects, such as the tiniest microcalcifications, but some digital detectors are superior in the mid-frequency range, which includes the calcifications which are most commonly seen and most likely to be malignant (Figure 1-14).

**Physics of Photon Scatter**

In an ideal imaging situation, the only physical process resulting in the attenuation of a photon would be the photoelectric effect. Each ray of an x-ray beam would pass through the patient in a straight line originating from an infinitesimal focal spot, and the distribution of x-ray intensities exiting the patient would then perfectly represent the two-dimensional projection of the object contrast. Unfortunately, however, the photoelectric effect is not the only x-ray interaction that occurs in the patient. Scattering of the x rays is a major contribution to the overall interaction cross-section in the diagnostic energy range, actually surpassing the photoelectric effect at energies above 26 keV in soft tissue. Scattering degrades image quality because the scattered photons undergo, not only a change in energy, but also a change in angle. Thus, a scattered photon is likely to emerge from the patient in the “wrong” location (i.e., not along the straight line representing the ideal ray). When a large number of scattered photons is detected, the resulting image suffers from a fog that reduces contrast and increases noise. Barrett and Swindell assert that "scatter is often the most important factor limiting image quality and diagnostic accuracy".

In the diagnostic energy range, there are two photon interactions which result in scattering: coherent (Rayleigh) scattering, and Compton scattering. Compton scattering is important at all energy ranges used in diagnostic radiology, whereas coherent
scattering is important primarily at low energies (the coherent scattering cross-section in soft tissue is less than 10% of the Compton scattering cross-section at photon energies greater than approximately 40 keV).\textsuperscript{4} Coherent scattering has often been ignored in the literature for this reason, so that most research investigating scatter focuses on Compton scattering. In the mammography energy range, however, coherent scattering comprises a significant portion of the total scatter production (Figure 1-15).\textsuperscript{4}

**Coherent Scattering**

Coherent scattering occurs when a photon interacts with an atom as a whole. Thus, the probability of coherent scattering is dependent on the atomic number of the material,\textsuperscript{23} since larger atomic numbers generally indicate larger atoms, which present larger targets. The coherent scattering cross-section $\sigma_R$ is described by the equation:

$$
\sigma_R = \pi r_e^2 \int_0^{\pi} \sin \theta (1 + \cos^2 \theta)[F(x, Z)]^2 d\theta \tag{1-7}
$$

where $r_e$ is the classical electron radius ($2.818 \times 10^{-15}$ m), $\theta$ is the scattering angle, and $F(x, Z)$ is the form factor, which is a function of the atomic number ($Z$) and the momentum transfer variable, $x$ (where $x = \sin (\theta / 2)/\lambda$, and $\lambda$ is the photon wavelength).\textsuperscript{24}

The coherent interaction is considered elastic because the photon does not lose energy in the interaction, and the atom moves slightly to conserve momentum. The photon is generally scattered only at small angles (i.e., with little change from its original trajectory). Because no energy is lost, coherent scattering has no effect on patient dose, but it does have some impact on image quality due to the change in photon trajectory, especially with the high-spatial resolution requirements of mammography. The impact of coherent scattering is more pronounced at low energies for two reasons: first, because
the cross-section is inversely proportional to the square of the energy; and second, because the scattering angle is greater at lower energies.\textsuperscript{23}

The average scattering angle for tissue as a function of photon energy can be calculated from Equation 1-7, using the effective atomic number of tissue for \( Z \) and published form factors such as those found in Podgoršak.\textsuperscript{24} The atomic compositions of adipose and glandular breast tissue, as found in ICRU Report 44, are shown in Table 1-1.\textsuperscript{25} The effective atomic number \( \bar{Z} \) is calculated using the equation:\textsuperscript{26}

\[
\bar{Z} = \sqrt[3]{a_1 Z_1^{3.5} + a_2 Z_2^{3.5} + \ldots + a_n Z_n^{3.5}} \ldots
\]  

(1-8)

where \( Z \) is the atomic number and \( a \) is the fractional number of electrons per gram corresponding to each value of \( Z \). Per Equation 1-8, the effective atomic number equals 6.5 for adipose tissue and 7.1 for glandular tissue. Figure 1-16 shows the average scattering angle for \( Z = 6 \) over photon energies ranging from 1-30 keV.

**Compton Scattering**

Compton scattering occurs when a photon interacts with an electron. Thus, the probability of Compton scattering is primarily dependent on the electron density of the material. The incoming photon loses some energy to the electron with which it interacts, which causes the electron to recoil and the photon to change direction. The relationship between the scattering angle and the loss of energy is described by the Compton equation:

\[
\hbar v' = \frac{\hbar v_0}{1 + \frac{\hbar v_0}{m_0 c^2 (1 - \cos \theta)}},
\]

(1-9)

where \( \hbar v_0 \) is the initial energy of the photon, \( \hbar v' \) is its final energy, \( \theta \) is the scattering angle, and \( m_0 c^2 \) is the rest mass energy of an electron (0.511 MeV). In the diagnostic
energy range (less than 150 keV), the distribution of scattering angles is essentially isotropic, which is certainly true in the mammography energy range.\textsuperscript{22}

The probability that a photon will undergo Compton scattering with an electron is described by the collision cross-section, $\sigma_C$. This cross section is calculated using the Klein-Nishina equation:

$$
\sigma_C = 2\pi r_e^2 \left\{ \frac{1+\alpha}{\alpha^2} \left[ \frac{2(1+\alpha)}{1+2\alpha} - \frac{\ln(1+2\alpha)}{\alpha} \right] + \frac{\ln(1+2\alpha)}{2\alpha} - \frac{1+3\alpha}{(1+2\alpha)^2} \right\},
$$

(1-10)

where $\alpha = h\nu / m_0 c^2$ and $r_e$ is the classical electron radius, equal to $2.818 \times 10^{-15}$ m.\textsuperscript{23} Once the cross-section is known, the Compton linear attenuation coefficient, $\mu_C$, can be calculated as follows:

$$
\mu_C = \sigma_C n_e,
$$

(1-11)

where $n_e$ is the electron density of the material.\textsuperscript{27} This value describes the rate of removal of photons from an x-ray beam due to Compton scattering. The fraction of photons ($N / N_0$) undergoing attenuation due to Compton scattering in a given thickness of material, $t$, is given by:

$$
\frac{N}{N_0} = e^{-\mu_C t}.
$$

(1-12)

It should be noted that the fraction of photons undergoing a Compton interaction is not equal to the fraction of incident photon energy lost to Compton interactions, due to the fact that the energy is shared between the scattered photon and Compton electron. Energy transferred to the electrons is locally deposited, whereas the energy retained by the scattered photon is “lost” from the immediate area. The energy-transfer cross-section $\sigma_{tr}$ describes the fraction of energy deposited locally by Compton electrons and is related to the collision cross section by the equation:
\[ \sigma_{tr} = \sigma_c \frac{T}{h \nu}, \]  

(1-13)

where \( T \) is the mean kinetic energy of the Compton electron.\(^{23}\) The quantity \( \sigma_{tr} \) is useful for predicting the contribution of Compton scattering to dose, because it describes the energy transferred to the tissue in which the x ray interacts. Compton scattering is nearly elastic in the diagnostic range of energies, and particularly so in the low-energy mammography range, so \( \sigma_{tr} \) is small, and there is essentially no direct impact on patient dose (although the indirect impact caused by the use of scatter rejection grids can be quite large). Thus, Compton scattering is primarily a concern for image quality.

The collision cross-section (\( \sigma_c \)) describes the total probability that a photon will undergo Compton scattering, and is thus useful for predicting the fraction of photons likely to contribute to image degradation by appearing in the wrong place in the image.

**Distribution of Scatter**

The presence of scatter in an image is sometimes described as a constant or "DC offset" to the primary image, as in the case of a uniform fog; however, this description is false. If scatter were a DC phenomenon, then subtraction of this DC term from the image could be accomplished simply via display windowing; however, simple subtraction does not fully compensate for scatter effects.\(^{28}\) Research by Boone et al. indicated that scatter distribution is a very low frequency, but not DC, phenomenon.\(^{28}\) Because scatter distribution does have a frequency component, a frequency-sensitive correction method such as deconvolution (or division in frequency space) is the appropriate technique for image processing-based attempts at scatter removal. Deconvolution is discussed in more detail later in this chapter.
Impact of Scatter on Image Quality

As stated previously, scatter produces a “fog” of photons in the image which do not provide useful information. These photons have a negative effect both on noise and on contrast. The total noise level in the image is higher because the total number of detected photons is higher, and \( \sigma = \sqrt{N} \). Although more detected photons typically corresponds to a higher SNR, scatter actually degrades the SNR because it essentially contributes a false signal (i.e., because it is in the wrong location). Thus, noise increases, but the true signal does not.

Contrast suffers due to the contribution of photons from outside the object of interest (Figure 1-17). The relationship between contrast and scatter is described by the equation:

\[
C_s = \frac{C_0}{1 + S/P},
\]

where \( C_s \) is the contrast with scatter present, \( C_0 \) is the contrast without scatter present, \( P \) is the amount of primary (non-scattered) radiation detected, and \( S \) is the amount of scattered radiation detected. Without the use of an antiscatter grid, it is typical to lose 30-50% of the subject contrast in mammography due to scatter.\(^4\) This loss of contrast is highly dependent on breast thickness, since the ratio of scattered to primary radiation in an image increases quickly with increasing breast thickness, more than doubling as breast thickness increases from 3 cm to 6 cm.\(^{29}\) The image contrast with a grid can be improved from this level by up to 40%, but at the cost of 2-3 times the breast dose.\(^4\)

Spatial resolution is also theoretically degraded by scatter. Spatial resolution is often described as a measure of the amount of blur in a system, and scatter certainly has the result of blurring an image. The point spread function (PSF) is defined as the
response of the system to a point stimulus, which can be represented by a very narrow pencil beam of radiation. This pencil beam necessarily interacts with any object it hits, producing scatter. More scatter is associated with more spread of the PSF. However, for the Hologic™ Selenia® system used in this project, testing under conditions which include a scattering medium indicates that the limiting spatial resolution is already at the level predicted by the Nyquist frequency of the detector dels. Thus, correcting for scatter in this system is unlikely to improve the limiting spatial resolution noticeably, although it may improve the MTF at lower frequencies.

Fundamentals of Image Processing

A digital image is essentially a two-dimensional array of numbers, with each number representing a pixel in the final image. Each number is mapped to a gray level for display purposes. Image processing is the manipulation of these numbers for the purpose of altering the image appearance. In medical images, some of the most common applications of image processing include image smoothing and edge enhancement. Unfortunately, these processes (and all forms of image manipulation) have negative attributes. Image smoothing reduces the appearance of noise but increases blur. Edge enhancement essentially does the opposite, by reducing blur but worsening the appearance of noise. Both of these processes require an understanding of convolution.

Convolution

Convolution of an image for the purpose of image processing first requires a kernel, which is an array of numbers (typically much smaller than the original image) used to manipulate the image. The central pixel of the kernel is placed over a pixel of interest in the image, so that the remainder of the pixels in the kernel correspond one-
to-one with the pixels surrounding the pixel of interest in the image (Figure 1-18). Thus, for the example of a 3x3 kernel, the eight pixels surrounding the pixel of interest have corresponding pixels in the kernel associated with them.

To convolve the kernel with the image, each pixel in the kernel is multiplied with its corresponding pixel in the image. These products are then added together, sometimes normalized, and assigned to the pixel of interest. Thus, in the case of a 3x3 kernel, nine pixels contribute to the resulting value of the pixel of interest. The values used in the kernel depend on the effect desired by the convolution: in general, positive values blur the image, whereas negative values sharpen it. The process of convolution is written mathematically as:

$$i_f = i_i * k,$$  \hspace{1cm} (1-15)

where $i_f$ denotes the final image, $i_i$ is the initial image, and $k$ is the kernel.

The process explained above describes convolution in the spatial domain. With large images, however, it can be very slow to perform spatial convolution due to the number of computations involved. Thus, convolutions are often performed in frequency space. To perform a convolution in frequency space, first the kernel must be padded with zeros so that it is the same size matrix as the image. Then, the two-dimensional Fourier transforms of the image and the kernel are found. The element-by-element product of the transformed kernel and image is calculated. Lastly, the inverse Fourier transform is applied to the result, giving the final image. The fast Fourier transform (FFT) process is much faster than the convolution process. This method is equivalent to spatial convolution because the convolution theorem states that convolution is equivalent to multiplication in frequency space:
\[ i_f * k = \mathcal{F}^{-1}\left[\mathcal{F}(i_f) \cdot \mathcal{F}(k)\right], \quad (1-16) \]

where \( \mathcal{F} \) denotes the Fourier transform and \( \mathcal{F}^{-1} \) denotes the inverse Fourier transform.

For the remainder of this discussion, the convention of employing lower case letters to denote the spatial domain (i.e., \( k \)) and upper case letters to denote the frequency domain (i.e., \( K \)), will be used. Thus, Equation 1-16 can be rewritten:

\[ i_f * k = \mathcal{F}^{-1}\left[I_f \cdot K\right] \quad (1-17) \]

**Deconvolution and the Wiener Filter**

Deconvolution is simply the inverse operation to convolution. Assuming an image has already been "degraded" by convolution with a kernel (i.e., in the case of scatter), and assuming the kernel is known, then deconvolution is just the process of "undoing" the convolution to restore the "true" image. Unfortunately, however, this process is easier said than done.

In an ideal case, deconvolution can be performed in frequency space simply by dividing the degraded image, \( I_d \), element-by-element by the kernel, to yield the true image, \( I_t \):

\[ I_t = I_d / K. \quad (1-18) \]

The problem with this method is that, as mentioned previously, the kernel is generally much smaller than the original image. It is padded by zeros to make the kernel matrix the same size as the image, but no true data exist in those pixels. Even if the kernel was calculated out to the size of the image, so that there were no zero values in the matrix, the values far from the center would almost always be extremely small. Thus, the use of Equation 1-18 would result in a large number of pixels at the periphery of the transformed image being divided by extremely small numbers (or zero), which creates
extremely large and error-prone numbers in the result. Because noise is a high-frequency phenomenon and thus is represented in the periphery of the transformed image, the noise becomes greatly amplified. This effect may be so severe that the noise dominates, completely obscuring the rest of the image.  

There are several approaches which can improve the results of the deconvolution. The simplest are called direct inverse filtering, which involve methods such as low-pass filtering and constrained division (where the division is not performed if the value of the kernel in that pixel is less than some threshold value). Ideally, these methods would remove both noise and blur. However, in practice, they are undesirable for use with noisy images, because noise is generally a broadband phenomenon, so the frequencies of noise and image detail overlap. In the case of a low-pass filter, it is impossible to remove the noise without losing detail. Constrained division prevents the amplification of most of the noise, but it also prevents the kernel from operating on the high-frequency information in the image.

Another solution for images that are inherently noisy (such as most medical images) is the Wiener filter. The Wiener filter assumes that the degraded image can be expressed in frequency space as:

\[ I_d = I_t \cdot K + N, \]  

(1-19)

where \( N \) represents the contribution of additive noise. The Wiener filter then estimates the true image by minimizing the mean square error between an estimated image and the true image. Of course, the true image is rarely known, so the way the filter actually works is by the equation:

\[ i_t(x, y) = i_d(x, y) - \left( \frac{\sigma_N^2}{\sigma_L^2} \right) [i_d(x, y) - m_L], \]  

(1-20)
where \( i_t(x, y) \) is the estimate of the true image at pixel location \((x, y)\), \( i_d(x, y) \) is the degraded image at the same location, \( \sigma_N^2 \) is the variance of the total image noise, \( \sigma_L^2 \) is the variance of pixels in a local neighborhood (defined by the user), and \( m_L \) is the mean pixel value within that neighborhood.\(^{31, 32}\) If the local variance is high, there is likely an edge or boundary of some sort in the local neighborhood, and the output of Equation 1-20 remains close to \( i_d(x, y) \) to preserve such information. If the local variance is low, the output is near the mean local value, which reduces noise.

The Wiener filter is most often employed in the frequency rather than the spatial domain, however. Theoretically, the equation which should be used to generate the estimated image is:

\[
\hat{I}_t(u, v) = \frac{1}{K(u, v)} \cdot \frac{|K(u,v)|^2}{|K(u,v)|^2 + (|N(u,v)|^2/|F(u,v)|^2)} \cdot I_d(u, v),
\]

(1-21)

where \((u, v)\) represents a location in frequency space, \( N \) is the noise power spectrum of the noise, and \( F \) is the noise power spectrum of the true image. However, the noise power spectra \( N \) and \( F \) are very rarely known. Luckily, the approximation

\[
\hat{I}_t(u, v) \approx \frac{1}{K(u, v)} \cdot \frac{|K(u,v)|^2}{|K(u,v)|^2 + C} \cdot I_d(u, v)
\]

(1-22)

often works extremely well.\(^{31}\) The value \( C \) is a constant chosen to balance sharpness and noise in the final image: smaller values of \( C \) result in more noise, whereas larger values of \( C \) result in more smoothing.\(^{30}\)

Wiener filtering is challenged when there are large signal variations throughout the image. Some areas contain more detail than other areas and may require different values of \( C \). A change in the noise content from one area of the image to another can also pose a problem, because the use of Equation 1-22 assumes that a single signal-to-noise ratio is applicable throughout the image.\(^{31}\) This limitation is particularly relevant to
mammography images, since a significant portion of the detector is often exposed to two major compartments: the breast and the unattenuated beam. A large variation in signal and noise exists between these two compartments. However, Dougherty states that if the kernel is "relatively small compared to the size of the image", then separate values of $C$ can be used for the various regions of the image. Since there are only two main areas in a mammography image that differ widely, the use of Wiener filtering should be a reasonable approach for the deconvolution of mammography images.

**The Problem of Ringing**

One other significant problem encountered in image processing is ringing, which is the presence of oscillations in the image caused by the use of the Fourier transform (Figure 1-19). These oscillations result from the inability of the Fourier transform to reproduce sharp edges for a finite frequency range. In the Fourier transform process the left side of the image is contiguous with the right side and the top is contiguous with the bottom. A rapid change in signal value from left side to right side or from top to bottom causes ringing along both edges (Figure 1-20). Edge ringing can be particularly severe in mammography because the difference in pixel values between the chest wall (low signal) on one side of the image and the air (high signal) on the other side can often be quite large (more than an order of magnitude is typically seen on the Hologic™ Selenia® system). This ringing is easily managed using artificial ramps leading from the large pixel values on one side of the image to the small pixel values on the other side. The Fourier transform is much better able to mimic this gradual change than it is the steep jump from one side to the other, and the number of pixels in the ramp can be adjusted as needed to reduce ringing to negligible levels. The ramp is simply removed from the final image once it has been transformed back into the spatial domain.
Similarly, a sharp change from low signal along the periphery of the breast to air at the anterior skin line causes ringing, but this type of ringing is not as easily managed. Image smoothing reduces the appearance of ringing, but it blurs the image, as well. The value of $C$ must be optimized to find the best compromise between noise reduction, ringing, and detail preservation (when using the Wiener filter for deconvolution). This optimization is performed by trial and error.

**Summary and Conclusion**

All of the above-described topics are important in this research project. This research models the Hologic™ Selenia® mammography unit, both in the x-ray spectra and in the geometry and hardware (e.g., grid and detector) of the equipment. These spectra interact with different thicknesses and sizes of breast-simulating materials, both in a physical model and in a calculational model. Data from these interactions are simulated physically and through Monte Carlo calculations to describe the scatter and primary radiation, and to create detected image data sets. The image processing techniques described above are used in this research project to accomplish the goal of improving mammography images by removing scatter from the image data sets, as described in Chapter 2.
Table 1-1. Elemental Composition and Density of Breast Tissues, according to ICRU Report 44.

<table>
<thead>
<tr>
<th>Material</th>
<th>Elemental composition</th>
<th>Density (g/cm$^3$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adipose tissue</td>
<td>H: 11.4%, C: 59.8%, N: 0.7%, O: 27.8%, Na: 0.1%, S: 0.1%, Cl: 0.1%</td>
<td>0.950</td>
</tr>
<tr>
<td>Glandular tissue</td>
<td>H: 10.6%, C: 33.2%, N: 3.0%, O: 52.7%, Na: 0.1%, P: 0.1%, S: 0.2%, Cl: 0.1%</td>
<td>1.020</td>
</tr>
</tbody>
</table>

[Adapted from *ICRU Report 44: Tissue Substitutes in Radiation Dosimetry and Measurement*. (International Commission on Radiation Units and Measurements, Bethesda, MD, 1989).]

Figure 1-1. Configuration of a typical digital mammography unit.
Figure 1-2. Geometry of general radiography equipment (left) and mammography equipment (right).

Figure 1-3. A linear grid (left) and a cellular grid (right)
Figure 1-4. Linear attenuation coefficients of breast tissues. [Adapted from J. T. Bushberg, J. A. Seibert, E. M. Leidholdt Jr. and J. M. Boone, *The Essential Physics of Medical Imaging*, 2nd ed. (Lippincott Williams & Wilkins, Philadelphia, 2002).]
Figure 1-5. Representative spectra for a 30 kVp beam with target/filter combinations of A) Mo/Mo and Mo/Rh and B) W/Rh and W/Ag. The scale on both axes has been kept the same to permit ease in comparison.
Figure 1-6. Cross-section of a direct digital detector

Figure 1-7. Interaction of x-ray in selenium produces an electron
Figure 1-8. Simplified schematic of detector wiring for four dels when viewing the detector from above. The actual image matrix contains 14,008,320 dels.

Figure 1-9. A mammography image (left) and corresponding histogram (right). The peak for the black background actually extends to over 5 million, but the graph has been truncated to better show the details of the rest of the histogram. The image markers indicated in the histogram correspond to the white letters indicating the view and technologist initials.
Figure 1-10. Example of a 2-dimensional NPS (left) and corresponding 1-dimensional NPS (right). The 1-D NPS is simply a cross-section from the 2-D NPS, indicated by the box. The cross-section is typically taken over only half the width of the 2-D NPS because it is symmetric. The bright white axes seen in the 2-D NPS are considered artifacts; the cross-section is taken from an area near but not exactly on the axes.

Figure 1-11. Response of an imaging system to an impulse. An analog detector generates a traditional point spread function. However, the “all or nothing” activation of a pixel in a digital signal precludes the use of the PSF to describe the response to an infinitesimal impulse.
Figure 1-12. A line pair phantom. Each frequency group consists of lead lines and gaps with the frequency numbered on the right (specified in line pairs per mm).

Figure 1-13. Signal modulation (red arrow) in response to a line-pair phantom. The phantom on the left has a low frequency, and the system is able to respond with a modulation of 1. The phantom on the right has a higher frequency, and blur limits the modulation to about 0.5.
Figure 1-14. Number of microcalcifications found as a function of malignancy and size. The size reported is the length of the microcalcification measured along its largest dimension, averaged over the smallest and largest microcalcifications in the cluster. The coefficient of variance is 2.7% for both groups. The malignant microcalcifications have an average length 37.9% greater than the benign microcalcifications. [Adapted from Shalom S. Buchbinder, et al., "Can the size of microcalcifications predict malignancy of clusters at mammography?," Academic Radiology 9, page 23, Figure 2. (2002).]
Figure 1-15. Cross-sections of coherent scattering, Compton scattering, and the photoelectric effect in water, for the mammography range of energies. [Adapted from J. T. Bushberg, J. A. Seibert, E. M. Leidholdt Jr. and J. M. Boone, *The Essential Physics of Medical Imaging*, 2nd ed. (Lippincott Williams & Wilkins, Philadelphia, 2002).]
Figure 1-16. Average scattering angle as a function of photon energy for coherent scatter, as calculated from Equation 1-7.

Figure 1-17. Scatter degrades contrast. If only primary radiation (thin black arrows) contributed to the image, a small dense object such as a lead disk (represented by the darker gray square) would block more of the primary rays than the less-dense (lighter gray) background, and contrast would be determined only by differences in attenuation. However, radiation scatters in the background material, and some of it is detected under the lead disk (thick red arrows), which degrades the contrast, because more signal is detected under the disk than expected.
Figure 1-18. Representation of a 3x3 kernel overlaid on an image. The central pixel of the kernel (dark red) corresponds to the pixel of interest in the image (dark gray). The remaining pixels of the kernel (red) correspond one-to-one with the pixels surrounding the pixel of interest in the image (gray).

Figure 1-19. An example of severe ringing in a phantom image.
Figure 1-20. The Fourier transform assumes that the image is periodic, thus essentially wrapping the image around from right-to-left (and also top-to-bottom, which is not depicted). Sharp edges both within the image and created by the image wrapping are likely to cause ringing after transformation. In some images, particularly MLO views, edges created by wrapping occur in both directions.
Essential Background and Definitions

The goal of this study is to develop a method to remove scattered radiation from a mammography image using a spatially-variant Wiener filter. This scatter correction program relies on the development of a scatter kernel, which is a point spread function (PSF) describing the way in which scatter spreads from an impulse. The scatter kernel is described by two variables: the scatter fraction (SF) and the mean radial extent (MRE).

The SF is equal to the amount of scattered radiation detected, $S$, divided by the total amount of radiation detected ($S$ plus primary radiation, $P$):

$$SF = \frac{S}{S+P}$$  \hspace{1cm} (2-1)

The MRE is the mean distance from the impulse at which scattered radiation is detected.

It should be noted that both of these quantities are dependent on the radiation actually detected, and thus are not fundamental physical quantities in the sense that they could be calculated from basic principles. In order to be detected, scatter generated in the phantom must travel in the direction of the detector, exit the phantom, pass through the grid (if present), enter the detector, and produce a signal. A large portion of the scatter actually produced in the phantom is never detected and does not contribute to the above quantities (or the image).

The selection of the SF and MRE are dependent on the clinical techniques and the apparent thickness of the breast on a pixel-by-pixel basis. The apparent thickness is
a quantity found by calculating the ratio of the value of each pixel in the breast to the mean value of a region of interest (ROI) placed in a region of the detector exposed to the unattenuated beam. The ratio is compared to tables that are tied to the clinical technique and generated using breast-equivalent phantoms in a range of thicknesses. The apparent thickness is equal to the phantom thickness which would produce an equal pixel-value ratio under the same clinical techniques. More details about the procedures used to determine the SF, MRE, and apparent thickness are found in Chapter 4.

**Primary Goal**

The primary goal of this study is to determine whether scatter correction using spatially-variant, thickness-dependent filtration can improve the image quality of clinical mammograms. The basic plan to assess this goal consists of the following steps:

1. Measure the SF and MRE of scatter propagation for all common target/filter combinations, over the range of clinically used tube potentials and breast thicknesses. An antiscatter grid is used for these measurements to simulate current mammography clinical practice.

2. Measure the attenuation of different thicknesses of BR12 (a commonly-used phantom that simulates breast tissue) to create a table of apparent thicknesses for all common target/filter combinations, over the range of clinically used tube potentials.

3. Assess the accuracy of the scatter fraction and mean radial extent measurements by performing Monte Carlo simulations. These simulations are not intended to replace the physical measurements, but they serve only to ensure that the physical measurements are reasonable.

4. Develop a Monte Carlo simulation to determine the likely effect of breast glandularity on the measurements in Steps 1 and 2. BR12 simulates a mix of 50% glandular and 50% adipose tissue, but breast composition can vary from mostly adipose to mostly glandular.

5. Measure the modulation transfer function (MTF), contrast-to-noise ratio (CNR), and noise power spectrum (NPS) of the imaging system, using all target/filter combinations.
combinations and a range of tube potentials. A 4-cm scattering medium is used to simulate realistic scatter conditions.

6. Write a program to perform spatially-variant filtration, using the scatter measurements obtained in Step 1, the apparent thickness measurements obtained in Step 2, and the scatter point spread function described in Equation 3-5.

7. Process the MTF, CNR, and NPS images obtained in Step 5 with the scatter correction program. Determine the effect of the program on image quality measurements.

8. Obtain IRB approval to remove patient images from the mammography systems. Process these images with the scatter correction program.

9. Have radiologists review the images before and after processing. Obtain their professional opinions regarding image quality and artifacts, with the goal of determining whether the scatter correction program improves clinical image quality.

**Secondary Goal**

The secondary goal of this study is to determine whether the scatter correction program corrects scatter sufficiently well to retain adequate image quality while allowing removal of the antiscatter grid. The motivation in this case would be to reduce patient dose rather than to improve image quality beyond what is currently available. To assess this possibility, Steps 1 to 7 are repeated as described above, only with the removal of the grid. Steps 8 and 9 will not be performed for this portion of the study, because FDA regulations require the use of a grid for the acquisition of non-magnification clinical images. The use of magnification images is not proposed for this portion of the study because it is not possible to obtain apparent thickness measurements from these images. This issue is discussed in more detail in Chapter 4.
Hypotheses

Relating to the Scatter Fraction

Hypothesis 1-1a. The SF is independent of the tube potential in the mammography range of 24 kV-34 kV.

Hypothesis 1-1b. The SF is independent of the target/filter combination for Mo/Mo, Mo/Rh, W/Rh, and W/Ag.

Hypothesis 1-1c. The SF is higher without an antiscatter grid than with a grid.

Hypothesis 1-2a. The SF increases with breast thickness.

Hypothesis 1-2b. The SF is unrelated to the breast glandularity for a fixed apparent thickness.

Hypothesis 1-2c. The SF is unrelated to the compressed breast area.

Relating to the Mean Radial Extent

Hypothesis 2-1a. The MRE increases as the tube potential increases in the mammography range of 24 kV-34 kV.

Hypothesis 2-1b. The MRE increases as the average beam energy increases due to the target/filter combination selected (in order from least to greatest: Mo/Mo, Mo/Rh, W/Rh, W/Ag).

Hypothesis 2-1c. The MRE is greater without an antiscatter grid than with a grid.

Hypothesis 2-2a. The MRE increases with breast thickness.

Hypothesis 2-2b. The MRE is unrelated to the breast glandularity for a fixed apparent thickness.

Hypothesis 2-2c. The MRE is unrelated to the compressed breast area.
Relating to the Apparent Thickness and Tissue Type

Hypothesis 3-1. The difference in apparent thickness between glandular tissue and adipose tissue is insufficient to affect the scatter correction algorithm.

Relating to the MTF

Hypothesis 4-1. The MTF is improved by the scatter correction program at low frequencies.

Hypothesis 4-2. The scatter correction program does not change the limiting spatial resolution, which is determined by the Nyquist frequency.

Hypothesis 4-3. The scatter correction program improves the MTF without a grid to become similar to the MTF with a grid.

Relating to the CNR

Hypothesis 5-1. The CNR is improved by the scatter correction program.

Hypothesis 5-2. The scatter correction program improves the CNR without a grid to become similar to the CNR with a grid.

Relating to the NPS

Hypothesis 6-1. The scatter correction program lowers the stochastic noise content of the images.

Hypothesis 6-2. The scatter correction program improves the NPS without a grid to become similar to the NPS with a grid.

Relating to Clinical Image Quality

Hypothesis 7-1. The radiologists prefer the scatter-corrected images to the uncorrected images.

Rationale

Hypotheses 1-1a and 1-1b are related, because both an increase in tube potential and a harder beam result in a higher average beam energy (Figure 1-5), which results in photons that are more penetrating. The SF is expected to be independent of
beam energy because the total scatter cross-section changes little (from about 0.38 to
0.23 cm⁻¹) over the mammography range.⁴ The sharp decrease in the photoelectric
cross-section with energy affects both primary and secondary radiation similarly,
because Compton scattering has only a small effect on the energy spectrum in the
mammography range and coherent scattering has no effect; thus, the ratio of scatter to
primary radiation remains essentially unchanged over the mammography energy range.
As for Hypothesis 1-1c, the SF is higher without a grid than with a grid because the grid
absorbs scatter preferentially to primary radiation. The SF is expected to increase with
breast thickness (Hypothesis 1-2a) simply because the probability of an interaction
increases with thickness; thus, both the attenuation of the primary beam and the extra
opportunities for scatter production contribute to the increased SF. Hypothesis 1-2b is
expected because there is very little difference between the physical characteristics of
adipose and glandular tissue. The difference in the Compton scattering cross-section
between two materials depends on their electron densities, which only differ by about
6% between pure adipose and pure glandular tissue,³⁵ calculated using the values in
Table 5-2 (and breasts are some combination of the two). Differences in the coherent
scattering cross-sections depend on the atomic numbers of the materials in question,
which are also very similar between fat and glandular tissue (the effective atomic
number is 6.5 for adipose tissue and 7.1 for glandular tissue). The differences in
attenuation between the two tissue types are primarily the result of changes to the
photoelectric cross-section, which should have a minimal effect on the SF as explained
above. Hypothesis 1-2c, which states that breast area will not affect the SF, is proposed
because the PSF is defined as the response to an impulse, and the response from each
impulse is, theoretically, independent of the response from neighboring impulses. The total amount of scatter reaching the detector will certainly increase as breast size increases, but that effect is the result of the summation of many impulse responses rather than a change in the SF of a single impulse.

Hypotheses 2-1a and 2-1b are related in the same way as Hypotheses 1-1a and 1-1b. The MRE is expected to increase as the average beam energy increases because photons with more energy are more penetrating. This higher energy allows scattered photons to be more likely to exit the breast and be detected, despite having a longer path length through the breast than primary photons. The use of a grid reduces the MRE (Hypothesis 2-1c) because the photons travelling at angles great enough to intercept the septa are mostly attenuated. Hypothesis 2-2a is explained because as the phantom thickness increases, interactions occurring farther from the detector have more room to spread, which increases the MRE (assuming the scatter kernel is small in relation to the detector size, so that the scatter does not spread beyond the detector). Hypothesis 2-2b is expected for the same reasons as Hypothesis 1-2b. A change in glandularity should result in a change in the photoelectric cross-section, but not in the distribution of the detected scatter. A decrease in the photoelectric cross-section reduces the probability that a photon exits the breast by some percentage, but this reduction would apply equally to all photons exiting the breast regardless of angle, so there should be no change in the MRE. The explanation for Hypothesis 2-2c is the same as that for Hypothesis 1-2c.

The difference in photoelectric attenuation between adipose and glandular tissue will cause the apparent thickness of tissue above a pixel to appear slightly greater as
the amount of glandular tissue above that pixel increases. However, this difference is
not expected to result in a difference in apparent thickness of more than a few
millimeters, and, based on research conducted by Ducote and Molloi, the MRE and SF
are not expected to change that quickly with respect to thickness. Thus, Hypothesis
3-1 states that the tissue glandularity will not interfere with the scatter correction
program.

Hypothesis 4-1 is expected because the scatter PSF is essentially a blurring
function, so removal of the PSF from the image should remove the blur, thus improving
the spatial resolution. However, the limiting spatial resolution observed on clinical units
is typically already at the Nyquist frequency, so the limiting spatial resolution cannot be
improved by the program (Hypothesis 4-2). If the scatter correction program removes
the effects of scatter as expected, then the MTF without a grid should improve at least
to the level of the uncorrected MTF with a grid (Hypothesis 4-3).

Scatter degrades contrast, so removal of the scatter should result in a contrast
improvement, thus improving the CNR per Hypothesis 5-1. If most of the scatter is
removed, then the CNR without a grid should be at least as good as the uncorrected
CNR with a grid (Hypothesis 5-2).

Scatter also increases noise because more photons are detected. As explained
in Chapter 1, these extra photons contribute only to noise and not to the useful signal.
Thus, removal of the scatter contribution should lower the amount of stochastic noise in
the image, as expected per Hypothesis 6-1. Like Hypotheses 4-3 and 5-2, if a large
portion of the scatter is removed by the scatter correction program, the NPS without the
grid should be very similar to that of unprocessed images obtained with a grid (Hypothesis 6-2).

All of the hypotheses relating to image quality predict improvements in image quality. If these hypotheses are true, then it stands to reason that the radiologists will prefer the corrected images to the uncorrected images due to the improvements in contrast, noise, and spatial resolution (Hypothesis 7-1).
Numerous attempts have been made to describe the physics and impact of scattered radiation in mammography systems. One of the first was the work of Barnes and Brezovich in 1978, in which they studied the scatter-to-primary ratio (SPR) in mammography for a range of tube voltages, phantom thicknesses, and field sizes. The phantom they used consisted of Lucite with thicknesses ranging from 3-6 cm (a 1 cm thickness of Lucite is equivalent to a 1.25 cm thickness of breast tissue with a composition of 65% adipose tissue and 35% glandular tissue). The mammography unit used a tungsten target with a 0.5 mm Al filter and did not employ a grid.

Barnes and Brezovich used the beam stop method to measure the SPR, which is an extremely common technique used by a number of researchers. In this method, the imaging system is set up with a phantom on the breast support surface to provide a scattering material. A small lead disk (beam stop) with thickness sufficient to block the primary beam is placed atop the phantom, and an exposure is made. Any signal measured underneath the disk must be the result of scattered radiation. A second exposure is made without the lead disk, which is the sum of primary and scatter radiation. From these two measurements, the scatter-to-primary ratio can be calculated. However, the measured value with the beam stop in place is necessarily dependent on the radius of the disk, because a larger radius means that the scatter has to travel farther to reach the detector. Thus, several disks of different radii are used, and an extrapolation is performed to determine the amount of scatter for a disk radius of zero. This extrapolated value is used in the SPR calculation.
The SPR measurements obtained by Barnes and Brezovich are shown in Table 3-1 and 3-2. They found that the SPR was insensitive to changes in tube voltage in the range they tested (27 kVp-42 kVp), which is slightly higher than the range most commonly used in current clinical practice (24 kVp-34 kVp). They also found that the SPR increased quickly with increasing phantom thickness (changing from about 0.40 at 3 cm to 0.85 at 6 cm).

The field size also had an effect on the SPR, with the effect becoming more pronounced as the phantom thickness increased; however, these field size data do not translate well to current clinical imaging for several reasons. First, Barnes and Brezovich used a circular phantom and a circular radiation field, with the detector in the center of the field. An increase in field size equated to a proportional increase in irradiated phantom volume, with scatter being produced in equal quantities from all directions. Current clinical mammography uses a rectangular field, with the breast at one edge of the field, and the choice of only two field sizes (for screening mammography) which generally extend beyond the breast margins. Switching from the small to the large field of view may irradiate more tissue for a large breast, but certainly not in proportion to the increase in field size. The bulk of the breast is also relatively distant from the newly irradiated area. The second reason why their field size data are not clinically relevant today is that their field sizes were very small, ranging from only 12.6 cm² to 154 cm², compared with the two field sizes in common clinical use (432 cm² and 696 cm²). Their data showed that the SPR increased with increasing field size, but that this rate of increase became progressively smaller as field size increased above 20 cm² (Figure 3-1). Thus, it is unlikely that the effect of field size has a major impact on
SPR in clinical mammography (and in modern mammography, which uses a grid, it is likely to be negligible).

Another early analysis of SPR was performed by Dance and Day using Monte Carlo methods. They simulated breasts as an equal mix by weight of water and fat, with thicknesses from 2 cm to 8 cm. The simulations were modeled both with and without a grid, using monoenergetic spectra ranging from 12.5 keV to 50 keV (Table 3-3). Unlike most other studies of SPR, they calculated SPR over the entire breast shadow rather than as a point measurement. They compared their non-grid results to those of Barnes and Brezovich by interpolation, with results shown in Figure 3-2. Using a wider, monoenergetic energy range, they determined that the SPR does in fact depend on photon energy, but noted that the dependence is very small in the range studied by Barnes and Brezovich (who used a W/Al spectrum with energies ranging from 27 kVp-42 kVp). The SPR without a grid increased rapidly below 20 keV, which the authors attributed to the increasing importance of scattered photons from the compression cone that reached the area of the film in the breast shadow. At lower beam energies, the relative contribution of these x rays (which did not traverse the breast) to those which did traverse the breast became greater due to attenuation effects. The researchers noted that this increase in SPR at low energies is important because the characteristic K x-ray energies of Mo are 17.4 keV and 19.6 keV.

The results of the simulations by Dance and Day with a grid showed even stronger energy dependence than without a grid, with the SPR's showing minima around 25-30 keV but increasing both at lower and higher energies (Figure 3-3). They explained the increase at higher energies to be the result of grid penetration; the grid
was a focused linear grid with lead strips 0.02 mm thick, 1.5 mm high, with a grid ratio of 5:1. The increase at lower energies was due to the same mechanism described above, but was more pronounced because Dance and Day modeled a 12 mm gap between the breast and image receptor with the grid, whereas there was no air gap without the grid. This additional gap allowed more scatter originating from the compression cone to reach the breast shadow. (Note, however, that the overall SPR with a grid was much lower than that without a grid despite the increased energy dependence.)

In 2000, Boone and Cooper performed an extensive series of Monte Carlo simulations to investigate mammography SPR. They tried two different methods: a “direct” simulation in which the geometry used was physically realistic, and a “convolution” method in which the incident x rays were modeled as a δ-function input. They also investigated the effect of breast composition on the SPR at 26 kVp using Mo/Mo; they simulated glandularities of 0%, 50%, and 100% and found very little difference among them (Figure 3-4).

Boone and Cooper compared their results to the backscatter-corrected results reported by Dance and Day (Figure 3-2, dashed lines) and found good agreement (Figure 3-5). They also compared their results using the convolution method with the physical measurements made by Barnes and Brezovich and found the difference to be 6.6% for a 3 cm thickness at 32 kVp and 7.5% for a 6 cm thickness at 32 kVp. The last verification of their results also compared the Monte Carlo-derived SPR's for a variety of thicknesses, breast compositions, and tube potentials to physical measurements published separately by Cooper et al. This comparison demonstrated excellent agreement between the two sets of measurements. Thus, their Monte Carlo calculations
were shown to be reliable, compared both to previous research and to modern physical measurements.

Boone et al. continued their investigations in another publication later in 2000. They limited their work to scatter in mammography without a grid, evaluating the effects of peak tube potential (kVp), spectrum (the two target and filter combinations Mo/Mo and Rh/Rh), breast composition, breast thickness, and field of view (FOV) on the SPR. The kVp dependence ranged from zero over the energy range of 22-40 kVp with a 2 cm-thick phantom, to a maximum of 18% for a 6 cm-thick phantom (Figure 3-6). The dependence on target/filter combination was also minor, with no difference noted using phantom thicknesses of 2 and 4 cm, and a maximum difference of only 3% using a phantom thickness of 8 cm. Predictably, breast thickness had a major impact on SPR (Figure 3-6), and this relationship between SPR and thickness was shown to be nearly linear. Breast composition, on the other hand, had very little effect on SPR. The position within the field of view had a larger impact on SPR (Figure 3-7).

The studies described above all focused on the SPR, but SPR alone does not fully describe the characteristics of scattered radiation. The scatter point spread function (PSF) depends, not only on the amount of scatter produced, but also on the distance it travels. Ducote and Molloi are the only researchers to date who have published research on the scatter fraction (SF) and scatter mean radial extent (MRE) in mammography. They used both physical measurements and Monte Carlo measurements, although not under identical conditions. The physical measurements were acquired with a grid in place (Table 3-4), using a tungsten-target digital mammography unit with a 50 μm rhodium filter, and they cover a range of tube
potentials and breast thicknesses. Their measurements were performed using the beam stop method described earlier. The phantom material they used was polymethyl methacrylate (PMMA), the thicknesses of which are shown in Table 3-4 as equivalent thicknesses of 50% glandular tissue, in order to facilitate comparison with the results of my study. (Because the effective atomic numbers of the two materials are similar, the thickness conversion was performed by multiplying the PMMA thickness by the ratio of the two densities. \(^{29}\))

Their Monte Carlo measurements were simulated without a grid, using 100% adipose tissue and 100% glandular tissue, each with a thickness of 5.1 cm (Table 3-5). Their results show that tube potential had almost no impact on SF but did affect the MRE. The MRE was lower at 28 kVp than at 24 kVp and 31 kVp, indicating that the MRE exhibited a minimum in the mid-range of mammography energies for the W/Rh spectrum. Comparing the results of the physical measurements (with a grid) to the Monte Carlo measurements (without a grid) suggests that the SF and MRE were roughly an order of magnitude larger without a grid than with a grid.

**Scatter Correction**

The publications described in the previous section all focused on producing descriptions of scattered radiation in mammography. Understanding the characteristics of scatter is a necessary first step towards the goal of mitigating its effects. One method of controlling scattered radiation already employed by clinical mammography systems is the use of an antiscatter grid. Another potential method investigated by numerous researchers is the use of image processing, described below.
Development of the Point Spread Function

In order to perform scatter correction via image processing, it is necessary to represent the scatter distribution with a mathematical function that can be used by the processing program. The representation used in all the publications about this type of correction is the point spread function (PSF), and it describes the way in which radiation from an impulse is detected by the imaging system. A number of researchers have attempted to characterize the PSF, leading to several different proposed forms. However, two variables are common to all forms of the PSF: the scatter magnitude (which is related to the scatter fraction) and distance of travel (mean radial extent).

In the following sections, descriptions of previous research about the PSF are divided into two categories: those leading to the form of the PSF used in my study, and those with other forms that are not directly related. An explanation detailing why the form used in my study was selected is discussed in at the end of this chapter.

The form of the PSF used in my study

The PSF used in my study originated in studies of light spread in photographic film. Frieser, one of the earliest researchers whose work ultimately contributed, developed a spread function to describe the response of photographic emulsions to a thin slit light source. He stated that the spread function (ψ) describing the exponential decrease of scattered light from the slit could be written as:

\[ \psi(\xi) = \frac{2.3}{k} 10^{-2|\xi|/k}, \]

where \( k \) was a constant describing the magnitude of the spread in the emulsion of interest, and \( \xi \) was the distance from the slit. However, he also noted that many photographic emulsions did not fit that model well because the diffusion was really
composed of two parts: non-diffused or only slightly-diffused light, which results in a sharp image, and diffused light, which results in a diffusion halo. Taking both of these factors into account resulted in the equation:

\[
\psi(\xi) = \frac{2.3}{k_1} \rho 10^{-2|\xi|/k_1} + \frac{2.3}{k_2} (1 - \rho) 10^{-2|\xi|/k_2},
\]

(3-2)

where \(\rho\) indicated “the share of the exposure of the first distribution”. Frieser also noted that \(k_1\) was very small in many cases, leading the first term on the right hand side to be a Dirac delta function.

In 1967, Gilmore independently derived a point spread function, also with the intention of describing the spread of light in photographic film emulsions. The spread was assumed to be the result of both scattering and radial diffusion of light within the film, with the predominance of one versus the other depending on the turbidity of the medium. Turbidity is the result of an irregular distribution of optical non-uniformities in a medium, often the result of defects or the deliberate inclusion of particles with indices of refraction different from the medium. These non-uniformities cause multiple scattering of light, resulting in an isotropic diffusion of energy. In a non-turbid medium, energy diffuses in a radial pattern and dissipates by absorption.

The first use of this PSF for a medical physics application was by Seibert, Nalcioglu, and Roeck in 1984. They needed a PSF to describe veiling glare in fluoroscopy, which is caused by light spreading in the image intensifier. Because the light was impinging on a thin, regular phosphor, they assumed that radial diffusion was the primary mechanism of light spread. Thus, they adapted the part of Gilmore’s equation describing radial spreading, and eliminated the portion describing isotropic spreading (Equation 3-3; see Appendix A for the derivation).
\[ h_G(r) = \frac{1}{2\pi r_0} e^{-r/r_0} \]  

Equation 3-3

The variable \( r \) represents the distance from the origin, \( r_0 \) represents the mean propagation distance of the light, and \( h_G(r) \) represents the portion of Gilmore’s PSF used. Seibert et al. then added a term to Equation 3-3 describing the fraction of light that maps directly onto the image intensifier without contributing to the PSF, giving a new PSF, \( h_S(r) \). The result was Equation 3-4, in which the first term represents the portion from direct mapping and the second term represents the portion which undergoes radial spreading. The variable \( \rho \) represents the fraction of light which spreads (note that this is the opposite to Frieser’s \( \rho \), which indicated the portion that did not spread), \( k \) represents the mean propagation distance of the light (replacing \( r_0 \), and returning to Frieser’s original notation), and the Dirac delta function indicates the portion of light that is directly mapped.

\[ h_S(r) = (1 - \rho) \frac{\delta(r)}{r} + \frac{\rho}{2kr} e^{-r/k} \]  

Equation 3-4

The derivation of this equation from Equation 3-3 is found in Appendix A.

The leap from using a PSF designed for light impinging on a non-turbid, nearly two-dimensional image receptor, to using it for describing x-ray scatter within a thick object, is not entirely intuitive. The key to making this connection is that the x rays are used to form a projection image. Assume an infinitely-thin "pencil" beam of x rays impinging on an object, followed by an air gap and then a detector (Figure 3-8). Some of the x rays undergo a scattering interaction and change direction. Assuming they do not undergo any additional scattering interactions, those photons scattered toward the detector travel in a straight line until they hit the detector. The two-dimensional projection of the scattered x rays from many different x-ray interactions over the thick
object can be treated as functionally equivalent to a radial spread within the detector (analogous to the film or image intensifier of previous research). The fraction of light which spreads in the detector $\rho$ is equivalent to the scatter fraction, $SF$ (defined as the fraction of energy exiting the bottom surface of the object which comes from scattered radiation); in both cases, this quantity represents a loss of energy from the primary beam. The mean propagation distance of the light $k$ is equivalent to the mean propagation distance of the scattered x rays, projected into the two-dimensional plane of the detector. Ducote and Molloi showed that Monte Carlo measurements of mammography scatter can be fit reasonably well to the PSF derived from a two-dimensional energy spread (correlation coefficient of 0.88 at 28 kVp for a 4 cm thickness of PMMA, without a grid).\textsuperscript{36} However, they did not acquire physical data under the same conditions as the Monte Carlo simulations, so they were not able to compare the accuracy of measured vs. simulated values for $\rho$ and $k$.

Ducote and Molloi were the first to investigate the use of the PSF for scatter correction in mammography.\textsuperscript{36} The PSF they used $h_D(r)$ was nearly identical to that in Equation 3-4; however, they noted that using the original PSF was "designed to restore scattered light to its proper location",\textsuperscript{36} meaning that the total integrated signal was the same before and after the scatter correction, which would have been ideal in the case of light spread in an image intensifier. In an image intensifier, the signal from both primary and secondary photons undergo spread, and there is no way to distinguish them. For scatter correction, on the other hand, it is preferable to remove the scatter completely and leave only the primary signal. While it might seem preferable to “collapse” the scatter back into the primary impulse to make better use of dose, this solution would
work only if the PSF were a perfect model of the scatter. With an imperfect model, it is safer to remove the scatter than to reassign it and keep it in the image, because an incorrect reassignment would result in “re-scattering” the scatter. By reassigning the scatter to a new location, there is some weight assigned to the scatter estimate in the PSF, so it contributes to the image, potentially in a new location. Removing the scatter means assigning zero weight to the scatter estimate. If the scatter is removed rather than reassigned, the worst case scenario with an imperfect model would be some scatter remaining in its original distribution or the removal of a little primary radiation.

Thus, Ducote and Molloi modified their PSF to correct for this effect (Equation 3-5).

\[ h_D(r) = \frac{\delta(r)}{r} + \frac{\rho}{(1-\rho)2kr} e^{-r/k} \]  

(3-5)

**Other forms of the point spread function**

It was noted above that the form the PSF used by Ducote and Molloi (Equation 3-5) was adapted from a PSF developed for the case of two-dimensional light spread in film. They showed by Monte Carlo simulation that the PSF described x-ray scatter relatively well, but there is no theoretical derivation of that PSF originating from the case of three-dimensional x-ray scatter. Boone and Seibert developed a PSF, noted here as \( h_{BS}(r) \), specific to the case of x-ray scatter in diagnostic radiology. Using the model shown in Figure 3-9, they derived a two-dimensional PSF of the form:

\[ h_{BS}(r) = K \int_0^t \mu_p(t-s) e^{\mu_s \alpha} (2 \sin \theta - \sin^3 \theta) \left( \frac{1}{s+g} \right) \left[ 1 + \left( \frac{r}{s+g} \right)^2 \right]^{-1} ds \]  

(3-6)

where \( t \) is the object thickness, \( g \) is the air gap, \( r \) is the radial distance, \( \theta \) is the scatter angle, \( s \) is the distance from the bottom of the object to the location of the scatter event, and \( \alpha \) is the path length of scatter through the object. Note that \( \alpha = [s/(s + g)]r^2 + \)
\((s + g)^2\)^{1/2} \text{ and } \theta = \tan^{-1}\left[\frac{r}{s + g}\right]. \ K \text{ is a constant which is determined by setting } SPR = \int_0^\infty g'(r) \, dr. \text{ The PSF was derived by consideration of four main contributions: the scatter source intensity, the probability of scattering as a function of angle, scatter attenuation, and detector cross section as a function of angle. Its accuracy was verified using Monte Carlo simulations, with excellent agreement according to the graphical comparisons published by Boone and Seibert.}^{45} \text{ However, the PSF was not developed with mammography applications in mind, so only Compton scattering was assumed to contribute to the PSF.}

In earlier research, Boone et al.\(^{28}\) also approximated the PSF of x-ray scatter in fluoroscopy as a Gaussian distribution of the form:

\[
h_B(r) = \left(\frac{1}{r}\right) e^{-\frac{1}{2} \left(\frac{r}{\sigma}\right)^2}, \quad (3-7)
\]

where \(h_B(r)\) represents the PSF, \(r\) is the distance from a \(\delta\)-function primary response, and \(\sigma\) is found by fitting measured PSF profiles to Equation 3-7. Although an approximately Gaussian form can be derived analytically using a single scatter approximation for a thin object,\(^{22}\) Seibert and Boone do admit that "the choice of a Gaussian form is somewhat arbitrary in the multiple scatter environment".\(^{46}\)

Boone et al. measured PSF profiles using several phantom thicknesses and lead apertures (as opposed to disks) of three different diameters. The measured profiles were shown to fit the Gaussian PSF with excellent agreement. However, Seibert and Boone found later\(^{46}\) that the parameter \(\sigma\) depended on field size, thus indicating non-linear behavior; they suggested that the Gaussian PSF may be an incorrect match for the true functional form, and mentioned that an exponential fit might be more appropriate, based on the results of Monte Carlo studies.
Smith and Kruger developed a PSF to describe scatter as a function of phantom thickness, field size, source-to-object distance (SOD), object-to-detector distance (ODD), and photon energy. The model was developed by analyzing Compton scatter and attenuation, but the PSF was limited by three assumptions: it had a constant geometric shape, a volume dependent on phantom thickness, and a width dependent on ODD. The PSF was approximated as a cone for simplicity, so the tails of the PSF predicted by Compton scattering were truncated. Only single scattering was modeled. The conical PSF was interesting from the standpoint that it described scatter as a function of so many important parameters, but it was not very accurate for air gaps less than 10 cm (such as those used in clinical screening mammography).

Related Attempts at Scatter Correction

Scatter correction using convolution-based techniques was first attempted in the early 1980’s with respect to fluoroscopy, and the bulk of the research in this area has been in that context. Most attempts focused on finding a single point-spread function (PSF) to describe x-ray scatter and/or light scatter (veiling glare) within the image intensifier. Because these corrections were meant to be applied during live fluoroscopy, and because the earliest calculations were hardware- rather than software-based, these corrections had to be simple and quickly calculated. Thus, one PSF was typically applied to the entire image, regardless of thickness differences and technical factors such as tube potential, field size, etc.

One general theory frequently encountered in this early research was that a measured radiographic image \( I_m \) is the sum of a “scatter image” \( I_s \) and a “primary image” \( I_p \), and that the scatter component of the image could be approximated by
convolving the primary image with a scattering kernel \((h_s)\), leading to the following equations:

\[
I_m = I_p + I_s \tag{3-8}
\]

\[
I_s = I_p * h_s. \tag{3-9}
\]

The primary image was then calculated by assuming that the measured image was a fair approximation of the primary image \((I_p \approx I_m)\):

\[
I_p = I_m - I_s \approx I_m - (I_m * h_s). \tag{3-10}
\]

Although the researchers who relied on this method all reported successful attempts at scatter correction using this assumption, the use of the scatter-degraded image as an approximation of the primary image is a limitation. A variation on the theory described above, first reported by Seibert et al. in 1985, altered the algorithm so that it was no longer necessary to use the measured image to approximate the primary image. Rather than using the scatter kernel \(h_s\) of Equation 3-9, a different convolution kernel was defined which represented a mapping of both the primary and scatter radiation (rather than just the scatter radiation alone). Equation 3-8 could then be re-written as

\[
I_m = I_p * h_{s+p}, \tag{3-11}
\]

where \(h_{s+p}\) is still called the scatter kernel, but is given a different subscript to clarify that it is distinct from the scatter kernel in Equation 3-9. Thus, the premise that the measured image is formed by an additive relationship between the primary and scatter images was replaced with the idea that the measured image is a convolution of the primary image with a scatter kernel, thus allowing the calculation of the primary image from the measured image with the simple deconvolution
\[ I_p = \mathcal{F}^{-1} \left( \frac{\mathcal{F}(I_m)}{\mathcal{F}(h_{s+p})} \right), \]  

(3-12)

where \( \mathcal{F} \) indicates the 2-D Fourier transform and \( \mathcal{F}^{-1} \) indicates the inverse Fourier transform.

Seibert et al. used this technique and the PSF shown in Equation 3-4 in an early attempt at mathematical deconvolution for the removal of veiling glare from image intensifiers.\(^{52} \) They assumed that the PSF was a spatially-invariant, rotationally-symmetric function, and they solved for the primary image using Equation 3-12.

The work done by Seibert et al. was intended only to investigate the feasibility of the image deconvolution and did not involve the use of patient or anthropomorphic images. They used only lead disks, and measured the contrast ratio (i.e., the ratio of the signal next to the disk to that under it) before and after the deconvolution. They also replicated the experimental results with computer simulations. Although the scenarios were not clinically representative, both their experiments and their computer simulations did indicate that the contrast ratio improved by roughly a factor of 10.

Seibert and Boone\(^ {46} \) later attempted deconvolution of x-ray scatter in fluoroscopy using Equation 3-12 with a Gaussian PSF. They assumed that the total PSF \( h_t(r) \) is the sum of primary \( h_p(r) \) and scatter \( h_s(r) \) components:

\[ h_t(r) = h_p(r) + h_s(r) \]  

(3-13)

and that the PSF is circularly symmetric. Using a Dirac delta function to represent the primary component and a Gaussian distribution to represent the scatter component, they wrote Equation 3-13 as:

\[ h_t(r) = k_1 (1 - \rho) \frac{\delta(r)}{r} + k_2 \frac{\rho}{\beta r} e^{-\frac{1}{2}(r/\sigma)^2}, \]  

(3-14)
where \( k_1 \) and \( k_2 \) are normalization constants, \( \beta \) is related to field size dependence, and all other variables are as described previously. The values of the normalization constants were found by setting the volume under \( h_t(r) \) equal to one, and

\[
\beta = \text{erf}\left(\frac{F}{\sqrt{2\sigma}}\right),
\]

where \( F \) is the field size and \( \text{erf} \) indicates the error function. The values of \( \rho \) and \( \sigma \) were found experimentally. Veiling glare was deconvolved from the image using the techniques described in Seibert et al.\(^{52}\) before deconvolution of the scatter.

The deconvolution algorithm was tested on images of copper disks placed over 10 cm lucite, using a worst-case scenario of no grid or air gap. The contrast improved from 15% to 43% after deconvolution. The CNR also increased by 27%, despite the increase in noise due to the high-pass attributes of the PSF.\(^{46}\)

In 2010, Ducote and Molloi applied the work of the earlier researchers to scatter correction in digital mammography.\(^{36}\) However, they modified the approach of the fluoroscopy researchers in an important way: they altered the algorithm from a single shift-invariant PSF applied to the whole image, to a pixel-by-pixel approach in which the scatter kernel size and shape is varied depending on the apparent breast thickness at each pixel (an approach that was applied to fluoroscopy by Ersahin, Molloi, and Qian in 1995\(^{54}\)).

The form of the scatter kernel used was that of Equation 3-5, which was a minor revision from the scatter kernel used by Seibert et al.\(^{52}\) Ducote and Molloi characterized the SF and MRE of the kernel using the beam stop methodology employed by Seibert et al\(^{44}\) (described in more detail in Chapter 4). The apparent thickness of the breast over each pixel (in 1 mm increments) was found using a ratio of the signal in the pixel to the
signal in a region exposed to the unattenuated beam (also described in more detail in Chapter 4). The image was then separated into separate "masks", each consisting of all the pixels of one apparent thickness. The Fourier transform of each mask image was found and divided by the Fourier transform of the appropriate scatter kernel. The SF and MRE used for each scatter kernel were chosen based on the apparent breast thickness and tube potential. After each mask was divided by the corresponding kernel, the inverse Fourier transform was found, and the masks were recombined to form the final image. Although Ducote and Molloi referred to this process as “deconvolution”, the process of decomposing the mammography image into many mask images and using a separate PSF for each mask image is not a true deconvolution. In my study, the term “spatially-variant filtration” is used to describe this process.

The results indicated an increase in CNR of approximately 5% for images acquired with a grid, using a W/Rh target/filter combination at energies of 24, 28, and 31 kVp. The amount of noise in the images showed little change, so the increase in CNR was attributed to an increase in contrast.

**Implications of These Publications for My Study**

Much work has been done to characterize radiation scatter, and several potential forms of the PSF have emerged. Several forms of the PSF were eliminated from consideration for use in my study due to concerns about their accuracy or practical applicability. The form proposed by Boone and Seibert\(^{45}\) (Equation 3-6) was rejected for several reasons. First, it assumes that Compton scattering is the only source of scatter. This major limitation prevents the use of this PSF in the mammography energy range, because, as explained in Chapter 1, coherent scattering is a major source of scattered radiation at low energies. Another major limitation to this PSF is that it cannot be easily
adapted to the use of a grid. Its practical applicability also limited by its complexity. Image processing using such a complicated PSF as the scatter kernel would be difficult and slow.

A Gaussian PSF was proposed by Boone et al. However, the validity of Equation 3-7 to describe a mammography PSF has never been addressed, and Seibert and Boone suggested in a later publication that the PSF might be better described by an exponential function than a Gaussian function. The method of measuring the PSF profiles and finding the corresponding values of \( \sigma \) is also relatively complex. Therefore, the Gaussian PSF was also rejected for use in my study.

Smith and Kruger developed a conical PSF which described scatter as a function of several different parameters. However, the model was not tested using phantom thicknesses or beam energies appropriate for mammography. Smith and Kruger’s model was also based only on Compton scatter and had no way to consider the effects of coherent scattering, which are so important to mammography.

The PSF chosen for my study is descended from that developed by Gilmore for use in photographic film and first adapted for medical physics use by Seibert et al. The specific form used is that of Ducote and Molloi (Equation 3-5), which has already been shown to provide a reasonably good PSF for mammography. It should be noted that this PSF is normalized to an arc length of \( \pi \), and must be scaled by \( \frac{1}{\pi} \) before use for image manipulation (Appendix A).

Previous research clearly indicates the potential for image processing-based scatter correction to improve contrast. The study performed by Ducote and Molloi indicates this potential for digital mammography in particular. However, their work
was intended more for an application to dual-energy mammography rather than the current practice of clinical mammography, so an expansion of their methods to include more clinically-relevant techniques is needed. My study aims to develop their methods more fully to include a wider range of target/filter combinations and tube potentials, and to investigate their impact on clinical images. My study also investigates the potential for their methods to be expanded to images acquired without a grid.

The impact of scatter correction on other aspects of image quality such as noise and spatial resolution have not yet been well-studied. My study also examines the effects on these parameters more closely.
Table 3-1. Results of SPR measurements by Barnes and Brezovich for SPR as a function of kVp and phantom thickness.

<table>
<thead>
<tr>
<th>Tube voltage (kVp)</th>
<th>Phantom thickness</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>3 cm</td>
</tr>
<tr>
<td>27</td>
<td>0.41</td>
</tr>
<tr>
<td>32</td>
<td>0.40</td>
</tr>
<tr>
<td>36</td>
<td>0.40</td>
</tr>
<tr>
<td>42</td>
<td>0.40</td>
</tr>
</tbody>
</table>

[Adapted from G. T. Barnes and I. A. Brezovich, "The intensity of scattered radiation in mammography," Radiology 126, pages 244 and 245, Tables II and III. (1978).]

Table 3-2. Results of SPR measurements by Barnes and Brezovich for SPR as a function of field size for 2 kVp's and 2 phantom thicknesses.

<table>
<thead>
<tr>
<th>Field diameter (cm)</th>
<th>32 kVp Phantom thickness</th>
<th>42 kVp Phantom thickness</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>3 cm</td>
<td>6 cm</td>
</tr>
<tr>
<td>4</td>
<td>0.32</td>
<td>0.54</td>
</tr>
<tr>
<td>6</td>
<td>0.37</td>
<td>0.65</td>
</tr>
<tr>
<td>10</td>
<td>0.39</td>
<td>0.80</td>
</tr>
<tr>
<td>14</td>
<td>0.40</td>
<td>0.86</td>
</tr>
</tbody>
</table>

[Adapted from G. T. Barnes and I. A. Brezovich, "The intensity of scattered radiation in mammography," Radiology 126, pages 244 and 245, Tables II and III. (1978).]

Table 3-3. Results of SPR measurements by Dance and Day for SPR as a function of keV and phantom thickness.

<table>
<thead>
<tr>
<th>Energy (keV)</th>
<th>Breast thickness</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>2 cm</td>
</tr>
<tr>
<td>12.5</td>
<td>0.327</td>
</tr>
<tr>
<td>15</td>
<td>0.269</td>
</tr>
<tr>
<td>17.5</td>
<td>0.271</td>
</tr>
<tr>
<td>20</td>
<td>0.275</td>
</tr>
<tr>
<td>22.5</td>
<td>0.267</td>
</tr>
<tr>
<td>25</td>
<td>0.272</td>
</tr>
<tr>
<td>30</td>
<td>0.263</td>
</tr>
<tr>
<td>35</td>
<td>0.246</td>
</tr>
<tr>
<td>40</td>
<td>0.228</td>
</tr>
<tr>
<td>50</td>
<td>0.217</td>
</tr>
</tbody>
</table>

All values have an error of less than 3% except where indicated. [Adapted from D. R. Dance and G. J. Day, "The computation of scatter in mammography by Monte Carlo methods," Phys. med. biol. 29, page 240, Table 1. (1984).]
Table 3-4. Values of scatter fraction (SF) and mean radial extent (k) acquired using physical measurements, with a grid.

<table>
<thead>
<tr>
<th>Phantom thickness (cm)</th>
<th>1.08</th>
<th>2.17</th>
<th>4.82</th>
<th>6.99</th>
<th>9.64</th>
</tr>
</thead>
<tbody>
<tr>
<td>24 kVp SF</td>
<td>0.02</td>
<td>0.04</td>
<td>0.08</td>
<td>0.11</td>
<td>0.13</td>
</tr>
<tr>
<td>MRE (mm)</td>
<td>2.3</td>
<td>2.8</td>
<td>5.7</td>
<td>7.5</td>
<td>11.5</td>
</tr>
<tr>
<td>28 kVp SF</td>
<td>0.02</td>
<td>0.04</td>
<td>0.09</td>
<td>0.11</td>
<td>0.16</td>
</tr>
<tr>
<td>MRE (mm)</td>
<td>2.2</td>
<td>2.9</td>
<td>5.0</td>
<td>6.1</td>
<td>8.3</td>
</tr>
<tr>
<td>31 kVp SF</td>
<td>0.02</td>
<td>0.04</td>
<td>0.08</td>
<td>0.12</td>
<td>0.16</td>
</tr>
<tr>
<td>MRE (mm)</td>
<td>2.8</td>
<td>3.4</td>
<td>5.4</td>
<td>7.6</td>
<td>10.2</td>
</tr>
</tbody>
</table>

The phantom thicknesses are expressed as an equivalent thickness of 50/50 breast tissue. [Adapted from J. L. Ducote and S. Molloi, "Scatter correction in digital mammography based on image deconvolution," Phys. med. biol. 55, page 1304, Table 3. (2010).]

Table 3-5. Values of scatter fraction (SF) and mean radial extent (MRE) acquired using Monte Carlo measurements at 28 kVp, without a grid, for 5.1 cm of adipose and glandular tissue.

<table>
<thead>
<tr>
<th>Adipose tissue</th>
<th>Glandular tissue</th>
</tr>
</thead>
<tbody>
<tr>
<td>SF</td>
<td>0.41</td>
</tr>
<tr>
<td>MRE (mm)</td>
<td>36</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Glandular tissue</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>SF</td>
<td>0.43</td>
</tr>
<tr>
<td>MRE (mm)</td>
<td>30</td>
</tr>
</tbody>
</table>
Figure 3-1. Dependence of scatter-to-primary ratio (SPR) on the area of the radiation field, for a 32 kVp W/Al x-ray beam, without a grid. [Adapted from G. T. Barnes and I. A. Brezovich, "The intensity of scattered radiation in mammography," Radiology 126, pages 244 and 245, Figure 6 and Table III. (1978).]
Figure 3-2. Comparison of SPR values obtained for a monoenergetic beam by Dance and Day (solid lines) to those measured by Barnes and Brezovich (short dotted lines), which have been converted from a W/Al spectrum to an average energy. The dashed lines show the SPR "corrected" to ignore sources of scatter outside the breast. [Adapted from D. R. Dance and G. J. Day, "The computation of scatter in mammography by Monte Carlo methods," Phys. med. biol. 29, page 241, Figure 1. (1984).]
Figure 3-3. SPR for a monoenergetic beam as a function of energy and breast thickness, with a grid. [Adapted from D. R. Dance and G. J. Day, "The computation of scatter in mammography by Monte Carlo methods," Phys. med. biol. 29, page 244, Figure 6. (1984).]
Figure 3-4. Scatter intensity as a function of breast composition, for a 26 kV Mo/Mo spectrum with a 5 cm breast and 10 mm air gap. The scatter intensity is the height of the PSF as a function of distance; the scatter intensity integrated over all distances is related to the SPR\textsuperscript{38} [Adapted from J.M. Boone and V.N. Cooper III, "Scatter/primary in mammography: Monte Carlo validation," Med. Phys. 27, page 1823, Figure 3c. (2000).]
Figure 3-5. SPR as a function of photon energy and breast thickness. The "Direct Monte Carlo" data are from a simulation of a clinically-realistic geometry with data acquisition using the beam-stop method. The "Impulse Monte Carlo" data are from the simulation of a single impulse. Also shown is a comparison to the data of Dance and Day. [Adapted from J.M. Boone and V.N. Cooper III, "Scatter/primary in mammography: Monte Carlo validation," Med. Phys. 27, page 1824, Figure 7. (2000).]
Figure 3-6. SPR as a function kVp, target/filter, and breast thickness. Monte Carlo simulations were performed with a 15 cm-diameter circular field of view, a 15 mm air gap, no grid, and a breast composed of 50% glandular tissue.

[Adapted from J.M. Boone, K.K. Lindfors, V.N. Cooper III, and J.A. Seibert, "Scatter/primary in mammography: Comprehensive results," Med. Phys. 27, page 2408, Figure 1(a). (2000).]
Figure 3-7. SPR as a function of distance from the center of a 80 mm-radius semicircular field for different breast thicknesses. The measurements were taken along a line near the center of mass of the semicircle. [Adapted from J.M. Boone, K.K. Lindfors, V.N. Cooper III, and J.A. Seibert, "Scatter/primary in mammography: Comprehensive results," Med. Phys. 27, page 2410, Figure 2. (2000).]
Figure 3-8. Scattering of x rays within an object. The projection of the scattered x ray is approximated as the two-dimensional energy spread within the detector, with a propagation distance $r$.

Figure 3-9. Scattering of x rays within an object, with notation as used in the PSF developed by Seibert and Boone [Adapted from J.A. Seibert and J.M. Boone, "An analytical model of the scattered radiation distribution in diagnostic radiology," Med. Phys. 15, page 721, Figure 1. (1988).]
CHAPTER 4
PHYSICAL SCATTER MEASUREMENTS

Introduction

This chapter details the acquisition methods, results, and discussion of the physical measurements. These measurements include all data related to determination of the scatter fraction (SF) and mean radial extent (MRE), as well as the data needed by the scatter correction program to assess the apparent thickness. Also included are descriptions of several ancillary investigations related to determining the effect of scatter from the detector cover, the effect of phantom size, and the feasibility of applying this scatter correction to magnification mammography.

Mammography Units

The mammography units used were both Hologic™ Selenia® models: Unit 1 had a molybdenum target with 30 µm molybdenum and rhodium filters, and Unit 2 had a tungsten target with 58 µm rhodium and silver filters. Other than the x-ray tube and filters, the units were identical. The source-to-detector distance was 66 cm, and source-to-breast support distance was 64 cm. The units used a HTC (High Transmission Cellular) focused grid consisting of copper septa with air interspaces. The grid frequency was 23 lines/cm and the grid ratio was 4:1.

The detectors were both model FFDM-L, which is a direct conversion detector made of amorphous selenium paired with thin film transistors. It has a pixel size of 70 µm and a maximum image size of 24 x 29 cm, which corresponds to a matrix size of 3420 x 4096 pixels. The image size can also be changed to 18 x 24 cm for a smaller field of view, which has a 2560 x 3328 matrix size. Raw pixel values are given a DC offset of 50 from the measured values to avoid the possibility of negative pixel values.
once the flat field calibration is performed. The DC offset was subtracted from all pixel values used in the calculations described in this chapter.

When the mammography units are first used after being logged off, a detector warm-up period (lasting nearly 1 hour) is begun automatically. This warm-up period was allowed to complete for all cases in which the units had not been used within the prior hour. It was also discovered during the course of the investigation that very low pixel values (i.e., those under lead disks) increased slightly for approximately the first 10 exposures when starting from a "cold" unit, even after the detector warm-up. After that, they stabilized. Thus, the first 10 exposures were discarded for each data collection session. It should be emphasized that this trend was not noted with pixel values in the range observed clinically, so it is not expected that this issue would cause problems with clinical image processing, even if the unit had not been used for some time prior to the acquisition of clinical images.

A brief test of detector ghosting was performed to determine the minimum amount of time necessary between exposures. First, an image of a lead disk atop a 4-cm thickness BR1234 (a phantom material which simulates 47% glandular / 53% adipose tissue and is produced in rectangular slabs measuring 12.5 cm x 10 cm) was acquired using clinical techniques, and the average pixel value beneath the disk was measured using a small ROI. Then, the lead disk was removed and a second image acquired. Last, the lead disk was replaced, a third image was acquired, and the average pixel value beneath the disk was recorded again. An increase in pixel value between the first and third exposure was assumed to be the result of ghosting. By varying the time between the second and third exposures, it was shown that ghosting is undetectable
after 1 minute. All measurements described in the remainder of this chapter were acquired with at least 1 minute and 15 seconds between exposures.

**Apparent Thickness Measurements**

The first step in the scatter correction algorithm, described in more detail in Chapter 6, was to determine the apparent tissue thickness at each pixel. The thickness was found by dividing the signal (pixel value) at the pixel of interest $S_a$ by the average signal of a small ROI in a region of the detector that received the full unattenuated exposure $S_u$ and taking the negative natural logarithm of the ratio, resulting in the "log signal ratio" (LSR):

$$LSR = -\ln \left( \frac{S_a}{S_u} \right).$$  \hspace{1cm} (4-1)

A 64 pixel x 64 pixel ROI in the anterior left corner of the detector was chosen for measurement of the unattenuated signal. Although the x-ray intensity is lower at this position than over the breast due to the heel effect, the flat field correction ensures that the pixel values are approximately equal for the unattenuated beam at both locations (the lower x-ray intensity does result in more noise, however). The grid does not retract completely for exposures acquired without the grid; it retracts only 18 cm, even if the large (24 cm depth) field of view is selected. Thus, the ROI was placed about 1 cm from the left edge and 1 cm from the edge of the grid in the retracted position (i.e., at 17 cm from the chest wall). The same position was used for all images, whether taken with or without the grid. A sample of measurements was acquired to see whether a significant difference was found between placing the ROI on the left or right side of the detector, and none was found. Thus, the fact that the measurements were acquired on the left
side should have no effect on clinical situations in which it would be preferable to place the ROI on the right side (i.e., for a right MLO view).

A “bank” of attenuation measurements was determined using BR12 in thicknesses of 2, 4, 6 and 8 cm. For each exposure, the signal was measured in the center of the BR12 and in the detector corner as described. Exposures were acquired using three tube potentials, with all target/filter combinations, with and without the grid. The mAs used for each technique combination was selected to produce a pixel value of at least 60 under an 8-cm thickness of BR12. A minimum value of 60 was chosen because the DC offset of 50 corresponded to essentially zero signal, and the standard deviation of the pixel values in the ROI was about 3; a value slightly greater than three standard deviations (i.e., 10) above the DC offset was chosen to assure that adequate signal had been detected. The mAs was then held constant for all thicknesses.

The LSR of each measurement pair (i.e., the $S_a$ under the phantom and the $S_u$ for that image) was calculated. These LSR values were graphed with respect to tube potential, and linear fits were applied to the LSR values for each set of techniques and thicknesses (Figure 4-1). The equations derived from the fits (Appendix B) were used to calculate LSR values for peak tube potentials ranging from 22-38 kVp, as shown in Figure 4-1. The calculated LSR values for each kV were then graphed with respect to BR12 thickness, and a quadratic fit was applied to the data for each combination of target, filter, field size, grid status, and peak tube potential (Figure 4-2). The coefficients for the fits are found in Appendix B.

For each target/filter and grid combination, the coefficients from the quadratic fits were used to generate a table of LSR values for all thicknesses (in increments of 1 mm).
The tables were loaded into Matlab® (version 7.10.0.499 (R2010a), MathWorks®, Natick, MA) for use by the scatter correction program and indexed by target/filter combination, tube potential, and thickness. The data collection process was performed with both the large and small paddles, and separate tables were generated for each of the two field sizes.

### Scatter Kernel Measurements

#### Methods

Lead disks were required for the scatter kernel measurements. Six disks were manufactured with an approximate thickness of 2 mm and radii of 2.3, 2.9, 3.6, 5.5, 7.5, and 10 mm. Only five disks were used for the kernel measurements. The 10 mm disk was not used after it was determined that the pixel values obtained under it were too low to be statistically significant when a grid was used (statistical significance was assumed to be met if a pixel value was greater than three standard deviations above the DC offset.)

Images of each of the lead disks were acquired with and without a grid for each of the target/filter combinations and BR12 thicknesses of 2 cm, 4 cm, 6 cm, and 8 cm. An image with no lead disk was also acquired in each case. Images were made for five different tube potentials for each combination of target, filter, grid status, and BR12 thickness, because the relationship between MRE and tube potential was found to be non-linear. The range of tube potentials selected covered the range of clinical tube potentials used.

Each image was analyzed by recording the mean and standard deviation of a small ROI centered under the disk (or in the same location, for the images without a disk). Each ROI was approximately 20x20 pixels. All values were corrected for the DC offset.
offset by subtracting 50 from the mean pixel value, and the LSR for each disk measurement was calculated using Equation 4-1. For each tube potential, thickness, and target/filter combination, the LSR’s were then graphed as a function of disk radius (Figure 4-3). The trend was obviously linear, and any individual measurement that showed a marked deviation from linearity, as compared with the other measurements in that series, was repeated two more times and the average of the measurements taken (Figure 4-4). Estimates of the measurement error for each measurement were calculated from the standard deviation of the pixel values reported in the ROI statistics. It should be noted that, while the mean pixel values did fluctuate to ~5%, the standard deviations reported were typically unchanged over multiple measurements.

The MRE for each parameter combination was then calculated as the reciprocal of the slope of the linear fit. The SF was calculated by taking the exponential of the negative of the y-intercept. The resulting data for each target/filter combination was entered into Matlab® and fitted as a function of tube potential and thickness using the Surface Fitting function (Figure 4-5). The resulting fits were of the form:

\[ y = C_1 + C_2 t + C_3 E + C_4 t^2 + C_5 t E + C_6 E^2 + C_7 t^3 + C_8 t^2 E + C_9 t E^2, \]  

(4-2)

where \( y \) is the MRE or SF value, \( t \) is the apparent thickness, and \( E \) is the tube potential. The coefficients \( C_1 - C_9 \) appropriate for the target/filter combination and grid status are shown in Appendix B. The use of Equation 4-2 allows the SF and MRE to be calculated for any thickness (in increments of 1 mm) and any tube potential (in increments of 1 kVp).
Results and Discussion

The SF and MRE data found using the method just described can be found in Tables 4-1 through 4-8. Graphs showing the MRE and SF as functions of tube potential and thickness are shown in Figures 4-6 through 4-9. The data related to the scatter fraction and mean radial extent are discussed separately below.

The scatter fraction

There is little difference in the SF among the target/filter combinations. With a grid, the SF increases from a minimum of about 0.05 for a 2 cm phantom at 24 kVp to a maximum of about 0.16 for an 8 cm phantom at 34 kVp (Figure 4-6). The SF's of Mo/Mo and Mo/Rh rise about 10% above those of W/Rh and W/Ag at 8 cm. There is a slight dependence on tube potential above 30 kVp for the larger phantom thicknesses, which causes ~10% increase in SF between 30 and 34 kVp. This dependence may be due to grid penetration, because the increase in SF is not seen without a grid. Without a grid, the SF increases from about 0.25 for a 2 cm phantom at 24 kVp to about 0.52 for an 8 cm phantom at 34 kVp (Figure 4-7). There is essentially no dependence on target/filter combination or tube potential without a grid. The independence from tube potential and target/filter combination shown by these data supports Hypotheses 1-1a and 1-1b.

The SF is higher by a factor of 3.0-5.2 (depending on phantom thickness) without a grid than with a grid, which supports Hypothesis 1-1c. The SF also more than doubles as breast thickness increases from 2 cm to 8 cm, both with and without a grid, which supports Hypothesis 1-2a.

Only Ducote and Molloi have previously published work describing the SF in mammography. They used only W/Rh, with a grid. Because they used acrylic rather
than BR12, the thicknesses of acrylic they used were converted here to equivalent thicknesses of BR12 for comparison with my data, as explained in Chapter 3. The values from my study used in the comparison were calculated from the Matlab®-generated surface fits of my data, since neither the tube potentials nor the phantom thicknesses used were the same between my study and theirs. My data demonstrate reasonably good agreement with theirs (Table 4-9): the average difference was 23%, with differences between individual measurements ranging from 1% to 65%. The largest differences (all greater than 60%) were at the 1 cm phantom thickness, which is not a tissue thickness that is common in clinical mammography (except at the nipple). The SF values were also very low at this thickness, so that a small absolute difference in the measurements resulted in a large percent difference. Excluding the 1 cm phantom thickness brought the average error down to 11%, which, considering the different phantom materials and the use of the surface fits to find matching tube potentials and thicknesses in my data, is not unreasonable. The best matches between the data sets were at the 4.8 and 7.0 cm thicknesses, with average differences of 7% and 6% respectively. Of the thicknesses considered, these are the ones most likely to be encountered in clinical mammography, because the average breast thickness has been reported to be a little over 5 cm.\textsuperscript{56, 57}

As described in Chapter 3, several other researchers have published data on the scatter-to-primary ratio (SPR) in the mammography range. The SPR is easily calculated once the scatter fraction is known. The definitions of the two are as follows:

\[
SPR = \frac{S}{P} \quad \text{(4-3)}
\]

\[
SF = \frac{S}{S+P} \quad \text{(4-4)}
\]
where $S$ is the contribution from scatter radiation and $P$ is the contribution from primary radiation. The quantity $S + P$ is equal to the average pixel value of the ROI measured over the BR12 without any lead disk intercepting the beam, so it is acquired as part of the data collection described in the previous section. The SPR was calculated for the purpose of comparison with other publications (Figures 4-10 and 4-11). As expected, the overall trends are very similar to those described for the SF; essentially all that has changed is that the SF values have been scaled. With a grid, the SPR ranges from about 0.06 at 2 cm to 0.2 at 8 cm. Without a grid, it ranges from approximately 0.3 at 2 cm to 1.1 at 8 cm.

The non-grid values can be compared to the work of several other researchers. The Monte Carlo-generated SPR results published by Boone et. al.\textsuperscript{40} (Figure 3-6) are compared to the Mo/Mo SPR values from my study in Table 4-10. The agreement between the data improves as phantom thickness increases, with average differences of 36% for a 2 cm thickness, 22% at 4 cm, 16% at 6 cm, and 5% at 8 cm. The agreement does not appear to depend on tube potential.

Table 4-11 contains a comparison of my data to the physical measurements published by Barnes and Brezovich\textsuperscript{29} for a W/Al spectrum. This target/filter combination was not used in my study, but because my data indicated little dependence on target/filter combination or tube potential, W/Ag was used for the comparison since it was the most similar spectrum to W/Al (Ag has a higher K-edge than Rh, whereas Al has no K-edge in the mammography energy range). The agreement between the two data sets is good, with errors ranging from -7.8% to 11.5% and averaging 0.8% (the average of the absolute value of the errors is 5.8%). The data from my study show less
variation with phantom thickness than do the data from Barnes and Brezovich, because my SPR values are higher than theirs at 3 and 4 cm and lower at 5 and 6 cm. Both studies indicate that there is no dependence on tube potential in the mammography range.

The other studies cited in Chapter 3 used Monte Carlo simulations of monoenergetic spectra to determine SPR values, and thus are not directly comparable to my values. However, the non-grid simulations reported by Boone et al. and Dance and Day (Figure 3-5) still show values reasonably similar to those obtained in my study. Both of those studies produced data with very little dependence on keV and average SPR values around 0.23 at 2 cm, 0.4 at 4 cm, 0.58 at 6 cm, and 0.72 at 8 cm. My study produced values which were all somewhat higher: 0.33 at 2 cm, 0.56 at 4 cm, 0.80 at 6 cm, and 1.06 at 8 cm. This difference is likely attributable to sources of scatter not included in the Monte Carlo calculations; evidence for this theory lies in Figure 3-2, which indicates a sharp increase in SPR below about 20 keV for the data points that were not corrected for sources of scatter outside the breast. My physical measurements include these sources of scatter, and the polyenergetic beam includes these lower energies.

The only simulations of SPR with a grid in the literature, which also depended on monoenergetic spectra, were those performed by Dance and Day (Figure 3-3). Their data demonstrate a large energy dependence and scatter-to-primary ratios somewhat higher than those seen in my study (Table 4-12). The relatively small energy dependence seen in my data might be explained by the polyenergetic spectra used, which would have averaged out the SPR values over the range of monoenergetic
spectra considered by Dance and Day. The higher values of SPR can be attributed to the grid and geometry they simulated: they used a linear grid as opposed to the cellular grid in my study, and modeled a 12 mm air gap as opposed to a 20 mm air gap. Both of these factors would have contributed to a higher SPR.

**The mean radial extent**

The MRE shows more variation as a function of target/filter combination than the SF. With a grid, W/Rh and W/Ag generally produce larger MRE's at breast thicknesses of 2 cm and 4 cm than do Mo/Mo and Mo/Rh, but they are all roughly equal at 6 cm, and the trend reverses at 8 cm (Figure 4-8). Without a grid, the W-target MRE's are consistently higher except for the lowest and highest tube potentials at the 8 cm thickness (Figure 4-9). All of the curves, both with and without the grid, demonstrate a marked concavity. The tube potential at which the minimum MRE is seen varies depending on the phantom thickness and target/filter combination, but for the most part, it appears to increase as the thickness increases (Table 4-13). The major exception to this trend is for the W-target spectra without a grid, in which the minimum increases from 2-6 cm and drops at 8 cm. The concavity tends to become more pronounced as the phantom thickness increases. Because of the complexity of the relationship between MRE and tube potential, Hypothesis 2-1a is not supported by these data. Hypothesis 2-1b is likewise not supported due to the complex relationship between the MRE and target/filter combination.

These relationships between MRE and tube potential are also reflected in the data published by Ducote and Molloi,36 which is the only previous publication that has addressed the MRE of scatter in mammography. A comparison of the MRE values found in their work to the MRE values from my study is shown in Table 4-14. As was the
case for the comparison of the SF, it was necessary to modify Ducote and Molloi’s MRE values from acrylic to estimated thicknesses of BR12, and use the surface fits to find data from my study which matched theirs with respect to tube potential and thickness. The values from my study are consistently higher; differences ranged from 4% to 46% (average of 21%) depending on the tube potential and phantom thickness. The worst of the discrepancies were at the lower extremes of phantom thickness (1.1 cm and 2.2 cm), and excluding these brought the average difference down to 14%, with a range of 4% to 28%.

The cause of this concave relationship was not discussed by Ducote and Molloi, but it is possibly related to an interplay between the differential cross-section of coherent scattering with respect to angle and the effects of attenuation. The dependence of scatter angle on x-ray energy changes quickly for coherent scatter in the low-energy range: it decreases from an average angle of about 27 degrees at 15 keV to 21 degrees at 30 keV (Figure 1-16).24 By simple geometry, this decrease in photon angle results in a decrease in radial travel of 2-12 mm depending on the height in the phantom (from 2-8 cm) at which scattering occurred. In this energy range, the average Compton scatter angle is essentially unchanged.22 Thus, the MRE would be expected to decrease as the x-ray energy increases due to the change in scatter angle. However, it is also true that, as energy increases, a photon becomes more penetrating and therefore more likely to exit the phantom and be detected. Thus, the MRE would be expected to increase as the x-ray energy increases due to the effects of attenuation. These competing processes are likely the cause of the observed relationship between tube potential and MRE. This idea is further supported by two observations: first, that
the minima moved to higher tube potentials as the phantom thickness increased (because a photon requires a higher energy to penetrate a thicker phantom); and second, that for a given thickness, the minima were generally lower for the more penetrating W-target spectra (which would not require as high a tube potential to penetrate the phantom as would the softer Mo-target spectra).

With a grid, the MRE increases from about 3.0-3.5 mm for a 2 cm phantom at 24 kVp to about 12-13 mm for an 8 cm phantom at 34 kVp. Without a grid, the MRE increases from a minimum of approximately 20 mm for a 2 cm phantom at 24 kVp to a maximum of 40-45 mm for an 8 cm phantom at 34 kVp. Both Hypotheses 2-1c and 2-2a are therefore supported by these data.

**Ancillary Investigations**

**Additional Kernel Measurements**

During the course of the project, the possibility was proposed that the area of the detector exposed to the unattenuated beam might have a significant effect on the scatter kernels, especially without the grid in place. The idea was that scatter from the detector cover or paddle might be able, in the absence of the grid, to travel a significant distance in the 2 cm space between the detector cover and the detector. If such an effect was noticeable, it might be corrected by developing a separate kernel describing scatter from the detector cover as a function of area exposed to the unattenuated beam. To test this hypothesis, lead sheets with circular cut-outs were made from a discarded lead apron (0.5 mm Pb-equivalent at 80 kV). The sheets were large enough to cover the entire detector, with a central hole cut out of each one in diameters of 2.0, 4.1, 6.2, 8.0, and 10.2 cm.
Exposures were made with each of the lead sheets in place, and the mean pixel value was measured using an ROI in the center of the circular hole. The large field of view and large paddle were used for all measurements. It was determined that the paddle height did not affect the measurements for any height between 2-8 cm (the range of compressed breast thicknesses commonly encountered). Imaging was performed at 24, 26, 28, 30, 32, and 34 kV, with and without the grid, for all target/filter combinations. The range of mAs values used was selected to keep pixel values within the holes as close to the clinical range as possible.

Although the pixel values did drop when only very small areas were exposed to the full beam, the pixel values remained relatively constant at diameters greater than 6.2 cm. This diameter represents an area of 30 cm$^2$. The area of the small (18 cm x 24 cm) field is 432 cm$^2$, and the large (24 x 29 cm) field is 696 cm$^2$. A situation in which less than 30 cm$^2$ of the detector surface is exposed to the unattenuated beam would be extremely unlikely in clinical mammography; thus, it was decided that developing a correction to the kernel to account for the area exposed to the unattenuated beam was unnecessary.

**The Effect of Phantom Size**

All scatter kernel measurements were performed with BR12, which is a rectangular phantom measuring 12.5 cm x 10 cm. Real breasts, of course, vary widely in size and shape, so it was necessary to determine whether a large error might be expected when applying the scatter kernels to breasts differing in size from the BR12. Measurements of scatter fraction and mean radial extent made using a single 2 cm-thick slab of BR12 were compared to those using enough 2 cm-thick slabs of BR12 to fully cover the detector (24 cm x 29 cm). The grid was also removed to allow for a
greater amount of scatter. This set-up simulated the worst-case scenario of the (highly
unlikely) situation in which a breast extends beyond the field of view on all sides, and no
grid is used. More measurements would have been performed with intermediate
phantom sizes if there had been a marked difference between this scenario and the use
of only one slab of BR12. However, the scatter kernel parameters were very similar
even in this extreme situation (Figure 4-12). The difference was only 3.4% in the radial
extent and 1.3% in the scatter fraction, with significant overlap of the error bars.
Therefore, it was determined that the effect of phantom size on the measurements was
negligible. Hypotheses 1-2c and 2-2c are therefore supported.

**Feasibility of Application to Magnification Mammography**

Magnification mammography is the only form of typical clinical mammography in
which a grid is not used as part of standard clinical practice. The use of a large air gap
eliminates much, but not all, of the scatter, so the idea of applying the scatter correction
to magnification mammography was proposed. Measuring the scatter fraction and mean
radial extent in a magnification geometry does not present any significant problem, but
determining the apparent thickness does. The method of apparent thickness calculation
described earlier in this chapter requires at least a small portion of the detector to be
exposed to the unattenuated beam, but magnification mammograms are collimated to a
small portion of the breast. Frequently, all of the beam undergoes some attenuation by
breast tissue.

One idea proposed to overcome this difficulty was to see whether there was
enough consistency in absolute pixel values to allow the determination of thickness
based solely on a single pixel value rather than a ratio. This procedure would require
the development of a bank a pixel values, as a function of imaging techniques including
target/filter combination, kVp, and mAs. Ideally, this bank of values would be applicable to all units of the same type and software version and not be dependent on individual unit calibrations. A bank that needed to be regenerated for each individual unit after every calibration would be unfeasible, since service engineers perform detector calibrations on a fairly frequent basis. Mammography technologists perform minor gain calibrations on a weekly basis, which could also affect the pixel values.

To test the feasibility of the pixel value bank, ROI measurements of average pixel values were made under identical conditions using two tungsten-target mammography units of the same age, at the same facility, calibrated at the same time by the same service engineer (which should theoretically be the best-case scenario). The pixel values differed by as much as 11%, which would translate into an apparent thickness difference of nearly 1 cm. Thus, determination of the apparent thickness using a single pixel value is not a feasible option; a ratio of values is needed to overcome the effects of calibration differences between units.

The unattenuated beam measurement could be acquired fairly simply if a small hole was drilled in the collimator. The entire detector is never used for magnification mammography, and it would be possible to place the hole in a location that would allow unattenuated photons to hit the detector without interfering with the clinical image. However, because the mammography units used in this research project are systems currently in clinical use, modification of such systems is not possible. Thus, the application of the scatter correction program to magnification mammography is beyond the scope of this project.
Table 4-1. Scatter fractions and 95% confidence intervals with and without a grid for Mo/Mo. The ratio of the SF with and without a grid is also given for each combination of thickness, tube potential, target, and filter.

<table>
<thead>
<tr>
<th>Thickness (cm)</th>
<th>Tube potential (kVp)</th>
<th>SF with grid Mo/Mo</th>
<th>SF without grid Mo/Mo</th>
<th>Ratio without grid/with grid</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>24</td>
<td>0.055 (0.044,0.065)</td>
<td>0.25 (0.23,0.27)</td>
<td>4.5</td>
</tr>
<tr>
<td>2</td>
<td>26</td>
<td>0.055 (0.045,0.065)</td>
<td>0.25 (0.23,0.27)</td>
<td>4.5</td>
</tr>
<tr>
<td>2</td>
<td>28</td>
<td>0.057 (0.046,0.067)</td>
<td>0.25 (0.23,0.27)</td>
<td>4.4</td>
</tr>
<tr>
<td>2</td>
<td>30</td>
<td>0.055 (0.045,0.065)</td>
<td>0.25 (0.23,0.27)</td>
<td>4.5</td>
</tr>
<tr>
<td>2</td>
<td>34</td>
<td>0.049 (0.040,0.058)</td>
<td>0.25 (0.24,0.28)</td>
<td>5.1</td>
</tr>
<tr>
<td>4</td>
<td>24</td>
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<td>0.37 (0.35,0.40)</td>
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<tr>
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<td>26</td>
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<td>0.37 (0.34,0.39)</td>
<td>4.6</td>
</tr>
<tr>
<td>4</td>
<td>28</td>
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<td>0.37 (0.34,0.39)</td>
<td>4.5</td>
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<tr>
<td>4</td>
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<td>0.37 (0.34,0.39)</td>
<td>4.5</td>
</tr>
<tr>
<td>4</td>
<td>34</td>
<td>0.086 (0.077,0.096)</td>
<td>0.37 (0.35,0.40)</td>
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<tr>
<td>6</td>
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<td>0.10 (0.096,0.11)</td>
<td>0.45 (0.43,0.48)</td>
<td>4.3</td>
</tr>
<tr>
<td>6</td>
<td>30</td>
<td>0.11 (0.097,0.11)</td>
<td>0.45 (0.43,0.49)</td>
<td>4.3</td>
</tr>
<tr>
<td>6</td>
<td>32</td>
<td>0.11 (0.10,0.12)</td>
<td>0.45 (0.42,0.48)</td>
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<td>6</td>
<td>34</td>
<td>0.12 (0.11,0.13)</td>
<td>0.45 (0.42,0.45)</td>
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<tr>
<td>8</td>
<td>26</td>
<td>0.14 (0.13,0.16)</td>
<td>0.52 (0.48,0.52)</td>
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<tr>
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<td>0.14 (0.13,0.16)</td>
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<td>3.7</td>
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<td>3.3</td>
</tr>
<tr>
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<td>34</td>
<td>0.17 (0.15,0.18)</td>
<td>0.52 (0.48,0.53)</td>
<td>3.1</td>
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Table 4-2. Scatter fractions and 95% confidence intervals with and without a grid for Mo/Rh. The ratio of the SF with and without a grid is also given for each combination of thickness, tube potential, target, and filter.

<table>
<thead>
<tr>
<th>Thickness (cm)</th>
<th>Tube potential (kVp)</th>
<th>SF with grid Mo/Rh</th>
<th>SF without grid Mo/Rh</th>
<th>Ratio without grid/with grid</th>
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<tbody>
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<td>0.25 (0.23,0.26)</td>
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</tr>
<tr>
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<td>0.25 (0.23,0.27)</td>
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<td>0.054 (0.045,0.062)</td>
<td>0.23 (0.22,0.25)</td>
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<tr>
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<td>0.080 (0.073,0.088)</td>
<td>0.36 (0.34,0.39)</td>
<td>4.5</td>
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<tr>
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<td>26</td>
<td>0.082 (0.075,0.089)</td>
<td>0.36 (0.34,0.39)</td>
<td>4.4</td>
</tr>
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<td>28</td>
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<td>0.37 (0.35,0.39)</td>
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<tr>
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<td>0.36 (0.35,0.39)</td>
<td>4.5</td>
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<td>0.45 (0.42,0.49)</td>
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<td>32</td>
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<td>0.45 (0.42,0.49)</td>
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<td>0.45 (0.42,0.45)</td>
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<tr>
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<td>0.52 (0.49,0.54)</td>
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<td>32</td>
<td>0.15 (0.13,0.17)</td>
<td>0.52 (0.49,0.54)</td>
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<td>34</td>
<td>0.16 (0.13,0.18)</td>
<td>0.52 (0.49,0.54)</td>
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</table>
Table 4-3. Scatter fractions and 95% confidence intervals with and without a grid for W/Rh. The ratio of the SF with and without a grid is also given for each combination of thickness, tube potential, target, and filter.

<table>
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<tr>
<th>Thickness (cm)</th>
<th>Tube potential (kVp)</th>
<th>SF with grid W/Rh</th>
<th>SF without grid W/Rh</th>
<th>Ratio without grid/with grid</th>
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</thead>
<tbody>
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<td>24</td>
<td>0.050 (0.040,0.060)</td>
<td>0.25 (0.23,0.26)</td>
<td>4.9</td>
</tr>
<tr>
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</tr>
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</tr>
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</tr>
<tr>
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</tr>
<tr>
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</tr>
<tr>
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<td>0.079 (0.069,0.089)</td>
<td>0.36 (0.34,0.38)</td>
<td>4.5</td>
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<td>0.44 (0.42,0.48)</td>
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<td>6</td>
<td>30</td>
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<td>0.44 (0.42,0.48)</td>
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<td>32</td>
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<td>0.45 (0.42,0.48)</td>
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<tr>
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<td>30</td>
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<td>32</td>
<td>0.15 (0.13,0.17)</td>
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<tr>
<td>8</td>
<td>34</td>
<td>0.17 (0.15,0.19)</td>
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Table 4-4. Scatter fractions and 95% confidence intervals with and without a grid for W/Ag. The ratio of the SF with and without a grid is also given for each combination of thickness, tube potential, target, and filter.

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<th>Thickness (cm)</th>
<th>Tube potential (kVp)</th>
<th>SF with grid W/Ag</th>
<th>SF without grid W/Ag</th>
<th>Ratio without grid/with grid</th>
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<td>0.25 (0.23, 0.25)</td>
<td>4.9</td>
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<td>0.25 (0.23, 0.25)</td>
<td>5.2</td>
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<td>0.25 (0.23, 0.25)</td>
<td>4.9</td>
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<tr>
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<td>0.44 (0.42, 0.44)</td>
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<td>0.44 (0.42, 0.44)</td>
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<td>0.13 (0.12, 0.15)</td>
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<td>0.13 (0.12, 0.15)</td>
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<td>0.16 (0.15, 0.18)</td>
<td>0.51 (0.48, 0.51)</td>
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Table 4-5. Mean radial extents and 95% confidence intervals with and without a grid for Mo/Mo. The ratio of the MRE with and without a grid is also given for each combination of thickness, tube potential, target, and filter. Values are in millimeters.

<table>
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<th>Thickness (cm)</th>
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<th>MRE with grid Mo/Mo</th>
<th>MRE without grid Mo/Mo</th>
<th>Ratio without grid/with grid</th>
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<td>20 (18,22)</td>
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Table 4-6. Mean radial extents and 95% confidence intervals with and without a grid for Mo/Rh. The ratio of the MRE with and without a grid is also given for each combination of thickness, tube potential, target, and filter. Values are in millimeters.

<table>
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<th>Thickness (cm)</th>
<th>Tube potential (kVp)</th>
<th>MRE with grid Mo/Rh</th>
<th>MRE without grid Mo/Rh</th>
<th>Ratio without grid/ with grid</th>
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<td>28 (25,34)</td>
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</tbody>
</table>
Table 4-7. Mean radial extents and 95% confidence intervals with and without a grid for W/Rh. The ratio of the MRE with and without a grid is also given for each combination of thickness, tube potential, target, and filter. Values are in millimeters.

<table>
<thead>
<tr>
<th>Thickness (cm)</th>
<th>Tube potential (kVp)</th>
<th>MRE with grid W/Rh</th>
<th>MRE without grid W/Rh</th>
<th>Ratio without grid/with grid</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>24</td>
<td>3.7 (3.2,4.1)</td>
<td>22 (19,24)</td>
<td>5.9</td>
</tr>
<tr>
<td>2</td>
<td>26</td>
<td>3.6 (3.1,4.1)</td>
<td>22 (19,24)</td>
<td>6.0</td>
</tr>
<tr>
<td>2</td>
<td>28</td>
<td>3.9 (3.4,4.4)</td>
<td>22 (19,24)</td>
<td>5.6</td>
</tr>
<tr>
<td>2</td>
<td>30</td>
<td>5.0 (4.3,5.7)</td>
<td>23 (20,25)</td>
<td>4.6</td>
</tr>
<tr>
<td>2</td>
<td>34</td>
<td>6.2 (5.4,7.1)</td>
<td>24 (21,27)</td>
<td>3.8</td>
</tr>
<tr>
<td>4</td>
<td>24</td>
<td>5.0 (4.6,5.4)</td>
<td>28 (26,31)</td>
<td>5.5</td>
</tr>
<tr>
<td>4</td>
<td>26</td>
<td>4.9 (4.5,5.4)</td>
<td>27 (25,30)</td>
<td>5.5</td>
</tr>
<tr>
<td>4</td>
<td>28</td>
<td>5.4 (4.9,5.8)</td>
<td>28 (25,31)</td>
<td>5.1</td>
</tr>
<tr>
<td>4</td>
<td>30</td>
<td>6.4 (5.8,6.9)</td>
<td>29 (27,33)</td>
<td>4.6</td>
</tr>
<tr>
<td>4</td>
<td>34</td>
<td>7.9 (7.2,8.6)</td>
<td>32 (29,37)</td>
<td>4.0</td>
</tr>
<tr>
<td>6</td>
<td>26</td>
<td>7.1 (6.3,7.8)</td>
<td>33 (29,38)</td>
<td>4.7</td>
</tr>
<tr>
<td>6</td>
<td>28</td>
<td>6.7 (6.0,7.5)</td>
<td>33 (29,38)</td>
<td>4.9</td>
</tr>
<tr>
<td>6</td>
<td>30</td>
<td>7.3 (6.5,8.1)</td>
<td>33 (29,38)</td>
<td>4.6</td>
</tr>
<tr>
<td>6</td>
<td>32</td>
<td>8.5 (7.6,9.4)</td>
<td>35 (31,40)</td>
<td>4.1</td>
</tr>
<tr>
<td>6</td>
<td>34</td>
<td>11 (9.6,12)</td>
<td>38 (34,38)</td>
<td>3.6</td>
</tr>
<tr>
<td>8</td>
<td>26</td>
<td>10 (8.6,11)</td>
<td>38 (32,38)</td>
<td>3.8</td>
</tr>
<tr>
<td>8</td>
<td>28</td>
<td>9.0 (7.7,10)</td>
<td>39 (33,39)</td>
<td>4.3</td>
</tr>
<tr>
<td>8</td>
<td>30</td>
<td>9.0 (7.9,10)</td>
<td>41 (34,41)</td>
<td>4.4</td>
</tr>
<tr>
<td>8</td>
<td>32</td>
<td>11 (10,13)</td>
<td>43 (36,43)</td>
<td>3.9</td>
</tr>
<tr>
<td>8</td>
<td>34</td>
<td>15 (13,17)</td>
<td>47 (40,47)</td>
<td>3.2</td>
</tr>
</tbody>
</table>
Table 4-8. Mean radial extents and 95% confidence intervals with and without a grid for W/Ag. The ratio of the MRE with and without a grid is also given for each combination of thickness, tube potential, target, and filter. Values are in millimeters.

<table>
<thead>
<tr>
<th>Thickness (cm)</th>
<th>Tube potential (kVp)</th>
<th>MRE with grid W/Ag</th>
<th>MRE without grid W/Ag</th>
<th>Ratio without grid/with grid</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>24</td>
<td>3.9 (3.4, 4.4)</td>
<td>22 (20, 22)</td>
<td>5.6</td>
</tr>
<tr>
<td>2</td>
<td>26</td>
<td>4.2 (3.7, 4.7)</td>
<td>22 (21, 22)</td>
<td>5.3</td>
</tr>
<tr>
<td>2</td>
<td>28</td>
<td>4.7 (4.1, 5.3)</td>
<td>23 (21, 23)</td>
<td>4.9</td>
</tr>
<tr>
<td>2</td>
<td>30</td>
<td>5.1 (4.5, 5.8)</td>
<td>24 (22, 24)</td>
<td>4.6</td>
</tr>
<tr>
<td>2</td>
<td>34</td>
<td>5.8 (5.1, 6.5)</td>
<td>25 (23, 25)</td>
<td>4.3</td>
</tr>
<tr>
<td>4</td>
<td>24</td>
<td>4.9 (4.5, 5.4)</td>
<td>29 (26, 29)</td>
<td>5.8</td>
</tr>
<tr>
<td>4</td>
<td>26</td>
<td>5.1 (4.7, 5.6)</td>
<td>29 (27, 29)</td>
<td>5.7</td>
</tr>
<tr>
<td>4</td>
<td>28</td>
<td>5.6 (5.1, 6.1)</td>
<td>30 (27, 30)</td>
<td>5.3</td>
</tr>
<tr>
<td>4</td>
<td>30</td>
<td>6.5 (5.9, 7.0)</td>
<td>31 (28, 31)</td>
<td>4.8</td>
</tr>
<tr>
<td>4</td>
<td>34</td>
<td>7.4 (6.7, 8.0)</td>
<td>32 (30, 32)</td>
<td>4.4</td>
</tr>
<tr>
<td>6</td>
<td>26</td>
<td>7.1 (6.6, 7.6)</td>
<td>35 (32, 35)</td>
<td>4.9</td>
</tr>
<tr>
<td>6</td>
<td>28</td>
<td>7.2 (6.7, 7.7)</td>
<td>35 (32, 35)</td>
<td>4.8</td>
</tr>
<tr>
<td>6</td>
<td>30</td>
<td>7.6 (7.0, 8.1)</td>
<td>35 (32, 35)</td>
<td>4.6</td>
</tr>
<tr>
<td>6</td>
<td>32</td>
<td>8.5 (7.9, 9.1)</td>
<td>36 (33, 36)</td>
<td>4.3</td>
</tr>
<tr>
<td>6</td>
<td>34</td>
<td>9.8 (9.1, 10)</td>
<td>39 (35, 39)</td>
<td>4.0</td>
</tr>
<tr>
<td>8</td>
<td>26</td>
<td>10 (8.7, 11)</td>
<td>40 (35, 40)</td>
<td>4.0</td>
</tr>
<tr>
<td>8</td>
<td>28</td>
<td>9.0 (8.2, 10)</td>
<td>41 (36, 41)</td>
<td>4.4</td>
</tr>
<tr>
<td>8</td>
<td>30</td>
<td>10 (8.7, 11)</td>
<td>41 (36, 41)</td>
<td>4.1</td>
</tr>
<tr>
<td>8</td>
<td>32</td>
<td>11 (9.0, 12)</td>
<td>42 (37, 42)</td>
<td>4.0</td>
</tr>
<tr>
<td>8</td>
<td>34</td>
<td>13 (11, 14)</td>
<td>43 (38, 43)</td>
<td>3.4</td>
</tr>
</tbody>
</table>
Table 4-9. Comparison of scatter fractions obtained in this study to those of Ducote and Molloi. All data are for W/Rh with a grid.

<table>
<thead>
<tr>
<th>Tube potential (kVp)</th>
<th>Phantom thickness (cm)</th>
<th>Ducote and Molloi</th>
<th>Leon</th>
<th>Percent difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>24</td>
<td>1.08</td>
<td>0.02</td>
<td>0.04</td>
<td>-64%</td>
</tr>
<tr>
<td></td>
<td>2.17</td>
<td>0.04</td>
<td>0.05</td>
<td>-29%</td>
</tr>
<tr>
<td></td>
<td>4.82</td>
<td>0.08</td>
<td>0.09</td>
<td>10%</td>
</tr>
<tr>
<td></td>
<td>6.99</td>
<td>0.11</td>
<td>0.12</td>
<td>8%</td>
</tr>
<tr>
<td></td>
<td>9.64</td>
<td>0.13</td>
<td>0.16</td>
<td>22%</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Tube potential (kVp)</th>
<th>Phantom thickness (cm)</th>
<th>Ducote and Molloi</th>
<th>Leon</th>
<th>Percent difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>28</td>
<td>24</td>
<td>0.02</td>
<td>0.04</td>
<td>-65%</td>
</tr>
<tr>
<td></td>
<td>26</td>
<td>0.04</td>
<td>0.05</td>
<td>-29%</td>
</tr>
<tr>
<td></td>
<td>28</td>
<td>0.09</td>
<td>0.12</td>
<td>8%</td>
</tr>
<tr>
<td></td>
<td>30</td>
<td>0.12</td>
<td>0.16</td>
<td>22%</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Tube potential (kVp)</th>
<th>Phantom thickness (cm)</th>
<th>Ducote and Molloi</th>
<th>Leon</th>
<th>Percent difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>31</td>
<td>24</td>
<td>0.02</td>
<td>0.04</td>
<td>-60%</td>
</tr>
<tr>
<td></td>
<td>26</td>
<td>0.04</td>
<td>0.05</td>
<td>-29%</td>
</tr>
<tr>
<td></td>
<td>28</td>
<td>0.09</td>
<td>0.12</td>
<td>14%</td>
</tr>
<tr>
<td></td>
<td>30</td>
<td>0.12</td>
<td>0.16</td>
<td>4%</td>
</tr>
<tr>
<td></td>
<td>34</td>
<td>0.16</td>
<td>0.17</td>
<td>5%</td>
</tr>
</tbody>
</table>

Table 4-10. Comparison of the SPR values obtained in this study to those of Boone et al. All data are for Mo/Mo without a grid. The values from Boone et al. are read from Figure 1(a) of that publication and thus are approximate.

<table>
<thead>
<tr>
<th>Phantom thickness (cm)</th>
<th>Tube potential (kVp)</th>
<th>SPR Boone et al.</th>
<th>SPR Leon</th>
<th>Percent difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>24</td>
<td>0.23</td>
<td>0.33</td>
<td>35%</td>
</tr>
<tr>
<td>2</td>
<td>26</td>
<td>0.23</td>
<td>0.33</td>
<td>35%</td>
</tr>
<tr>
<td>2</td>
<td>28</td>
<td>0.23</td>
<td>0.33</td>
<td>36%</td>
</tr>
<tr>
<td>2</td>
<td>30</td>
<td>0.23</td>
<td>0.33</td>
<td>36%</td>
</tr>
<tr>
<td>2</td>
<td>34</td>
<td>0.23</td>
<td>0.33</td>
<td>37%</td>
</tr>
<tr>
<td>4</td>
<td>24</td>
<td>0.46</td>
<td>0.58</td>
<td>24%</td>
</tr>
<tr>
<td>4</td>
<td>26</td>
<td>0.46</td>
<td>0.58</td>
<td>23%</td>
</tr>
<tr>
<td>4</td>
<td>28</td>
<td>0.47</td>
<td>0.58</td>
<td>22%</td>
</tr>
<tr>
<td>4</td>
<td>30</td>
<td>0.47</td>
<td>0.58</td>
<td>20%</td>
</tr>
<tr>
<td>4</td>
<td>34</td>
<td>0.48</td>
<td>0.58</td>
<td>19%</td>
</tr>
<tr>
<td>6</td>
<td>24</td>
<td>0.68</td>
<td>0.83</td>
<td>20%</td>
</tr>
<tr>
<td>6</td>
<td>26</td>
<td>0.69</td>
<td>0.83</td>
<td>18%</td>
</tr>
<tr>
<td>6</td>
<td>28</td>
<td>0.70</td>
<td>0.83</td>
<td>17%</td>
</tr>
<tr>
<td>6</td>
<td>30</td>
<td>0.71</td>
<td>0.82</td>
<td>14%</td>
</tr>
<tr>
<td>6</td>
<td>32</td>
<td>0.72</td>
<td>0.82</td>
<td>13%</td>
</tr>
<tr>
<td>8</td>
<td>24</td>
<td>1.00</td>
<td>1.06</td>
<td>5.6%</td>
</tr>
<tr>
<td>8</td>
<td>26</td>
<td>1.01</td>
<td>1.09</td>
<td>7.2%</td>
</tr>
<tr>
<td>8</td>
<td>28</td>
<td>1.02</td>
<td>1.08</td>
<td>6.5%</td>
</tr>
<tr>
<td>8</td>
<td>30</td>
<td>1.02</td>
<td>1.06</td>
<td>4.3%</td>
</tr>
<tr>
<td>8</td>
<td>32</td>
<td>1.03</td>
<td>1.06</td>
<td>3.3%</td>
</tr>
</tbody>
</table>
Table 4-11. Comparison of the SPR values obtained in this study to those of Barnes and Brezovich. All data are without a grid. Barnes and Brezovich used a W/Al spectrum at 27, 32, and 36 kVp; this study used a W/Ag spectrum at 27, 33, and 36 kVp.

<table>
<thead>
<tr>
<th>Tube voltage (kVp)</th>
<th>3 cm phantom thickness</th>
<th>4 cm phantom thickness</th>
<th>5 cm phantom thickness</th>
<th>6 cm phantom thickness</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>B&amp;B</td>
<td>Leon</td>
<td>% Difference</td>
<td>B&amp;B</td>
</tr>
<tr>
<td>27</td>
<td>0.41</td>
<td>0.45</td>
<td>8.2%</td>
<td>0.55</td>
</tr>
<tr>
<td>32/33</td>
<td>0.40</td>
<td>0.44</td>
<td>10.6%</td>
<td>0.54</td>
</tr>
<tr>
<td>36</td>
<td>0.40</td>
<td>0.45</td>
<td>11.5%</td>
<td>0.54</td>
</tr>
</tbody>
</table>
Table 4-12. Comparison of the with-grid SPR values measured in this study to those simulated by Dance and Day. The data from Dance and Day is restricted to spectra of less than 36 keV; the range of spectra used for my data was 24-36 kVp, with all target/filter combinations included.

<table>
<thead>
<tr>
<th>Thickness (cm)</th>
<th>SPR range D&amp;D</th>
<th>SPR range Leon</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>0.055-0.095</td>
<td>0.049-0.060</td>
</tr>
<tr>
<td>4</td>
<td>0.091-0.20</td>
<td>0.082-0.099</td>
</tr>
<tr>
<td>6</td>
<td>0.11-0.27</td>
<td>0.11-0.15</td>
</tr>
<tr>
<td>8</td>
<td>0.17-0.35</td>
<td>0.15-0.21</td>
</tr>
</tbody>
</table>

Table 4-13. Tube potentials at which the minimum MRE is observed, for all target/filter combinations, with and without a grid.

<table>
<thead>
<tr>
<th>Thickness (cm)</th>
<th>With grid</th>
<th>Without grid</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mo/Mo</td>
<td>Mo/Rh</td>
</tr>
<tr>
<td>2</td>
<td>26.3</td>
<td>27.0</td>
</tr>
<tr>
<td>4</td>
<td>27.4</td>
<td>27.5</td>
</tr>
<tr>
<td>6</td>
<td>28.2</td>
<td>28.9</td>
</tr>
<tr>
<td>8</td>
<td>27.4</td>
<td>28.6</td>
</tr>
</tbody>
</table>
Table 4-14. Comparison of mean radial extents obtained in this study to those of Ducote and Molloi. All data are for W/Rh with a grid. Values are in millimeters.

<table>
<thead>
<tr>
<th>Phantom thickness (cm)</th>
<th>1.08</th>
<th>2.17</th>
<th>4.82</th>
<th>6.99</th>
<th>9.64</th>
</tr>
</thead>
<tbody>
<tr>
<td>24 kVp Ducote and Molloi</td>
<td>2.30</td>
<td>2.80</td>
<td>5.70</td>
<td>7.50</td>
<td>11.50</td>
</tr>
<tr>
<td>Leon</td>
<td>2.99</td>
<td>3.62</td>
<td>5.96</td>
<td>8.51</td>
<td>12.08</td>
</tr>
<tr>
<td>Percent difference</td>
<td>26%</td>
<td>26%</td>
<td>4%</td>
<td>13%</td>
<td>5%</td>
</tr>
<tr>
<td>28 kVp Ducote and Molloi</td>
<td>2.20</td>
<td>2.90</td>
<td>5.00</td>
<td>6.10</td>
<td>8.30</td>
</tr>
<tr>
<td>Leon</td>
<td>3.52</td>
<td>3.84</td>
<td>5.57</td>
<td>7.77</td>
<td>11.05</td>
</tr>
<tr>
<td>Percent difference</td>
<td>46%</td>
<td>28%</td>
<td>11%</td>
<td>24%</td>
<td>28%</td>
</tr>
<tr>
<td>31 kVp Ducote and Molloi</td>
<td>2.80</td>
<td>3.40</td>
<td>5.40</td>
<td>7.60</td>
<td>10.20</td>
</tr>
<tr>
<td>Leon</td>
<td>4.13</td>
<td>4.44</td>
<td>6.24</td>
<td>8.58</td>
<td>12.18</td>
</tr>
<tr>
<td>Percent difference</td>
<td>38%</td>
<td>27%</td>
<td>14%</td>
<td>12%</td>
<td>18%</td>
</tr>
</tbody>
</table>
Figure 4-1. LSR values obtained during the apparent thickness measurements. Shown is the graph of Mo/Mo, using the large paddle and a grid. The diamonds and associated error bars indicate data points acquired on the mammography unit, whereas the stars indicate the points calculated for each tube potential using the fit equations. A separate graph was generated for every combination of target/filter, grid status, and paddle size (16 total graphs). The other 15 graphs were similar in appearance to this one and are not shown. The data from the fits can be found in Table B-1.
Figure 4-2. The fitted LSR values from Figure 4-1, replotted as a function of thickness instead of kVp. The coefficients from these fitted lines were used to create a table of LSR’s for all values of kVp and thickness. $R^2$ is equal to 1 for all curves. A separate graph was generated for every combination of target/filter, grid status, and paddle size (16 total graphs). The other graphs were similar in appearance to this one and are not shown. The data from the fits can be found in Table B-2.
Figure 4-3. Graph of LSR measurements as a function of disk radius for 4 cm BR12, Mo/Mo, with grid. Fitted linear equations and associated $R^2$ values are shown. A separate graph was generated for every combination of target/filter, grid status, and BR12 thickness (32 total graphs). The other 31 graphs were similar in appearance to this one and are not shown, but the fits all had $R^2$ values greater than 0.99. The data from the fits of all 32 graphs was used to produce the SF and MRE values shown in Tables 4-1 through 4-8.
Figure 4-4. Sample graph of LSR as a function of disk radius, showing A) a measurement (circled) not conforming well to the best fit line, and B) the result after three measurements at that disk radius were averaged. The $R^2$ goodness-of-fit also shows improvement.
Figure 4-5. Surface fits for A) the scatter fraction and B) mean radial extent for Mo/Mo, with a grid. A separate surface fit was obtained for each target/filter combination (4 SF fits and 4 MRE fits). The remaining graphs look very similar to these and are not shown. The coefficients generated by Matlab® for each of the fits can be found in Table B-3.
Figure 4-6. Scatter fraction as a function of tube potential, for each of the target/filter combinations and thicknesses, with a grid.
Figure 4-7. Scatter fraction as a function of tube potential, for each of the target/filter combinations and thicknesses, without a grid.
Figure 4-8. Mean radial extent as a function of tube potential, with a grid, for A) the Mo/Mo and Mo/Rh spectra and B) the W/Rh and W/Ag spectra.
Figure 4-9. Mean radial extent as a function of tube potential, without a grid, for A) the Mo/Mo and Mo/Rh spectra and B) the W/Rh and W/Ag spectra.
Figure 4-10. The SPR as a function of tube potential for each of the target/filter combinations, with a grid.
Figure 4-11. The SPR as a function of tube potential for each of the target/filter combinations, without a grid.
Figure 4-12. Effect of phantom size on the scatter kernel measurements. Techniques used were 24 kV, 40 mAs, W/Rh, without a grid.
CHAPTER 5
MONTE CARLO SIMULATIONS

The Monte Carlo simulations had two purposes in my research. The first was to assess the accuracy of the measured scatter fraction (SF) and mean radial extent (MRE) found from the physical measurements discussed in Chapter 4. The second was to determine what effect tissue glandularity might have on the SF, MRE, and apparent thickness measurements.

Computer Hardware and Software

MCNPX (version 2.6.0, Radiation Safety Information Computational Center, Oak Ridge, TN) was used for the Monte Carlo simulations. A custom-built personal computer running Windows® 7 (Microsoft®, Redmond, WA) was used to run the simulations. Because processing time limitations were a major concern that affected the design of the Monte Carlo simulations, it is relevant to include a list of the most important hardware specifications (Table 5-1).

Design of Geometry and Materials

The mammography unit geometry was simulated to conform to a clinical Selenia® system as closely as possible (Figure 5-1). The SID was 66 cm, with 2 cm between the breast support and the selenium detector. The breast support was 1 mm-thick carbon fiber and the compression paddle was 2 mm-thick polycarbonate. Details of the elemental compositions and densities used in the simulations can be found in Table 5-2.

The Hologic™ Selenia® grid is a focused cellular copper grid with air interspaces. Due to the difficulty of simulating a focused cellular grid, the cellular grid was approximated by two focused linear grids oriented perpendicularly to each other. The thickness and height of the septa, line density, and grid-to-detector distance were
modeled using proprietary information provided by Hologic™. Grid motion was not simulated; however, the effect of grid attenuation was considered and found to have a minimal effect on the SF and MRE. Details of that simulation will be described in more detail later in this chapter.

Breasts with thicknesses of 2 cm, 4 cm, 6 cm, and 8 cm were modeled using the elemental compositions and densities listed for adipose tissue and breast (mammary gland) tissue found in ICRU Report 44. Most simulations were performed using a 50% glandular /50%adipose (hereafter referred to simply as 50/50) breast composition, although experiments to assess the effects of breast density were also performed using compositions of 30/70 and 70/30. The elemental compositions and densities of each breast composition were calculated using the appropriate weighted average for adipose and glandular tissue. The breast material was modeled as a slab that covered the entire detector to increase simulation speed. The impact of using such a large slab for comparison with the much smaller BR-12 was considered in a separate simulation, to be described later in this chapter, and found to be minimal.

The x-ray spectra were not calculated using MCNPX in order to conserve processing time and to avoid setting up the fairly complicated geometry of the x-ray tube. Instead, the unfiltered spectra for the Mo and W targets were first calculated using the MAMSPEC program created by Boone, Fewell, and Jennings (available online from the American Institute of Physics website). This program produces spectra in the range of 18-42 kV, which are filtered only by the inherent filtration of the target and the beryllium window. As downloaded, the program produced spectra only in increments of 5 kV, but the data files contained the data to produce spectra in increments of 1 kV. The
program source code (written in the C language (Bell Labs/Lucent Technologies, Murray Hill, NJ)) was included in the software package, so the source code was modified to produce spectra in increments of 1 kV and recompiled. The output of these files consisted of un-normalized data in the form of number of photons per 0.5 keV interval. These data were imported into Microsoft® Excel® in order to calculate the percent contribution of each 0.5 kV interval to the total number of photons, which is the format used by MCNPX. The correct spectrum for each simulation was then copied into the appropriate MCNPX input file. Filtration of the spectrum was performed in MCNPX by the inclusion of a slab of Mo, Rh, or Ag as appropriate (30 µm thick for Mo/Mo and Mo/Rh, and 58 µm thick for W/Rh and W/Ag).

**Verification of the Scatter Kernel Data**

**Simulation Design**

The first simulation design attempted was essentially identical to that used for the experimental measurements: the radiation field was emitted from a 0.3 mm focal spot located at the position of the physical focal spot, the x-ray field covered the entire detector, and the measurements were obtained by placing tallies (i.e., user-defined radiation detectors) under simulated lead disks. However, this method proved to be impractical due to the excessive calculation times needed to obtain adequate statistics, even after a variety of variance reduction strategies were applied.

Upon further contemplation, it became apparent that replicating the experimental set-up was unnecessary. In theory, the scatter kernel is the response function to a perfect impulse of particles. The experimental set-up using the lead disks is a clever way to get around the limitations of a physical system that is incapable of producing an
infinitesimally narrow stream of particles, but MCNPX has no such limitations. Rather than simulating the experimental set-up, the flexibility of MCNPX was used to produce the scatter kernel directly, by simulating a perfect impulse and measuring the response.

The two components of the scatter kernel that had to be verified, the MRE and SF, were assessed separately. The same basic set-up was used for both measurements: a single impulse directed at a location on the detector centered from left-to-right and 5.5 cm from the chest wall edge (which corresponds to the location of the ROI's in the physical measurements). Only the tallies and analyses of the results differed between the SF and MRE measurements, as described below. All tally types used in these simulations measured energy deposition, which is the quantity detected by the a-Se x-ray detector being modeled.\textsuperscript{38} For all simulations, a sufficient number particles were used to reduce the relative error below 5% and obtain passing scores in the statistical checked performed by the MCNPX program. Both Compton and coherent scattering interactions were simulated.

The only variance reduction technique included in the code was the use of cell importances for geometry splitting.\textsuperscript{60} This technique works by increasing the number of particles that reach cells close to the tally, but decreasing their weight (i.e., contribution to the final tally) proportionally, depending on the ratio of the cell importances. For example, if a particle moves from a cell with an importance of 8 to a cell with an importance of 16, the particle is split into 2 particles (the ratio 16/8), and both particles are followed independently from that point. However, the weight of these particles is decreased by the same ratio, with each assigned a weight of half that of the original particle. Thus, the total number of particles which might reach a tally is increased,
resulting in a decreased variance. However, the total energy deposition recorded by the
tallies is not affected because of the reduction in weight.

Several sets of simulations were performed. The first set consisted of simulations
intended to be roughly equivalent to clinical techniques. These simulations were
performed for breast thicknesses of 2, 4, 6, and 8 cm using tube potentials similar to
those found in the facility technique charts for those thicknesses (24, 28, 30, and 32 kVp
respectively). Although the selection of target/filter combination for a clinical technique
also depends on thickness, all target/filter combinations were simulated with all the
thicknesses listed above in order to produce complete data sets. The effect of tube
potential on the fit was then isolated by simulating 24, 26, 28, 30, and 32 kVp Mo/Mo
and W/Rh spectra with the 4 cm thickness. Mo/Mo and W/Rh were chosen out of the
four target/filter combinations to conserve processing time and because they represent
two very different spectra (i.e., the presence of large characteristic K x-ray peaks from
the Mo target, as opposed to the lack of such peaks in the W-target spectrum). The
effect of thickness was isolated by simulating phantoms of 2, 4, 6, and 8 cm with 30 kVp
Mo/Mo and W/Rh spectra.

The Monte Carlo codes for the simulations performed using a 28 kVp Mo/Mo
spectrum and a phantom thickness of 4 cm, with and without a grid, are shown in
Appendix C. The simulations performed using other thicknesses and spectra entailed
only minor modifications to these codes.

**Scatter fraction**

To measure the SF, a single tally covering the entire surface of the detector was
used. The tally was *F1* type, which reports the energy flux through a surface. This tally
type also allows the data to be grouped by angle of impact, which was used to separate
particles that were definitely scattered from those that were likely to be primary (i.e., not scattered) because they had the same angle of impact as the impulse. Although it is possible that a photon could undergo multiple scatterings that would result in an angle of impact identical to the angle of the impulse, the number of such particles would be very small in relation to the total number of primary particles. Thus, their contribution to the energy flux attributed to primary radiation was assumed to be negligible. The ratio of the energy flux from scattered particles to the total energy flux (scattered + primary) gives the SF. Because the number of particles was selected to keep the relative error of each measurement to less than 5%, the extrapolated error of the SF is ~7%.

The results of the simulations are compared to the scatter fractions generated from the physical measurements in Tables 5-3 through 5-5, without a grid, and Tables 5-6 through 5-8, with a grid. The absolute differences between the measured and simulated scatter fractions are very similar both with and without a grid: they range from -0.04 to +0.01 (average -0.01) without a grid and -0.03 to +0.07 (average 0.02) with a grid.

Without a grid, the Monte Carlo-derived SF ranged from 0.28 to 0.51, compared to a range of 0.25 to 0.52 in the measured SF. In all cases, the measured SF was within the 95% confidence interval of the Monte Carlo data. The SF increased as thickness increased, with essentially no dependence on tube potential or target/filter combination. The conformance between the physical and Monte Carlo measurements was best at a thickness of 6 cm for the Mo-target spectra and 8 cm for the W-target spectra, although the deviation was not great at any thickness.
With a grid, the Monte Carlo-derived SF ranged from a low of 0.065 (for a 28 kVp W/Ag spectrum and 4 cm thickness) to a high of 0.089 (for a 32 kVp Mo/Mo spectrum and 8 cm thickness). This range is less than that seen in the physical data, which demonstrated a low of 0.049 and a high of 0.15. The Monte Carlo SF’s were greater than the measured SF’s at 2 cm, roughly equal at 4 cm, and less at 6 cm and 8 cm, indicating less variation in SF as a function of thickness than was found with the physical measurements. The measured SF’s were within the 95% confidence intervals of the Monte Carlo data for most of the 4 cm-thickness simulations, but were outside for 2, 6, and 8 cm for all combinations of tube potential, target, and filter. The trends seen in the clinical techniques also differed between the Monte Carlo and the measured SF values; the measured SF’s all increased as thickness increased, but the Monte Carlo SF’s demonstrated a minimum at 4 cm, for all target/filter combinations.

The agreement with a grid is not as good as the agreement without a grid, but it is still not unreasonable considering the complexity of the simulation. Some of the discrepancy may be due to inaccuracies in the simulation of the grid, which, as explained earlier, was simulated using two perpendicular linear grids rather than a cellular grid. Some of the inaccuracy is likely due to other sources of systematic error in the simulation or in the acquisition of the physical measurements. One possibility is a difference in the absorption characteristics of the ideal detector used in the Monte Carlo simulation, which records all energy flux entering the surface of the detector, compared to the physical detector absorption and translation to pixel values. The observation that the absolute differences between the measured and Monte Carlo SF values are very similar with and without a grid lends support to the theory of a source of systematic error
apart from the grid design. Because the SF’s with a grid are so much smaller than the SF's without a grid, the errors comprise a larger fraction of the SF, resulting in worse agreement between the two data sets.

**Mean radial extent**

The MRE is an indication of how far the scattered radiation spreads from the point of impact; this quantity was determined by using ring-shaped tallies placed in concentric circles around the impulse (Figure 5-2). The thickness of each ring tally was 2 mm for the non-grid simulations. With a grid, 1 mm-thick tallies were used closer to the impulse, where the slope of the PSF changed rapidly, but these were changed to thicknesses of 5 mm at radii greater than 10 mm due to the limited flux and the slow rate of change at these distances. The minimum average tally radius for all simulations was 1 mm and the maximum radius was 55 mm. These tallies were of type F6, which reports results in units of MeV/g, so the fact that each detector ring had a different volume did not affect the measurements because the reported results were normalized to the mass included in each tally. The PSF was technically measured in a form that was integrated over 360 degrees (i.e., $\text{PSF}_{\text{measured}} = \int_0^{2\pi} \text{PSF} \cdot rdr\theta$). The true measurement for a tally defined by rings of radii $r_n$ and $r_{n+1}$ can be obtained from the integrated PSF by division by $\pi(r_{n+1}^2 - r_n^2)$, but this calculation has already been performed on the reported data due to use of the F6 tally type. Thus, no additional correction was performed.

Using the Matlab® (version 7.10.0.499 (R2010a), MathWorks®, Natick, MA) Curve Fitting Toolbox, the data acquired from the simulations were then fit to the PSF:

$$h(r) = \frac{c}{kr} e^{-r/k} \quad (5-1)$$
where $C$ was a constant which was allowed to vary to improve the fit, and the other variables are as described previously. The value $k$ was set equal to the MRE acquired from the surface fit of the physical data (described in Chapter 4) for the appropriate set of techniques. Although $C$ is, in theory, related to the SF, the fit was not used to determine the SF because the tallies provided values for energy absorbed (in MeV/g) that were not normalized. However, the accuracy of the comparison between the measured and simulated MRE was not affected by this lack of normalization. The reported values for the $R^2$ goodness-of-fit for each data set are shown in Tables 5-9 though 5-14. Graphs of the data and fitted PSF curves can be seen in Figures 5-3 and 5-4.

The Monte Carlo simulations using the grid conformed very well to the PSF equation, with all $R^2$ values greater than 0.96. Without a grid, the $R^2$ values ranged from 0.65 to 0.86, indicating that the PSF does not conform as well to the non-grid Monte Carlo data. The fits without a grid were better for the smaller breast thicknesses, decreasing from an average $R^2$ value of 0.84 at 2 cm to an average $R^2$ value of 0.67 at 8 cm. The goodness of fit did not depend on tube potential.

Interestingly, experimentation with different functions indicated that the data could be fit almost perfectly to a bi-exponential function of the form:

$$f(x) = ae^{-bx} + ce^{-dx}$$  \hspace{1cm} (5-2)

where $a$, $b$, $c$, and $d$ were variables determined by the Matlab® curve fitting algorithm (Figure 5-5). It seemed plausible that the reciprocal values of $b$ and $d$ could be analogous to the mean radial extent ($k$) in the PSF, suggesting that the true PSF was a function of two processes with different mean radial extents (Tables 5-15 and 5-16).
A possible explanation that fits this theory is that one exponential describes the contribution of coherent scatter, and one describes Compton scatter. This idea was tested by running the clinical-technique Mo/Mo simulations again, but removing the simulation of coherent scatter via the use of the "NOCOH" option that is built into MCNPX. All scatter events in this simulation were therefore the result of Compton scattering only; essentially, the coherent scattering cross-section was simply removed from the total scattering cross-section, so that the total scattering cross-section was equal to the Compton scattering cross-section. The contribution of coherent scattering was then isolated by subtracting the Compton scatter results from the original summed (coherent + Compton scatter) results. This process was performed both with and without a grid; the results of both simulations are shown in Figure 5-6.

Matlab® was used to fit exponential curves to the non-grid data for the 2 cm, 24 kV simulation (Figure 5-7). The Compton scattering data are obviously not the result of an exponential process at all distances, but limiting the fit to distances of at least 10 mm, where the Compton contribution is most significant, seemed reasonable. The resulting fitted coefficients were then compared to the coefficients obtained from the original bi-exponential fit (Table 5-17), with good agreement: a difference of only 0.45 mm in the first MRE and 0.2 mm in the second MRE. It is apparent from Figure 5-7 that the contribution from Compton scattering produces the term with the larger MRE, and coherent scattering contributes the shorter MRE.

The results of this investigation indicate that the separate contributions of coherent and Compton scatter are the likely cause of the bi-exponential phenomenon. The reason the data with a grid can be fit well with a single exponential term (Equation
5-1) is because the grid eliminates most of the Compton contribution--in other words, the scattered radiation detected with a grid in place is almost entirely the result of coherent scatter. Evidence for this lies in the ratio of the coefficients $a$ and $c$: this ratio is in the range of 1-3 without a grid, but 13-143 with a grid. Thus, the coherent term is at least an order of magnitude greater than the Compton term when a grid is used, but they are roughly equal without a grid.

One mystery that is not explained is that the $1/r$ term used in Equation 5-1 is not used in the bi-exponential fit. Attempts to add in this term generated worse rather than better fits, and the fits are already near-perfect without it. A theoretical explanation for this lack of the $1/r$ term is not obvious.

**Effects of Breast Composition**

There are two issues which were investigated with respect to breast composition: first, whether the SF or MRE changed due to glandularity differences, and second, how much error in the determination of apparent thickness might result from glandularity differences.

**Scatter Fraction and Mean Radial Extent**

The effects of breast composition on the SF and MRE were assessed by repeating the simulations for a 6 cm breast thickness at 30 kVp and all the target/filter combinations, but substituting the 30/70 and 70/30 breast compositions for the 50/50 mix used in the prior simulations. (Although Boone simulated glandularity at the extremes of 0% and 100%, the range used in my study was limited to that more commonly seen in clinical mammography.) Simulations were performed both with and without the grid. The impact on SF was less than 1% for all measurements (Table 5-18).
The results for the MRE, as demonstrated by the graphs of several points along the scatter kernels (Figure 5-8) also show little discernable impact.

**Apparent Thickness**

The effect of breast composition on apparent thickness was determined by acquiring a tally with 6 cm of simulated breast in the beam, then removing the breast material and acquiring a second tally using otherwise identical conditions. The apparent thickness was then calculated using the process described in Chapter 4, with the tally results in place of ROI values. The apparent thickness was calculated for 50/50, 30/70, and 70/30 breast compositions, and the percent error was calculated relative to the 50/50 composition (Table 5-19). The error averaged -5.2% for the fatty breast composition and +5.2% for the dense breast composition, with essentially identical results regardless of whether a grid was used. The change in the scatter kernel (SF or MRE) over the approximately 3 mm that error represents is negligible, which supports Hypothesis 3-1.

Neither the apparent thickness nor the scatter kernel are appreciably affected by breast composition. Thus, is it reasonable to expect that the scatter correction algorithm performs equally well for breasts of any density.

**Ancillary Investigations**

**Grid Attenuation**

As mentioned previously, grid motion was not modeled in the simulation. In order to avoid non-uniformities resulting from the grid in a clinical image, it is a reasonable assumption that all areas of the detector spend an equal proportion of the exposure time covered by a grid line when a moving grid is used. Because those grid lines must always be somewhere, it follows that the proportion of time a given pixel is covered by a
grid line is equal to the proportion of the grid that is actually composed of copper (as opposed to the air interspaces). This proportion is approximately 10%; the exact percentage was used in the below analysis, but is withheld from this publication because it is proprietary to Hologic™.

Thus, in a clinical system, a ray aimed at a particular pixel will be attenuated by a grid line 10% of the time. To determine what effect this attenuation would have on the scatter kernel, a Monte Carlo simulation was performed using a set-up identical to that described previously, except that the focal spot and tallies were shifted just enough to intersect a nearby grid line. A weighted average of the tally results (90% without the grid line and 10% with the grid line) was calculated. There was no effect on the resulting SF or MRE. This result made sense because the grid attenuated the incident ray by several orders of magnitude, essentially just resulting in a 10% reduction of the number of total incident particles. Neither the scatter fraction nor the mean radial extent are dependent on the total number of particles, so there was no discernable effect. The number of photons scattered from the grid and subsequently detected was also very small, presumably because relatively few scattered photons fell within the limited range of angles necessary to exit the cross-hatched grid.

**Phantom Size**

The simulations all employed a phantom that was as large as the detector, whereas the physical measurements used a phantom which was only 12.5 cm x 10 cm. It was already shown in Chapter 4 that the physical measurements were not affected by phantom size. To show that the Monte Carlo values were not affected, an alternate geometry using breast material the same size as the BR12 phantom was simulated. The remainder of the simulated geometry was identical to that previously described,
with a single F6 tally placed 5.5 cm from the chest wall and centered left-to-right. The tally results were compared to those from a second simulation with parameters identical except for the phantom size, which covered the entire detector. There was no difference in tally values between the two simulations.

**Conclusion**

Overall, the Monte Carlo simulations support the reasonableness of the SF and MRE values obtained from the physical measurements. They also suggest that breast glandularity will not present a problem for the scatter correction program. However, the data suggest that a bi-exponential PSF equation (Equation 5-2) would more accurately fit the data, particularly without a grid, than does the form used in this research (Equation 5-1). The two exponential terms likely come from the separate contributions of coherent and Compton scattering. Because it is not possible to separate these scatter sources during the acquisition of physical measurements, a scatter correction program using the bi-exponential PSF would have to use data solely from simulations.

The primary goal of this study is to assess the impact of the scatter correction program on clinical mammography, which uses a grid, and Equation 5-1 still fit the with-grid data extremely well. Thus, the mono-exponential PSF is used for the scatter correction program because the data are supported by the physical measurements and because the simpler form allows faster computation speeds. The composition of a full Monte Carlo-generated data set for use of the bi-exponential PSF is a topic for future research, particularly for applications of the scatter correction program to images acquired without a grid.
Table 5-1. Computer hardware specifications

<table>
<thead>
<tr>
<th>Component</th>
<th>Specification</th>
</tr>
</thead>
<tbody>
<tr>
<td>Processor</td>
<td>AMD Phenom™ II X6 1090T, 3200 MHz, 6 cores</td>
</tr>
<tr>
<td>RAM</td>
<td>DDR3 2000, 8 GB</td>
</tr>
<tr>
<td>Hard Drive</td>
<td>1 TB SATA, 7200 rpm</td>
</tr>
<tr>
<td>Operating System</td>
<td>Microsoft® Windows® 7 Professional, 64 bit</td>
</tr>
</tbody>
</table>

Table 5-2. Materials used in Monte Carlo simulations

<table>
<thead>
<tr>
<th>Material</th>
<th>Elemental Composition</th>
<th>Density (g/cm³)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Carbon Fiber</td>
<td>Approximated as pure C</td>
<td>1.45</td>
</tr>
<tr>
<td>Polycarbonate</td>
<td>C₅H₈O₂</td>
<td>1.20</td>
</tr>
<tr>
<td>Amorphous Selenium</td>
<td>Se</td>
<td>4.27</td>
</tr>
<tr>
<td>Molybdenum</td>
<td>Mo</td>
<td>10.28</td>
</tr>
<tr>
<td>Rhodium</td>
<td>Rh</td>
<td>12.41</td>
</tr>
<tr>
<td>Silver</td>
<td>Ag</td>
<td>10.49</td>
</tr>
<tr>
<td>Adipose Tissue</td>
<td>H: 11.4%, C: 59.8%, N: 0.7%, O: 27.8%, Na: 0.1%, S: 0.1%, Cl: 0.1%</td>
<td>0.950</td>
</tr>
<tr>
<td>Glandular Tissue</td>
<td>H: 10.6%, C: 33.2%, N: 3.0%, O: 52.7%, Na: 0.1%, P: 0.1%, S: 0.2%, Cl: 0.1%</td>
<td>1.020</td>
</tr>
</tbody>
</table>
Table 5-3. SF values for representative clinical techniques, without a grid, showing physical measurements, Monte Carlo simulations with 95% confidence intervals, and differences between the two.

<table>
<thead>
<tr>
<th>Target</th>
<th>Filter</th>
<th>Thickness (cm)</th>
<th>kVp</th>
<th>SF Measured</th>
<th>SF Monte Carlo</th>
<th>Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo</td>
<td>Mo</td>
<td>2</td>
<td>24</td>
<td>0.25</td>
<td>0.29 (0.25,0.32)</td>
<td>-0.04</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>28</td>
<td>0.37</td>
<td>0.38 (0.32,0.43)</td>
<td>-0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>0.45</td>
<td>0.45 (0.39,0.51)</td>
<td>0.00</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>32</td>
<td>0.52</td>
<td>0.51 (0.44,0.58)</td>
<td>0.01</td>
</tr>
<tr>
<td>Mo</td>
<td>Rh</td>
<td>2</td>
<td>24</td>
<td>0.25</td>
<td>0.28 (0.24,0.32)</td>
<td>-0.03</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>28</td>
<td>0.36</td>
<td>0.37 (0.32,0.43)</td>
<td>-0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>0.45</td>
<td>0.45 (0.39,0.51)</td>
<td>0.00</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>32</td>
<td>0.52</td>
<td>0.51 (0.44,0.58)</td>
<td>0.01</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>2</td>
<td>24</td>
<td>0.25</td>
<td>0.28 (0.24,0.32)</td>
<td>-0.03</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>28</td>
<td>0.36</td>
<td>0.37 (0.32,0.43)</td>
<td>-0.02</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>0.44</td>
<td>0.45 (0.39,0.51)</td>
<td>-0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>32</td>
<td>0.51</td>
<td>0.51 (0.44,0.58)</td>
<td>0.00</td>
</tr>
<tr>
<td>W</td>
<td>Ag</td>
<td>2</td>
<td>24</td>
<td>0.25</td>
<td>0.28 (0.24,0.32)</td>
<td>-0.03</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>28</td>
<td>0.36</td>
<td>0.37 (0.32,0.42)</td>
<td>-0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>0.44</td>
<td>0.45 (0.39,0.51)</td>
<td>-0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>32</td>
<td>0.51</td>
<td>0.51 (0.44,0.58)</td>
<td>0.00</td>
</tr>
</tbody>
</table>
Table 5-4. SF values for a 4 cm phantom thickness and tube potentials ranging from 24-32 kVp, without a grid, showing physical measurements, Monte Carlo simulations with 95% confidence intervals, and differences between the two.

<table>
<thead>
<tr>
<th>Target</th>
<th>Filter</th>
<th>kVp</th>
<th>SF Measured</th>
<th>SF Monte Carlo</th>
<th>Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo</td>
<td>Mo</td>
<td>24</td>
<td>0.37</td>
<td>0.38 (0.33,0.43)</td>
<td>-0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td>26</td>
<td>0.37</td>
<td>0.38 (0.33,0.43)</td>
<td>-0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td>28</td>
<td>0.37</td>
<td>0.38 (0.32,0.43)</td>
<td>-0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td>30</td>
<td>0.37</td>
<td>0.37 (0.32,0.43)</td>
<td>-0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td>32</td>
<td>0.37</td>
<td>0.37 (0.32,0.42)</td>
<td>-0.01</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>24</td>
<td>0.36</td>
<td>0.38 (0.32,0.43)</td>
<td>-0.02</td>
</tr>
<tr>
<td></td>
<td></td>
<td>26</td>
<td>0.36</td>
<td>0.37 (0.32,0.43)</td>
<td>-0.02</td>
</tr>
<tr>
<td></td>
<td></td>
<td>28</td>
<td>0.36</td>
<td>0.37 (0.32,0.43)</td>
<td>-0.02</td>
</tr>
<tr>
<td></td>
<td></td>
<td>30</td>
<td>0.36</td>
<td>0.37 (0.32,0.43)</td>
<td>-0.02</td>
</tr>
<tr>
<td></td>
<td></td>
<td>32</td>
<td>0.36</td>
<td>0.38 (0.32,0.43)</td>
<td>-0.02</td>
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</table>
Table 5-5. SF values for a tube potential of 30 kVp and phantom thicknesses ranging from 2-8 cm, without a grid, showing physical measurements, Monte Carlo simulations with 95% confidence intervals, and differences between the two.

<table>
<thead>
<tr>
<th>Target</th>
<th>Filter</th>
<th>Thickness (cm)</th>
<th>SF Measured</th>
<th>SF Monte Carlo</th>
<th>Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo</td>
<td>Mo</td>
<td>2</td>
<td>0.25</td>
<td>0.28 (0.24,0.32)</td>
<td>-0.03</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>0.37</td>
<td>0.37 (0.32,0.43)</td>
<td>-0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>0.45</td>
<td>0.45 (0.39,0.51)</td>
<td>0.00</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>0.52</td>
<td>0.51 (0.44,0.58)</td>
<td>0.01</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>2</td>
<td>0.25</td>
<td>0.28 (0.24,0.32)</td>
<td>-0.03</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>0.36</td>
<td>0.37 (0.32,0.43)</td>
<td>-0.02</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>0.44</td>
<td>0.45 (0.39,0.51)</td>
<td>-0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>0.51</td>
<td>0.51 (0.44,0.58)</td>
<td>0.01</td>
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</table>
Table 5-6. SF values for representative clinical techniques, with a grid, showing physical measurements, Monte Carlo simulations with 95% confidence intervals, and differences between the two.

<table>
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<th>Target</th>
<th>Filter</th>
<th>Thickness (cm)</th>
<th>kVp</th>
<th>SF Measured</th>
<th>SF Monte Carlo</th>
<th>Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo</td>
<td>Mo</td>
<td>2</td>
<td>24</td>
<td>0.054</td>
<td>0.082 (0.070,0.093)</td>
<td>-0.028</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>28</td>
<td>0.080</td>
<td>0.076 (0.065,0.086)</td>
<td>0.004</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>0.11</td>
<td>0.076 (0.065,0.086)</td>
<td>0.032</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>32</td>
<td>0.15</td>
<td>0.089 (0.076,0.10)</td>
<td>0.066</td>
</tr>
<tr>
<td>Mo</td>
<td>Rh</td>
<td>2</td>
<td>24</td>
<td>0.054</td>
<td>0.076 (0.065,0.086)</td>
<td>-0.022</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>28</td>
<td>0.082</td>
<td>0.072 (0.062,0.082)</td>
<td>0.010</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>0.11</td>
<td>0.075 (0.064,0.085)</td>
<td>0.033</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>32</td>
<td>0.15</td>
<td>0.086 (0.074,0.098)</td>
<td>0.064</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>2</td>
<td>24</td>
<td>0.051</td>
<td>0.072 (0.062,0.082)</td>
<td>-0.021</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>28</td>
<td>0.078</td>
<td>0.071 (0.061,0.080)</td>
<td>0.007</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>0.11</td>
<td>0.075 (0.065,0.086)</td>
<td>0.032</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>32</td>
<td>0.15</td>
<td>0.087 (0.075,0.099)</td>
<td>0.058</td>
</tr>
<tr>
<td>W</td>
<td>Ag</td>
<td>2</td>
<td>24</td>
<td>0.049</td>
<td>0.071 (0.061,0.081)</td>
<td>-0.023</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>28</td>
<td>0.079</td>
<td>0.065 (0.056,0.074)</td>
<td>0.013</td>
</tr>
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<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>0.11</td>
<td>0.071 (0.061,0.081)</td>
<td>0.036</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>32</td>
<td>0.14</td>
<td>0.083 (0.071,0.094)</td>
<td>0.059</td>
</tr>
</tbody>
</table>
Table 5-7. SF values for a tube potential of 30 kVp and phantom thicknesses ranging from 2-8 cm, with a grid, showing physical measurements, Monte Carlo simulations with 95% confidence intervals, and differences between the two.

<table>
<thead>
<tr>
<th>Target</th>
<th>Filter</th>
<th>kVp</th>
<th>SF Measured</th>
<th>SF Monte Carlo</th>
<th>Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo</td>
<td>Mo</td>
<td>24</td>
<td>0.077</td>
<td>0.081 (0.070,0.092)</td>
<td>-0.004</td>
</tr>
<tr>
<td></td>
<td></td>
<td>26</td>
<td>0.078</td>
<td>0.078 (0.067,0.089)</td>
<td>-0.001</td>
</tr>
<tr>
<td></td>
<td></td>
<td>28</td>
<td>0.080</td>
<td>0.076 (0.065,0.086)</td>
<td>0.004</td>
</tr>
<tr>
<td></td>
<td></td>
<td>30</td>
<td>0.082</td>
<td>0.073 (0.063,0.083)</td>
<td>0.009</td>
</tr>
<tr>
<td></td>
<td></td>
<td>32</td>
<td>0.084</td>
<td>0.072 (0.062,0.082)</td>
<td>0.013</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>24</td>
<td>0.078</td>
<td>0.072 (0.062,0.083)</td>
<td>0.005</td>
</tr>
<tr>
<td></td>
<td></td>
<td>26</td>
<td>0.077</td>
<td>0.071 (0.061,0.081)</td>
<td>0.006</td>
</tr>
<tr>
<td></td>
<td></td>
<td>28</td>
<td>0.078</td>
<td>0.071 (0.061,0.080)</td>
<td>0.007</td>
</tr>
<tr>
<td></td>
<td></td>
<td>30</td>
<td>0.079</td>
<td>0.071 (0.061,0.080)</td>
<td>0.008</td>
</tr>
<tr>
<td></td>
<td></td>
<td>32</td>
<td>0.082</td>
<td>0.075 (0.064,0.085)</td>
<td>0.007</td>
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</table>
Table 5-8. SF values for a tube potential of 30 kVp and phantom thicknesses ranging from 2-8 cm, with a grid, showing physical measurements, Monte Carlo simulations with 95% confidence intervals, and differences between the two.

<table>
<thead>
<tr>
<th>Target</th>
<th>Filter</th>
<th>Thickness (cm)</th>
<th>SF Measured</th>
<th>SF Monte Carlo</th>
<th>Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo</td>
<td>Mo</td>
<td>2</td>
<td>0.055</td>
<td>0.075 (0.064,0.085)</td>
<td>-0.020</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>0.082</td>
<td>0.073 (0.063,0.083)</td>
<td>0.009</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>0.11</td>
<td>0.076 (0.065,0.086)</td>
<td>0.032</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>0.15</td>
<td>0.083 (0.072,0.095)</td>
<td>0.063</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>2</td>
<td>0.051</td>
<td>0.069 (0.059,0.078)</td>
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</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
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<td>0.008</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>0.11</td>
<td>0.075 (0.065,0.086)</td>
<td>0.032</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>0.14</td>
<td>0.082 (0.071,0.094)</td>
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</table>
Table 5-9. Fit of Monte Carlo data to Equation 5-1 using measured MRE values for representative clinical techniques, with a grid.

<table>
<thead>
<tr>
<th>Target</th>
<th>Filter</th>
<th>Thickness (cm)</th>
<th>kVp</th>
<th>Measured MRE</th>
<th>R² value</th>
</tr>
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<tbody>
<tr>
<td>Mo</td>
<td>Mo</td>
<td>2</td>
<td>24</td>
<td>3.36</td>
<td>0.97</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>28</td>
<td>4.71</td>
<td>0.99</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>7.67</td>
<td>1.00</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>32</td>
<td>11.44</td>
<td>1.00</td>
</tr>
<tr>
<td>Mo</td>
<td>Rh</td>
<td>2</td>
<td>24</td>
<td>3.32</td>
<td>0.98</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>28</td>
<td>4.54</td>
<td>0.99</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>6.81</td>
<td>1.00</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>32</td>
<td>10.31</td>
<td>1.00</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>2</td>
<td>24</td>
<td>3.50</td>
<td>0.98</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>28</td>
<td>4.91</td>
<td>0.99</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>7.07</td>
<td>0.96</td>
</tr>
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<td>32</td>
<td>10.49</td>
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</tr>
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<td>W</td>
<td>Ag</td>
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<td>24</td>
<td>3.83</td>
<td>0.97</td>
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<td>28</td>
<td>5.29</td>
<td>0.97</td>
</tr>
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<td></td>
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<td>30</td>
<td>7.53</td>
<td>1.00</td>
</tr>
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<td></td>
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<td>32</td>
<td>10.35</td>
<td>1.00</td>
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Table 5-10. Fit of Monte Carlo data to Equation 5-1 using measured MRE values for a 4 cm phantom thickness and tube potentials ranging from 24-32 kVp, with a grid.

<table>
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<th>Measured MRE</th>
<th>R² value</th>
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</thead>
<tbody>
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</tr>
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<td></td>
<td>26</td>
<td>4.76</td>
<td>0.99</td>
</tr>
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<td></td>
<td></td>
<td>28</td>
<td>4.71</td>
<td>0.99</td>
</tr>
<tr>
<td></td>
<td></td>
<td>30</td>
<td>5.06</td>
<td>0.99</td>
</tr>
<tr>
<td></td>
<td></td>
<td>32</td>
<td>5.81</td>
<td>1.00</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>24</td>
<td>5.13</td>
<td>0.99</td>
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<td>26</td>
<td>4.87</td>
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<td>28</td>
<td>4.91</td>
<td>0.99</td>
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<td>5.25</td>
<td>0.99</td>
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Table 5-11. Fit of Monte Carlo data to Equation 5-1 using measured MRE for a tube potential of 30 kVp and phantom thicknesses ranging from 2-8 cm, with a grid.

<table>
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<th>Target</th>
<th>Filter</th>
<th>Thickness (cm)</th>
<th>Measured MRE</th>
<th>R² value</th>
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</thead>
<tbody>
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<td>2</td>
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<td>0.98</td>
</tr>
<tr>
<td></td>
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<td>4</td>
<td>5.06</td>
<td>0.99</td>
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<td></td>
<td></td>
<td>6</td>
<td>7.67</td>
<td>1.00</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>2</td>
<td>4.14</td>
<td>0.98</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>5.25</td>
<td>0.99</td>
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<td></td>
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<td>6</td>
<td>7.07</td>
<td>0.96</td>
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</tbody>
</table>

Table 5-12. Fit of Monte Carlo data to Equation 5-1 using measured MRE values for representative clinical techniques, without a grid.

<table>
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<th>Target</th>
<th>Filter</th>
<th>Thickness (cm)</th>
<th>kVp</th>
<th>Measured MRE</th>
<th>R² value</th>
</tr>
</thead>
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<td>Mo</td>
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<td>19.30</td>
<td>0.82</td>
</tr>
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<td></td>
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<td>28</td>
<td>24.15</td>
<td>0.75</td>
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<td>30</td>
<td>30.88</td>
<td>0.72</td>
</tr>
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<td></td>
<td>8</td>
<td>32</td>
<td>41.41</td>
<td>0.65</td>
</tr>
<tr>
<td>Mo</td>
<td>Rh</td>
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<td>24</td>
<td>20.29</td>
<td>0.84</td>
</tr>
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<td></td>
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<td>28</td>
<td>25.12</td>
<td>0.76</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>30.84</td>
<td>0.72</td>
</tr>
<tr>
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<td></td>
<td>8</td>
<td>32</td>
<td>38.34</td>
<td>0.67</td>
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<td>W</td>
<td>Rh</td>
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<td>21.63</td>
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<td>28</td>
<td>27.56</td>
<td>0.77</td>
</tr>
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<td></td>
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<td>30</td>
<td>33.49</td>
<td>0.73</td>
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<td>32</td>
<td>42.10</td>
<td>0.68</td>
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<td>W</td>
<td>Ag</td>
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<td>24</td>
<td>21.89</td>
<td>0.86</td>
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<td>28</td>
<td>29.11</td>
<td>0.79</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30</td>
<td>35.40</td>
<td>0.72</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>32</td>
<td>41.41</td>
<td>0.68</td>
</tr>
</tbody>
</table>
Table 5-13. Fit of Monte Carlo data to Equation 5-1 using measured MRE values for a 4 cm phantom thickness and tube potentials ranging from 24-32 kVp, without a grid.

<table>
<thead>
<tr>
<th>Target</th>
<th>Filter</th>
<th>kVp</th>
<th>Measured MRE</th>
<th>R² value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo</td>
<td>Mo</td>
<td>24</td>
<td>23.51</td>
<td>0.74</td>
</tr>
<tr>
<td></td>
<td></td>
<td>26</td>
<td>23.56</td>
<td>0.74</td>
</tr>
<tr>
<td></td>
<td></td>
<td>28</td>
<td>24.15</td>
<td>0.75</td>
</tr>
<tr>
<td></td>
<td></td>
<td>30</td>
<td>25.28</td>
<td>0.77</td>
</tr>
<tr>
<td></td>
<td></td>
<td>32</td>
<td>26.95</td>
<td>0.78</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>24</td>
<td>27.83</td>
<td>0.77</td>
</tr>
<tr>
<td></td>
<td></td>
<td>26</td>
<td>27.52</td>
<td>0.75</td>
</tr>
<tr>
<td></td>
<td></td>
<td>28</td>
<td>27.56</td>
<td>0.77</td>
</tr>
<tr>
<td></td>
<td></td>
<td>30</td>
<td>27.95</td>
<td>0.75</td>
</tr>
<tr>
<td></td>
<td></td>
<td>32</td>
<td>28.68</td>
<td>0.79</td>
</tr>
</tbody>
</table>

Table 5-14. Fit of Monte Carlo data to Equation 5-1 using measured MRE for a tube potential of 30 kVp and phantom thicknesses ranging from 2-8 cm, with a grid.

<table>
<thead>
<tr>
<th>Target</th>
<th>Filter</th>
<th>Thickness (cm)</th>
<th>Measured MRE</th>
<th>R² value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo</td>
<td>Mo</td>
<td>2</td>
<td>21.06</td>
<td>0.84</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>25.28</td>
<td>0.77</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>30.88</td>
<td>0.72</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>39.79</td>
<td>0.68</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>2</td>
<td>21.70</td>
<td>0.85</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>27.95</td>
<td>0.75</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>33.49</td>
<td>0.73</td>
</tr>
<tr>
<td></td>
<td></td>
<td>8</td>
<td>40.30</td>
<td>0.68</td>
</tr>
</tbody>
</table>
Table 5-15. Fit of Monte Carlo data to Equation 5-2 without a grid. The ratio of the coefficient $a$ to the coefficient $c$ indicates the relative contribution of each exponential term. The reciprocals of the coefficients $b$ and $d$ give MRE values for each term.

<table>
<thead>
<tr>
<th>Thickness (cm)</th>
<th>Technique</th>
<th>$a$</th>
<th>$b$</th>
<th>$c$</th>
<th>$d$</th>
<th>Ratio of $a/c$</th>
<th>MRE 1 (mm)</th>
<th>MRE 2 (mm)</th>
<th>$R^2$ value</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>24 kVp, Mo/Mo</td>
<td>1.9E-04</td>
<td>0.2</td>
<td>7.2E-05</td>
<td>0.07</td>
<td>2.69</td>
<td>5.33</td>
<td>14</td>
<td>1.00</td>
</tr>
<tr>
<td>4</td>
<td>28 kVp, Mo/Mo</td>
<td>7.6E-05</td>
<td>0.2</td>
<td>4.1E-05</td>
<td>0.06</td>
<td>1.86</td>
<td>5.8</td>
<td>16</td>
<td>1.00</td>
</tr>
<tr>
<td>6</td>
<td>30 kVp, Mo/Mo</td>
<td>3.1E-05</td>
<td>0.2</td>
<td>1.8E-05</td>
<td>0.05</td>
<td>1.75</td>
<td>5.9</td>
<td>18</td>
<td>1.00</td>
</tr>
<tr>
<td>8</td>
<td>32 kVp, Mo/Mo</td>
<td>1.4E-05</td>
<td>0.2</td>
<td>1.3E-05</td>
<td>0.05</td>
<td>1.13</td>
<td>4.4</td>
<td>18</td>
<td>1.00</td>
</tr>
</tbody>
</table>

Table 5-16. Fit of Monte Carlo data to Equation 5-2 with a grid. The ratio of the coefficient $a$ to the coefficient $c$ indicates the relative contribution of each exponential term. The reciprocals of the coefficients $b$ and $d$ give MRE values for each term.

<table>
<thead>
<tr>
<th>Thickness (cm)</th>
<th>Technique</th>
<th>$a$</th>
<th>$b$</th>
<th>$c$</th>
<th>$d$</th>
<th>Ratio of $a/c$</th>
<th>MRE 1 (mm)</th>
<th>MRE 2 (mm)</th>
<th>$R^2$ value</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>24 kVp, Mo/Mo</td>
<td>4.0E-04</td>
<td>1.2</td>
<td>2.8E-06</td>
<td>0.08</td>
<td>142.6</td>
<td>0.83</td>
<td>13</td>
<td>1.00</td>
</tr>
<tr>
<td>4</td>
<td>28 kVp, Mo/Mo</td>
<td>1.1E-04</td>
<td>0.8</td>
<td>1.7E-06</td>
<td>0.08</td>
<td>65.25</td>
<td>1.3</td>
<td>13</td>
<td>1.00</td>
</tr>
<tr>
<td>6</td>
<td>30 kVp, Mo/Mo</td>
<td>3.7E-05</td>
<td>0.6</td>
<td>1.2E-06</td>
<td>0.08</td>
<td>30.24</td>
<td>1.7</td>
<td>13</td>
<td>1.00</td>
</tr>
<tr>
<td>8</td>
<td>32 kVp, Mo/Mo</td>
<td>1.7E-05</td>
<td>0.5</td>
<td>1.3E-06</td>
<td>0.08</td>
<td>13.40</td>
<td>2.0</td>
<td>13</td>
<td>1.00</td>
</tr>
</tbody>
</table>

Table 5-17. Comparison of the mean radial extents from the bi-exponential fit of total scatter to the exponential fits of the isolated contributions of coherent and Compton scattering, for 2 cm, 24 kV Mo/Mo, without a grid.

<table>
<thead>
<tr>
<th></th>
<th>Coherent scattering</th>
<th>Compton scattering</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bi-exponential (combined) fit</td>
<td>5.33</td>
<td>13.7</td>
</tr>
<tr>
<td>Exponential (isolated) fit</td>
<td>5.78</td>
<td>13.9</td>
</tr>
<tr>
<td>Percent difference</td>
<td>8.1%</td>
<td>1.4%</td>
</tr>
</tbody>
</table>
Table 5-18. Impact of different glandularities on the SF with and without a grid. The Monte Carlo-derived scatter fractions are accurate to within ±7%.

<table>
<thead>
<tr>
<th>Target/Filter Combination</th>
<th>30/70 glandularity with grid</th>
<th>70/30 glandularity with grid</th>
<th>30/70 glandularity without grid</th>
<th>70/30 glandularity without grid</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo/Mo</td>
<td>0.2%</td>
<td>0.0%</td>
<td>-0.5%</td>
<td>0.4%</td>
</tr>
<tr>
<td>Mo/Rh</td>
<td>-0.1%</td>
<td>0.0%</td>
<td>-0.5%</td>
<td>0.5%</td>
</tr>
<tr>
<td>W/Rh</td>
<td>-0.4%</td>
<td>0.5%</td>
<td>-0.8%</td>
<td>0.4%</td>
</tr>
<tr>
<td>W/Ag</td>
<td>-0.6%</td>
<td>0.7%</td>
<td>-0.8%</td>
<td>0.3%</td>
</tr>
</tbody>
</table>

Table 5-19. Impact of different glandularities on the apparent thickness with and without a grid.

<table>
<thead>
<tr>
<th>Target/Filter Combination</th>
<th>30/70 glandularity with grid</th>
<th>70/30 glandularity with grid</th>
<th>30/70 glandularity without grid</th>
<th>70/30 glandularity without grid</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo/Mo</td>
<td>-5.8%</td>
<td>5.9%</td>
<td>-5.9%</td>
<td>5.7%</td>
</tr>
<tr>
<td>Mo/Rh</td>
<td>-5.7%</td>
<td>6.1%</td>
<td>-5.6%</td>
<td>5.6%</td>
</tr>
<tr>
<td>W/Rh</td>
<td>-4.8%</td>
<td>4.8%</td>
<td>-4.9%</td>
<td>4.5%</td>
</tr>
<tr>
<td>W/Ag</td>
<td>-4.5%</td>
<td>4.7%</td>
<td>-4.6%</td>
<td>4.3%</td>
</tr>
</tbody>
</table>
Figure 5-1. Design of the simulation.

Figure 5-2. Schematic of ring tallies (gray circles) placed around the impulse.
Figure 5-3. Fit of the PSF to the Monte Carlo data for a 4 cm phantom thickness and technique of 28 kVp Mo/Mo, with a grid. The graphs for the other techniques with a grid appear very similar to this one and are not displayed here; however, the goodness of fit values for all the fits can be found in Tables 5-9 through 5-11.
Figure 5-4. Fit of the PSF to the Monte Carlo data for a 4 cm phantom thickness and technique of 28 kVp Mo/Mo, without a grid. The graphs for the other techniques without a grid appear very similar to this one and are not displayed here; however, the goodness of fit values for all the fits can be found in Tables 5-12 through 5-14.
Figure 5-5. Fit of Equation 5-2 to the Monte Carlo data for a 4 cm phantom thickness and technique of 28 kVp Mo/Mo, without a grid. The graphs for the other techniques without a grid appear very similar to this one and are not displayed here; however, the goodness of fit values for all the fits can be found in Table 5-15.
Figure 5-6. Comparison of the PSF's generated by Compton scattering and total scattering A) with a grid and B) without a grid for a 2 cm phantom and 24 kVp Mo/Mo spectrum.
Figure 5-7. Matlab® fits of the PSF's generated by the isolated effects of Compton scattering and coherent scattering (distances less than 10 mm were excluded for the Compton fit)
Figure 5-8. Comparison of scatter kernels by glandularity A) with a grid and B) without a grid. The tally results are normalized to the value of the 0.2 cm tally.
CHAPTER 6
SCATTER CORRECTION PROGRAM

This chapter describes the design and optimization of the scatter correction program. Matlab® (version 7.10.0.499 (R2010a), MathWorks®, Natick, MA) with the Image Processing Toolbox and Parallel Processing Toolbox was used for all image processing. The code for the main program and all custom functions (i.e., those functions written for this project, rather than those functions that come with Matlab®) discussed in this chapter can be found in Appendix D. Processing time limitations were as much a concern for the scatter correction program as they were for the Monte Carlo simulation; Table 5-1 contains a list of the most important hardware specifications.

IRB approval was obtained to acquire all patient images used for this study.

**Basic Program Design**

The first step in the scatter correction program is to read the DICOM® header of the image to be processed. The main program calls a custom function "imginfo" that reports back the kV, grid status, and view (including laterality). This function also finds the image size and target/filter combination from the DICOM® header and loads the appropriate tables for use with the apparent thickness, mean radial extent (MRE), and scatter fraction (SF) calculations. The DICOM® image is then read into a Matlab® variable, producing an array equal in size to the number of detector elements used for that field of view (i.e., 2560 x 3328 for the small field of view or 3328 x 4096 for the large field of view). The images from the Hologic™ Selenia® have a border on the left and right sides of the image consisting of 12 rows of pixels per side that do not contain image information. The values in these pixels would cause artifacts in the scatter-corrected images, so they are removed from the image before further processing. The
DC offset of 50 (described in Chapter 4) is also subtracted from the pixel values before further processing.

**Apparent Thickness Determination**

The main program then calls the custom function "thickmap" to produce a map of apparent thicknesses from the image data. The first step in this process is to create a region of interest (ROI) measuring 31 x 31 pixels in either the anterior right or anterior left corner of the image. The choice of ROI location depends on the laterality and is intended to minimize the likelihood of including breast tissue in the ROI: images of the left breast have the ROI in the anterior right corner, whereas images of the right breast have the ROI in the anterior left corner. The average pixel value in the ROI is calculated, and the log signal ratio (LSR) is found for each pixel in the image by use of the formula:

\[
LSR = -\ln \left( \frac{\text{pixel value}}{\text{ROI average}} \right).
\]  
(6-1)

Once the LSR is determined for each pixel of an image, the program reads the previously-loaded thickness table to find the tissue thickness that most closely corresponds to that LSR (rounding down to the nearest 1 mm). The map of apparent thicknesses is then returned to the main program. A thickness map for a sample image is shown in Figure 6-1.

For the next step in processing, the apparent thickness map is broken down into mask images, each consisting of all the pixels corresponding to one apparent thickness (Figure 6-2). To reduce ringing from the Fourier transform due to differences in the pixel values between opposite edges of the image (as explained in Chapter 1), a ramp is constructed leading from the pixel values on one side of the mask to the pixel values on
the opposite side. The ramp is created in both the left-right and anterior-chest wall
directions.

**Scatter Kernel Calculation**

For each 1 mm-thickness, the scatter kernel parameters (MRE and SF) are
calculated. Both values are calculated using the general equation:

\[
y = C_1 + C_2 t + C_3 E + C_4 t^2 + C_5 t E + C_6 E^2 + C_7 t^3 + C_8 t^2 E + C_9 t E^2,
\]

where \( y \) is MRE or SF value, \( t \) is the apparent thickness, and \( E \) is the tube potential.

Equation 6-2 is simply the form of the surface fit generated by Matlab® for the measured
MRE and SF data (see Chapter 4 for more detail). The tables loaded by the "imginfo"
function contain the coefficients \( C_1 - C_9 \) appropriate for the target/filter combination and
grid status.

The scatter kernel parameters are passed to another custom function,
"scatterkernel," which generates a 2-dimensional point spread function (PSF) equal in
size to the image (Figure 6-3). The PSF is passed back to the main program, where the
Fourier transform is calculated.

**Application of the Wiener Filter**

The next step in the program is the application of the Wiener filter to the mask
image. This calculation is performed using the Wiener filter discussed in Chapter 1:

\[
\hat{I}_t(u, v) \approx \frac{1}{K(u, v)} \cdot \frac{|K(u, v)|^2}{|K(u, v)|^2 + C} \cdot I_d(u, v)
\]

where \( \hat{I}_t(u, v) \) is the estimate of the true (i.e., scatter-corrected) image at location \((u, v)\)
in frequency space, \( I_d(u, v) \) is the degraded (i.e., measured) image at the same
location, \( K(u, v) \) is the value of the PSF at that location, and \( C \) is a constant chosen to
balance sharpness and noise in the final image. The method for choosing the value $C$ is described later in this chapter.

The filtration is not performed on areas of the image corresponding to the unattenuated beam, because leaving this area unprocessed prevents the appearance of ringing artifacts at the edges of the breast and does not appear to affect the image quality otherwise. The results of the filtration for all apparent thicknesses greater than 2 mm are used to produce the final image.

**Final Steps**

After the filtration step, the inverse Fourier transform of each deconvolved image is calculated, the images corresponding to each thickness are added together, and the ramp is cut off. The pixel values are shifted so that the lowest pixel value of the corrected image is equal to the lowest pixel value of the original image. This step prevents negative pixel values, which cannot be displayed by the diagnostic workstation. Pixel values greater than the highest pixel value in the original image are replaced by the highest pixel value of the original image (these high pixel values correspond to the black background of the image, so truncating them does not affect the appearance of the breast). The border rows on the right and left edges of the image are also added back on.

The final image is then converted to a two-byte integer type and saved as a DICOM® file. Sample images are shown in Figures 6-4 and 6-5. The DICOM® header from the original image is transferred to the new image. For unknown reasons, Matlab® makes several changes the DICOM® header, even when instructed to copy the header without changes. Thus, the header of the processed file is returned to the Hologic™ format using Sante DICOM® Editor (version 3.1.16, Santesoft LTD, Athens, Greece).
Selection of the Constant $C$

As mentioned previously, the Wiener filter depends on a constant $C$, which is an approximation of the ratio of the NPS of the stochastic noise to the NPS of the image. The value of $C$ was chosen with the input of three radiologists (radiologist credentials are discussed in Chapter 8). Five clinical images were processed four times each, with the only difference between the images in each set being the $C$ value used. Four different $C$ values were applied in each image set: 2.5, 5, 7.5, and 10. The five image sets were shown to one of the radiologists, and he was asked to rank them in order of preference. He consistently ranked them in order of $C$ value (with 2.5 as the best and 10 as the worst), although he was not aware of how the image processing differed among the images.

This result was used as a preliminary test to identify the appropriate range of $C$ values. The image sets were processed again, this time with $C$ values of 1, 2, 3, and 4. The images were shown to two different radiologists, neither of whom was the radiologist used in the preliminary test, and they were asked to choose their favorite image. Both radiologists chose $C$ equal to 1 for all image sets. The appearance of noise in the images processed with a $C$ value of 1 is noticeably the most pronounced. However, the images processed with a $C$ of 1 are also the sharpest images, and the impact of the $C$ value on the appearance of fine detail is clearly noticeable on the 5-megapixel diagnostic review workstations. The radiologists were asked about the appearance of noise in the images, and both indicated that it did not bother them. In fact, neither of them noticed the image noise until it was pointed out. Thus, it would...
appear that the radiologists participating in this study filter out image noise when interpreting the image, but they depend strongly on image sharpness.

In addition to radiologist preference, the NPS results discussed in the next chapter support the use of a $C$ equal to 1 because it was shown that the image noise is composed mostly of stochastic noise; the contribution to the total noise from nonstochastic sources was about one and a half orders of magnitude below the contribution of stochastic sources. Thus, it makes sense that the ratio of the NPS from stochastic sources to the NPS of the image (i.e., $C$) would be equal to 1.

**Investigations Related to the Algorithm Design**

The spatially-variant algorithm is a unique and fairly complex approach to scatter reduction, and the question of whether similar results could be obtained with a simpler method was raised. Two methods were suggested: a simple deconvolution using the average thickness of the breast, and a mathematical manipulation of the histogram.

To test the simple deconvolution, the breast was segmented from the black background in a clinical image, and the average thickness was calculated. The scatter kernel corresponding to this thickness was deconvolved from the image by division of the Fourier transform of the image by the Fourier transform of the scatter kernel. A comparison of the original image, spatially-variant scatter-corrected image, and deconvolved image shows that the contrast is inferior using the simple deconvolution (Figure 6-6). The appearance of noise is also greater with the simple deconvolution.

The next step was to see if the scatter correction process could be bypassed by use of histogram manipulation. To test this, the histograms of the images in Figures 6-6a and 6-6b were compared (Figure 6-7). The peak corresponding to the black background was removed for this analysis. The histogram for the unprocessed image
had a mean of 1004.1 and a standard deviation 800.8, whereas the histogram for the scatter-corrected image had a mean of 641.2 and a standard deviation of 512.9. The mean and standard deviation of the histogram in Figure 6-8a were then matched to those of Figure 6-8b by compressing the spectrum and shifting the mean pixel value. The compression was performed by multiplying the values above and below the peak by 0.67 and 1.33, respectively (those values were determined by trial and error). The shift was performed by subtracting 120.2 from all the pixel values, which was the difference between the two means. This process produced a histogram with a mean of 641.2 and a standard deviation of 514.9. The resulting histogram and corresponding image are shown in Figure 6-9. Severe contouring artifacts related to the grayscale compression are apparent in the image, indicating that histogram manipulation is not an adequate substitute for the scatter correction algorithm.

**Discussion**

The impact of the scatter correction program on the images is noticeable, not only in the gross image appearance, but also in the window/level settings required to produce that appearance. In the pre-processed raw images, it is not possible to view all the image detail at one time with a reasonable contrast (i.e., the window/level settings that provide the best contrast in the thinner areas of the breast do not overlap with those ideal for the thicker areas). After scatter correction, much more of the breast can be visualized with good contrast at a single window/level setting, which signifies better image contrast. For example, Figure 6-4a is shown displayed with a window of 2712 and a level of 1356. Figure 6-4b, which is displayed so that it appears as similar as possible to Figure 6-4a, uses display settings with a window of 1600 and a level of 800. That the breast tissue is able to be displayed in a much smaller window indicates that
the range of pixel values required to display the image information has grown smaller, which in turn means that the image contrast has improved. To further demonstrate the effect of the window and level settings, the unprocessed image is shown in Figure 6-4c with window/level settings equal to those used for the processed image. With the narrower window, not all of the breast tissue can be visualized.

It was reported previously (in the dissertation proposal) that there was one known artifact associated with the application of scatter correction: metal objects and large calcifications appeared "hollow," so that the outline of the object was visible but the center was a gray level similar to the background, rather than bright white. It was discovered that this artifact was caused by an error in the Matlab® program, in which the absolute value of the final image was taken unnecessarily. Very dense objects result in negative pixel values before corrections are added to the final image, so taking the absolute value of that image resulted in those negative values inappropriately becoming positive. This error was corrected, and a DC offset was added to all the pixel values to ensure that there are no negative values in the final image. This offset sets the lowest pixel value in the final image equal to the lowest pixel value in the original image.

The processing time for the scatter correction program depends on image size and the value of the maximum apparent thickness. On the personal computer described previously, with the use of parallel processing to utilize the multicore processor, typical processing times were about 5-10 minutes. Although this time is long for clinical applications, the use of a modern, dedicated computer and more streamlined code could reduce this time substantially. Advancements in computing power over the next few years would also be expected to reduce processing time. Therefore, it is not
unreasonable to expect that this scatter correction program could be implemented with clinically-acceptable processing times now or in the near future.
Figure 6-1. Thickness map for a sample image. Black represents zero thickness and white represents the maximum thickness (in this case, corresponding to a calcification).

Figure 6-2. Mask image showing pixels corresponding to an apparent thickness of 4.7 cm, for the image shown in Figure 6-1.
Figure 6-3. Sample PSF for 28 kVp Mo/Mo, with a grid. The actual kernel size is as large as the mammogram image size, but the range has been restricted for display purposes to better show the PSF.
Figure 6-4. Clinical mammogram with a grid A) before scatter correction, with a window of 2712 and level of 1356, B) after scatter correction, with a window of 1600 and level of 800, and C) before scatter correction, with a window of 1600 and a level of 800. The images A and B have been windowed and leveled to appear as similar as possible. Using the post-scatter correction window/level for the uncorrected image (Figure C) results in missing tissue at the breast edges, because the uncorrected image can only be displayed in its entirety with a large window, which results in poor image contrast.
Figure 6-5. Image of an anthropomorphic phantom A) with a grid, before scatter correction, B) with a grid, after scatter correction, C) without a grid, before scatter correction, and D) without a grid, after scatter correction. The images have been windowed and leveled to appear as similar as possible.
Figure 6-6. Clinical image shown A) before any processing, B) after spatially-variant scatter correction, and C) after scatter correction using a simple deconvolution of the scatter kernel corresponding to the average thickness. Images B and C have been windowed and leveled to appear as similar as possible.
Figure 6-7. Histograms corresponding to A) the unprocessed image shown in Figure 6-6a, and B) the scatter-corrected image shown in Figure 6-6b.
Figure 6-8. Results of histogram manipulation on A) the histogram and B) the resulting image.
CHAPTER 7
IMAGE QUALITY MEASUREMENTS

The measurement of image quality before and after processing with the scatter correction program is essential for assessing the program's impact. Image quality was initially assessed with three measurements: the noise power spectrum (NPS), the contrast-to-noise ratio (CNR), and the modulation transfer function (MTF). This chapter describes the procedures for measuring each of these quantities and the results associated with all three prior to processing. The measurements after processing include only the NPS and CNR. The MTF was not measured post-processing because images of the line pair phantom used for the MTF measurement suffered from ringing that affected the data (see Chapter 1 for a description of ringing). A substitute, semi-quantitative assessment of spatial resolution was used instead of MTF due to this issue.

As mentioned in Chapter 6, the scatter correction program uses the Wiener filter, which depends on a parameter $C$ that controls the noise suppression properties of the filter. The higher the value of $C$, the more the filter suppresses noise, at the cost of blurring the image. The NPS, CNR, and spatial resolution are therefore dependent on the value of $C$ selected. The post-processing measurements presented here are for a value of $C$ equal to 1. The method used to select this value is described in Chapter 6.

Assessment of Image Quality

The NPS, CNR, and MTF were first assessed using raw images acquired as described in the sections below. Measurements were performed both on Unit 1 (Mo target with Mo and Rh filters) and Unit 2 (W target with Rh and Ag filters), using a 4.0 cm-thick acrylic slab as a scattering medium. The images used for each of the
analyses were then processed with the scatter correction program described in Chapter 6, and measurements were repeated on the corrected images. The only modification to the scatter correction program (for phantom images only) was a change in the method of apparent thickness determination, which was necessary because the acrylic phantom used for image acquisition covered the entire detector. Thus, it was not possible to find the apparent thickness using the ratio of the pixel value to that of an area exposed to the unattenuated beam. The thickness of the phantom was known, however, so the apparent thickness function (used for patient mammography images) was bypassed and the thickness was defined directly in the program.

Because the scatter kernel parameters were measured with BR12 rather than acrylic, a separate test was performed to find an acrylic-to-BR12 thickness "conversion". For this comparison, the acrylic slab was placed on the breast support surface diagonally, so that the anterior corners of the detector were uncovered. The apparent thickness was determined using the ratio of the attenuated and unattenuated pixel values, as described previously, and was found to be approximately 4.2 cm using W/Rh at 24, 28, and 32 kVp. Analysis of the apparent thickness data given in Chapter 4 indicates that the apparent thickness changes by less than 1 mm across the range of tube potentials and target/filter combinations tested, so the conversion measurement was not performed with the other target/filter combinations and tube potentials. The acrylic slab was programmed to have a thickness of 4.2 cm instead of 4.0 cm.

**Noise Power Spectrum**

**Procedures**

Before acquiring images for the NPS calculation, the detector flat-field calibration required as part of the weekly quality control was performed according to the procedure
outlined in the Radiologic Technologist section of the Hologic™ Quality Control Manual (version MAN-01476 Rev 001). A test for artifacts was then performed according to the procedure specified in the Medical Physicist section of the Hologic™ QC manual. Two small negative-density, smudge-like artifacts, which were not removed by the flat-field calibration, were observed on Unit 1 images. Neither artifact was in a location which interfered with any region of interest (ROI) used for the NPS assessment. No artifacts were observed on Unit 2 images.

All images were acquired with the acrylic slab placed on the breast support surface. The acrylic completely covered the active area of the detector. Other researchers have recommended placing the attenuator at an elevated position to reduce scatter; however, because the purpose of the NPS measurement in this case was to assess the effect of the scatter reduction algorithm, the acrylic was placed directly on the breast support surface to simulate clinically-realistic scatter conditions. Images were acquired at 24, 28, and 32 kVp with all target/filter combinations, with and without a grid. The tube current-time product (mAs) for each combination of exposure parameters was determined by making an initial exposure of the acrylic phantom using automatic exposure control (AEC), and then using the closest manual mAs setting to that determined by the AEC.

According to Williams, the total 2D NPS ($S_{tot}(u,v)$) can be calculated by averaging noise power spectra obtained from many different images, which provides a spectrum that contains contributions from both stochastic (i.e., quantum) noise and nonstochastic (i.e., structured) noise. The nonstochastic sources can be isolated by first averaging many images, and then finding the spectrum of the averaged image.
\( S_{\text{stoch}}(u,v) = S_{\text{tot}}(u,v) - S_{\text{avg}}(u,v) \).

The scatter correction algorithm was expected to affect primarily the stochastic portion of the NPS because it would reduce the influence of scattered photons, which are a source of stochastic noise, but it was unknown to what degree stochastic noise sources contributed to the total NPS. The 2D NPS was calculated for both \( S_{\text{tot}}(u,v) \) and \( S_{\text{stoch}}(u,v) \) in order to quantify the effect on both stochastic sources of noise and total noise. Rather than taking many images, a large ROI of 320x320 pixels was split into 25 small 64x64 pixel ROI's, and all of the small ROI's were then used in aggregate, either by finding the NPS's first and then averaging them to find \( S_{\text{tot}}(u,v) \), or by averaging the ROI's first and then finding the average NPS, to find \( S_{\text{avg}}(u,v) \).\(^{63-65}\) Nine large 320x320 ROI's were used for the assessment of the NPS at different areas of the image, with locations shown in Figure 7-1. Because the grid does not retract all the way, the location of the anterior-edge ROI's was determined by the maximum imaged area of the non-grid image (18 cm x 29 cm). The same ROI positions were used for the grid and non-grid images. Images were acquired using all target/filter combinations at 24, 28, and 32 kV.

The two-dimensional (2D) NPS was determined using the equation:\(^{66}\)

\[
NPS(u,v) = \frac{\Delta_x \Delta_y}{MN_xN_y} \langle |\mathcal{F}(I(x,y))|^2 \rangle
\]

(7-1)

where \( \Delta_x \) and \( \Delta_y \) are the pixel size in mm in the x- and y-directions, respectively; \( N_x \) and \( N_y \) are the number of pixels in the ROI in the x- and y-directions, respectively; \( M \) is the number of ROI's averaged; \( I(x,y) \) is the portion of the image contained in the ROI; \( \mathcal{F} \) represents the Fourier Transform; and the pointed brackets represent the ensemble average produced from many ROI's. Equation 7-1 produces an NPS with units of...
(pixel value)$^2$ mm$^2$, so the NPS was normalized by dividing by the square of the average pixel value. The Matlab® program used to find the NPS can be found in Appendix E, and a sample 2D NPS is shown in Figure 7-2.

A one-dimensional (1D) NPS curve was produced from each 2D NPS by taking four rows on both sides of one axis of the 2D NPS (for a total of 8 rows) and averaging them. The isotropy of the 2D NPS was tested by comparing 1D NPS’s taken along both axes of $S_{\text{tot}}(u,v)$ and $S_{\text{avg}}(u,v)$ for one of the images; the difference between the 1D NPS’s from the two axes was only 2% for $S_{\text{tot}}(u,v)$ and 3% for $S_{\text{avg}}(u,v)$, so the 2D NPS was considered to be isotropic, and 1D curves were produced parallel to only one axis in subsequent analyses. It should be noted that the axes of the 2D NPS are considered artifacts of low-frequency phenomena such as the heel effect (in one direction) and inverse-square law effect (in both directions), and thus the axes are excluded from the 1D NPS sections. Zero frequency is therefore excluded from the 1D graphs of the NPS. The range of data shown is from 0.22 mm$^{-1}$ to 7.14 mm$^{-1}$, which is the Nyquist frequency of the detector (calculated using Equation 1-6 and a pixel size of 0.07 mm).

The coefficient of variation (COV) for the 1D NPS can be calculated using the equation:

$$COV \approx \frac{1.0}{\sqrt{n}}$$ (7-2)

where $n$ is the number of independent measurements associated with a single NPS value. In this case, $n$ is equal to the number of ROI’s (25) multiplied by the number of rows averaged (8). For the procedure described above, the COV is approximately 7%.

The Matlab® program written to determine the NPS was applied to the raw images acquired from the two Selenia® units to obtain pre-correction measurements.
Then, the images used for the NPS evaluation were processed with the scatter correction program, and the NPS was measured a second time.

Results and Discussion

The 1D NPS curves shown in Figure 7-3 demonstrate the total, stochastic, and nonstochastic noise contributions for a 28 kVp Mo/Mo spectrum, with and without a grid. The contribution of nonstochastic noise is about one-and-a-half orders of magnitude below the contribution of stochastic noise, and this ratio does not change much as a result of scatter correction. For example, for 28 kV Mo/Mo at position 5, the ratio of the stochastic to the nonstochastic noise at 0.22 mm$^{-1}$ was 37.44 before scatter correction, and 37.99 after scatter correction. The scatter correction program reduces all sources of noise, with the amount of reduction independent of frequency. There is also no significant difference in noise reduction as a result of grid status.

Because nonstochastic noise is primarily related to factors independent of the x-ray spectrum (i.e., beam and detector non-uniformities, electronic noise, and image processing, as explained in Chapter 1), only the total NPS was calculated at position 5 (center of the detector) for the remaining tube potentials and target/filter combinations. The total NPS measurements for 24, 28, and 32 kVp spectra, before and after scatter correction, are shown in Figure 7-4. They indicate that the NPS’s with and without a grid are not significantly different before scatter correction—the difference between the two sets is only 5% averaged over all frequencies, compared to the 7% COV. These graphs also indicate that the scatter correction program has a similar effect on the NPS for all tube potentials and target/filter combinations: with a grid, there is ~60% reduction in noise for a Mo target and ~70% for a W target; without a grid, the noise is reduced ~60% for Mo/Mo, W/Rh, and W/Ag, and ~70% for Mo/Rh. Because the COV is 7% and
the differences among the target/filter combinations are only about 10% (which is less than two standard deviations, or 14% of the value for each data point) the differences do not meet significance at the 95% confidence level. Therefore, target/filter combination plays no significant role in the effect of the scatter correction program on the NPS.

Figure 7-5 shows the total NPS as a function of ROI position, with and without a grid, for Mo/Mo at 28 kVp. Any variation in the NPS as a function of position was expected to be due to effects independent of the x-ray spectrum, such as the inverse-square law and heel effects, so the NPS’s associated with other tube potentials and target/filter combinations were not analyzed as a function of position. Predictably, the NPS curves demonstrated greater stochastic noise at positions 7 and 9, where photon flux is the lowest due to the inverse square law and the heel effect. Positions 2 and 5 were least affected by stochastic noise, also as expected, because they receive the greatest number of photons. This trend did not depend on grid presence, and it was not altered by the scatter correction program.

Hypothesis 6-1, stating that the scatter correction algorithm lowers the stochastic noise content of the images, is clearly true. Hypothesis 6-2 states that the scatter correction program improves the NPS without a grid to become similar to the NPS with a grid; this hypothesis is based on the premise that the NPS’s with and without a grid differ to begin with, which is shown by these data to be false. The NPS’s are very similar both with and without a grid prior to the scatter correction, and they are again very similar after the correction.
Contrast-to-Noise Ratio

Procedures

The CNR was assessed in two ways. The first attempt to measure CNR, described in the dissertation proposal, utilized an acrylic disk as a low-contrast object. The results of this attempt were eventually discarded due to artifacts caused by the way the scatter correction algorithm interacted with the sharp edges of the disk (more detail about this issue is found in the Results and Discussion section). The second attempt to measure CNR involved the use of an acrylic sphere in place of the disk. The edges of the sphere presented a much more gradual shift in pixel values than the disk, and also were more similar to objects typically seen clinically. Use of the sphere did not cause an artifact.

Disk method

The CNR was determined by covering the detector with the acrylic phantom described previously, then using a 4 mm-thick, 1 cm-diameter acrylic disk as a low contrast object. The center of the disk was placed approximately at each of the nine locations shown in Figure 7-1, and images were acquired with and without a grid, for all target/filter combinations and three tube potentials (24, 28, and 32 kVp for Mo/Mo and W/Rh, and 26, 30, and 34 kVp for Mo/Rh and W/Ag). The mAs for each combination of exposure parameters was the same as that used for the NPS images. One set of measurements (28 kVp with W/Rh) was acquired a total of 4 times in order to determine the standard deviation in the measurements.

The CNR was determined using the equation:

\[
\frac{S_{bg} - S_d}{\sigma_{bg}}, \quad (7-3)
\]
where $S_{bg}$ is the average pixel value in a 64x64 ROI placed in the background area next to the disk, $\sigma_{bg}$ is the standard deviation of the pixel values in that ROI, and $S_d$ is the average pixel value of a 64x64 ROI placed over the disk. In relation to the ROI over the disk, the ROI used for the background measurements was displaced in the direction perpendicular to the anode-cathode axis of the x-ray tube to avoid any impact from the heel effect. The initial measurements were made directly on the acquisition workstation. After the scatter correction program was applied to the CNR images, the CNR for each image was then re-measured using ImageJ (Version 1.43u, National Institutes of Health, Bethesda, MD). The measurements acquired using ImageJ were compared to the measurements acquired using the acquisition workstation for a small sample of the pre-correction images and found to yield identical results; thus, the use of different programs for the pre- and post-correction images had no impact on the measured values.

**Sphere method**

A 5 mm-diameter acrylic sphere used as a low-contrast object in place of the disk, but otherwise the experimental setup was the same. Images were acquired in position 5 with all four target/filter combinations, with and without a grid, for tube potentials of 24, 28, and 32 kVp. Images were acquired for positions 1-9 using Mo/Mo at 28 kVp, with and without a grid. One set of measurements (28 kVp Mo/Mo, with and without a grid) was acquired a total of 5 times in order to determine the standard deviation in the measurements. These repeated measurements were performed at position 9 because it had the highest noise (i.e., the standard deviation of pixel values) and was assumed to
represent the worst-case scenario for the standard deviation of the CNR. The standard deviation was assessed both before and after scatter correction.

Equation 7-3 was used to calculate the CNR, and all measurements were made using ImageJ. The signal over the sphere was measured by placing a circular ROI over the sphere, with the edges of the ROI matching the edges of the sphere as closely as possible. The ROI therefore contained values corresponding to the varied thicknesses on the sphere, rather than the single thickness of the disk. The same ROI was then positioned about one diameter away from the sphere for the background measurement.

**Results and Discussion**

After processing the disk images, it was discovered that a profile taken across the image of the disk displayed a hump in the center of the disk. This hump was not seen prior to scatter correction (Figure 7-6). A description of a brief investigation conducted regarding the cause of the hump can be found in Appendix F. Because the hump caused a loss of contrast between the disk and the background, the measurements acquired using the disk were not utilized for the CNR analyses. It should also be noted that the CNR results presented in the dissertation proposal used a \( C \) value of 20, however, and this value was selected before the radiologists were asked to choose the best \( C \) value. The use of this high value for \( C \) caused excessive smoothing of the image, which resulted in artificially high CNR values (even in combination with the loss of contrast caused by the hump). Thus, the results presented at the proposal were discarded for multiple reasons associated with the disk artifacts and use of a value for \( C \) (\( C = 20 \)) that was very different from the preferred value (\( C = 1 \)).
The results and discussion presented in this section therefore relate to the sphere measurements. The data are discussed in terms of trends before scatter correction and trends after scatter correction. The absolute difference between CNR values before and after scatter correction was not compared directly, because CNR comparisons are valid only between systems with identical MTF's, and the systems are no longer identical once the processing is applied. Although MTF was not measured in this study because the effect of the algorithm caused ringing in the phantom images (discussed later in this chapter), the selection of the parameter $C$ was observed to change the spatial resolution of the images, and it is known that the selection of $C$ balances sharpness and noise in the image. Thus, the scatter correction must change and does change the MTF.

The CNR is the difference between two signal levels in an image, divided by the average noise associated with those signals. Thus, it is closely related to the signal-to-noise ratio (SNR), and in fact, is sometimes termed the signal difference-to-noise ratio (SDNR). The reason why the CNR depends on the MTF is because the general form of the signal-to-noise ratio (SNR) is:

$$\text{SNR}(f) \propto \frac{S_P(f) MTF(f)}{W(f)},$$

(7-4)

where $S_P(f)$ is the spatial frequency spectrum of the image and $W(f)$ is the Wiener spectrum (NPS). The common definition of SNR as the signal divided by the noise is a simplification of the above equation. Commonly, the average pixel value in an ROI is used as a measurement of signal, but when the MTF's of two systems differ, the meaning of the pixel values differs because the way in which the input signal is recorded as an output signal (i.e., the modulation transfer) changes. Thus, a
comparison of pixel values between two systems can be used only when both systems have identical MTF's.

Additionally, the use of the square root of the signal as equal to the noise is an approximation, based on the ideal case in which quantum (Poisson) noise is the only noise source, and the output noise is equal to the input noise. This second point is particularly important, because the noise in the output signal is also dependent on the transfer characteristics of the system (MTF).69,72 Because the scatter correction algorithm uses the Wiener filter, which is an algorithm that reduces the contribution of noise to the image, this filter has the effect of altering the relationship between input and output noise. Thus, the numerical values of the CNR before and after scatter correction cannot be compared directly; however, the effect on trends before scatter correction and trends after scatter correction can be compared.

In the following sections, all trends were analyzed using a one-sample ANOVA test, calculated using Microsoft® Excel®. Statistical significance was assumed for all values of $p$ less than 0.05.

**Trends prior to scatter correction**

The CNR's without a grid ranged from 63-95% of those with a grid, showing, as expected, that CNR is worse when the grid is removed (Tables 7-1 and 7-2). The standard deviation of the CNR measurements using a 28 kVp Mo/Mo spectrum was found to be about 0.46% of the CNR with a grid and 2.2% without a grid.

The measurements performed at all sphere positions (Figure 7-7) indicated that the differences among positions were significant (with $p$ values of 0.0006 with a grid and 0.01 without a grid). All CNR values obtained as a function of sphere position can be found in Table 7-1. With the grid, there was a 19% difference between the highest CNR
(position 2) and lowest CNR (position 9) with respect to position. There was a 32% difference between the highest CNR (position 6) and lowest CNR (position 8) without a grid, indicating more variation in CNR with respect to position without a grid than with a grid.

With a grid, a more detailed analysis showed that the difference was not significant for the following combinations of positions: positions 2 and 3; positions 1, 4, and 6; and positions 7 and 9. The pre-correction CNR was highest at positions 2 and 3. Positions 9 and 7 had the lowest CNR’s, with the remaining positions in the middle of the distribution. This trend is partially explained by the greater photon flux at the chest wall edge of the detector and the lesser flux in the anterior corners (heel effect and inverse square law). However, it would make sense for positions 1 and 3 to be approximately equal, and this was not the case.

The reason why position 1 differed from position 3 was investigated, and it was found that there was both a slight reduction in contrast (difference between the mean pixel value over the sphere and the mean pixel value of the background) and a small increase in noise (standard deviation of the pixel values in the background) at position 1 compared to position 3. Analysis of the images showed that the edge of the sphere at position 1 was 11.9 mm from the chest wall edge of the detector, whereas the sphere at position 3 was 14.5 mm from the chest wall edge. This discrepancy was due to the difficulty in positioning the sphere in an exact location: the sphere tended to roll before coming to a stop, and the edge of the detector was difficult to see accurately through the acrylic slab. It was not thought that such a small difference would matter to the CNR measurements, but it did. Small ROI’s were positioned at locations progressively closer
to the detector edge, and it was discovered that the standard deviation became larger
approaching the edge. This change was sufficient to alter the CNR; the ratio of the CNR
at position 3 to the CNR at position 1 was 1.06, but had the noise content been equal
between positions 1 and 3, the ratio would have been only 1.02. The increase in noise
and the reduction in contrast seen between the two positions are both likely to be the
result of scatter from the edge of the detector.

Without the grid, however, the pattern was very different. There was no difference
among positions 1, 3, 4, and 6; and among positions 2, 5, 7, and 9. The CNR was
highest at positions 3, 4 and 6, and lowest at position 8. Positions 2 and 5 showed
values of CNR that were also near the bottom. This pattern is likely the result of a trade-
off between photon flux and scatter. The increased photon flux near the center of the
detector tends to decrease the relative contribution of noise in that area, but the center
of the detector also receives scatter from all directions, which reduces contrast and
increases noise. Thus, the highest CNR’s were found at the edges of the image, where
less scatter is received. The CNR’s were lowest in the middle of the image, where the
most scatter is received. It should also be noted that, while the CNR’s at positions 1, 3,
4 and 6 were not significantly different without a grid, the CNR at position 1 was still
lower than expected due to its proximity to the detector edge. (The sphere was not
moved between the acquisitions with and without a grid.)

For the remainder of the analyses, only position 5 was used, because it was the
location at which scatter was most isotropic (thus providing the best match to the PSF
model), and because it was also a position likely to be under the center of the breast.
Figure 7-8a and Table 7-2 show the CNR’s for each tube potential and target/filter
combination. (Table 7-8b displays the CNR data after scatter correction, which is discussed in the next section.) It should be noted that, although only a single phantom thickness of 4 cm was used for the CNR measurements, many of the kVp and target/filter combinations assessed would not be used clinically with a 4 cm breast thickness. Instead, they would be used over the range of clinical thicknesses, which are typically from about 2 cm to 8 cm. It should also be noted that, although all four target/filter combinations were analyzed for the CNR measurements, only Mo/Mo and Mo/Rh were used in the radiologist review of clinical images presented in the next chapter.

With a grid, the CNR decreased as the tube potential increased for all target/filter combinations, which is expected due to the decreasing photoelectric cross-section at higher photon energies. This effect was statistically significant for all target/filter combinations. The CNR for Mo/Mo fell by 16% from the lowest to the highest tube potential, 7% for Mo/Rh, 17% for W/Rh, and 22% for W/Ag. Without a grid, the trend of decreasing CNR with increasing tube potential was also true for Mo/Mo (12% decrease), W/Rh (19% decrease), and W/Ag (25% decrease). With Mo/Rh, the difference among tube potentials was not statistically significant without a grid.

The differences in CNR among target/filter combinations were also analyzed for each tube potential. At all tube potentials, the differences in CNR before scatter correction were statistically significant. For all tube potentials, with and without a grid, the highest CNR's were seen with Mo/Mo, and the lowest were seen with W/Ag. With a grid, Mo/Rh demonstrated the lowest CNR at 24 kVp and was not significantly different from W/Rh at 28 kVp; for all other cases, the CNR's decreased in the order: Mo/Mo,
Mo/Rh, W/Rh, and W/Ag. This trend makes sense, as the average beam energy increases in the same order, and an increase in beam energy causes a decrease in contrast.

In an attempt to explain the trends related to CNR, the CNR values for position 5 were graphed as a function of the MRE for the appropriate target/filter/tube potential combination (Figure 7-9). A larger MRE results in scatter travelling farther within the image and thus should produce a lower CNR. The SF was not considered because it underwent only very small changes (on the order of 0.01 both with a grid and without a grid) over the range of spectra tested. Before scatter correction, the slope of the best fit line with a grid was -1.1; without a grid, it was -0.34. The Pearson correlation coefficient was calculated, and the results of the correlation were $r = -0.67$ with a grid and $r = -0.89$ without a grid, both of which are significant below $p = 0.05$. Thus, a decrease in CNR is significantly correlated with an increase in MRE.

**Trends after scatter correction**

The CNR's without a grid ranged from 50-62% of those with a grid (Tables 7-3 and 7-4). This difference is larger than that seen before scatter correction, but may be due, at least partially, to the smaller CNR values seen after scatter correction (i.e., the CNR's ranged from 4.2-9.4 before scatter correction, and 1.35-3.44 after scatter correction). The standard deviations of the CNR measurements using a 28 kVp Mo/Mo spectrum after scatter correction were found to be about 0.89% of the CNR with a grid and 1.4% without a grid.

It should be noted that the numerical values of the CNR's decreased after scatter correction, compared to the values before scatter correction (Figures 7-7 and 7-8). The CNR’s after scatter correction ranged from 26-60% (average 44%) of those before
scatter correction. For example, for the case of the measurement using a 28 kVp Mo/Mo spectrum at position 5, with a grid, the CNR decreased from 8.69 before scatter correction to 2.91 after scatter correction. The change in CNR did not depend on position of the sphere with a grid, but did depend on position without a grid. Without a grid, the scatter-corrected CNR averaged 37% of the initial CNR at positions 2, 5, and 8, but averaged 29% of the initial CNR at the other positions. This trend caused the relationship between CNR and position without a grid to change after scatter correction, as explained later in this section.

This difference is caused by a decrease in the contrast, as measured by dividing the difference of the mean pixel values of the sphere and the background by the mean pixel value of the background. The contrast after scatter correction ranged from 12-23% (average 18%) of the contrast before scatter correction. For example, for the same measurement conditions cited previously, the contrast decreased from 0.15 before scatter correction (with a mean pixel value of 444.6 in the sphere and 524.9 in the background) to 0.031 after scatter correction (mean pixel value of 414.3 in the sphere and 424.5 in the background). The relatively large change in pixel value in the background, compared to the smaller change in the sphere, was consistent among all the measurements. The change in contrast did not depend on position of the sphere with a grid, but did depend on position without a grid. Without a grid, the scatter-corrected contrast was about 16% of the initial contrast at positions 2, 5, and 8, but averaged 13% of the initial contrast at the other positions.

As expected from the results of the NPS analysis, the noise improved after scatter correction, with the standard deviations of the pixel values in the background
averaging only 49% of those before scatter correction. For the same example measurement cited previously, the standard deviation decreased from 9.25 before scatter correction to 4.61 after scatter correction. The change in noise was not dependent on sphere position, regardless of grid presence. Because the CNR decreased despite the lower standard deviation, it is apparent that the decrease in contrast dominated the improvement in noise.

However, as explained earlier in this chapter, comparisons between CNR values in this manner are not valid when two systems have different MTF’s. These numbers do not imply that the CNR perceived by a person viewing a scatter-corrected image is inferior to that of an uncorrected image; the visual perception of an object is not described completely by a single number. The "meaning" of the CNR value has a frequency dependence. The visibility of an object depends on the contributions of the subject contrast, NPS, and MTF, as well as display factors such as window and level settings. The differences in magnitude between the CNR values before and after scatter correction are mentioned only to explain the different scales used to demonstrate the trends shown in Figures 7-7 and 7-8. The trends among CNR values can still be analyzed and compared, even though the magnitude of the CNR values cannot.

The CNR trends prior to scatter correction with respect to sphere position were similar to the CNR trends post-scatter correction for the images acquired with the grid, as shown in Figure 7-7 and Table 7-3. The differences among positions were significant (with p values of 0.004 with a grid and 0.01 without a grid). The difference between the maximum value (position 2) and minimum value (position 9) with respect to sphere
position was 17%, which is very similar to the 19% difference between those positions seen before scatter correction.

Without a grid, there was much less variation in CNR as a function of sphere position after scatter correction; the differences between the maximum value (position 2) and minimum value (position 7) decreased from 32% before scatter correction to 18% after scatter correction. Thus, the total variation with respect to position without a grid became similar to that seen with a grid. With a grid, positions 1, 2, 3, and 5 were statistically equivalent; positions 4 and 6 were equivalent; and positions 7 and 9 were equivalent. Without a grid, positions 1 through 6 were equivalent, and positions 7 and 9 were equivalent.

It appears that the CNR at the positions that were formerly lowest (positions 2, 5, and 8) were increased to the level of the other detector positions (Figure 7-7). This observation supports the theory that the CNR at these locations was originally lowest due to the contribution of scatter, the effects of which have been removed after scatter correction. After scatter correction, the distribution of the CNR's with respect to position followed roughly the same pattern with and without a grid, as opposed to the very different patterns seen before scatter correction. This pattern is consistent with that which would be expected from the heel effect and inverse square law, rather than scatter contribution.

The CNR's after scatter correction for each tube potential and target/filter combination, for sphere position 5, are shown in Figure 7-8b and Table 7-4. With a grid, the CNR still decreased as the tube potential increased for W/Rh (a total decrease of 13% from 24 to 32 kVp). In the case of Mo/Mo, Mo/Rh, and W/Ag, the decreases were
17%, 12%, and 14% respectively, but the CNR's at 28 and 32 kVp became nearly identical (no statistically significant difference). Without a grid, the trend of decreasing CNR with increasing tube potential held only for W/Rh (20% decrease) and W/Ag (25% decrease). The differences among the tube potentials without a grid were not statistically significant for any combination of tube potentials using Mo/Mo or Mo/Rh.

With respect to target/filter combination, the trends in CNR seen before scatter correction no longer held after scatter correction. With a grid, the CNR decreased in the order: Mo/Mo, W/Rh, Mo/Rh, and W/Ag. Without a grid, the order of CNR increase changed for each tube potential. The differences among target/filter combinations did not meet statistical significance for 28 kVp without a grid; however, the differences among target/filter combinations were significant for all tube potentials with a grid and for 24 kVp and 32 kVp without a grid.

Theoretically, if the scatter correction program worked perfectly, there should be no correlation between the post-scatter correction CNR values and MRE. Figure 7-9 shows a plot of the CNR values as a function of MRE; the slope with a grid is equal to -0.13 (compared to -1.1 before scatter correction), and the slope without a grid is equal to -0.024 (compared to -0.34 before scatter correction). The correlation coefficients are $r = -0.24$ with a grid and $r = -0.37$ without a grid. Although these values are not equal to zero, they indicate only weak correlations that are not statistically significant. These results indicate that the scatter correction program does, in fact, remove the correlation between CNR and MRE.

Hypothesis 5-1 stated that the CNR is improved by scatter correction. This hypothesis was formulated without realizing that the CNR's before and after scatter
correction could not be directly compared, as described earlier in the section discussing the results of the CNR investigation; thus, it cannot be proved or disproved.

Hypothesis 5-2 stated that the CNR without a grid is improved by the scatter correction algorithm to a level similar to that with a grid; this statement is shown to be false. Scatter correction removes the correlation between CNR and MRE, and it removes the effect of scatter on the pattern of CNR values with respect to position on the detector. It does not, however, make the CNR values equivalent. There are several points to consider with respect to Hypothesis 5-2. First, because AEC was used to determine the mAs used for image acquisition, the images without a grid were acquired with about half the mAs of those with the grid. Because so much of the signal at the detector was scatter, these images were inherently more noisy, and they may have been acted on by the Wiener filter differently than those acquired with a grid. The $C$ value was not optimized separately without a grid, due to the fact that clinical images could not be acquired without a grid. Thus, it may be the case that a different $C$ is optimal for the case of no grid. Lastly, it was found in Chapter 5 that the PSF fit the scatter data much better with a grid than without a grid, and that a bi-exponential function was necessary to describe the scatter kernel without a grid more accurately. Thus, it may be necessary to implement a bi-exponential PSF in order to fully realize the CNR improvement without a grid.

**Modulation Transfer Function**

**Procedures**

The MTF was determined by use of a line pair phantom and the method described by Droege and Morin, which depends on measurements of the standard deviation $M’$ measured for each frequency group (Figure 7-10). Also needed are
measurements of the mean value $M$ and standard deviation $N$ of uniform “light” and “dark” areas, where the light area is comprised of the same thickness of lead used to make the bars in the phantom, and the dark area does not contain lead (Figure 7-10). The MTF for each frequency is calculated using Equations 7-4 through 7-7.

$$MTF(f) = \frac{\pi \sqrt{2} M(f)}{4 M_0}$$  \hspace{1cm} (7-4)

$$M = \sqrt{M''^2 - N^2}$$  \hspace{1cm} (7-5)

$$N^2 = \frac{N_{\text{Light}}^2 - N_{\text{Dark}}^2}{2}$$  \hspace{1cm} (7-6)

$$M_0 = \frac{M_{\text{Light}} - M_{\text{Dark}}}{2}$$  \hspace{1cm} (7-7)

The exposures used to calculate the MTF were acquired with the detector covered by the acrylic phantom. The high-contrast line pair resolution pattern used contained frequency groups in a range from 2 to 10 line pairs per mm (lp/mm) (model 07-521, Fluke® Biomedical, Everett, WA). The spatial resolution steps used in this study were 2, 2.5, 3.15, 4, 5, 6.3, 7.1, 8, 9, and 10 lp/mm. The techniques used for the MTF calculations included Mo/Mo and Mo/Rh, both with and without the grid, with tube voltages of 24, 28, and 32 kVp. The mAs for each combination of exposure parameters was the same as that used for the NPS images. The paddle was in place for all exposures.

Most of the MTF measurements were made 6 cm from the chest wall edge of the detector, centered from right to left. The pattern was oriented so that the lines were perpendicular to the anode-cathode axis (i.e., parallel to the chest wall edge of the detector). However, because the projected focal spot changes size in the direction parallel to the anode-cathode axis (in this case, in the anterior-chest wall direction), the
dependence of the MTF on position was investigated by repeating the MTF measurements for one set of exposure parameters (28 kVp Mo/Mo), with and without a grid, at four different distances from the chest wall: 1 cm (position 1), 6 cm (position 2), 11 cm (position 3), and 16 cm (position 4). Due to the size of the resolution pattern, each resolution step was exposed separately, with the pattern moved for each exposure so that each ROI could be placed without a shift in location between frequency groups.

Results and Discussion

The differences in the MTF as a function of position were minor, with the MTF at position 4 roughly 0.05 higher than the MTF at position 1 regardless of frequency or grid status (Figure 7-11). The dependence on the grid was also minor, with the MTF about 0.05 higher with the grid than without at all frequencies. The MTF showed no dependence on tube potential or target/filter combination (Figure 7-12).

The scatter correction program was applied to the MTF images in the same way described previously. Because the line pair phantom is very thin, the apparent thickness of the acrylic slab was used for the entire image, which also prevented the calculation of a very large apparent thickness for the lead lines. Use of the apparent thickness tables for lead would have resulted in the use of inappropriate values for the SF and MRE (those corresponding to a thickness of 10 cm, the maximum thickness in the tables) and the potential for artifacts. After processing the images with the scatter correction program, an attempt was made to re-measure the MTF, but it soon became apparent that the scatter correction program resulted in ringing in the phantom image, which made it impossible to acquire reliable MTF measurements (Figure 7-13). Ringing is a problem for the line pair phantom images because the phantom is composed of sharp
edges and a very dense material, but ringing was not noticed on any of the clinical images.

As an alternative, semi-quantitative measure of spatial resolution, images of the ACR mammography phantom (model 18-220, Fluke® Biomedical, Everett, WA) were acquired (Figure 7-14). The ACR phantom is composed of a tissue-equivalent material, and contains specks with diameters of 0.54, 0.40, 0.32, 0.24, and 0.16 mm, which are intended as a visual measure of spatial resolution. Unlike the lead bars of the line-pair phantom, the specks in the ACR phantom are of a clinically-representative size, shape, and material. The specks are organized into groups of 6 specks of each diameter, with a score assigned to the number of speck groups seen; a group is scored as 1.0 if at least 4 of the 6 specks are seen, and 0.5 if 2 or 3 specks are seen.5 The Selenia® Quality Control manual states that a passing score for the Selenia® system is 4.0 (i.e., visibility of the 0.24 mm speck group).61 The images were acquired with the phantom centered along the chest wall edge of the detector, as it is used for clinical quality control. Images were acquired with Mo/Mo and Mo/Rh using tube potentials of 24, 28, and 32 kVp, with and without a grid. The mAs was determined by the AEC system.

The phantom was scored for each the raw images, and then the images were processed with the scatter correction program. The images were windowed and leveled so that the image appearance was approximately matched before and after scatter correction, and the images were viewed on a one-to-one scale (i.e., one pixel of the image was equal to one pixel on the monitor). The phantom was then scored again. Although some ringing of the phantom edges was noted after scatter correction, it was
not in a location that impeded visualization of the speck groups. No ringing was visible in the images of the individual specks.

The number of speck groups seen was equal to 4.0 for all images acquired, both before and after scatter correction (Figure 7-14). In fact, all 6 specks were seen for the 4th speck group in all images. None of the specks in the 5th group were clearly seen in any image. Thus, it appears that the scatter correction program is unlikely to degrade the spatial resolution to an unacceptable level. This measure is not a substitute for the MTF, however; nor does it permit a conclusive analysis to support or deny any of the hypotheses related to the MTF (Hypotheses 4-1, 4-2, and 4-3).

Summary

As mentioned in Chapter 6, the Wiener filter parameter $C$ is a compromise between blurring and noise suppression. The value used was that chosen by radiologists, and also that value that made the most sense based on the NPS data. The NPS, CNR, and spatial resolution are dependent on the value of $C$ selected, however, so the post-processing measurements presented here are easily changed if $C$ is changed.

The data shown in this section clearly indicate that the scatter correction program improves the noise content of the images. The MTF could not be quantitatively measured, and the CNR values could not be directly compared before and after scatter correction due to the effect of the algorithm on spatial resolution. However, the trends related to CNR as a function of position suggest that the effects of scatter have been reduced, particularly without a grid. The correlation between CNR and MRE also loses statistical significance as a result of scatter correction, which suggests that the algorithm
has the intended effect of scatter reduction. A more telling and more relevant test of contrast and spatial resolution is the radiologist preference study presented in Chapter 8.
Table 7-1. CNR values as a function of sphere position, before scatter correction, for Mo/Mo at 28 kVp.

<table>
<thead>
<tr>
<th>Position</th>
<th>CNR with grid</th>
<th>CNR without grid</th>
<th>Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>8.44</td>
<td>7.10</td>
<td>0.84</td>
</tr>
<tr>
<td>2</td>
<td>8.98</td>
<td>6.48</td>
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</tr>
<tr>
<td>3</td>
<td>8.92</td>
<td>7.78</td>
<td>0.87</td>
</tr>
<tr>
<td>4</td>
<td>8.38</td>
<td>7.80</td>
<td>0.93</td>
</tr>
<tr>
<td>5</td>
<td>8.69</td>
<td>6.55</td>
<td>0.75</td>
</tr>
<tr>
<td>6</td>
<td>8.47</td>
<td>8.00</td>
<td>0.95</td>
</tr>
<tr>
<td>7</td>
<td>7.55</td>
<td>6.65</td>
<td>0.88</td>
</tr>
<tr>
<td>8</td>
<td>7.95</td>
<td>5.79</td>
<td>0.73</td>
</tr>
<tr>
<td>9</td>
<td>7.41</td>
<td>6.77</td>
<td>0.91</td>
</tr>
</tbody>
</table>

Table 7-2. CNR values as a function of target/filter combination and tube potential, before scatter correction, for position 5.

<table>
<thead>
<tr>
<th>Target</th>
<th>Filter</th>
<th>kV</th>
<th>CNR with grid</th>
<th>CNR without grid</th>
<th>Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mo</td>
<td>Mo</td>
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<td>9.40</td>
<td>6.99</td>
<td>0.74</td>
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<tr>
<td></td>
<td></td>
<td>28</td>
<td>8.69</td>
<td>6.55</td>
<td>0.75</td>
</tr>
<tr>
<td></td>
<td></td>
<td>32</td>
<td>7.91</td>
<td>6.12</td>
<td>0.77</td>
</tr>
<tr>
<td>Mo</td>
<td>Rh</td>
<td>24</td>
<td>8.41</td>
<td>5.89</td>
<td>0.70</td>
</tr>
<tr>
<td></td>
<td></td>
<td>28</td>
<td>8.29</td>
<td>5.91</td>
<td>0.71</td>
</tr>
<tr>
<td></td>
<td></td>
<td>32</td>
<td>7.80</td>
<td>5.99</td>
<td>0.77</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>24</td>
<td>8.79</td>
<td>5.82</td>
<td>0.66</td>
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<tr>
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<td></td>
<td>28</td>
<td>8.28</td>
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<tr>
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<td></td>
<td>32</td>
<td>7.31</td>
<td>4.70</td>
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</tr>
<tr>
<td>W</td>
<td>Ag</td>
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<td>8.51</td>
<td>5.56</td>
<td>0.65</td>
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<tr>
<td></td>
<td></td>
<td>28</td>
<td>7.23</td>
<td>4.70</td>
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<tr>
<td></td>
<td></td>
<td>32</td>
<td>6.60</td>
<td>4.16</td>
<td>0.63</td>
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</table>
Table 7-3. CNR values as a function of sphere position, after scatter correction, for Mo/Mo at 28 kVp.

<table>
<thead>
<tr>
<th>Position</th>
<th>CNR with grid</th>
<th>CNR without grid</th>
<th>Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2.82</td>
<td>1.65</td>
<td>0.58</td>
</tr>
<tr>
<td>2</td>
<td>2.94</td>
<td>1.70</td>
<td>0.58</td>
</tr>
<tr>
<td>3</td>
<td>2.93</td>
<td>1.63</td>
<td>0.56</td>
</tr>
<tr>
<td>4</td>
<td>2.75</td>
<td>1.52</td>
<td>0.55</td>
</tr>
<tr>
<td>5</td>
<td>2.91</td>
<td>1.65</td>
<td>0.57</td>
</tr>
<tr>
<td>6</td>
<td>2.69</td>
<td>1.65</td>
<td>0.61</td>
</tr>
<tr>
<td>7</td>
<td>2.53</td>
<td>1.40</td>
<td>0.55</td>
</tr>
<tr>
<td>8</td>
<td>2.67</td>
<td>1.50</td>
<td>0.56</td>
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<tr>
<td>9</td>
<td>2.48</td>
<td>1.42</td>
<td>0.57</td>
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Table 7-4. CNR values as a function of target/filter combination and tube potential, after scatter correction, for position 5.

<table>
<thead>
<tr>
<th>Target</th>
<th>Filter</th>
<th>kV</th>
<th>CNR with grid</th>
<th>CNR without grid</th>
<th>Ratio</th>
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<tbody>
<tr>
<td>Mo</td>
<td>Mo</td>
<td>24</td>
<td>3.44</td>
<td>1.74</td>
<td>0.50</td>
</tr>
<tr>
<td></td>
<td></td>
<td>28</td>
<td>2.91</td>
<td>1.65</td>
<td>0.57</td>
</tr>
<tr>
<td></td>
<td></td>
<td>32</td>
<td>2.87</td>
<td>1.69</td>
<td>0.59</td>
</tr>
<tr>
<td>Mo</td>
<td>Rh</td>
<td>24</td>
<td>3.02</td>
<td>1.55</td>
<td>0.51</td>
</tr>
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<td></td>
<td></td>
<td>28</td>
<td>2.66</td>
<td>1.56</td>
<td>0.59</td>
</tr>
<tr>
<td></td>
<td></td>
<td>32</td>
<td>2.67</td>
<td>1.66</td>
<td>0.62</td>
</tr>
<tr>
<td>W</td>
<td>Rh</td>
<td>24</td>
<td>3.09</td>
<td>1.80</td>
<td>0.58</td>
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<td></td>
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<td>2.85</td>
<td>1.62</td>
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<td>32</td>
<td>2.69</td>
<td>1.44</td>
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<td>W</td>
<td>Ag</td>
<td>24</td>
<td>2.96</td>
<td>1.80</td>
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<td>2.62</td>
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<td></td>
<td>32</td>
<td>2.56</td>
<td>1.35</td>
<td>0.53</td>
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Figure 7-1. Location of the 9 ROI’s used for the NPS and CNR measurements.

Figure 7-2. Sample 2D NPS.
Figure 7-3. Sample 1D NPS curves showing the total NPS, NPS from stochastic sources only, and NPS from nonstochastic sources only, A) with a grid, and B) without a grid, before and after scatter correction. Techniques used were 28 kVp with a Mo/Mo spectrum. The total and stochastic curves overlap.
Figure 7-4. 1D NPS curves showing the total NPS for tube potentials of 24, 28, and 32 kVp, before and after scatter correction, for A) Mo/Mo with a grid, B) Mo/Mo without a grid, C) Mo/Rh with a grid, D) Mo/Rh without a grid, E) W/Rh with a grid, F) W/Rh without a grid, G) W/Ag with a grid, and H) W/Ag without a grid.
Figure 7-4. Continued.
Figure 7-4. Continued.
Figure 7-4. Continued.
Figure 7-5. Total NPS for a 28 kVp Mo/Mo spectrum as a function of position on detector, for the positions shown in Figure 7-1, A) with a grid, and B) without a grid.
Figure 7-6. Plot profile across the low-contrast disk A) before scatter correction and B) after scatter correction. The hump displayed after scatter correction is an artifact, and is discussed in more detail in Appendix F.
Figure 7-7. Comparison of CNR values for the sphere positions shown in Figure 7-1, for a 28 kVp Mo/Mo spectrum, A) before scatter correction, B) after scatter correction, and C) after scatter correction, with an expanded scale. The error for each measurement was assumed to be equal to that measured for the 28 kV Mo/Mo data point at position 9. The error bars indicate the 95% confidence intervals.
Figure 7-8. CNR for each combination of target, filter, grid status, and tube potential, for sphere position 5, A) before scatter correction, B) after scatter correction, and C) after scatter correction, with an expanded scale. The error for each measurement was assumed to be equal to that measured for the 28 kV Mo/Mo data point at position 9. The error bars indicate the 95% confidence intervals.
Figure 7-9. CNR’s before and after scatter correction plotted versus MRE, A) with a grid and B) without a grid. The equations describing the best fit lines are shown.
Figure 7-10. The line pair phantom used, with dark area, light area, and a frequency group labeled. There are 15 frequency groups total.
Figure 7-11. MTF as a function of position, A) with a grid and B) without a grid. Distances from the chest wall are as follows: Position 1 = 1 cm, Position 2 = 6 cm, Position 3 = 11 cm, and Position 4 = 16 cm.
Figure 7-12. MTF as a function of tube potential, A) with a grid and B) without a grid.
Figure 7-13. Ringing of the line pair phantom caused by the Fourier transform. The ringing is most apparent visually around the thick border of the phantom, but it affects the line pairs, as well.
Figure 7-14. Image of the ACR phantom acquired with Mo/Mo at 28 kVp, with a grid A) before scatter correction and B) after scatter correction. The images have been windowed and leveled so that the appearance of the gray background is similar between the two images. The images are magnified to show the 4th speck group C) before scatter correction and D) after scatter correction.
CHAPTER 8
RADIOLOGIST PREFERENCE STUDY

The final step in the project was an assessment by radiologists of the impact of the scatter correction program on clinical image quality. The purpose of the radiologist review was to compare scatter-corrected images to raw (unprocessed) images for an image set consisting of clinical patient mammograms. Because clinical images were used, this portion of the study used only images acquired with a grid in place.

Study Design and Methodology

IRB approval was granted to acquire a set of 100 clinical patient images from a hospital. The images were from five different Mo-target Selenia® units (no W-target units were located at the facility for which IRB approval was obtained). All images were standard screening mammograms. Patients with breast implants were not included. The images were selected to obtain a roughly even mix of cranio-caudal (CC) and mediolateral oblique (MLO) views and right and left breasts. Both small and large field sizes were included, although not in equal numbers because of the relatively infrequent use of the small field of view at the facility. Breasts of all common thicknesses and densities were included. The raw images were burned to CD from the Selenia® unit and anonymized using the Sante DICOM® Editor (version 3.1.16, Santesoft LTD, Athens, Greece). The CD's containing identifiable patient information were then destroyed.

The raw images were processed with the scatter correction program. The use of raw images was necessary because the Hologic™ image processing manipulates the actual pixel values rather than just the image appearance. Ideally, the scatter-corrected images would be processed with the proprietary Hologic™ image processing before viewing by the radiologists, and the corrected images compared to the standard clinical
(processed) Selenia® images. The benefit of this scenario would be that the image appearance would be similar to that to which the radiologists are accustomed, so they would be able to assess the diagnostic utility of the images. It is possible to remove the raw images from the Selenia®, perform the scatter correction on them, and then have them undergo the Hologic™ processing directly on the Selenia® unit or at the workstation. However, when this process was attempted, it was discovered that the Hologic™ processing includes some form of manipulation which is incompatible with the scatter correction, as evidenced by artifacts at the breast edges and over-enhancement of detail (Figure 8-1). The details of the proprietary Hologic™ processing are unknown, and it was not feasible either to replicate the Hologic™ processing or to “reverse engineer” it in a way which would make the scatter-corrected images compatible. Thus, the radiologists performed the image comparison using the raw Selenia® images compared to the scatter-corrected images. The purpose of this comparison was only to assess whether the raw images had improved, not to compare them to standard clinical images, because the scatter correction alone was not sufficient to produce images with an appearance similar to that produced by the commercial processing. Thus, radiologist preference, rather than diagnostic utility, was assessed.

From the set of images described above, a subset of 50 images was selected for processing by the scatter correction program. Images with obviously interesting features such as masses or calcifications were preferentially included, and the remainder of the subset was constructed to provide a relatively even distribution of breast thicknesses, breast densities, mammographic views, and field sizes (Appendix G).
A diagnostic review workstation (consisting of two Barco 5-megapixel monitors with Hologic™ SecurView software, version 7.0.1) was used to view the images for comparison. The Hologic™ workstations are designed to apply Hologic™ image processing to raw images automatically when they are imported into the database. This processing was avoided with the assistance of a Hologic™ service engineer, who, at the direction of a Hologic™ physicist, disabled the "LORAD IP" license in the workstation manager. Because raw images are never imported into the workstation at the facility, this action had no clinical impact. In addition to disabling the license, the images also had to have all date-related DICOM® tags ("Study Date" (0008,0020), "Series Date" (0008,0021), "Acquisition Date" (0008,0022), and "Image Date" (0008,0023)) in the DICOM® header dated within 5 days of the date the images were imported.

Each scatter-corrected image was presented side-by-side with its uncorrected raw image, with the location of the images on the right or left randomized. This arrangement was produced by randomly labeling one image as "CC" and the other as "MLO" in the DICOM® header (tags "Series Description" (0008,103E), "Protocol Name" (0018,1030), "View Position" (0018,5101), and unnamed tag (0019,1083)). The laterality of the original image was preserved in the DICOM® header. A hanging protocol was set up to display full-size CC and MLO images of the same laterality next to each other on the two monitors.

The images were presented to the radiologists in two sets of 25 image pairs, with the two sets viewed in separate sessions to prevent fatigue. Three MQSA-qualified radiologists viewed the images and were asked to assign a preference score to each
image pair as follows: 1, strongly prefer image on left; 2, slightly prefer image on left; 3, no preference; 4, slightly prefer image on right; 5, strongly prefer image on right,74 for each of the following categories:75

1. Detection of the pectoral muscle  
2. Visibility of the nipple  
3. Visibility of the skin  
4. Sharpness  
5. Contrast  
6. Visibility of microcalcifications  
7. Visibility of masses  
8. Overall image quality

Numbers 1, 2, 3, 6 and 7 were rated “zero” if not applicable or if the radiologist said the feature being scored was not visible in either of the paired images.

The majority of image-comparison “reader” studies encountered in the literature used ratings of image quality based on a numerical scale, generally from 1 to 3, 4 or 5, with each number associated with a description of image quality such as “poor,” “fair,” “good,” and “excellent.”74-84 Sometimes these descriptions were further related to the diagnostic utility of the images (i.e., “poor” corresponding to a nondiagnostic image, “fair” corresponding to an image with diagnostic quality somewhat impaired, etc.)74, 81, 84 However, this scoring method was not used for my study because the radiologists were not accustomed to interpreting raw images. Using a scheme similar to those described above may have resulted in a large number of images receiving “poor” or “nondiagnostic” ratings. The scatter correction program does not replace the remainder of the manufacturer processing needed to make a raw image reach diagnostic quality.

Magnification and window/leveling were permitted during the reading session. Before the first image set was assessed, a trial session consisting of paired images was conducted to accustom the radiologists to the rating method and the appearance of raw
images. The images used for the trial session were not included in the 50 paired images that were assessed. The trial session continued until the radiologists felt comfortable with the rating method and images.

For image analysis, the preference scores were modified from the scheme detailed above as follows: 1, strongly prefer uncorrected image; 2, slightly prefer uncorrected image; 3, no preference; 4, slightly prefer scatter-corrected image; 5, strongly prefer scatter-corrected image. Thus, if the scatter-corrected image was on the left and the radiologist assigned a score of “1”, the score recorded for analysis became “5” (strongly prefer scatter-corrected image). If the scatter-corrected image was on the right and the radiologist assigned a score of “1”, the score recorded for analysis remained “1”.

The scores in each category were analyzed using the one-sample Wilcoxon signed rank test, as other similar studies have done, with comparison of the scores to a median of 3 (indicating no preference) and $p < 0.05$ considered significant. Scores of zero were not included in the analysis. Because the goal of this assessment was to determine whether the scatter-corrected images were significantly different from the uncorrected images, either for better or for worse, a two-sided test was used. The test was performed for each radiologist individually and for all radiologists’ scores together, for each category.

Additionally, the differences in scores among the three radiologists were assessed for each category using a Kruskal-Wallis test with $p < 0.05$ considered significant. The Kruskal-Wallis test was chosen because a nonparametric test was needed to compare scores among more than two groups. The categories often had unequal numbers of
non-zero scores (because the radiologists did not agree on what was visible and not visible in the images), so a test for unmatched samples was required. OriginPro (version 8.6, OriginLab® Corporation, Northampton, MA) was used for all statistical analyses.

**Results and Discussion**

The average scores were greater than 3 for all categories and all radiologists, indicating that the scatter-corrected rather than original images were favored in all cases (Table 8-1). The results of the statistical analysis indicate that these results are highly significant; all $p$ values were much less than the significance level of 0.05 (Table 8-2). The scores were highest for sharpness, contrast, and overall image quality (average 4.9), and lowest for the pectoral muscle (average 4.0), nipple (average 4.1), and skin (average 4.0). In fact, the score for overall image quality tracked closely with that for sharpness and contrast for all radiologists, regardless of the scores in the other categories. Radiologist 1 commented that the appearance of the skin and nipple do not matter clinically.

The main goal of the algorithm was to improve contrast, so it is encouraging that the score for this category is so high. The appearance of calcifications (average score 4.7) and masses (average score 4.3) also improved, which suggests that the algorithm might improve diagnostic accuracy.

The scores for sharpness, contrast, and overall image quality were very consistent, and there were no significant differences among radiologists in those categories. All the other categories showed significant differences among radiologists (Table 8-3). The scores for the skin appearance were the most different: one radiologist strongly preferred the skin appearance after scatter correction (average 4.9), while the other two had comparatively little preference (averages of 3.5 and 3.7). The scores for
the nipple also showed a wide variation among radiologists, with scores of 3.6, 4.1, and 4.7. It is interesting that the categories which produced the most disagreement were those that one radiologist stated were not important. The categories most indicative of image quality (sharpness, contrast, and overall image quality) had a high level of agreement. Also interesting is that fact that Radiologist 3 tended to give lower scores than the other radiologists for most categories, but had the highest score for overall image quality.

The high scores for sharpness, contrast, and overall image quality lend support to the hypotheses that spatial resolution and contrast are improved, which is important since Hypotheses 4-1 through 5-1 could not be proved by the image quality measurements described in Chapter 7. This finding also suggests that the reason the CNR values after scatter correction were less than those of the original images is because the MTF has improved as a result of the scatter correction algorithm; systems with better MTF's require a lower CNR for a given level of image quality.\textsuperscript{68} Because the image quality was shown to have improved significantly despite the lower CNR values seen in Chapter 7, the logical conclusion is that the MTF has improved. This conclusion is also supported by the significant increase in sharpness, because sharpness is associated with spatial resolution and the MTF.

The favorable results of this radiologist preference study indicate that the scatter correction algorithm does cause a significant improvement in the raw images. Although they do not match the commercially-processed images in appearance, the scatter-corrected raw images could be used as input into the commercial processing, potentially
improving the final diagnostic image. An improvement in the final image could potentially lead to improved diagnostic accuracy.

It should be noted that this radiologist review study did not include images from W-target mammography units. However, the scatter data from the W-target unit were similar to those from the Mo-target unit, and the agreement between the physical measurements and the Monte Carlo data was equally good for both target materials. It is unlikely that the target material would significantly impact the effectiveness of the scatter correction program.
Table 8-1. Average scores for radiologist preference in each rated category.

<table>
<thead>
<tr>
<th></th>
<th>Radiologist 1</th>
<th>Radiologist 2</th>
<th>Radiologist 3</th>
<th>All radiologists</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pectoral muscle</td>
<td>4.3</td>
<td>4.2</td>
<td>3.5</td>
<td>4.0</td>
</tr>
<tr>
<td>Nipple</td>
<td>4.1</td>
<td>4.7</td>
<td>3.6</td>
<td>4.1</td>
</tr>
<tr>
<td>Skin</td>
<td>3.5</td>
<td>4.9</td>
<td>3.7</td>
<td>4.0</td>
</tr>
<tr>
<td>Sharpness</td>
<td>4.9</td>
<td>4.9</td>
<td>4.9</td>
<td>4.9</td>
</tr>
<tr>
<td>Contrast</td>
<td>4.9</td>
<td>4.9</td>
<td>4.9</td>
<td>4.9</td>
</tr>
<tr>
<td>Calcifications</td>
<td>4.5</td>
<td>4.6</td>
<td>4.9</td>
<td>4.7</td>
</tr>
<tr>
<td>Masses</td>
<td>4.6</td>
<td>4.5</td>
<td>3.8</td>
<td>4.3</td>
</tr>
<tr>
<td>Overall</td>
<td>4.9</td>
<td>4.9</td>
<td>5.0</td>
<td>4.9</td>
</tr>
</tbody>
</table>

Table 8-2. The $p$ values for each category for differences in score from a median score of 3 (indicating no preference). A $p$ value less than 0.05 is considered significant (all are significant). The $p$ values displayed as zero were below the minimum value reported by the statistical software program.

<table>
<thead>
<tr>
<th></th>
<th>Radiologist 1</th>
<th>Radiologist 2</th>
<th>Radiologist 3</th>
<th>All radiologists</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pectoral muscle</td>
<td>3.1E-05</td>
<td>2.4E-07</td>
<td>1.07E-02</td>
<td>5.16E-10</td>
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<tr>
<td>Nipple</td>
<td>3.7E-05</td>
<td>3.7E-09</td>
<td>4.34E-06</td>
<td>8.48E-14</td>
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<tr>
<td>Skin</td>
<td>3.5E-03</td>
<td>1.8E-15</td>
<td>2.51E-08</td>
<td>2.22E-16</td>
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<tr>
<td>Sharpness</td>
<td>1.8E-15</td>
<td>1.8E-15</td>
<td>3.55E-15</td>
<td>0.00E+00</td>
</tr>
<tr>
<td>Contrast</td>
<td>1.8E-15</td>
<td>1.8E-15</td>
<td>3.55E-15</td>
<td>0.00E+00</td>
</tr>
<tr>
<td>Calcifications</td>
<td>7.5E-09</td>
<td>1.5E-08</td>
<td>7.45E-09</td>
<td>0.00E+00</td>
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<tr>
<td>Masses</td>
<td>9.3E-10</td>
<td>9.8E-04</td>
<td>1.56E-02</td>
<td>1.60E-10</td>
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<tr>
<td>Overall</td>
<td>1.8E-15</td>
<td>1.8E-15</td>
<td>3.55E-15</td>
<td>0.00E+00</td>
</tr>
</tbody>
</table>

Table 8-3. The $p$ values for each category for difference among radiologists. A $p$ value less than 0.05 is considered significant.

<table>
<thead>
<tr>
<th></th>
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<tr>
<td>Pectoral muscle</td>
<td>1.52E-03</td>
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<td>Nipple</td>
<td>2.35E-06</td>
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<td>Skin</td>
<td>8.58E-13</td>
</tr>
<tr>
<td>Sharpness</td>
<td>1.43E-01</td>
</tr>
<tr>
<td>Contrast</td>
<td>1.70E-01</td>
</tr>
<tr>
<td>Calcifications</td>
<td>1.68E-02</td>
</tr>
<tr>
<td>Masses</td>
<td>1.00E-02</td>
</tr>
<tr>
<td>Overall</td>
<td>1.70E-01</td>
</tr>
</tbody>
</table>
Figure 8-1. Scatter-corrected image after processing with the proprietary Hologic™ algorithms. The Hologic™ processing over-enhances the breast edges and detail, turning them a bright white.
This aim of this study is to reduce the presence of scatter in digital mammography by developing a point spread function (PSF) to describe the scatter and subsequently using a spatially-variant Wiener filter to remove scatter described by the PSF from the image. The PSF is a model of the way in which scatter spreads from an impulse. The form of the PSF is shown in Equation 3-5, and was used in earlier work by Ducote and Molloi. \(^{36}\)

The PSF depends on a number of factors, including the target/filter combination, tube potential, presence of absence of an antiscatter grid, and the apparent thickness of each pixel in the image. The apparent thickness is a function of the ratio of the signal in a pixel to the unattenuated signal, and is used as a way to estimate the thickness of breast tissue above each pixel.

The algorithm employed in this study is unique because it takes into account such a wide variety of factors in order to produce the most accurate PSF for each pixel. The PSF is then removed from the image on a pixel-by-pixel basis.

**Physical Measurements**

The parameters used to describe the PSF, scatter fraction (SF) and mean radial extent (MRE) were obtained by recording the signal acquired under lead disks of varying diameters and plotting these values as a function of disk radius. A tissue-equivalent phantom, BR12, was used as a scattering material. The SF and MRE measurements were acquired using four phantom thicknesses, five tube potentials, and four target/filter combinations, both with and without an antiscatter grid. The data acquired were used to find eight surface functions describing the SF and MRE as a
function of tube potential and thickness: one for each target/filter and grid combination. Thus, the final PSF developed was a function of target/filter combination, tube potential, tissue thickness, and the presence or absence of a grid.

The data acquired show that the SF is independent of tube potential in the mammography range, thus supporting Hypothesis 1-1a. The SF is also independent of target/filter combination for the four combinations tested (Mo/Mo, Mo/Rh, W/Rh, and W/Ag), so Hypothesis 1-1b is also supported. The SF is higher by a factor of 3.0-5.2 (depending on phantom thickness) without a grid than with a grid, which supports Hypothesis 1-1c. The SF also more than doubles as breast thickness increases from 2 cm to 8 cm, both with and without a grid, which supports Hypothesis 1-2a. Hypothesis 1-2b was investigated by Monte Carlo simulations, which is discussed in the next section. Hypothesis 1-2c, which states that the SF is unrelated to the compressed breast area, is also shown to be true. This last hypothesis was tested by measuring the SF with the entire detector covered with BR12 (an extreme situation), and comparing the results to those obtained when only one slab of BR12 is used.

Hypothesis 2-1a states that the MRE increases as the tube potential increases; this hypothesis is not supported, because the relationship between tube potential and MRE is complex. The shape of the curves relating MRE to tube potential are concave, with the kV corresponding to the minimum MRE depending on phantom thickness and target/filter combination. Hypothesis 2-2b, which states that the MRE increases as the average beam energy increases due to the target/filter combination selected, is likewise not supported due to the complexity of the relationships. However, the trends in MRE with respect to grid presence clearly indicate that the MRE is greater without a grid than
with a grid, thus supporting Hypothesis 2-1c; the ratio of the MRE without a grid is about 3-6 times higher than that with a grid, depending on tube potential, target/filter combination, and thickness. The trend of increasing MRE with increasing breast thickness is also clearly shown: over the range of 2-8 cm, the MRE approximately doubles without a grid and nearly triples with a grid. Therefore, Hypothesis 2-2a is supported, as well. Hypothesis 2-2b was investigated by Monte Carlo simulations and is discussed in the next section. Hypothesis 2-2c states that the MRE is unrelated to the compressed breast area. This was tested in the same way as Hypothesis 1-2c, and is likewise found to be true.

**Monte Carlo Simulations**

The results of the physical measurements were then verified with Monte Carlo simulations. The geometrical setup of the simulated mammography machine replicated the experimental conditions as closely as possible, including an accurate simulation of all distances involved, materials used, spectra, and grid design. The main departure from the experimental setup was the use of an impulse of radiation rather than a full-field situation with simulated lead disks. This design was chosen for two reasons: first, because the PSF is defined as the response to an impulse, this arrangement is a way of obtaining data in a more direct manner than that used for the physical measurements; and second, because the full-field situation would have required prohibitively long processing times in order to obtain satisfactory statistics under the lead disks.

The SF was determined by assessing the angle of impact of the detected events; if a particle did not enter the detector with an angle equal to the angle of incidence of the impulse, it was assumed to be scattered. The accuracy of the MRE was determined by placing ring-shaped tallies around the impulse and fitting those data to the PSF.
The agreement between the physical measurements and the results of the Monte Carlo simulations is good. The SF values are very similar both with and without a grid: the absolute differences between the measured and simulated SF’s range from -0.04 to +0.01 (average -0.01) without a grid and -0.03 to +0.07 (average 0.02) with a grid. The Monte Carlo data for MRE agree with the physical data with goodness-of-fit values ranging from 0.96-1.00 with a grid, and 0.65-0.86 without a grid. Interestingly, the data suggest that a bi-exponential PSF equation would more accurately fit the data, particularly without a grid, than does the form used in this research. The two exponential terms likely come from the separate contributions of coherent and Compton scattering. Because it is not possible to separate these scatter sources during the acquisition of physical measurements, a scatter correction program using the bi-exponential PSF would have to rely on data solely from simulations.

Monte Carlo simulations were also used to assess the likely impact of breast glandularity on the SF and MRE. The breast composition assumed for the physical measurements and the rest of the Monte Carlo measurements was 50% glandular and 50% adipose (50/50), but for this portion of the study, 30/70 and 70/30 compositions were also considered. Little impact is seen for either the SF or the MRE, thus supporting Hypotheses 1-2b and Hypothesis 2-2b. Hypothesis 3-1 states that the difference in apparent thickness between glandular and adipose tissue would be insufficient to affect
the scatter correction program, which is also found to be true. These findings suggest that breast glandularity does not present a problem for the scatter correction program.

**The Scatter Correction Program**

A computer program was written in Matlab® to remove the scatter from raw (unprocessed) mammography images by use of a Wiener filter. The SF and MRE used are dependent on information pulled from the DICOM® header regarding target/filter combination, tube potential, and grid status. The SF and MRE are also dependent on the apparent thickness of each pixel in the image. Thus, each pixel in the image is assigned an apparent thickness corresponding to the amount of tissue directly above it, which in turn is used to determine the appropriate parameters for the PSF. The image is decomposed into mask images consisting of all the pixels of each apparent thickness (in 1 mm increments). The spatially-variant filtration is carried out in frequency space on each mask image separately, and the images are then recombined to form the final image.

**Image Quality Measurements**

The image quality before and after scatter correction was assessed by measurement of spatial resolution, contrast-to-noise ratio (CNR), and the noise power spectrum (NPS).

Before scatter correction, differences in the MTF as a function of position in the chest wall-anterior direction were small. The dependence on the grid was also minor, with the MTF about 0.05 higher with the grid than without at all frequencies. The MTF shows no dependence on tube potential or target/filter combination. However, the MTF after scatter correction could not be calculated due to ringing of the line pair phantom caused by the algorithm. Thus, the hypotheses relating to the MTF (Hypotheses 4-1,
4-2, and 4-3) could not be proved or disproved. During the course of the project, however, it was discovered that the algorithm has a noticeable impact of the perceived sharpness of the image. The images after scatter correction appear sharper than those before scatter correction. (This increase in sharpness is confirmed by the radiologist preference study, discussed in the next section.) This finding suggests that the MTF is potentially improved by the scatter correction program.

Trends in CNR were analyzed with respect to target/filter combination, tube potential, grid status, and position on the detector. With a grid, the trends depending on position on the detector are consistent with what would be expected as a result of the heel effect and inverse square law, and remain unchanged after scatter correction. Without a grid, however, a larger variation is seen as a function of position, and the pattern is dominated by the influence of scatter. After scatter correction, the variations with respect to position without a grid decrease and follow roughly the same pattern as the trends with a grid, suggesting that the influence of scatter is reduced.

Before scatter correction and with a grid, the CNR decreases as the tube potential increases for all target/filter combinations, which is expected due to the decreasing photoelectric cross-section at higher photon energies. Without a grid, the trend of decreasing CNR with increasing tube potential is also true for all except Mo/Rh. The pattern with respect to target/filter combination is not dependent on grid status, and in most cases, the CNR decreases in the order: Mo/Mo, Mo/Rh, W/Rh, and W/Ag. After scatter correction, the trends with respect to kVp become less identifiable, and the trends with respect to target/filter combination do not hold at all.
In an attempt to explain the trends related to CNR, the CNR values were graphed as a function of the MRE for the appropriate target/filter/tube potential combination, and the Pearson correlation coefficient was calculated. A larger MRE results in scatter travelling farther within the image, and thus, should produce a lower CNR. Before scatter correction, a decrease in CNR is significantly correlated with an increase in MRE. After scatter correction, the correlation between CNR and MRE is removed, which suggests that the impact of scatter has been reduced.

During the course of the project, it was found (by literature review) that CNR values can be compared directly only for systems with identical MTF’s. Because the algorithm affects sharpness, it likely changes the MTF of the images. Thus, it is not valid to make a direct comparison between the CNR’s obtained before and after scatter correction. This fact was not known when Hypothesis 5-1 was written. Hypothesis 5-1 states that the CNR is improved by the scatter correction program, and this hypothesis cannot be proved or disproved. Hypothesis 5-2 states that the CNR without a grid is improved by the scatter correction algorithm to a level similar to that with a grid, which is shown to be false. Scatter correction removes the correlation between CNR and MRE, and it removes the effect of scatter on the pattern of CNR values with respect to position on the detector. It does not, however, make the CNR values equivalent. This finding may be due to a number of factors: the images without a grid were acquired with about half the mAs (tube current-time product) of those with a grid; the algorithm was not optimized separately for images without a grid due to the lack of clinical images without a grid; and the fit of the PSF to the scatter data is better with a grid than without a grid,
so a different PSF (a bi-exponential function, as discussed above) may be needed to fully realize CNR improvement without a grid.

The image noise decreases after scatter correction by 60-70%, with little dependence on tube potential, target/filter combination, or grid status. These findings support Hypothesis 6-1. Hypothesis 6-2 states that the scatter correction program improves the NPS without a grid to become similar to the NPS with a grid; this hypothesis is based on the premise that the NPS’s with and without a grid differ to begin with, which is shown by these data to be false. The NPS’s are very similar both with and without a grid prior to the scatter correction, and they are again very similar after the correction. Noise was also assessed as a function of position on the detector, and the scatter correction program does not alter the trends seen as a function of position.

Radiologist Preference Study

IRB approval was acquired to collect the raw data from 100 clinical images obtained by routine mammography, process them with the scatter correction program, and use these images in a reviewer preference study. Fifty pairs of original and scatter-corrected images were shown side-by-side (with the position on left or right randomized) on a diagnostic review workstation. Three MQSA-qualified radiologists were asked to score the images in the following categories: appearance of the pectoral muscle, appearance of the nipple, appearance of the skin, sharpness, contrast, visibility of microcalcifications, visibility of masses, and overall image quality. The scoring system was from 1 to 5, with 1 indicating a strong preference for the image on the left, 3 indicating no preference, and 5 indicating a strong preference for the image on the right.

With knowledge of the positions on the left or right of each image, the scores were reassigned so that 1 corresponded to a strong preference for the original image and 5
indicated a strong preference for the scatter-corrected image. The scores for each
category were then analyzed using a one-sample Wilcoxon test. All radiologists had a
significant preference for the scatter-corrected images in all categories. The scores
were particularly high for sharpness, contrast, and image quality, with all three
averaging 4.9 out of 5. Hypothesis 7-1, that the radiologists prefer the scatter-corrected
image to the uncorrected images, is shown to be true.

Final Conclusions

This study had two goals. The primary goal was to determine whether scatter
correction using spatially-variant filtration can improve the image quality of clinical
mammograms. This goal was accomplished, and the image quality of clinical
mammograms was found to have improved, as shown by the radiologist preference study.

The secondary goal of this study was to determine whether the scatter correction
program corrects scatter sufficiently well to retain adequate image quality while allowing
removal of the antiscatter grid. This goal was to be assessed only from the image
quality measurements, because clinical images acquired without a grid are not
available. Due to the inability to obtain MTF measurements or to directly compare CNR
measurements before and after scatter correction, this idea could neither be proved nor
disproved.

A summary of all the conclusions related to each hypothesis is found in Table 9-1.

Possibilities for Future Research

This project raises several topics for future research:

1. Application of the scatter correction program to magnification imaging, which
does not use a grid. This idea was not pursued in this study because
modification of the mammography equipment would be required in order to obtain apparent thickness measurements.

2. Further development of the bi-exponential PSF, which appears to provide a better fit for the non-grid Monte Carlo data. The SF and MRE used for a bi-exponential fit would have to be obtained entirely from simulations, rather than physical measurements, so an entirely new data set would be required.

3. Conducting a radiologist preference study with patient images acquired without a grid. Because the FDA requires the use of a grid for clinical mammography, images would have to come from an IRB-approved research study involving each participant's informed consent and the acquisition of extra images (beyond the clinical mammograms). If improvement in the gridless images is demonstrated, it might be possible to remove the grid entirely, which would potentially allow for a reduction in absorbed dose to the breast.

4. This radiologist review study did not include images from W-target mammography units. Therefore, although the scatter data were similar between the Mo-target and W-target units, and the physical measurements and Monte Carlo data agreed equally well, the application of the scatter correction program to images from W-target units should be tested in order to verify the process for clinical mammography.

5. Application of the scatter correction program to digital breast tomosynthesis (DBT) images. During the time in which this study was being conducted, DBT was FDA-approved for use on patients. This modality involves the acquisition of images from multiple angles over the breast, and allows the reconstruction of 3-dimensional data sets consisting of "slices" through the breast. The current implementation of DBT does not use a grid, and therefore these images might have even more to gain from scatter correction than do conventional mammography images.
<table>
<thead>
<tr>
<th>Hypothesis Number</th>
<th>Hypothesis</th>
<th>Conclusion</th>
<th>Location in Text</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-1a</td>
<td>The SF is independent of the tube potential in the mammography range of 24 kV-34 kV.</td>
<td>Supported</td>
<td>Chapter 4, Scatter Kernel Measurements, Results and Discussion, The scatter fraction</td>
</tr>
<tr>
<td>1-1b</td>
<td>The SF is independent of the target/filter combination for Mo/Mo, Mo/Rh, W/Rh, and W/Ag.</td>
<td>Supported</td>
<td>Chapter 4, Scatter Kernel Measurements, Results and Discussion, The scatter fraction</td>
</tr>
<tr>
<td>1-1c</td>
<td>The SF is higher without an antiscatter grid than with a grid.</td>
<td>Supported</td>
<td>Chapter 4, Scatter Kernel Measurements, Results and Discussion, The scatter fraction</td>
</tr>
<tr>
<td>1-2a</td>
<td>The SF increases with breast thickness.</td>
<td>Supported</td>
<td>Chapter 4, Scatter Kernel Measurements, Results and Discussion, The scatter fraction</td>
</tr>
<tr>
<td>1-2b</td>
<td>The SF is unrelated to the breast glandularity for a fixed apparent thickness.</td>
<td>Supported</td>
<td>Chapter 5, Effects of Breast Composition, Scatter Fraction and Mean Radial Extent</td>
</tr>
<tr>
<td>1-2c</td>
<td>The SF is unrelated to the compressed breast area.</td>
<td>Supported</td>
<td>Chapter 4, Ancillary Investigations, The Effect of Phantom Size</td>
</tr>
<tr>
<td>2-1a</td>
<td>The MRE increases as the tube potential increases in the mammography range of 24 kV-34 kV.</td>
<td>Not Supported¹</td>
<td>Chapter 4, Scatter Kernel Measurements, Results and Discussion, The mean radial extent</td>
</tr>
<tr>
<td>2-1b</td>
<td>The MRE increases as the average beam energy increases due to the target/filter combination selected (in order from least to greatest: Mo/Mo, Mo/Rh, W/Rh, W/Ag).</td>
<td>Not supported¹</td>
<td>Chapter 4, Scatter Kernel Measurements, Results and Discussion, The mean radial extent</td>
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<tr>
<td>2-1c</td>
<td>The MRE is greater without an antiscatter grid than with a grid.</td>
<td>Supported</td>
<td>Chapter 4, Scatter Kernel Measurements, Results and Discussion, The mean radial extent</td>
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<tr>
<td>2-2a</td>
<td>The MRE increases with breast thickness.</td>
<td>Supported</td>
<td>Chapter 4, Scatter Kernel Measurements, Results and Discussion, The mean radial extent</td>
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<tr>
<td>2-2b</td>
<td>The MRE is unrelated to the breast glandularity for a fixed apparent thickness.</td>
<td>Supported</td>
<td>Chapter 5, Effects of Breast Composition, Scatter Fraction and Mean Radial Extent</td>
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<tr>
<td>2-2c</td>
<td>The MRE is unrelated to the compressed breast area.</td>
<td>Supported</td>
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Table 9-1. Continued.

<table>
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<tr>
<th>Hypothesis Number</th>
<th>Hypothesis</th>
<th>Conclusion</th>
<th>Location in Text</th>
</tr>
</thead>
<tbody>
<tr>
<td>3-1</td>
<td>The difference in apparent thickness between glandular tissue and adipose tissue is insufficient to affect the scatter correction algorithm.</td>
<td>Supported</td>
<td>Chapter 5, Effects of Breast Composition, Apparent Thickness</td>
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<tr>
<td>4-1</td>
<td>The MTF is improved by the scatter correction program at low frequencies.</td>
<td>Inconclusive²</td>
<td>Chapter 7, Modulation Transfer Function, Results and Discussion</td>
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<tr>
<td>4-2</td>
<td>The scatter correction program does not change the limiting spatial resolution, which is determined by the Nyquist frequency.</td>
<td>Inconclusive²</td>
<td>Chapter 7, Modulation Transfer Function, Results and Discussion</td>
</tr>
<tr>
<td>4-3</td>
<td>The scatter correction program improves the MTF without a grid to become similar to the MTF with a grid.</td>
<td>Inconclusive²</td>
<td>Chapter 7, Modulation Transfer Function, Results and Discussion</td>
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<td>The CNR is improved by the scatter correction program.</td>
<td>Not testable³</td>
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<td>The scatter correction program improves the CNR without a grid to become similar to the CNR with a grid.</td>
<td>Not Supported⁴</td>
<td>Chapter 7, Contrast-to-Noise Ratio, Results and Discussion, Trends after scatter correction</td>
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<tr>
<td>6-1</td>
<td>The scatter correction program lowers the stochastic noise content of the images.</td>
<td>Supported</td>
<td>Chapter 7, Noise Power Spectrum, Results and Discussion</td>
</tr>
<tr>
<td>6-2</td>
<td>The scatter correction program improves the NPS without a grid to become similar to the NPS with a grid.</td>
<td>Not Supported⁵</td>
<td>Chapter 7, Noise Power Spectrum, Results and Discussion</td>
</tr>
<tr>
<td>7-1</td>
<td>The radiologists prefer the scatter-corrected images to the uncorrected images.</td>
<td>Supported</td>
<td>Chapter 8, Results and Discussion</td>
</tr>
</tbody>
</table>

¹The relationship between MRE and beam energy is complex.
²The MTF post-scatter correction could not be tested with the high-contrast line pair phantom due to ringing of the phantom caused by the scatter correction program.
³CNR comparisons can be made only between systems with identical MTF’s.
⁴The CNR without a grid was lower than the CNR with a grid.
⁵The NPS’s with and without a grid were already similar before scatter correction, and were again similar after scatter correction.
Gilmore's Derivation of the Point Spread Function

Assume a polar coordinate system. The steady-state energy flow, $E(r, \theta)$, from a point in space into an annulus surrounding it, per unit time and per unit arc length (Figure A-1), is described by the equation:

$$E(r, \theta) \propto \frac{\theta r E(r) - \theta (r + \Delta r) E(r + \Delta r)}{\Delta r}$$ (A-1)

The goal is to find $q(r)$, the energy transferred across an arc length of unit length (i.e., $\theta r = 1$). Starting with the energy flow into an annulus ($\theta = 2\pi$) and normalizing it by dividing by $2\pi r$ gives:

$$q(r) \propto \frac{2\pi r E(r) - 2\pi (r + \Delta r) E(r + \Delta r)}{2\pi \Delta r},$$ (A-2)

which simplifies to:

$$q(r) \propto \frac{r E(r) - (r + \Delta r) E(r + \Delta r)}{r \Delta r}.$$ (A-3)

Next, $E(r + \Delta r)$ can be expanded to $E(r) + \Delta r \left( \frac{\partial E}{\partial r} \right)$, giving:

$$q(r) \propto \frac{r E(r) - (r + \Delta r) \left( E(r) + \Delta r \left( \frac{\partial E}{\partial r} \right) \right)}{r \Delta r}.$$ (A-4)

Expansion, simplification, and discarding of the non-linear term in $\Delta r$ lead to:

$$q(r) \propto \left[ \frac{E(r)}{r} + \frac{\partial E}{\partial r} \right],$$ (A-5)

and letting $k$ be equal to the proportionality constant yields:
\[ q(r) = -k \left[ \frac{E(r)}{r} + \frac{\partial E}{\partial r} \right]. \quad (A-6) \]

Making the substitution \( \frac{E(r)}{r} + \frac{\partial E}{\partial r} = \frac{1}{r} \frac{\partial}{\partial r} [r(E(r))] \) results in:

\[ q(r) = -k \frac{1}{r} \frac{\partial}{\partial r} [r(E(r))], \quad (A-7) \]

which represents the steady-state flow of energy into a unit area.

Now, consider the net energy flow \( (q_{\text{net}}) \) into a small area \( r\Delta r \Delta \theta \) (Figure A-2), which is given by:

\[ q_{\text{net}} = q(r)r\Delta \theta - q(r + \Delta r)(r + \Delta r)\Delta \theta, \quad (A-8) \]

where the first arc length crossed is given by \( r\Delta \theta \) and the second arc length crossed is given by \( (r + \Delta r)\Delta \theta \). The term \( q(r + \Delta r) \) can be expanded to \( q(r) + \Delta r \left( \frac{\partial q}{\partial r} \right) \), yielding:

\[ q_{\text{net}} = q(r)r\Delta \theta - (r + \Delta r)\Delta \theta \left[ q(r) + \Delta r \left( \frac{\partial q}{\partial r} \right) \right]. \quad (A-9) \]

Expansion, simplification, and discarding of the non-linear term in \( \Delta r \) lead to:

\[ q_{\text{net}} = -\Delta r \Delta \theta \left[ q(r) + r \left( \frac{\partial q}{\partial r} \right) \right], \quad (A-10) \]

and making the substitution \( \frac{\partial}{\partial r} [r(q(r))] = q(r) + r \frac{\partial q}{\partial r} \) results in:

\[ q_{\text{net}} = -\Delta r \Delta \theta \frac{\partial}{\partial r} [r(q(r))]. \quad (A-11) \]

Recognizing that the total energy in an area, \( r\Delta r \Delta \theta E(r) \), is proportional to the net energy flow, allows these quantities to be set equal to one another with the inclusion of a proportionality constant, \( b \).

\[ -\Delta r \Delta \theta \frac{\partial}{\partial r} [r(q(r))] = br \Delta r \Delta \theta E(r) \quad (A-12) \]

\[ -\frac{\partial}{\partial r} [r(q(r))] = br E(r) \quad (A-13) \]

Substituting Equation A-6 into Equation A-13 gives:
\[
\frac{\partial}{\partial r} \left[ kr \left( \frac{E(r)}{r} + \frac{\partial E}{\partial r} \right) \right] = b r E(r), \quad (A-14)
\]

and applying the derivative to the term in brackets results in:
\[
k \left[ 2 \frac{\partial}{\partial r} E(r) + r \frac{\partial^2}{\partial r^2} E(r) \right] = b r E(r). \quad (A-15)
\]

Moving the term on the right hand side to the left and dividing through by \( rk \) produces:
\[
\frac{\partial^2}{\partial r^2} E(r) + \frac{2}{r} \frac{\partial}{\partial r} E(r) - \frac{b}{k} E(r) = 0. \quad (A-16)
\]

Replacing \( \frac{b}{k} \) with \( \frac{1}{r_0^2} \) results in:
\[
\frac{\partial^2}{\partial r^2} E(r) + \frac{2}{r} \frac{\partial}{\partial r} E(r) - \frac{1}{r_0^2} E(r) = 0, \quad (A-17)
\]

which can be solved by recognizing that making the substitutions
\[
E(r) = r^{-1/2} Z_{1/2}(r/r_0) \quad (A-18)
\]

and \( x = r/r_0 \) will turn it into a modified Bessel equation of order 1/2 as follows:
\[
Z''_{1/2} + \frac{1}{x} Z'_{1/2} - \left( 1 + \frac{1}{4x^2} \right) Z_{1/2} = 0. \quad (A-19)
\]

A Bessel equation of this form has the solution:
\[
Z_{1/2} = C_1 K_{1/2} + C_2 I_{1/2}, \quad (A-20)
\]

where \( C_1 \) and \( C_2 \) are arbitrary constants and \( K_{1/2} \) and \( I_{1/2} \) are modified Hankel and

Bessel functions, respectively. However, because the amount of energy decreases in

magnitude the farther it is from the origin, \( C_2 \) must be equal to zero, since \( I_{1/2} \) is an

increasing function of \( x \) (and, therefore, \( r \)). Substituting Equation A-20, with \( C_2 = 0 \), into

Equation A-18, results in:
\[
E(r) = C_1 r^{-1/2} K_{1/2}(r/r_0), \quad (A-21)
\]

the solution of which is:
\[ E(r) = \frac{c_1}{r} \sqrt{\frac{r_0 \pi}{2}} e^{-r/r_0}. \quad (A-22) \]

If the total point spread function is normalized such that \( \int_0^{2\pi} \int_0^\infty E(r)rdrd\theta = 1 \), then Equation A-22 can be normalized to:

\[ E(r) = \frac{1}{2\pi r_0} e^{-r/r_0}, \quad (A-23) \]

which is equal to the PSF \( h_G(r) \) of Equation 3-3.

**Modification of the PSF to Include Direct Mapping Term**

Equation A-23 describes the spread of energy through a two-dimensional medium. However, when an impulse of radiation is applied to a detector, not all of the energy spreads; some is directly mapped to the point of impact, which is described by the two-dimensional Dirac delta function, \( \frac{\delta(r)}{\pi r} \). The fraction that spreads is called the scatter fraction, \( \rho \). The fraction that maps directly is therefore equal to \( 1 - \rho \), and Equation A-23 can be modified to include the direct mapping term as follows:

\[ E(r) = (1 - \rho) \frac{\delta(r)}{\pi r} + \frac{\rho}{2\pi r_0} e^{-r/r_0} \quad (A-24) \]

For this equation to be valid in a physical sense, it must be true that the sum of the radiation which maps directly and that which spreads be equal to 1. This normalization can be shown by integration over \( rdrd\theta \).

\[ \int_0^{2\pi} \int_0^\infty E(r)rdrd\theta = \frac{(1-\rho)}{\pi} \int_0^{2\pi} \int_0^\infty \delta(r)drd\theta + \frac{\rho}{2\pi r_0} \int_0^{2\pi} \int_0^\infty e^{-r/r_0} drd\theta \quad (A-25) \]

\[ = 2(1 - \rho) \int_0^\infty \delta(r)dr + \frac{\rho}{r_0} \int_0^\infty e^{-r/r_0} dr \quad (A-26) \]

\[ = (1 - \rho) \int_{-\infty}^\infty \delta(r)dr + \frac{\rho}{r_0} \int_0^\infty e^{-r/r_0} dr \quad (A-27) \]

\[ = (1 - \rho) + \rho \]

\[ = 1 \quad (A-29) \]
Removal of the factor of $\frac{1}{\pi}$ from Equation A-24 and replacement of the variable name $r_0$ by $k$ gives the scatter kernel $h_S(r)$ as published by Seibert et al:

$$h(r) = (1 - \rho) \frac{\delta(r)}{r} + \frac{\rho}{2kr} e^{-r/k}. \quad (A-30)$$

This is the form of the PSF given in Equation 3-4.
Figure A-1. Energy flow across an annulus (section).

Figure A-2. Energy flow across a small area $r\Delta r\Delta\theta$. 
<table>
<thead>
<tr>
<th>Thickness</th>
<th>2 cm</th>
<th>4 cm</th>
<th>6 cm</th>
<th>8 cm</th>
<th>2 cm</th>
<th>4 cm</th>
<th>6 cm</th>
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</thead>
<tbody>
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<td>-0.0703</td>
<td>-0.1143</td>
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<td>-0.0642</td>
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<td>0.9982</td>
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<td>R-squared</td>
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<td>0.9997</td>
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<td>0.9999</td>
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Table B-2. Coefficients for the quadratic equations describing apparent thickness LSR as a function of thickness, for tube potentials ranging from 22 – 38 kVp.

### Mo/Mo, large paddle, with grid

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<thead>
<tr>
<th>Tube potential</th>
<th>22</th>
<th>23</th>
<th>24</th>
<th>25</th>
<th>26</th>
<th>27</th>
<th>28</th>
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<tbody>
<tr>
<td>Quadratic term</td>
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<td>-0.0199</td>
<td>-0.0204</td>
<td>-0.0209</td>
<td>-0.0214</td>
<td>-0.0219</td>
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<td>-0.0229</td>
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<td>Linear term</td>
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<td>0.7300</td>
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<td>0.6978</td>
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<tr>
<td>Intercept</td>
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### Mo/Rh, large paddle, with grid

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<td>-0.0176</td>
<td>-0.0185</td>
<td>-0.0194</td>
<td>-0.0203</td>
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<tr>
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<td>0.6656</td>
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<td>0.6566</td>
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<tr>
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<td>0.1740</td>
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### Mo/Mo, large paddle, no grid

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W/Rh, large paddle, with grid

W/Ag, large paddle, with grid
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Table B-3. Coefficients of the surface fit for SF as a function of tube potential and thickness, for each target/filter combination, with and without a grid. The fit is of the form given by Equation 4-2.

<table>
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<tr>
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<th>Mo/Rh</th>
<th>W/Rh</th>
<th>W/Ag</th>
<th>Mo/Mo</th>
<th>Mo/Rh</th>
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Table B-4. Coefficients of the surface fit for MRE as a function of tube potential and thickness, for each target/filter combination, with and without a grid. The fit is of the form given by Equation 4-2.

<table>
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<tr>
<th>Coefficient</th>
<th>Mo/Mo</th>
<th>Mo/Rh</th>
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<th>W/Ag</th>
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APPENDIX C
MONTE CARLO CODE

28 kVp, Mo/Mo, 4 cm Phantom Thickness, Without a Grid

c 28 kV Mo/Mo 4cm
1  204 -0.001225 1 -2 10 -41 -3 4 imp:p=1 $air above breast
2  204 -0.001225 -6 :5 :-1 :2 :3 :-4 imp:p=0 $outside world
3  281  -1.45 -8 7 1 -2 -3 4 imp:p=1024 $breast support
4  253  -0.985 -14 1 8 -2 -3 4 imp:p=512 $breast bottom .5 cm
5  282  -1.2 12 -10 -2 1 -3 4 imp:p=1 $paddle
6  204 -0.001225 22 1 -2 -7 -3 4 imp:p=2048 $below breast support
7  204 -0.001225 9 -12 1 -2 -3 4 $air between paddle and breast
   (11 :12 :-9 ) imp:p=2
8  204 -0.001225 -11 9 -12 imp:p=2 $ disk (originally lead, replaced with air)
9  256  -4.27 -13 -22 24 imp:p=2048 $detector
10 253  -0.985 14 -15 -2 1 -3 4 imp:p=256 $breast .5-1 cm from bottom
11 253  -0.985 15 -16 -1 2 -3 4 imp:p=128 $breast 1-1.5 cm from bottom
12 253  -0.985 16 -17 1 -2 -3 4 imp:p=64 $breast 1.5-2 cm from bottom
13 253  -0.985 17 -18 1 -2 -3 4 imp:p=32 $breast 2-2.5 cm from bottom
14 253  -0.985 18 -19 1 -2 -3 4 imp:p=16 $breast 2.5-3 cm from bottom
15 253  -0.985 19 -20 -2 1 -3 4 imp:p=8 $breast 3-3.5 cm from bottom
16 253  -0.985 20 -9 1 -2 -3 4 imp:p=4 $breast 3.5-4 cm from bottom
18 256  -4.27 -22 24 1 -2 -3 4 622 imp:p=2048 $detector plane
19 204 -0.001225 -24 -2 6 1 -3 4 imp:p=0 $below detector
40 257  -10.28 41 -40 -3 1 -2 4 imp:p=1 $filter
41 204 -0.001225 1 -2 40 -5 -3 4 imp:p=1 $air above filter
595  256  -4.27 -595 13 -22 24 imp:p=2048 $ ring 1
596  256  -4.27 -596 595 -22 24 imp:p=2048 $ ring 2
597  256  -4.27 -597 596 -22 24 imp:p=2048 $ ring 3
598  256  -4.27 -598 597 -22 24 imp:p=2048 $ ring 4
599  256  -4.27 -599 598 -22 24 imp:p=2048 $ ring 5
600  256  -4.27 -600 599 -22 24 imp:p=2048 $ ring 6
601  256  -4.27 -601 600 -22 24 imp:p=2048 $ ring 7
602  256  -4.27 -602 601 -22 24 imp:p=2048 $ ring 8
603  256  -4.27 -603 602 -22 24 imp:p=2048 $ ring 9
604  256  -4.27 -604 603 -22 24 imp:p=2048 $ ring 10
605  256  -4.27 -605 604 -22 24 imp:p=2048 $ ring 11
606  256  -4.27 -606 605 -22 24 imp:p=2048 $ ring 12
607  256  -4.27 -607 606 -22 24 imp:p=2048 $ ring 13
608  256  -4.27 -608 607 -22 24 imp:p=2048 $ ring 14
609  256  -4.27 -609 608 -22 24 imp:p=2048 $ ring 15
610  256  -4.27 -610 609 -22 24 imp:p=2048 $ ring 16
611  256  -4.27 -611 610 -22 24 imp:p=2048 $ ring 17
612  256  -4.27 -612 611 -22 24 imp:p=2048 $ ring 18
613  256  -4.27 -613 612 -22 24 imp:p=2048 $ ring 19
1  px 0 $chest wall edge
2  px 24 $anterior edge
3  py 15 $left edge
4  py -15 $right edge
5  pz 70 $top of geometry
6  pz -1 $1 cm below detector
7  pz 1.88 $bottom of breast support
8  pz 2 $top of breast support / bottom of breast
9  pz 6 $top of breast / bottom of disk
10  pz 6.4 $top of paddle
11  c/z 5 0 1
12  pz 6.2 $bottom of paddle / top of disk
13  c/z 5.515 0.01 0.005 $tally at point of impact
14  pz 2.5
15  pz 3
16  pz 3.5
17  pz 4
18  pz 4.5
19  pz 5
20  pz 5.5
22  pz 0 $top of detector
24  pz -0.02 $bottom of detector
40  pz 64.003 $top of filter
41  pz 64 $bottom of filter
595  c/z 5.515 0.01 0.1
596  c/z 5.515 0.01 0.3
597  c/z 5.515 0.01 0.5
598  c/z 5.515 0.01 0.7
599  c/z 5.515 0.01 0.9
600  c/z 5.515 0.01 1.1
601  c/z 5.515 0.01 1.3
602  c/z 5.515 0.01 1.5
603  c/z 5.515 0.01 1.7
604  c/z 5.515 0.01 1.9
605  c/z 5.515 0.01 2.1
606  c/z 5.515 0.01 2.3
mode  p
m204  7000.04p  -0.755636 $air
   8000.04p  -0.231475 18000.04p  -0.012889
m280  1000.04p  -0.111915 $water
   8000.04p  -0.888085
m281  6000.04p              1 $carbon fiber
m282  1000.04p              8 $polycarbonate
   6000.04p              5 8000.04p              2
m252  82000.04p            -1 $lead, den=-11.35
m253  1000.04p        -0.11 $breast tissue 50/50 den=-.985
   6000.04p        -0.465 7000.04p        -0.0185 8000.04p        -0.4025
   11000.04p        -0.001 15000.04p        -0.0005 16000.04p        -0.0015
   17000.04p        -0.001
m254  1000.04p        -0.1116 $breast tissue 30G/70A den=-.971
   6000.04p        -0.5182 7000.04p        -0.0139 8000.04p        -0.3527
   11000.04p        -0.001 15000.04p        -0.0003 16000.04p        -0.0013
   17000.04p        -0.001
m255  1000.04p        -0.1084 $breast tissue 70G/30A den=-.999
   6000.04p        -0.4118 7000.04p        -0.0231 8000.04p        -0.4523
   11000.04p        -0.001 15000.04p        -0.0007 16000.04p        -0.0017
   17000.04p        -0.001
m256  34000.04p             1 $amorphous selenium den=-4.27
m257  42000.04p             1 $molybdenum den=-10.28
m258  45000.04p             1 $rhodium  den=-12.41
m259  47000.04p             1 $silver   den=-10.49
m261  13000.04p             1 $aluminum den=-2.70
m262  1000.04p             8 $polyethylene terephthalate (PET) den=-1.37
   6000.04p              10 8000.04p
nps 1000000
28 kVp, Mo/Mo, 4 cm Phantom Thickness, With a Grid

c 28 kV Mo/Mo 4cm

1 204 -0.001225 1 -2 10 -41 -3 4 imp:p=1 $air above breast
2 204 -0.001225 5 -1 :2 -6 :3 -:4 imp:p=0 $outside world
3 281 -1.45 -8 7 1 -2 -3 4 imp:p=1024 $breast support
4 253 -0.985 -14 1 8 -2 -3 4 imp:p=512 $breast bottom .5 cm
5 282 -1.12 12 -10 -2 1 -3 4 imp:p=1 $paddle
6 204 -0.001225 70 1 -2 -7 -3 4 imp:p=1024 $air between breast support and grid
7 204 -0.001225 9 -12 1 -2 -3 4 $air between paddle and breast
(11 :12 :-9 ) imp:p=2
8 204 -0.001225 -11 9 -12 imp:p=2 $ disk (originally lead, replaced with air)
9 256 -4.27 -13 -22 24 imp:p=2048 $detector
10 253 -0.985 14 -15 -2 1 -3 4 imp:p=256 $breast .5-1 cm from bottom
11 253 -0.985 15 -16 1 -2 -3 4 imp:p=128 $breast 1-1.5 cm from bottom
12 253 -0.985 16 -17 1 -2 -3 4 imp:p=64 $breast 1.5-2 cm from bottom
13 253 -0.985 17 -18 1 -2 -3 4 imp:p=32 $breast 2-2.5 cm from bottom
14 253 -0.985 18 -19 1 -2 -3 4 imp:p=16 $breast 2.5-3 cm from bottom
15 253 -0.985 19 -20 -2 1 -3 4 imp:p=8 $breast 3-3.5 cm from bottom
16 253 -0.985 20 -9 1 -2 -3 4 imp:p=4 $breast 3.5-4 cm from bottom
18 256 -4.27 -22 24 1 -2 -3 4 622 imp:p=2048 $detector plane
### PARALLEL GRID SEGMENTS

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<td>-3 to -5 cm</td>
</tr>
<tr>
<td>60</td>
<td>-148 147</td>
<td>-5 to -7 cm</td>
</tr>
<tr>
<td>61</td>
<td>-147 146</td>
<td>-7 to -9 cm</td>
</tr>
<tr>
<td>62</td>
<td>-146 145</td>
<td>-9 to -11 cm</td>
</tr>
<tr>
<td>63</td>
<td>-145 144</td>
<td>-11 to -13 cm</td>
</tr>
<tr>
<td>64</td>
<td>-144 143</td>
<td>-13 to -15 cm</td>
</tr>
</tbody>
</table>

### CROSS GRID SEGMENTS

<table>
<thead>
<tr>
<th>Segment</th>
<th>Grid Points</th>
<th>Grid Dimensions</th>
</tr>
</thead>
<tbody>
<tr>
<td>65</td>
<td>171 -271</td>
<td>0-2 cm</td>
</tr>
<tr>
<td>66</td>
<td>172 -272</td>
<td>2-4 cm</td>
</tr>
<tr>
<td>67</td>
<td>173 -273</td>
<td>4-6 cm</td>
</tr>
<tr>
<td>68</td>
<td>174 -274</td>
<td>6-8 cm</td>
</tr>
<tr>
<td>69</td>
<td>175 -275</td>
<td>8-10 cm</td>
</tr>
<tr>
<td>70</td>
<td>176 -276</td>
<td>10-12 cm</td>
</tr>
</tbody>
</table>
*fill=4 (0 0 0 90 90 90 0 90 99.61 90 350.39)
71 0 177 -277 4 -3 51 -70 imp:p=2048 $cross grid 12-14 cm
*fill=4 (0 0 0 90 90 90 0 90 101.31 90 348.69)
72 0 178 -278 -3 4 51 -70 imp:p=2048 $cross grid 14-16 cm
*fill=4 (0 0 0 90 90 90 0 90 102.99 90 347.01)
73 0 179 -279 4 -3 51 -70 imp:p=2048 $cross grid 16-18 cm
*fill=4 (0 0 0 90 90 90 0 90 104.66 90 345.34)
74 0 180 -280 -3 4 51 -70 imp:p=2048 $cross grid 18-20 cm
*fill=4 (0 0 0 90 90 90 0 90 106.29 90 343.71)
75 0 181 -281 -3 4 51 -70 imp:p=2048 $cross grid 20-22 cm
*fill=4 (0 0 0 90 90 90 0 90 107.9 90 342.1)
76 0 182 -282 4 -3 51 -70 imp:p=2048 $cross grid 22-24 cm
*fill=4 (0 0 0 90 90 90 0 90 109.49 90 340.51)
c *************LATTICE CELLS**********************************************
90 260 -8.94 1 -2 -91 90 350 -351 u=2 imp:p=1 $90 degree grid line
91 204 -0.001225 #90 u=2 imp:p=1 $90 degree grid gap
92 0 -92 93 -351 350 -2 1 fill=2 u=1 lat=1 imp:p=1
93 260 -8.94 4 -3 -95 94 350 -351 u=3 imp:p=1 $90 degree cross grid line
94 204 -0.001225 #93 u=3 imp:p=1 $90 degree grid gap
95 0 -96 97 -351 350 -3 4 u=4 lat=1 fill=3 imp:p=1 $cross grid 0-2 cm
c *************CELLS BETWEEN CROSS GRID SEGMENTS*************************
100 204 -0.001225 1 -171 51 -3 4 imp:p=2048
101 204 -0.001225 271 -172 51 -3 4 imp:p=2048
102 204 -0.001225 272 -173 51 -3 4 imp:p=2048
103 204 -0.001225 273 -174 51 -3 4 imp:p=2048
104 204 -0.001225 274 -175 51 -3 4 imp:p=2048
105 204 -0.001225 275 -176 51 -3 4 imp:p=2048
106 204 -0.001225 276 -177 51 -3 4 imp:p=2048
107 204 -0.001225 277 -178 51 -3 4 imp:p=2048
108 204 -0.001225 278 -179 51 -3 4 imp:p=2048
109 204 -0.001225 279 -180 51 -3 4 imp:p=2048
110 204 -0.001225 280 -181 51 -3 4 imp:p=2048
111 204 -0.001225 281 -182 51 -3 4 imp:p=2048
112 204 -0.001225 282 -2 -70 -3 4 imp:p=2048
c *************CELLS BETWEEN PARALLEL GRID SEGMENTS*********************
120 204 -0.001225 -3 258 -51 1 -2 imp:p=2048
121 204 -0.001225 -158 257 50 -51 1 -2 imp:p=2048
122 204 -0.001225 -157 256 50 -51 1 -2 imp:p=2048
123 204 -0.001225 -156 255 50 -51 1 -2 imp:p=2048
124 204 -0.001225 -155 254 50 -51 1 -2 imp:p=2048
125 204 -0.001225 -154 253 50 -51 1 -2 imp:p=2048
126 204 -0.001225 -153 252 50 -51 1 -2 imp:p=2048
127 204 -0.001225 -152 151 50 -51 1 -2 imp:p=2048
128 204 -0.001225 -150 -159 50 -51 1 -2 imp:p=2048
129 204 -0.001225 259 -160 50 -51 1 -2 imp:p=2048
130 204 -0.001225 260 -161 50 -51 1 -2 imp:p=2048
131 204 -0.001225 261 -162 50 -51 1 -2 imp:p=2048
132 204 -0.001225 262 -163 50 -51 1 -2 imp:p=2048
133 204 -0.001225 263 -164 50 -51 1 -2 imp:p=2048
134 204 -0.001225 264 -165 50 -51 1 -2 imp:p=2048
135 204 -0.001225 265 4 50 -51 1 -2 imp:p=2048
595 256 -4.27 -595 13 -22 24 imp:p=2048 $ ring 1
596 256 -4.27 -596 595 -22 24 imp:p=2048 $ ring 2
597 256 -4.27 -597 596 -22 24 imp:p=2048 $ ring 3
598 256 -4.27 -598 597 -22 24 imp:p=2048 $ ring 4
599 256 -4.27 -599 598 -22 24 imp:p=2048 $ ring 5
600 256 -4.27 -600 599 -22 24 imp:p=2048 $ ring 6
601 256 -4.27 -601 600 -22 24 imp:p=2048 $ ring 7
602 256 -4.27 -602 601 -22 24 imp:p=2048 $ ring 8
603 256 -4.27 -603 602 -22 24 imp:p=2048 $ ring 9
604 256 -4.27 -604 603 -22 24 imp:p=2048 $ ring 10
605 256 -4.27 -605 604 -22 24 imp:p=2048 $ ring 11
606 256 -4.27 -606 605 -22 24 imp:p=2048 $ ring 12
607 256 -4.27 -607 606 -22 24 imp:p=2048 $ ring 13
608 256 -4.27 -608 607 -22 24 imp:p=2048 $ ring 14
609 256 -4.27 -609 608 -22 24 imp:p=2048 $ ring 15
610 256 -4.27 -610 609 -22 24 imp:p=2048 $ ring 16
611 256 -4.27 -611 610 -22 24 imp:p=2048 $ ring 17
612 256 -4.27 -612 611 -22 24 imp:p=2048 $ ring 18
613 256 -4.27 -613 612 -22 24 imp:p=2048 $ ring 19
614 256 -4.27 -614 613 -22 24 imp:p=2048 $ ring 20
615 256 -4.27 -615 614 -22 24 imp:p=2048 $ ring 21
616 256 -4.27 -616 615 -22 24 imp:p=2048 $ ring 22
617 256 -4.27 -617 616 -22 24 imp:p=2048 $ ring 23
618 256 -4.27 -618 617 -22 24 imp:p=2048 $ ring 24
619 256 -4.27 -619 618 -22 24 imp:p=2048 $ ring 25
620 256 -4.27 -620 619 -22 24 imp:p=2048 $ ring 26
621 256 -4.27 -621 620 -22 24 imp:p=2048 $ ring 27
622 256 -4.27 -622 621 -22 24 imp:p=2048 $ ring 28

1 px 0 $ chest wall edge
2 px 24 $ anterior edge
3 py 15 $ left edge
4 py -15 $ right edge
5 pz 70 $ top of geometry
6 pz -1 $ 1 cm below detector
7 pz 1.88 $ bottom of breast support
8 pz 2 $ top of breast support / bottom of breast
9 pz 6 $ top of breast / bottom of disk
10 pz 6.4 $ top of paddle
11 c/z 5 0 1
12 pz 6.2 $ bottom of paddle / top of disk
13 c/z 5.515 0.01 0.005 $tally at point of impact
14 pz 2.5
15 pz 3
16 pz 3.5
17 pz 4
18 pz 4.5
19 pz 5
20 pz 5.5
22 pz 0 $top of detector
24 pz -0.02 $bottom of detector
40 pz 64.003 $top of filter
41 pz 64 $bottom of filter
50 pz 1 $parallel grid bottom
51 pz 1.25 $parallel grid top / cross grid bottom
70 pz 1.5 $cross grid top
90 py 0 $90 degree grid line 1
91 py 0.0012 $90 degree grid line 2
92 py 0.01325
93 py -0.01325
94 px 1 $90 degree cross grid line 1
95 px 1.0012 $90 degree cross grid line 2
96 px 1.01325
97 px 0.98675

c ************PARALLEL GRID SEGMENTS***********************************
150 py -1
151 py 1
152 p 0 1 1.251 1 1 1.251 1 1.0384093250982636 0 $start 1-3 cm parallel grid cell
153 p 0 3 1.251 1 3 1.251 1 3.0768912491303004 0
154 p 0 5 1.251 1 5 1.251 1 5.11529889508697 0
155 p 0 7 1.251 1 7 1.251 1 7.153923630610599 0
156 p 0 9 1.251 1 9 1.251 1 9.192393401324786 0
157 p 0 11 1.251 1 11 1.251 1 11.23077128165043 0
158 p 0 13 1.251 1 13 1.251 1 13.269117958802419 0
159 p 0 -1 1.251 1 -1 1.251 1 -1.0384093250982636 0
160 p 0 -3 1.251 1 -3 1.251 1 -3.0768912491303004 0
161 p 0 -5 1.251 1 -5 1.251 1 -5.11529889508697 0
162 p 0 -7 1.251 1 -7 1.251 1 -7.153923630610599 0
163 p 0 -9 1.251 1 -9 1.251 1 -9.192393401324786 0
164 p 0 -11 1.251 1 -11 1.251 1 -11.23077128165043 0
165 p 0 -13 1.251 1 -13 1.251 1 -13.269117958802419 0

c ************CROSS GRID SEGMENTS*************************************
171 p 0 0.15 0 1 1.5 0.0230401578464344 0 0 $start 0-2cm cross grid cell
172 p 2 0 1.5 2 1 1.5 2.0691639917791018 0 0
173 p 4 0 1.5 4 1 1.5 4.115418709725438 0 0
174 p 6 0 1.5 6 1 1.5 6.161627826664412 0 0
c **********PARALLEL GRID SEGMENTS***********************************
252   p  0  3  1.251  1  3  1.251  1 3.0384093250982636 0  $end 1-3 cm parallel grid cell

253   p  0  5  1.251  1  5  1.251  1 5.0768912491303004 0
254   p  0  7  1.251  1  7  1.251  1 7.11529889508697 0
255   p  0  9  1.251  1  9  1.251  1 9.153923630610599 0
256   p  0 11  1.251  1 11  1.251  1 11.192393401324786 0
257   p  0 13  1.251  1 13  1.251  1 13.23077128165043 0
258   p  0 14.9461764082395 1.251  1 14.9461764082395 1.251  1 15  1
259   p  0  -3  1.251  1 -3  1.251  1 -3.0384093250982636 0
260   p  0  -5  1.251  1 -5  1.251  1 -5.0768912491303004 0
261   p  0  -7  1.251  1 -7  1.251  1 -7.11529889508697 0
262   p  0  -9  1.251  1 -9  1.251  1 -9.153923630610599 0
263   p  0  -11 1.251  1 -11 1.251  1 -11.192393401324786 0
264   p  0  -13 1.251  1 -13 1.251  1 -13.23077128165043 0
265   p  0 -14.9461764082395 1.251  1 -14.9461764082395 1.251  1 -15  1

c **********CROSS GRID SEGMENTS************************************
271   p  2  0  1.5  2  1  1.5  2 2.0230401578464344 0 0  $end 0-2cm cross grid cell
272   p  4  0  1.5  4  1  1.5  4 4.0691639917791018 0 0
273   p  6  0  1.5  6  1  1.5  6 6.115418709725438 0 0
274   p  8  0  1.5  8  1  1.5  8 8.161627826664412 0 0
275   p 10  0  1.5 10  1  1.5 10 10.207608545549499 0 0
276   p 12  0  1.5 12  1  1.5 12 12.253975306378356 0 0
277   p 14  0  1.5 14  1  1.5 14 14.300001838539491 0 0
278   p 16  0  1.5 16  1  1.5 16 16.346026544469307 0 0
279   p 18  0  1.5 18  1  1.5 18 18.392398567249684 0 0
280   p 20  0  1.5 20  1  1.5 20 20.438346524861556 0 0
281   p 22  0  1.5 22  1  1.5 22 22.484486799018757 0 0
282   p 24.44240270650108 0 0  24 0 1.25  24 1 1.25
350   pz  -10.11  $parallel grid bottom lattice
351   pz  10.11  $parallel grid top / cross grid bottom lattice

595   c/z  5.515 0.01 0.01
596   c/z 5.515 0.01 0.02
597   c/z 5.515 0.01 0.03
598   c/z 5.515 0.01 0.04
599   c/z 5.515 0.01 0.05
600   c/z 5.515 0.01 0.06
601   c/z 5.515 0.01 0.07
602  c/z 5.515 0.01 0.08
603  c/z 5.515 0.01 0.09
604  c/z 5.515 0.01 .1
605  c/z 5.515 0.01 .2
606  c/z 5.515 0.01 .3
607  c/z 5.515 0.01 .4
608  c/z 5.515 0.01 .5
609  c/z 5.515 0.01 .6
610  c/z 5.515 0.01 .7
611  c/z 5.515 0.01 .8
612  c/z 5.515 0.01 .9
613  c/z 5.515 0.01 1
614  c/z 5.515 0.01 1.5
615  c/z 5.515 0.01 2
616  c/z 5.515 0.01 2.5
617  c/z 5.515 0.01 3
618  c/z 5.515 0.01 3.5
619  c/z 5.515 0.01 4
620  c/z 5.515 0.01 4.5
621  c/z 5.515 0.01 5
622  c/z 5.515 0.01 5.5

mode p
m204 7000.04p -0.755636 $air
  8000.04p -0.231475 18000.04p -0.012889
m280 1000.04p -0.111915 $water
  8000.04p -0.888085
m281 6000.04p 1 $carbon fiber
m282 1000.04p 8 $polycarbonate
  6000.04p 5 8000.04p 2
m252 82000.04p -1 $lead, den=-11.35
m253 1000.04p -0.11 $breast tissue 50/50 den=-.985
  6000.04p -0.465 7000.04p -0.0185 8000.04p -0.4025
  11000.04p -0.001 15000.04p -0.0005 16000.04p -0.0015
  17000.04p -0.001
m254 1000.04p -0.1116 $breast tissue 30G/70A den=-.971
  6000.04p -0.5182 7000.04p -0.0139 8000.04p -0.3527
  11000.04p -0.001 15000.04p -0.0003 16000.04p -0.0013
  17000.04p -0.001
m255 1000.04p -0.1084 $breast tissue 70G/30A den=-.999
  6000.04p -0.4118 7000.04p -0.0231 8000.04p -0.4523
  11000.04p -0.001 15000.04p -0.0007 16000.04p -0.0017
  17000.04p -0.001
m256 34000.04p 1 $amorphous selenium den=-4.27
m257 42000.04p 1 $molybdenum den=-10.28
m258 45000.04p 1 $rhodium den=-12.41
m259  47000.04p          1  $silver  den=-10.49
m260  29000.04p          1  $copper den=-8.94
nps 5000000
sdef cel=41 dir=1 erg=d4 par=2 pos=0.215 0.01 66 vec=5.3 0 -66
si4   0 0.0005 0.0010 0.0015 0.0020 0.0025 0.0030 0.0035 0.0040 0.0045 &
0.0050 0.0055 0.0060 0.0065 0.0070 0.0075 0.0080 0.0085 0.0090 0.0095 &
0.0100 0.0105 0.0110 0.0115 0.0120 0.0125 0.0130 0.0135 0.0140 0.0145 &
0.0150 0.0155 0.0160 0.0165 0.0170 0.0175 0.0180 0.0185 0.0190 0.0195 &
0.0200 0.0205 0.0210 0.0215 0.0220 0.0225 0.0230 0.0235 0.0240 0.0245 &
0.0250 0.0255 0.0260 0.0265 0.0270 0.0275 0.0280 0.0285 0.0290 0.0295 &
0.0300 0.0305 0.0310 0.0315 0.0320 0.0325 0.0330 0.0335 0.0340 0.0345 &
0.0350 0.0355 0.0360 0.0365 0.0370 0.0375 0.0380 0.0385 0.0390 0.0395 0.0400
sp4   0 0 0 0 0 0 0 0 0.000732254 0.00124264 0.00268064 0.005416271 &
0.009169642 0.01401635 0.018488673 0.021967001 0.027492928 0.030782854 &
0.032194261 0.033679512 0.034913287 0.034433489 0.034443664 0.034174534 &
0.033076579 0.03243535 0.031222965 0.030154734 0.029256559 0.028242332 &
0.027201623 0.026110112 0.025043767 0.024070241 0.035398398 0.134087813 &
0.023003476 0.018837398 0.018118004 0.03596692 0.020965146 0.011338263 &
0.01064399 0.00907468 0.009121725 0.008364939 0.007488645 0.006718401 &
0.006138353 0.005244792 0.004462042 0.003688576 0.003118617 0.002316715 &
0.001204792 0.000852019 0.000271247 0 0 0 0 0 0 0 0 0 0 0 0 0 0 0 0 0 0 0 &
0 0 0 0 0 0
C1 -0.996806812 -.996778878 1 T
f06:p   9
f16:p 595 596 597 598 599 600 601 602 603 604 605 606 607 608 609 610 &
611 612 613 614 615 616 617 618 619 620 621 622
*f1:p 22
print
The Main Program

```matlab
function [] = main_SC(filename, finalfile)

tic;
% call function to read DICOM header and return kV, grid status, view,
% and appropriate files for radial extent, scatter fraction, thickness maps
% [kV, REcoeff, SFcoeff, tdata, tindex, grid, view]=imginfo(filename);

img=dicomread(filename);  % read file
[rows0,columns0]=size(img);   % get size of original file
img = double(img);
img=img(1:rows0-12,:);   % cut off border pixels which do not contain image info
img=img(13:end,:);   % cut off border pixels which do not contain image info
img=img-50;   % correct for DC offset
[rows,columns]=size(img);   % get size of new file

% call function to generate map of apparent thickness values
[imgT,avgROI] = thickmapP(img, tdata, tindex, view);
disp('Finish thickmap '); toc;

maxT = max(max(imgT)); % find maximum thickness
s=nonzeros(imgT);  % find all pixels not equal to zero thickness
modeT=mode(s);  % find mode of non-zero thicknesses

pad = 500;  % ramp between left/right and top/bottom to avoid ringing

% memory preallocations
mask = zeros(size(imgT));  % for thickness mask
imask = zeros(rows+pad, columns+pad); % for thickness mask + ramp
FimgPartial = zeros(rows+pad, columns+pad);  % for Fourier transform of the partial image
imgPartial = zeros(rows+pad, columns+pad);  % for Fourier transform of the partial image
imgSum = zeros(rows+pad, columns+pad);  % for Fourier transform of the partial image
Fkernel = zeros(rows+pad, columns+pad);  % for Fourier transform of the kernel
Fmask = zeros(rows+pad, columns+pad);  % for Fourier transform of the thickness map
imgResult = ones(size(imgT));  % for combined image
kernel = ones(rows+pad, columns+pad);  % for kernel
rampx = zeros(rows+pad, pad);  % for ramp extending # of rows
rampy = zeros(pad,columns+ pad);  % for ramp extending # of columns
noRE = false(rows+pad, columns+pad);
flag=0;

for index = 0:1:(maxT*10)  % for all thicknesses detected in the image
    t=index/10;  % convert from mm to cm
```
mask = ismember(imgT,t);  % create a mask of values of a certain thickness t

mask = mask.*img;  % replace ones in mask with actual pixel values
[a,b] = find(mask);  % find non-zero values of the mask

% Create ramp from opposing sides to reduce ringing
imask = padarray(mask,[pad pad],['replicate','post']);
[irow,icol] = size(imask);
for x=1:pad
    rampx(:,x) = (imask(:,1)-imask(:,(icol-pad))).*(x/pad) + imask(:,(icol-pad));
end
imask(:,(icol-pad):icol)=rampx;
for x=1:pad
    rampy(x,:) = (imask(1,:)-imask((irow-pad),:)).*(x/pad) + imask((irow-pad),:);
end
imask((irow-(pad-1):irow,:)=rampy;

Fmask = ((fft2(imask)));  % get Fourier transform of mask

% calculate RE and SF for each thickness, based on loaded files
RE = REcoeff(1)+REcoeff(2)*t+REcoeff(3)*kV+REcoeff(4)*t^2+REcoeff(5)*t*kV+REcoeff(6)*kV^2+REcoeff(7)*t^3+REcoeff(8)*t^2*kV+REcoeff(9)*t*kV^2;
SF = SFcoeff(1)+SFcoeff(2)*t+SFcoeff(3)*kV+SFcoeff(4)*t^2+SFcoeff(5)*t*kV+SFcoeff(6)*kV^2+SFcoeff(7)*t^3+SFcoeff(8)*t^2*kV+SFcoeff(9)*t*kV^2;

[kernel] = scatterkernel(SF,RE,irow,icol);  % call function to get kernel

if t < 0.2  % do not do deconvolution if t = 0, to reduce ringing
    noRE = ismember(kernel, 1);
    kernel = (noRE.*kernel);
end;

% Deconvolve mask with kernel to create partial primary image
Fkernel=((fft2(kernel)));  % get Fourier transform of kernel

% Use Wiener filter for deconvolution
if strcmp('HTC_IN',grid)==1  % if there is a grid
    if t<0.2  % if t < 2 mm
        K0 = 0;
        FimgPartial = Fmask.*(abs(Fkernel).^2./(abs(Fkernel).^2+K0)./Fkernel);
    else
        K = 1;
        FimgPartial = Fmask.*(abs(Fkernel).^2./(abs(Fkernel).^2+K))./Fkernel;
    end
else
    strcmp('HTC_OUT',grid)==1  % if there is not a grid
    if t<0.2  % if t < 2 mm

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K = 0;
    FimgPartial = (Fmask.*Fkernel.^2)./(Fkernel.*(Fkernel.^2+K));
  else
    K = 1;
    FimgPartial = Fmask.*(abs(Fkernel).^2./(abs(Fkernel).^2+K)./Fkernel);
  end
end

imgPartial = ifft2(FimgPartial); % take inverse Fourier transform of the partial image
    imgSum = imgSum + imgPartial; % add partial image contribution to final image
end
toc;

[height, width] = size(imgSum);

imgResult = (fftshift(imgSum));
    imgResult = imgResult(:,:,1:(icol-pad)); % cut off ramp
    imgResult = imgResult((1:(irow-pad)),:); % cut off ramp

    img_min = double(min(min(img))); % find minimum of original image
    imgResult_min = min(min(imgResult)); % find minimum of final image
    imgResult = imgResult - imgResult_min + img_min; % ensure there are no pixel values < 0

    imgResult = imgResult + 50; % add DC offset back on
    imgResult(imgResult > max(max(img))) = max(max(img)); % do not let maximum values exceed max of original image

    imgFinal = ones(rows0,columns0,'double').*16383;
    imgFinal(13:rows+12,:) = imgResult; % add back in the border rows deleted earlier

%Write final image to new DICOM file, keeping original DICOM header
    imgFinal = uint16(imgFinal);
    metadata = dicominfo(filename);
    dicomwrite(imgFinal,finalfile,metadata,'CreateMode','copy','WritePrivate',true);
clear;
end

The Function "imginfo"

function [kV, REcoeff, SFcoeff, tdata, tindex, grid, view] = 
    imginfo(filename)
    %reads the DICOM header and loads appropriate data based on information in
    %header
    info = dicominfo(filename);
kV = info.KVP;
target = info.AnodeTargetMaterial;
filter = info.FilterMaterial;
grid = info.Grid;
width = info.Width;
height = info.Height;
view = info.SeriesDescription;

if height == 4096
    paddle = 'large';
elseif height == 3328
    paddle = 'small';
else
    error('Problem with paddle size')
end

if strcmp(grid,'HTC_IN')==1
    if strcmp(target,'MOLYBDENUM')==1
        if strcmp(filter,'MOLYBDENUM')==1
            if strcmp(paddle, 'large')==1
                load('MoMoGLPthick.mat');
                tdata = MoMoGLPthick(:,kV-20);
                tindex = MoMoGLPthick(:,1);
            else
                load('MoMoGSPthick.mat');
                tdata = MoMoGSPthick(:,kV-20);
                tindex = MoMoGSPthick(:,1);
            end
        end
    load('RE_MoMoG.mat');
    load('SF_MoMoG.mat');
    REcoeff = RE_MoMoG;
    SFcoeff = SF_MoMoG;
    elseif strcmp(filter,'RHODIUM')==1
        if strcmp(paddle, 'large')==1
            load('MoRhGLPthick.mat');
            tdata = MoRhGLPthick(:,kV-20);
            tindex = MoRhGLPthick(:,1);
        else
            load('MoRhGSPthick.mat');
            tdata = MoRhGSPthick(:,kV-20);
            tindex = MoRhGSPthick(:,1);
        end
    load('RE_MoRhG.mat');
    load('SF_MoRhG.mat');
    REcoeff = RE_MoRhG;
    SFcoeff = SF_MoRhG;
    else
        error('Problem reading filter')
    end
    elseif strcmp(target,'TUNGSTEN')==1
        if strcmp(filter,'RHODIUM')==1
            if strcmp(paddle, 'large')==1
                load('WRhGLPthick.mat');
                tdata = WRhGLPthick(:,kV-20);
                tindex = WRhGLPthick(:,1);
            else

290
load('WRhGSPthick.mat');
tdata = WRhGSPthick(:,kV-20);
tindex = WRhGSPthick(:,1);
end
load('RE_WRhG.mat');
load('SF_WRhG.mat');
REcoeff = RE_WRhG;
SFcoeff = SF_WRhG;
elseif strcmp(filter,'SILVER')==1
  if strcmp(paddle,'large')==1
    load('WAgGLPthick.mat');
tdata = WAgGLPthick(:,kV-20);
tindex = WAgGLPthick(:,1);
  else
    load('WAgGSPthick.mat');
tdata = WAgGSPthick(:,kV-20);
tindex = WAgGSPthick(:,1);
  end
load('RE_WAgG.mat');
load('SF_WAgG.mat');
REcoeff = RE_WAgG;
SFcoeff = SF_WAgG;
else
  error('Problem reading filter')
end
elseif strcmp(grid,'HTC_OUT')==1
  if strcmp(target,'MOLYBDENUM')==1
    if strcmp(filter,'MOLYBDENUM')==1
      if strcmp(paddle,'large')==1
        load('MoMoNGLPthick.mat');
tdata = MoMoNGLPthick(:,kV-20);
tindex = MoMoNGLPthick(:,1);
      else
        load('MoMoNGSPthick.mat');
tdata = MoMoNGSPthick(:,kV-20);
tindex = MoMoNGSPthick(:,1);
      end
      load('RE_MoMoNG.mat');
      load('SF_MoMoNG.mat');
      REcoeff = RE_MoMoNG;
      SFcoeff = SF_MoMoNG;
    elseif strcmp(filter,'RHODIUM')==1
      if strcmp(paddle,'large')==1
        load('MoRhNGLPthick.mat');
tdata = MoRhNGLPthick(:,kV-20);
tindex = MoRhNGLPthick(:,1);
      else
        load('MoRhNGSPthick.mat');
tdata = MoRhNGSPthick(:,kV-20);
tindex = MoRhNGSPthick(:,1);
      end
      load('RE_MoRhNG.mat');
      load('SF_MoRhNG.mat');
      REcoeff = RE_MoRhNG;
    else
      error('Problem reading target')
    end
  else
    error('Problem reading filter')
  end
else
  disp('ok 2')
end
SFcoeff = SF_MoRhNG;
else
    error('Problem reading filter')
end
elseif strcmp(target,'TUNGSTEN')==1
    if strcmp(filter,'RHODIUM')==1
        if strcmp(paddle, 'large')==1
            load('WRhNGLPthick.mat');
            tdata = WRhNGLPthick(:,kV-20);
            tindex = WRhNGLPthick(:,1);
        else
            load('WRhNGSPthick.mat');
            tdata = WRhNGSPthick(:,kV-20);
            tindex = WRhNGSPthick(:,1);
        end
    end
    load('RE_WRhNG.mat');
    load('SF_WRhNG.mat');
    REcoeff = RE_WRhNG;
    SFcoeff = SF_WRhNG;
elseif strcmp(filter,'SILVER')==1
    if strcmp(paddle, 'large')==1
        load('WAgNGLPthick.mat');
        tdata = WAgNGLPthick(:,kV-20);
        tindex = WAgNGLPthick(:,1);
    else
        load('WAgNGSPthick.mat');
        tdata = WAgNGSPthick(:,kV-20);
        tindex = WAgNGSPthick(:,1);
    end
    load('RE_WAgNG.mat');
    load('SF_WAgNG.mat');
    REcoeff = RE_WAgNG;
    SFcoeff = SF_WAgNG;
else
    error('Problem reading filter')
end
else error('Problem reading target')
end
else
    error('Problem reading grid status')
end

% tdata format: 1st column gives thickness in cm. 2nd column is 22kV, 3rd
% column is 23 kV, etc, up to 38 kV

% The Function "thickmap"

function [imgT,avgROI] = thickmap(img, tdata, tindex, view)

% Find ROI for normalization
[row,col] = size(img); % find number of rows and columns in image
if strcmp(view,'R CC')==1 % for right CC views
    normX = 500; % set ROI at top left corner, 500 pixels in
    from the edge
    normY = 500;
elseif strcmp(view,'R MLO')==1 % for right MLO views
    normX = 500; % set ROI at top left corner, 500 pixels in
    from the edge
    normY = 500;
elseif strcmp(view,'L CC')==1 % for left CC views
    normX = col-500; % set ROI at top right corner, 500
    pixels in from the edge
    normY = 500;
elseif strcmp(view,'L MLO')==1 % for left MLO views
    normX = col-500; % set ROI at top right corner, 500
    pixels in from the edge
    normY = 500;
else
    error('Problem with ROI')
end

normROI = img(normY-15:normY+15,normX-15:normX+15); % ROI is 31x31 pixels
avgROI = mean(mean(normROI)); % find average pixel
value in ROI
if avgROI <= 0, error('Problem with ROI'), end;

[thickest,a] = size(tdata); % find the largest thickness in the data set (10 cm)

imgN = -log(img/avgROI); % calculate log signal of each value
imgT = zeros(size(imgN)); % allocate memory for thickness map

parfor y = 1:col % for all pixels in the image
    temp=zeros(row,1);
    for x = 1:row
        if(imgN(x,y)>tdata(thickest)) % if value is greatest than largest
            in the thickness data set, assign maximum thickness
            temp(x,1) = tindex(thickest);
        else % otherwise, assign thickness 1 mm
            i=0;
            for z=1:size(tdata)
                if tdata(z)>imgN(x,y)
                    i=z
                    break;
                end
            end
            if i == 1 % if value corresponding to 0
                thickness is higher, assign 0 thickness
                temp(x,1) = 0;
            elseif i>1 % otherwise, assign thickness 1 mm
                below that triggered as higher
                temp(x,1) = tindex(i-1);
            else
...
The Function "scatterkernel"

function [kernel] = scatterkernel(SF,k,rows,columns,grid)
% This function returns the scatter kernel given SF (scatter fraction)
% and k (radial extent).
quad1 = zeros(1001,1001,'double');        % create lower right quadrant
for i = 1:1001
    for j = 1:1001
        x = i-1;
        y = j-1;
        r = sqrt((.07*x)^2+(.07*y)^2); % find distance in mm, 1 pixel = 0.07
        mm
        if r == 0
            quad1(i,j) = 1;            % "delta" function at zero distance
        else
            quad1(i,j) = SF/((1-SF)*2*3.1415*k*r) * exp(-r/k); % use
            scatter kernel formula
            if quad1(i,j)<.005, quad1(i,j)=0; end; % if value is too
            low, use 0
        end;
    end;
end;
quad2 = flipud(quad1);                  % create upper right quadrant
quad4 = fliplr(quad1);                  % create lower left quadrant
quad3 = fliplr(quad2);                  % create upper left quadrant
upper = [quad3 quad2];                  % add together upper quadrants
lower = [quad4 quad1];                  % add together lower quadrants
smallkernel = [upper;lower];             % add together upper and lower
% halves
smallkernel(1001,:) = [ ];                % delete doubled data at "seam"
smallkernel(:,1001) = [ ];
kernel=zeros(rows,columns);  % final kernel size is equal to image size
% place PSF at center of final kernel
kernel((rows/2 - 999):(rows/2 + 1001), (columns/2 - 999):(columns/2 + 1001))
= smallkernel(:,1:);
[height, width] = size(kernel);
function [] = NPS(filename, finalfile)

img = dicomread(filename);  %Read in file
img=img-50;  % Correct for DC offset
Pos = zeros(320,320,9);

%Get large ROI's at 9 positions on the detector
Pos(1:320,1:320,1) = img(3705:4024,2938:3257);
Pos(1:320,1:320,2) = img(1888:2207,2938:3257);
Pos(1:320,1:320,3) = img(72:391,2938:3257);
Pos(1:320,1:320,4) = img(3705:4024,1885:2204);
Pos(1:320,1:320,5) = img(1888:2207,1885:2204);
Pos(1:320,1:320,6) = img(72:391,1885:2204);
Pos(1:320,1:320,7) = img(3705:4024,832:1151);
Pos(1:320,1:320,8) = img(1888:2207,832:1151);
Pos(1:320,1:320,9) = img(72:391,832:1151);

% Memory preallocations
temp = zeros(64,64);
PosAvg = zeros(64,64,9);
Stot = zeros(64,64,9);
Savg = zeros(64,64,9);
Sstoch = zeros(64,64,9);
Stot1D = zeros(32,1,9);
Savg1D = zeros(32,1,9);
Sstoch1D = zeros(32,1,9);

%calculate Stot (avg of many spectra for each position) and Savg (spectrum
%of average ROI's at each position)
for z = 1:9
   for i=1:64:320
      for j = 1:64:320
         temp = Pos(i:i+63,j:j+63,z);
         Stot(:,:,z) = Stot(:,:,z) + (abs(fftshift(fft2(temp))).^2);
         PosAvg(:,:,z) = PosAvg(:,:,z) + (temp);
      end
   end
   PosAvg(:,:,z) = PosAvg(:,:,z)/25;
   Pixelavg = mean(mean(PosAvg(:,:,z)));
   Stot(:,:,z) = ((Stot(:,:,z))/25*(0.07^2/64^2))/Pixelavg^2;  %normalize
   Stot
   Savg(:,:,z) = ((abs(fftshift(fft2(PosAvg(:,:,z))))).^2*(0.07^2/64^2))/Pixelavg^2;  %find
   Savg
   Sstoch(:,:,z) = Stot(:,:,z) - Savg(:,:,z);  %find Sstoch
   Sstoch
   %Find 1D spectra by averaging 8 slices around the central axis
   Stot1D(:,1,z) =
   (Stot(33:64,27,z)+Stot(33:64,28,z)+Stot(33:64,29,z)+Stot(33:64,30,z)+Stot(33:
   64,36,z)+Stot(33:64,37,z)+Stot(33:64,38,z)+Stot(33:64,39,z))/8;
Savg1D(:,1,z) =
(Savg(33:64,27,z)+Savg(33:64,28,z)+Savg(33:64,29,z)+Savg(33:64,30,z)+Savg(33:
64,36,z)+Savg(33:64,37,z)+Savg(33:64,38,z)+Savg(33:64,39,z))/8;
Sstoch1D(:,1,z) =
(Sstoch(33:64,27,z)+Sstoch(33:64,28,z)+Sstoch(33:64,29,z)+Sstoch(33:64,30,z)+
Sstoch(33:64,36,z)+Sstoch(33:64,37,z)+Sstoch(33:64,38,z)+Sstoch(33:64,39,z))/
8;
end

%Save 1D spectra
save(finalfile, 'Stot1D','Savg1D','Sstoch1D');
end
In an attempt to explain the "hump" artifact seen in the disk images, the disk was modeled in one dimension as a rect function (Figure F-1). The difference (i.e., the contrast) between the top (background) and bottom (disk) of the rect function was equal to 1. The Fourier transform of the rect function was acquired, and was divided by the Fourier transform of the point spread function (PSF) used in this study. Values for the scatter fraction (SF) and mean radial extent (MRE) representative of those found without a grid were chosen to describe this PSF. Thus, the PSF was deconvolved from the rect function. The resulting image, after the inverse Fourier transform was performed, demonstrates a hump very similar to that seen in the disk images (Figure F-2). There is also a loss of contrast: the difference between the background level and the center of the hump is now 0.9068. This indicates that the hump shape is not simply the result of overshoot at the edges, but represents a true decrease in contrast.

Using a rect function to represent the disk is a simplification of the situation, because in reality, the image of the disk would first be convolved with a PSF representing the scatter and focal spot effects. If this PSF matched the PSF used for deconvolution perfectly, there would be no artifact: the scatter contribution would simply be removed. Therefore, it is likely that the cause of the hump is a mismatch between the true scatter PSF and the modeled PSF. Some degree of mismatch is to be expected, as no model is perfect, and it is already known from Chapter 5 that the Monte Carlo data does not fit the PSF equation perfectly. To model this situation, the rect function was
first convolved with a Lorentzian function (Equation F-1), before being deconvolved with the scatter PSF.

\[ L(x, \gamma) = \frac{\gamma}{x^2 + \gamma^2} \] (F-1)

The Lorentzian was chosen because it has a form very similar, but not exactly identical, to the scatter PSF. The scale parameter \( \gamma \) was chosen to produce a function with a spread similar to that of the scatter PSF. In the case of \( \gamma = 0.05 \), the Lorentzian function displays slightly less spread than the scatter PSF (Figure F-3). The resulting image, after convolution with the Lorentzian and deconvolution with the scatter PSF, shows a similar hump (Figure F-4). The contrast in this case is 0.9482. Altering the Lorentzian to use \( \gamma = 0.1 \), which matches the scatter PSF even more closely, results in a contrast of 0.9899. Thus, the contrast loss is dependent on the match between the Lorentzian and the scatter PSF.

Choosing the scale parameter \( \gamma \) to be equal to 1 creates a spread greater than the scatter PSF (Figure F-5). The result is a blurring of the edges of the rect function and increase rather than loss of contrast (contrast = 1.0385). No hump is observed (Figure F-6).

It is likely that the hump is caused by a mismatch between the true scatter PSF and the modeled scatter PSF, and that the spread of the true PSF is slightly less than that modeled by the scatter PSF. This discrepancy is not great enough to create an observable artifact in clinical images, but with the artificial sharp edges created by the disk, the artifact becomes apparent.
Figure F-1. Rect function used to simulate the disk in one dimension.

Figure F-2. The result of deconvolution of the rect function and scatter PSF.
Figure F-3. The Lorentzian function with $\gamma = 0.05$ and the scatter PSF.

Figure F-4. Result after convolution of the rect function with the Lorentzian function shown in Figure F-3, followed by subsequent deconvolution with the scatter PSF.
Figure F-5. The Lorentzian function with $\gamma = 1$ and the scatter PSF.

Figure F-6. Result after convolution of the rect function with the Lorentzian function shown in Figure F-5, followed by subsequent deconvolution with the scatter PSF.
### APPENDIX G

**CHARACTERISTICS OF IMAGES USED IN RADIOLOGIST PREFERENCE STUDY**

Table G-1. Characteristics of the images used in the radiologists preference study.

<table>
<thead>
<tr>
<th>Image</th>
<th>Thickness (mm)</th>
<th>kVp</th>
<th>Filter</th>
<th>Laterality</th>
<th>View</th>
<th>FOV</th>
<th>Density*</th>
<th>Calc**</th>
<th>Mass**</th>
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* Density according to BI-RADS® criteria, where category 1 is less than 25% glandular tissue, category 2 is between 25% and 50% glandular tissue, category 3 is between 50% and 75% glandular tissue, and category 4 is more than 75% glandular tissue.

** Calcifications and masses were assumed to be present if at least 2 out of the 3 radiologists indicated they were visible.
LIST OF REFERENCES


BIOGRAPHICAL SKETCH

Stephanie Leon was born in 1982 and grew up in South Florida. She attended Boyd H. Anderson High School, where she received the International Baccalaureate diploma and graduated valedictorian in 2000. After attending Boston University her freshman year, she transferred back home to the University of Florida with the intention of obtaining a degree in psychology and going to medical school.

She discovered medical physics while volunteering in the Interventional Radiology department of Shands Hospital in 2003, and she realized her calling when she found she was more interested in why the x-ray machine was using 80,000 volts than in what the radiologist was doing during the procedure. She decided that a career in medical physics was the perfect way to combine her interest in science with her desire to help others. She was fortunate to receive an internship in medical physics with Dr. Manuel Arreola, Director of Medical Physics for the UF Department of Radiology, starting in December 2003. She received an AAPM Undergraduate Summer Fellowship in 2004, which allowed her to spend a summer at M.D. Anderson Cancer Center in Houston, TX. She graduated in April 2005 with Bachelor of Science degrees in physics and psychology.

She entered graduate school in 2005 and worked for one and a half years as a Graduate Assistant in medical physics. In January 2007, she started employment as a diagnostic medical physicist in the UF Department of Environmental Health and Safety, while simultaneously finishing her graduate work. She worked with Dr. Libby Brateman on her master's project (*Issues Encountered in the Conversion from Film-Screen to Digital Mammography at a Free-Standing Imaging Center*) and received her Master of
Science degree in August 2007. She completed the qualifying examination for her doctoral research in November 2007.

In late 2009, she received a job offer from the University of Texas Medical School at Houston, and she started employment there as a radiological physicist in the Department of Diagnostic and Interventional Imaging. Soon after starting her new job, her supervisor, Dr. Louis Wagner, encouraged her to pursue her Ph.D. She was re-admitted to the University of Florida in May 2010 and conducted her research long-distance with Dr. Brateman as her committee chair and Dr. Wagner as her local co-chair. She will be receiving her Ph.D. in December 2012.

She became certified in Radiation Protection by the American Board of Science in Nuclear Medicine in 2010. She received certification in Diagnostic Medical Physics by the American Board of Radiology in 2011. She is licensed by the State of Texas to practice Diagnostic Medical Physics and Medical Health Physics.

She is married to her high school sweetheart, Juan Carlos Leon, and currently lives in Pearland, TX.