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INVESTIGATION AND IMAGING CHARACTERISTICS OF A CMOS
SENSOR BASED DIGITAL DETECTOR COUPLED TO A RED EMITTING

FLUORESCENT SCREEN

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The dominant powder scintillator in most medical imaging modalities for decades is Gd$_2$O$_2$S:Tb due to the very good intrinsic properties and overall efficiency. Except for Gd$_2$O$_2$S:Tb there are alternative powder phosphor scintillators like Lu$_2$SiO$_5$:Ce and Gd$_2$O$_2$S:Eu that has been suggested for use in various medical imaging modalities. Gd$_2$O$_2$S:Eu emits red light and can be combined mainly with digital imaging devices like CCDs and CMOS based detectors. The purposes of the present thesis, is to investigate the fundamental imaging performance of a high resolution CMOS based imaging sensor combined with custom made Europium (Eu$^{3+}$) activated Gd$_2$O$_2$S screens in terms of Modulation Transfer Function (MTF), Normalized Noise Power Spectrum (NNPS), Detective Quantum Efficiency (DQE), Noise Equivalent Quanta (NEQ) and Information Capacity (IC) covering the mammography and general radiography energy ranges.

The CMOS sensor was coupled to two Gd$_2$O$_2$S:Eu scintillator screens with coating thicknesses of 33.3 and 65.1 mg/cm$^2$, respectively, which were placed in direct contact with the photodiode array. The CMOS photodiode array, featuring 1200x1600 pixels with a pixel pitch of 22.5 $\mu$m, was used as an optical photon detector. In addition to frequency dependent parameters (MTF, NPS, DQE) characterizing image quality, image information content was assessed through the application of information capacity (IC). The MTF was measured using the slanted-edge method to avoid aliasing while the Normalized NPS (NNPS) was determined by two-dimensional (2D) Fourier transforming of uniformly exposed images. Both parameters were assessed by irradiation under the RQA-5 protocol (70kVp digital-radiography) recommended by the International Electrotechnical Commission Reports.
62220-1 and the W/Rh, W/Ag beam qualities (28kVp digital-mammography). The DQE was assessed from the measured MTF, NNPS and the direct entrance surface air-Kerma (ESAK) obtained from X-ray spectra measurement with a portable cadmium telluride (CdTe) detector.

The spectral matching factor between the optical spectra emitted by the Gd$_2$O$_2$:Eu and the Gd$_2$O$_2$:Tb screens and the CMOS optical sensor, evaluated in the present study, was 1 and 0.95 respectively. The ESAK values ranged between 11.2-87.5 $\mu$Gy, for RQA-5, and between 65.8-334 $\mu$Gy, for W/Rh, W/Ag beam qualities. It was found that the detector response function was linear for the exposure ranges under investigation. Under radiographic conditions the MTF of the present system was found higher than previously published MTF data for a 48 $\mu$m CMOS sensor, in the low up to medium frequency ranges. DQE was found comparable, while the NNPS appeared to be higher in the frequency range under investigation (0–10 cycles/mm). NEQ reached a maximum (73563 mm$^{-2}$) in the low frequency range (1.8 cycles/mm), under the RQA 5 (ESAK: 11.2 $\mu$Gy) conditions. IC values were found to range between 1730-1851 bits/mm$^2$. Under mammographic conditions MTF, NNPS and NEQ were found comparable to data previously published for the 48 $\mu$m CMOS sensor while the DQE was found lower. The corresponding IC values were found ranging between 2475 and 2821 bits/mm$^2$.

The imaging performance of europium (Eu$^{3+}$) activated Gd$_2$O$_2$S screens in combination to the CMOS sensor, investigated in the present study, was found comparable to those of Terbium (Tb) activated Gd$_2$O$_2$S screens (combined with the CMOS sensor). It can be thus claimed that red emitting phosphors could be suitably used in digital imaging systems, where the Silicon (Si) based photodetectors are more sensitive to longer wavelength ranges, and particularly in the red wavelength range.
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INTRODUCTION

A.1 The problem

It was recognized early on that the digitization of medical information would advance the efficiency of diagnostic technology (Suzuki et al 2000, Liang et al 2008). However, the digitization of image data, is dependent on advances in technologies such as input (imaging conditions, powder scintillator/ digital sensor optimization), processing, transmission, storage and display (Bruckmann and Uhl 2000, Suzuki et al 2000, Efstathopoulos et al 2001, Koscis et al 2003). Insufficient advances in such technologies have effectively limited the digitization of image data for medical use (Suzuki et al 2000).

However the quest for the ideal scintillator compromising optimum imaging performance with high efficiency and low cost is still in the front for various research groups and the results are published in numerous works.

The development of new digital diagnostic imaging techniques and modalities, such as digital breast tomosynthesis (DBT) (Dobbins and Godfrey 2003), dual energy (DE) imaging (Alvarez et al 2004), scanning slot digital mammography detectors (Mainprize et al 2002), energy selective techniques (Patatoukas et al 2006, Alvarez 2010) and digital fluoroscopy systems (Antonuk et al 2009), require efficient scintillators to reduce patient dose. Digital imaging systems with smaller sizes for the detector elements (pixels or dels) are also required to improve image resolution and the visibility of microcalcifications, without noise increase and DQE degradation. Furthermore, in many digital imaging systems based on crystalline Silicon (Si) optical detectors such as CCDs and CMOS sensors (Nikl 2006, Chotas et al 1999, Kim et al 2001, Bigas et al 2006, Kim et al 2005) the green light emitted by Terbium-activated phosphors is not very efficiently detected (Gurwich 1995). This is because a large number of Si based devices, incorporated in X-ray imaging systems are not adequately sensitive to these wavelengths (500-550 nm) (Gurwich 1995). Since most Si based photodetectors are more sensitive to longer wavelength ranges, and particularly in the red wavelength range, the properties of red emitting phosphors (Rowlands and Yorkston 2000, Rodnyi et al 2001, Zych 2006, Nikl 2006) and in particular, of the Europium (Eu$^{3+}$) activated Gd$_2$O$_2$S has been investigated in terms of optical output (Zych 2006), emission efficiency (Michail et al 2008, Michail et al 2010, Michail et al 2011, Wiatrowska et al 2010) and imaging performance (Gurwich 1995, Michail et al 2008, Michail et al 2010, Michail et al 2011). Such phosphors are cost effective and can be easily prepared by inserting Europium ion activator (Eu$^{3+}$) in
rare earth based host matrices. In addition Europium-activated phosphors have a decay time of the order of one millisecond (slightly higher than Gd\textsubscript{2}O\textsubscript{2}:Tb) which is acceptable for applications that do not involve high framing rates (Lempicki et al. 2002, Nagarkar et al. 2003). These include digital radiography, and mammography. Gd\textsubscript{2}O\textsubscript{2}:Eu has been previously employed in single pulse dual energy radiography (Boone et al. 1992), in digital mammography and diffraction enhanced breast imaging with CCD arrays (Gambaccini et al. 1996, Taibi et al. 1997, Gambaccini et al. 1998, Harris et al. 2003). Furthermore Gd\textsubscript{2}O\textsubscript{2}:Eu has been previously used as phosphor insert in phantoms for limited-angle x-ray luminescence tomography (XLT) (Carpenter et al. 2011) and hybrid x-ray luminescence/optical imaging (Carpenter et al. 2010). Previously, the image quality of a detector based on a commercial Gd\textsubscript{2}O\textsubscript{2}:Tb screen coupled to an active CMOS sensor has been investigated by Michail and found comparable with CCD and passive CMOS sensors (Michail et al 2011). In the present study this CMOS sensor was coupled to two different custom made Gd\textsubscript{2}O\textsubscript{2}:Eu phosphor screens, prepared in our laboratory, and the whole detector was evaluated under general radiography and mammography conditions.

The investigation of the imaging performance of this detector (i.e. CMOS coupled to Gd\textsubscript{2}O\textsubscript{2}:Eu phosphor screens) was achieved by experimental assessment of the Signal Transfer Property (STP), the Modulation Transfer Function (MTF), the Normalized Noise Power Spectrum (NNPS), the Detective Quantum Efficiency (DQE), the Noise Equivalent Quanta (NEQ) and the Information Capacity (IC). The experimental method was based on the guidelines published by the International Electrotechnical Commission (IEC) (Medical Electrical Equipment-Characteristics of Digital X-Ray Imaging Devices 2005), which has standardized the methodology for measuring DQE in digital detectors. In addition the results of the present work have
been compared with the results of previously studied CMOS sensors coupled to Gd$_2$O$_2$S:Tb phosphor screens (Michail et al 2011, Cho et al 2008).
CHAPTER B

DIGITAL RADIOGRAPHY

B.1. Introduction

Digital radiography detector systems were first implemented for medical applications in the mid-1980s, but the promise of digital imaging was not realized until the early 1990s, in conjunction with the establishment of first generation picture archiving and communications systems (PACS). At the time, film screen detectors have nearly replaced by computed radiography systems (CR), charge coupled device (CCD) and Active-matrix flat-panel imager (AMFPI) systems. These technologies for digital image acquisition appeared in the mid-1990s with the use of large field of view (FOV) X-ray phosphors and optical lens assemblies to focus the X-ray induced light output onto a small-area charge coupled device (CCD) photodetector array, as well as rectangular CCD arrays used with slot-scan geometries. Active-matrix flat-panel imager (AMFPI) systems appeared in the late 1990s and early 2000s, employing either an X-ray-to-light converter with photodiode array, or a semiconductor material to directly convert incident X-rays into signals. Both CCD and flat-panel based detectors use an "active" readout of the image following acquisition to present the image immediately without further interaction by the technologist. Other technologies such as complementary metal-oxide semiconductor (CMOS) detectors were introduced in the same time frame as AMFPI’s, and so far are making great progress in medical imaging, making it competitive with the other digital imaging detectors. Beyond the digital detector characteristics are considerations for software for pre- and post-processing of the digital image data, the user and modality interfaces, display monitors and calibrations. Many unique acquisition capabilities, such as dual-
energy image tissue decomposition and limited angle digital tomosynthesis, are important when considering future applications of specific importance (Seibert 2009).

Luminescent materials, either in the form of powder scintillators (phosphor screens) or in the form of ceramics or single-crystals, are incorporated in many medical imaging radiation detectors. Powder scintillators have been successfully employed in a large number of X-ray radiography devices (from conventional and digital X-ray radiography, mammography and X-ray computed tomography to digital dental radiography and portal imaging) (Arnold 1979, Yaffe and Rowlands 1997, Boone 2000, Van Eijk 2002, Nikl 2006). Among various materials, rare-earth-ion doped Oxyorthosilicates provide excellent scintillating properties and thus they have been extensively investigated for the development of new scintillators (Lee et al 2006). Terbium (Tb)-activated phosphors (i.e. Gd$_2$O$_2$S:Tb, La$_2$O$_2$S:Tb and Y$_2$O$_2$S:Tb) have been up to now accepted to be of the most efficient X-ray-to-light converters (Arnold 1979, Gurwich1995, Kandarakis et al 1996, Liaparinos et al 2007) employed in mammography and radiography. Currently the most widely used phosphors are Gd$_2$O$_2$S:Tb and CsI:Tl.

Gd$_2$O$_2$S:Tb has been proven very useful in conventional radiography screen-film systems, where precise matching the spectral sensitivity of the X-ray film to the emission of the phosphor is of primary consideration in order to obtain the highest speed for the screen-film combination. However, in the past decade an increasing tendency to introduce digital radiography and mammography systems has been shown in various publications (Rodnyi et al 2001, Van Den Bergh and Leblans 2005).

In some digital imaging systems, based on crystalline Silicon (Si) optical detectors (CCDs, photodiodes, novel CMOS sensors), the green light emitted by Terbium-activated phosphors phosphors (as well as by Thalium activated CsI) is not very
efficiently detected (Gurwich 1995). This is because a large number of Si based
devices, incorporated in X-ray imaging systems are not adequately sensitive to these
wavelengths (500-550 nm); only 45-55 % of the light produced by Gd₂O₂S:Tb or
Y₂O₂S:Tb is registered by the Si photodiode (Gurwich 1995).

Since most Si and CMOS based photodetectors are more sensitive to longer
wavelengths, and particularly in the red wavelength range, it would be of interest to
investigate the emission efficiency of red emitting phosphors (Rodnyi et al 2001, Van
Den Bergh and Leblans 2005). For this purpose, Europium (Eu)-activated phosphors,
emitting at wavelengths towards the red region of the light spectrum, could be used
instead of green emitting Tb-activated ones (Gambaccini et al 1996). Furthermore the
performance of many Europium doped scintillators, and particularly Gd₂O₂S:Eu, have
been previously found comparable to Terbium-activated phosphors (both showing
190% of the CdWO₄’s optical output). Gd₂O₂S:Eu has high light yield (60000
photons/MeV) and luminescence efficiency, high density (7.3 g/cm³), high radiation
detection index (ρZₜeff =10³×10⁶) and finally decay time of the order of 1 ms (slightly
higher than Gd₂O₂S:Tb), which is acceptable for most X-ray radiography applications
that do not involve high framing rates (Lempicki et al 2002, Okumura et al 2002,
Nagarkar et al 2003, Michail et al 2008). These include stationary digital and
conventional general radiography and mammography. Gd₂O₂S:Eu has been
previously employed in single pulse dual energy radiography (Boone et al 1992), in
digital mammography and diffraction enhanced breast imaging with CCD arrays
B.2. Advantage of Digital Radiography

An advantage of DR over film/screen (F/S) radiography is the increased dynamic range and linear response of DR images (Artz 1997). According to Vuylsteke & Schoeters (1994), F/S has an exposure range in the order of 100:1. In contrast, DR has a dynamic range of nearly 1000:1 (Vuylsteke and Schoeters, 1994). These dynamic range advantages also exist for direct and indirect digital radiography.

A further advantage of DR over F/S is that it provides the facility to perform post-processing functions or manipulate the image after it has been acquired. Contrast magnification and split screen viewing can be manipulated on the monitor screen to aid in diagnosis. It has been noted that the value of contrast enhancement is actually task-specific (Dunn and Kantor 1993).

B.3. Disadvantage of Digital Radiography

Several disadvantages of DR have been reported. The spatial resolution of an imaging system is its ability to accurately represent small objects. Spatial frequencies in F/S general radiography are limited by factors external to the image recording system. Some external factors are the finite size of the focal spot where x-rays are produced and the geometric relationship between the focal spot, the object being imaged and the image recording system (Bushberg et al 2002, Bushong 2001, Curry et al 1990). A digital image’s spatial resolution is determined by the number of rows and columns of pixels in the image (Baxes, 1994). The spatial frequencies of a DR image are also limited by the external factors discussed above, as well as the workstation capabilities used for diagnosis. The spatial resolution of a DR image can be measured in pixels per millimetre or in line pairs per millimetre (Bushberg et al 2002, Bushong 2001). The spatial resolution of DR image recording systems is lower than that of F/S image...
recording systems (Schaetzing et al 1990, Artz, 1997, Bushberg et al 2002, Bushong 2001, Weiser 1997). This lower spatial resolution impacts on the overall spatial resolution of radiographic examinations. Specifically, the sampling frequency of the DR recording system is the limiting factor in determining the spatial resolution of the DR system (Bushberg et al 2002, Bushong 2001).

Direct radiographic image noise results from either internal system noise or from scattered radiation. Inherent system noise is greater in DR than in F/S and is reduced through software algorithms prior to the display of the image. CR and other DR systems are more sensitive to scattered radiation than F/S. The amount of scattered radiation is a function of the energy of the X-ray beam and the anatomy which attenuates the x-ray beam. Scatter can be reduced prior to reaching the image receptor in both F/S and DR systems. The effect of scattered radiation in DR can be further reduced through the use of scatter correction algorithms (Bushberg et al 2002, Bushong 2001).

Artifacts are a potential problem that can reduce the quality of images. DR has artifacts that are not found in F/S systems. Artifacts in CR fall into three main categories (Cesar et al 2001):

- Image plate artifacts
- Plate reading artifacts
- Image processing artifacts

B.4. Digital Radiography and X-Ray Dose

Digital radiography has been the subject of many studies evaluating the radiation doses received by patients as compared to doses received in F/S radiographic
examinations. The consensus of the findings is that the dose received by the patient in CR is similar to that of F/S radiographic examinations. In mammography especially the imaging capabilities and system setups of DR are generally equivalent with the corresponding CR [attached]. In paediatric radiographic examinations, doses received by the child were lower in F/S than in CR imaging examinations according to Nickoloff (2002).

A general consensus of recent articles is that the wide latitude of CR compared to F/S has not provided the expected dose reductions for CR. It has been suggested that quantum noise from low exposures used in CR systems limits its dose reduction capabilities. However, dose advantages are gained in direct DR imaging examinations over those of CR and F/S. The reduced dose of direct DR is attributable to the lower noise levels of direct DR over CR and the wider latitude of direct DR compared to F/S (Chotas et al 1993, Dobbins et al 2002, Marshall 2001, Marshall et al 1994).

B.5. Photostimulable Storage Phosphor Detectors (PSP)

More commonly recognized as CR, the PSP detector "system" is comprised of 2 main components. The detector is usually a cassette based storage phosphor that absorbs X-ray energy transmitted through the patient and temporarily stores the X-ray latent image as a 2-dimensional array of electrons trapped. The imaging plate "reader" is comprised of a scanning laser beam to stimulate the trapped electrons and produce "photostimulated luminescence" of a different wavelength that is optically separable from the stimulation wavelength. The reader also includes a light guide and photomultiplier assembly that extracts and processes the stored X-ray content to a sampling resolution on the order of 100 microns, digital electronics to create the
corresponding digital image, and an erasure stage to eliminate any residual signal and prepare for the next exposure. From the reader, all images proceed to a quality control workstation for image evaluation, annotation and transfer to PACS. Most often, the storage phosphor is layered on a flexible or solid substrate in a cassette enclosure, which allows for the ability to directly replace a screen-film cassette in a conventional radiography room. Thus there is the flexibility and portability of a cassette with digital radiography acquisition capability using existing X-ray equipment; this is CR's greatest asset. CR cassettes of various size and number, together with a high-speed imaging plate reader can service multiple X-ray rooms, resulting in a relatively low initial acquisition cost. However, the technologist must handle the cassettes and process the imaging plates in a manner similar to film, which can reduce patient throughput in a busy clinical room and increase labor costs, as the time to handle the exposed imaging plate to the reader and output of the X-ray image can often take several minutes (about 45 to 60 seconds to "read" the plate with the moving laser beam). Other expenses to consider are the need for high-frequency (e.g., 170 lines per inch) antiscatter grids for stationary (non-Bucky) applications such as portable bedside imaging and fixed grid cassette holders. Long-term costs include cassette and imaging plate longevity, maintenance and cleaning of the imaging plates and reader assembly, replacement of damaged detectors, and continuous oversight with a quality control program (Seibert et al 2006)
B.6. Charge-Coupled Device Detectors (CCD)

The design of a charge-coupled device (CCD) based DR system is straightforward. The detector is comprised of a large FOV (e.g., 43 cm by 43 cm) scintillator that converts absorbed X-ray energy into light. It also includes an optical lens assembly to focus the light onto the photosensitive CCD array, and a CCD camera to integrate, scan and output the corresponding light image. While there were initially several configurations in early systems, today's CCD based detector is typically comprised of a single compound optical lens and a high resolution CCD camera comprised of 9 million pixels (3000×3000 pixels) to 16 million pixels (4000×4000 pixels). When referred back to the image plane, this results in image pixel sizes of ~0.10 to ~0.14 μm. The photosensitive area of the CCD chip is actually quite small, on the order of 2.5 cm × 2.5 cm to 4.0 cm × 4.0 cm, which is required to maintain extremely high charge coupling efficiency and low noise operation during the readout of the image. Thus, there is a large optical demagnification that is necessary to focus the full FOV light image onto the CCD sensor. One physical difficulty is the inefficiency of light collection caused by the dispersed light emission from the phosphor, resulting in only a small fraction that can be focused onto the CCD, thus potentially reducing the
statistical integrity of information carried by the X-ray photons and increasing overall noise in the image. This is determined by the demagnification factor, conversion efficiency, luminance and directionality of the light emission. A non-structured phosphor such as gadolinium oxysulfide has a high light dispersion and corresponding low fraction of light that can be focused on the CCD, while a structured phosphor such as cesium iodide (CsI) produces a more forward-directed light output, so that the lens-light collection efficiency, and thus the SNR in the output image, is better for a given incident X-ray exposure. Newer, advanced CCD systems with a CsI phosphor have proven to be reasonably efficient, particularly when using higher kilovolt peak (kVp) techniques that produce more light photons per absorbed X-ray photon. One minor disadvantage in some positioning situations is the relatively large and bulky enclosure of a CCD based DR system, necessitated by placing the CCD out of the direct X-ray beam and using mirror optics to reflect the light to the photosensor array (Seibert 2009).

Linear CCD arrays optically coupled to a scintillator by fiberoptic channel plates (often with a demagnification taper of 2:1 to 3:1) are used in slot-scan geometries. A significant advantage is pre- and post-patient collimation that limits X-ray scatter and allows grid free operation with equivalent image quality (in terms of SNR) of a large area FOV at 2 to 4 times less patient dose. Disadvantages include the extended exposure time required for image acquisition with potential motion artifacts and reduced X-ray tube efficiency. Nevertheless, imaging systems based on slot-scan acquisition have provided excellent clinical results for dedicated chest and full-body trauma imaging (Seibert 2009).
B.7. Complementary Metal-Oxide Semiconductors Detectors (CMOS)

CMOS light-sensitive arrays are based upon a crystalline silicon matrix and are essentially random access memory "chips" with built-in photodiodes, storage capacitors, and active readout electronics, operating at low voltage (3 to 5 volts) for image acquisition and readout. The ability to randomly address any detector element on the chip enables opportunities for automatic exposure control (AEC) capabilities that are not easily performed with a CCD photo detector. However, electronic noise has been a problem that has slowed the introduction of this technology (Seibert 2009).

In recent years, interest in CMOS sensors has increased rapidly, driven largely by consumer electronics and the demand for sensors with lower cost, lower power consumption and miniaturization (Fossum 1997, Lo 1998). Cost effectiveness is considered by some to be the main advantage of CMOS detectors due to their manufacturability (Zhang et al 2005, Elbakri et al 2009).

Baysal and Toker (2005) reported on the development of a CMOS mammography cassette for specimen and biopsy imaging with MTF, DQE and ACR phantom scores comparable to industry standards. Arvanitis et al (2007) performed a
detailed study of the characteristics of a CMOS sensor with 25 \( \mu m \) dels and 13 cm \( \times \) 13 cm field of view. The study characterized various sources of noise in absolute units of electrons and reported MTF, NPS and DQE results that showed the potential of active-pixel sensor CMOS. Passive-pixel sensors (PPS) and active-pixel sensors (APS) are the two broad classes of CMOS detectors. In the PPS design, originally invented in 1967 (Weckler 1967), each detector element consists of a photodiode and an access transistor. Activating the access transistor connects the photodiode to the data line for readout and digitization (Elbakri et al 2009). A sensor with an active amplifier within each pixel is referred to as an Active Pixel Sensor (APS). For such sensors power dissipation is minimal and is generally less than a CCD, since each amplifier is only active during readout (Billas 2011).

B.8. Active Matrix Flat Panel Imagers Detectors (AMFPI)

AMFPI technologies are based on thin-film-transistor (TFT) arrays, made from amorphous silicon, upon which lithographic etching and material evaporation on the micron scale produces the electronic components and connections necessary for detector operation. The flat-panel substrate is divided into individual detector element (del) compartments, arranged in a row and column matrix, typically with a spacing dimension of 70 microns to 250 microns, depending on the detector specifications. Components within each del include a thin-film-transistor (essentially an electronic switch), a charge collection electrode and a storage capacitor. Electronic interconnections including gate lines (rows) and drain lines (columns) are connected to each of the TFTs to control the on/off status of each of the dels and to provide the conduction paths to the charge amplifiers, respectively. All of the TFTs are closed during an exposure to collect X-ray induced charge proportional to the incident X-ray
fluence in each del. After the exposure, active readout of the array occurs one row at a time by activating the respective gate line, which turns on the TFTs and allows the stored charge to flow along the columns from each del capacitor via drain lines to the corresponding charge amplifier. Banks of amplifiers simultaneously amplify the charge, convert to a proportional voltage, digitize signals in parallel from each row of the detector matrix and produce a corresponding row of integer values in the digital image matrix (Figure 4). This process is repeated for each row in the matrix. Detector readout speed is governed by the intrinsic lag characteristics of the X-ray converter material, switching speed of the TFT array electronics and the number of independent charge amplifier arrays that can function in parallel. Limits to spatial resolution for TFT arrays include fill factor, electronic interconnections and manufacturing yield. Fill-factor is a term describing the active charge collection area to the total area of the del, and ideally is 100% for most efficient collection of X-ray information. However, because electronic components and connection lines occupy space on the substrate, with conventional TFT designs the fill factor is less than ideal, this is more severe for smaller del areas of ~0.1 \( \mu m \) (e.g., <50%), and can ultimately limit the minimum size of the del (highest spatial resolution achievable) (Seibert 2009).
CHAPTER B

B.9. Image Quality

B.9.1. Pixel Size, Matrix and Detector Size

Digital images consist of an array of picture elements, often referred to as pixels. The two dimensional collection of pixels in the image is called the matrix, which is usually expressed as length (in pixels) by width (in pixels). Maximum achievable spatial resolution (Nyquist frequency, given in cycles per millimeter) is defined by pixel size and spacing. The smaller the pixel size (or the larger the matrix) is, the higher the maximum achievable spatial resolution. The overall detector size determines if the detector is suitable for all clinical applications. Larger detector areas are needed for chest imaging than for imaging of the extremities. In cassette-based systems, different sizes are available (Michail 2010).

B.9.2. Dynamic Range

Dynamic range is a measure of the signal response of a detector that is exposed to X-rays (Spahn 2005). In conventional screen-film combinations, the dynamic range gradation curve is S shaped within a narrow exposure range for optimal film
blackening; thus, the film has a low tolerance for an exposure that is higher or lower than required, resulting in failed exposures or insufficient image quality. For digital detectors, dynamic range is the range of X-ray exposure over which a meaningful image can be obtained. Digital detectors have a wider and linear dynamic range, which, in clinical practice, virtually eliminates the risk of a failed exposure. Another positive effect of a wide dynamic range is that differences between specific tissue absorptions (e.g., bone vs. soft tissue) can be displayed in one image without the need for additional images. On the other hand, because detector function improves as radiation exposure increases, special care has to be taken not to overexpose the patient by applying more radiation than is needed for a diagnostically sufficient image (Spahn 2005).

**B.9.3. Spatial Resolution**

Spatial resolution refers to the minimum resolvable separation between high-contrast objects. In digital detectors, spatial resolution is defined and limited by the minimum pixel size. Increasing the radiation applied to the detector will not improve the maximum spatial resolution. On the other hand, scatter of X-ray quanta and light photons within the detector influence spatial resolution. Therefore, the intrinsic spatial resolution for selenium-based direct conversion detectors is higher than that for indirect conversion detectors (Michail 2010).

**B.9.4. Modulation Transfer Function (MTF)**

Modulation transfer function (MTF) is the capacity of the detector to transfer the modulation of the input signal at a given spatial frequency to its output (Hendee 1970). At radiography, objects having different sizes and opacity are displayed with different gray-scale values in an image. MTF has to do with the display of contrast and object
size. More specifically, MTF is responsible for converting contrast values of different-sized objects (object contrast) into contrast intensity levels in the image (image contrast). For general imaging, the relevant details are in a range between 0 and 2 cycles/mm, which demands high MTF values. MTF is a useful measure of true or effective resolution, since it accounts for the amount of blur and contrast over a range of spatial frequencies (Michail 2010).

**B.9.5. Noise Power Spectrum (NPS)**

The noise power spectrum (NPS) is a spectral decomposition of the variance. As such, the NPS of a digital radiographic image provides an estimate of the spatial frequency dependence of the pixel-to-pixel fluctuations present in the image (Williams et al 1999). Such fluctuations are due to the shot (quantum) noise in the X-ray quanta incident on the detector, and any noise introduced by the series of conversions and transmissions of quanta in the cascaded stages between detector input and output. Examples of the latter are the signal and dark current shot noise, the reset noise, the amplifier noise, the white noise of source follower, the quantization noise, the fixed pattern noise due to pixel-to-pixel variation in sensitivity or photoresponse nonuniformity, the dark current fixed pattern noise and the offset differences between pixel and column parallel signal readout. The fixed pattern noise is the result of nonuniformities in the process causing small variations in device physical dimensions, and differences in pixel and column voltage thresholds (Williams et al 1999, Arvanitis et al 2007). The NPS is a much more complete description of image noise than is quantification of integrated (total) noise via simple measurement of the rms pixel fluctuations, because it gives information on the distribution in frequency space of the noise power. An understanding of the frequency content of image noise can
provide insight regarding its clinical impact. For example, in mammography, excess high-frequency noise may render the detection of microcalcifications impossible. If desired, the total variance can be obtained by integrating the NPS over spatial frequency (Williams et al 1999).

**B.9.6. Detective Quantum Efficiency (DQE)**

Detective quantum efficiency (DQE) is one of the fundamental physical variables related to image quality in radiography and refers to the efficiency of a detector in converting incident X-ray energy into an image signal. DQE is calculated by comparing the signal-to-noise ratio at the detector output with that at the detector input as a function of spatial frequency (Dainty and Shaw 1974). DQE is dependent on radiation exposure, spatial frequency, MTF, NPS and detector material. High DQE values indicate that less radiation is needed to achieve identical image quality; increasing the DQE and leaving radiation exposure constant will improve image quality. The ideal detector would have a DQE of 1, meaning that all the radiation energy is absorbed and converted into image information. During the past few years, various methods of measuring DQE have been established, making the comparison of DQE values difficult if not impossible. In 2003, the IEC62220-1 (Marshall 2006a) standard was introduced to standardize DQE measurements and make them comparable (Medical Electrical Equipment 2003, Saunders et al 2005).

**B.9.7. Information Capacity (IC)**

The concept of image information capacity (IC) has been introduced within the context of Shannon's information theory, in order to assess image information content (Shannon 1948, Wagner et al 1979, Panayiotakis 1984, Kandarakis et al 2001). However, little work relevant to medical imaging has been published up to now.
(Michail et al 2011, Kanamori 1968, Maiorchuk et al 1974, Brown et al 1979, Shaw 1979, Evans 1981, Kanamori and Matuoto 1984, Michail et al 2009, Gregg 1967, Cavouras et al 2000). According to Shannon’s theory the image information capacity, per unit of image area, is given as a function of the output signal to noise ratio. Information capacity shows an estimation of the system performance using a single value index, instead to the spatial frequency dependent parameters. Information capacity can be applied in newly developed technologies such as telemammography (Mahesh 2004), which is an advancement of digital mammography, since it allows underserved and geographically remote populations to access the latest in breast care.
CHAPTER C

MATERIALS AND METHODS

C.1. Experimental Setup

C.1.1. Phosphor Screen Preparation

The phosphor Gd$_2$O$_2$S:Eu for the experiments, were purchased in powder form (Phosphors Technology Ltd, England, code UKL63/N-R1 with mean grain size of approximately 8 μm and volume density of 7.3. The phosphor was used in the form of thin layers (test screens) to simulate the intensifying screen employed in X-ray mammography and radiography. The screens prepared by sedimentation of the powder phosphors on Borosilicate glass substrate 22x22 mm$^2$ with thickness of 0.14 mm (Waldemar Knittel- GmbH). Sodium Orthosilicate (Na$_2$SiO$_3$) was used as binding material between the powder grains. The sedimentation procedure was achieved by using a mixture consisting of 500 ml of de-ionized water, 10 ml of Na$_2$SiO$_3$, and the appropriate amount of phosphor powder in a glass tube of 100 cm height. The Borosilicate glass substrate was placed at the bottom of the tube (Giakoumakis et al 1990, Kandarakis et al 1997).

Sedimentation has been a widely accepted technique for preparation of radiographic phosphor screens with good homogeneity in various dimensions (i.e. area and thickness) and spatial resolution (Sadowsky 1949, Shigeo and William 1998). The screens obtained, with a packing density of the order of 50% which is common in commercial phosphor screens (Sadowsky 1949, Shigeo and William 1998).

C.1.2. CMOS Sensor

For the detector under investigation, two Gd$_2$O$_2$S:Eu phosphor screens, with coating thicknesses of 33.3 mg/cm$^2$, for mammography, and 65.1 mg/cm$^2$, for radiography,
were manually coupled to an optical readout device including a CMOS Remote RadEye HR photodiode pixel array (Remote RadEye Systems Rad-Icon Imaging Corporation a division of Dalsa). The CMOS photodiode array consists of 1200×1600 pixels with 22.5 μm pixel spacing. The Gd₂O₂S:Eu screens were overlaid onto the active area of the CMOS photodiode array, with the phosphor layer directly coupled to the photodiode array to prevent image degradation caused by the glass substrate. The photodiode array consists of an N-well diffusion on p-type epitaxial Silicon, and held by using a thin Polyurethane foam layer for compression between the screen and a 1-mm-thick Graphite cover. A component view is shown in Figure 5. Experiments, on the CMOS optical sensor, were carried out in both X-ray mammography and radiography energy ranges. Three beam qualities were used; 28 kV W/Rh and W/Ag for mammography and 70 kV (RQA 5) for radiography according to the IEC standards (Medical Electrical Equipment-Characteristics of Digital X-Ray Imaging Devices 2005). IEC standards X-ray spectra were achieved by adding 2 mm Al and 21 mm Al filtration respectively. Half value layers were calculated and found 7.1 mm for radiography and 0.77 and 0.89 mm for W/Rh and W/Ag mammographic conditions respectively.

FIG. 4. Gd₂O₂S:Eu screen coupled to RadEye HR CMOS sensor.
A Hologic Selenia X-ray tube, available with two target/filter combinations (W/Rh and W/Ag), was used for the mammographic beam qualities and a Del Medical Eureka X-ray tube with a rotating Tungsten anode and 1.5 mm Aluminum (Al) inherent filtration for the RQA 5 beam quality.

In this study the source-to-detector distances (SDD) between the X-ray focal spot and the surface of the detector was set to 185 and 66 cm for the radiographic and mammographic energies, respectively (Medical Electrical Equipment-Characteristics of Digital X-Ray Imaging Devices 2005). The added filtration was placed as close as possible to the source.

C.1.3. X-Ray Spectra Measurement

A portable Amptek XR-100T X-ray spectrometer (X-ray & Gamma Ray Detector, XR-100T-CdTe, Amptek, Bedford), based on a cadmium telluride (CdTe) crystal solid-state detector was used for direct diagnostic X-Ray spectra measurements (L. Abbene et al 2007). According to manufacturer’s data, the energy resolution of the CdTe detector at 122 keV, obtained with a source, was less than 1.2 keV at the full width at half maximum (FWHM) (X-ray & Gamma Ray Detector, XR-100T-CdTe, Amptek, Bedford, MA). The CdTe detector was calibrated for energy scales, linearity checks and energy resolution by using $\gamma$-ray calibration sources of $^{125}\text{I}$ and $^{99\text{m}}\text{Tc}$. The CdTe was placed at a focus to detector distance of 166 and 48 cm for the radiographic and the mammographic energies respectively. X-ray spectra were corrected by the inverse square law at the CMOS detector plane for all beam qualities. In order to minimize pile-up distortions, dedicated collimation systems (1 mm thick collimators with 400 $\mu$m and 100 $\mu$m diameters, respectively for the radiographic and mammographic X-ray spectra) were used. In addition the measured X-ray spectra
were corrected for the efficiency of the CdTe detector. The air Kerma value $K_\alpha$ (in mGy) at the surface of the detector was calculated according to (L. Abbene et al 2007):

$$K_\alpha = 0.00869 \times \sum_{E_{\text{min}}}^{E_{\text{max}}} (1.83 \times 10^{-6} \times \Phi_0(E) \times E \times (\mu_{en}(E) / \rho_{air}))$$  

(1)

Where $\Phi_0$ is the measured X-ray spectrum value (photons/mm$^2$) at energy $E$ (keV). $(\mu_{en}(E) / \rho_{air})$ (cm$^2$/g) is the X-ray mass energy absorption coefficient of air at energy $E$ obtained from the literature (J. R. Greening 1985). The exposure rate at the entrance surface of the CMOS photodiode array detector was measured for a range of tube current-time products (mAs). The measured X-ray spectra were corrected for the efficiency of the CdTe detector.

C.2. Image quality

C.2.1. Spectral Compatibility

The spectral compatibility between the emitted scintillator light and the spectral sensitivity of the CMOS optical sensors was calculated by the spectral matching factor ($\alpha_S$) according to (1):

$$\alpha_S = \int S_P(\lambda)S_D(\lambda)d\lambda / \int S_P(\lambda)d\lambda$$  

(2)

where $S_P(\lambda)$ is the emitted light spectrum of the phosphor and $S_D(\lambda)$ is the spectral sensitivity of the optical detector coupled to the phosphor (Michail et al 2010). The $S_P(\lambda)$ of the powder phosphor screens was measured under X-ray excitation by an
C.2.2. Signal Transfer Property (STP)

The Signal Transfer Property (STP), or detector response, states the relationship between mean pixel value (MPV) and ESAK at the detector surface. This relationship was obtained by plotting pixel values (PV) versus ESAK at the detector, as described in the IEC method (Medical Electrical Equipment-Characteristics of Digital X-Ray Imaging Devices 2005). A sequence of uniform images was acquired at different exposure levels. MPV was evaluated in a 1 × 1 cm region of interest (ROI). The mean pixel value (MPV) and the standard deviation within that region were measured. The system’s response curve was fitted using a linear equation of the form:

\[ MPV = \alpha + b \times K \]

Where \( \alpha \) and \( b \) are fit parameters. From the slope of the system’s response curve, the value of the gain factor (G) was obtained. This value is relating MPV to the incident exposure at the detector (in digital units per \( \mu \text{Gy} \)) (Neitzel et al 2004). The magnitude of the pixel offset at zero air Kerma was also estimated (Samei et al 1998).

C.2.3. Modulation Transfer Function (MTF)

The MTF was measured using the slanted-edge technique, following the procedures described in IEC standard (Marshall 2006a, Dobbins 2000, Illers et al 2005). A PTW Freiburg tungsten edge test device was used to obtain the slanted edge images in both radiographic and mammographic energies. The edge test device consists of a 1 mm thick W edge plate (100×75 mm\(^2\)) fixed on a 3 mm thick lead plate. Images of the edge, placed at a slight angle, were obtained under the radiographic and mammographic imaging conditions. Three exposure levels have to be chosen for the
measurements. From these levels, the medium one should be the 'normal' level routinely employed in the clinical practice (Medical Electrical Equipment-Characteristics of Digital X-Ray Imaging Devices 2005). The edge spread function (ESF) was calculated by the extraction of an 1×1 cm² ROI with the edge roughly at the center. The angle of the edge was then determined using a simple linear least squares fit and the 2D image data were re-projected around the angled edge (Marshall 2006a) to form an ESF with a bin spacing of 0.1 pixels. The ESF was smoothed with a median filter of five bins to reduce high frequency noise. The median filter is much less sensitive to extreme values (outliers) than averaging filters. Therefore it is more efficient to remove these outliers without reducing the sharpness of the image. Then the ESF was differentiated to obtain the line spread function (LSF) (Greening 1985, Boone 2001). Finally, the normalized LSF was Fourier transformed to give the pre-sampling MTF.

C.2.4. Normalized Noise Power Spectrum (NNPS)

The NPS was calculated according to IEC 62220-1-2. (Medical Electrical Equipment-Characteristics of Digital X-Ray Imaging Devices 2005) For each ROI, the PV were converted into air kerma units with equation (4). The slowly varying spatial background effects including the heel effect were corrected by fitting and subtracting a two-dimensional second-order polynomial to the original acquired image data. The area of analysis was subsequently divided into sub-images of 1024×1024 pixels. Half overlapping ROIs with a size of 128×128 pixels were then taken from the sub-images (Medical Electrical Equipment-Characteristics of Digital X-Ray Imaging Devices 2005). A total of 128 ROIs were taken from each flood image. For all the ROIs taken from each image 2D fast Fourier transform (FFT) of each ROI was calculated and
added to the NPS ensemble. NNPS was obtained by dividing NPS by the square of the corresponding $K_\alpha$ and afterwards the ensemble average was obtained.

**C.2.5. Detective Quantum Efficiency (DQE)**

The DQE of the system was calculated by the following equation:

$$DQE(u) = \frac{MTF^2(u)}{K_\alpha \times q \times NNPS(u)}$$

Where $q$ is the number of photons per unit Kerma ($\mu$Gy) per mm$^2$, determined by dividing the number of photons per mm$^2$ (measured with the portable X-ray spectrometer) with the corresponding air Kerma value ($\mu$Gy) (Abbene et al 2007). Values of 21738 $\text{photons} \times \mu\text{Gy}^{-1} \times \text{mm}^{-2}$, for the RQA 5, and 6161 and 5422 $\text{photons} \times \mu\text{Gy}^{-1} \times \text{mm}^{-2}$, for mammographic W/Ag and W/Rh beam qualities respectively were calculated from our direct X-ray spectra measurements, instead of using tabulated data.

**C.2.6. Noise Equivalent Quanta (NEQ)**

A measure of the lower number of exposure quanta required by an ideal detector to yield the same $SNR_{out}^2$ as the practical detector working at a given exposure level can be expressed by the noise equivalent quanta (NEQ) (Shaw 1978). The latter is the combined effects of signal and noise, in terms of spatial frequency. Also provides an index of the signal-to-noise ratio (SNR) associated with the diagnostic value of a medical image. NEQ can be written as follows (Dobbins 2000):

$$NEQ(u) = DQE(u) \times \Phi_0$$
C.2.7. Information Capacity (IC)


\[
IC = \pi \int_0^\infty \log_2 \left[1 + NEQ(u)\right] u du
\]  

(6)

IC shows an estimation of the system performance using a single value index, instead of the spatial frequency dependent parameters (Kandarakis et al 2001c).
CHAPTER D

RESULTS AND DISCUSSION

D.1. Optical Properties and MTF of Borosilicate Glass Substrate

The optical properties of the Borosilicate glass substrate have been measured via a Perkin-Elmer Lambda 15 UV/VIS Spectrophotometer. In the visible range the optical transparency was found very high, of the order of 93%, while absorption was found very low (0.029). These values show that the substrate has very high transparency, low reflectivity and low absorption in the wavelength range of the phosphor emission.

In addition the MTF of the thin glass substrate was experimentally investigated. The MTF was measured: a) with the glass substrate directly coupled to the photodiode array (MTF_1) and b) with the phosphor directly coupled to the photodiode array (MTF_2). In both cases the slanted-edge technique was used (Michail et al 2011). By taking into account that the MTF of an imaging system is the product of separate MTF components of an imaging system, thus the MTF of the glass substrate (MTF_{glass}) is (Michail 2010):

\[
MTF_{glass}(u) = \frac{MTF_1(u)}{MTF_2(u)}
\] (7)

Figure 5 shows the MTF_1 and the MTF_2 of the detector. From this figure can be depicted an obvious degradation of the MTF due to thin glass substrate. MTF_1 is 40% lower than MTF_2 on average, in the whole spatial frequency range. When the glass substrate is directly coupled to the photodiode array (MTF_1), the light photons emitted
by the phosphor interact with the glass substrate leading to a subsequent degradation of the MTF. When the phosphor is directly coupled to the photodiode array (MTF\textsubscript{2}) the light photons are directly detected by the photodiode array without interaction with the glass substrate. In this case there is no degradation of the MTF caused by the glass substrate. Furthermore figure 4 shows the MTF of the glass substrate (MTF\textsubscript{glass}) calculated by (1). The glass substrate has a low impact on resolution of the image at low frequencies (12% on average, up to 2.7 cycles/mm) and a higher impact at medium and high frequencies (70% on average).

\begin{figure}
\centering
\includegraphics[width=0.5\textwidth]{MTF.png}
\caption{Comparison of the MTFs curves for both cases. MTF of the glass substrate is also shown.}
\end{figure}

**D.2. Spectra Compatibility**

The spectral compatibilities between the optical spectra emitted by Gd\textsubscript{2}O\textsubscript{2}S:Eu and Gd\textsubscript{2}O\textsubscript{2}S:Tb screens and the sensitivity of the CMOS optical sensor were calculated for the wavelengths range from 375 nm up to 675 nm by evaluating the spectral matching factor (relation (3)). In particular, it was found that this factor is higher in the case of Gd\textsubscript{2}O\textsubscript{2}S:Eu screens coupled to the CMOS sensor (matching factor: 1), than in the case of the Gd\textsubscript{2}O\textsubscript{2}S:Tb screens (matching factor:0.95). This shows that red emitting
phosphors have better compatibility with the CMOS sensors and, apparently, with the rest of Silicon based sensors. Such results should be taken into account in evaluating phosphor and scintillator materials since their effective efficiency can be significantly altered depending on the optical sensor incorporated in the detector.

D.3. X-Ray Spectra

Figures 6 and 7 show measured spectra produced by the X-ray tubes (radiographic with W/Al target-filter combination and mammographic with W/Ag and W/Rh target-filter combinations), used in the present study. The tube settings were: 20, 63 and 157.5 mAs at 70 kVp (RQA 5) and 20 mAs, 60 mAs and 120 mAs at 28 kV respectively. The system reproducibility was verified by measuring X-ray spectra several times. Entrance surface air Kerma was calculated by relation (2) using the Amptek XR-100T X-ray spectrometer measurements and was found 11.2, 34.1 and 87.5 $\mu$Gy for the RQA 5 and 65.8, 197.4 and 394.1 $\mu$Gy and 55.7, 167.0 and 334.0 $\mu$Gy for mammographic conditions W/Ag and W/Rh, respectively.
Table I. Parameters of the mammographic and radiographic beam qualities.

<table>
<thead>
<tr>
<th>Radiation Beam Parameters</th>
<th>Mammographic quality (W/Rh)</th>
<th>Mammographic quality (W/Ag)</th>
<th>Radiographic quality (RQA 5)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anode/filtration combination</td>
<td>W/Rh</td>
<td>W/Ag</td>
<td>W/Al</td>
</tr>
<tr>
<td>Tube voltage (KV)</td>
<td>28</td>
<td>28</td>
<td>70</td>
</tr>
<tr>
<td>Added filtration (mm Al)</td>
<td>2</td>
<td>2</td>
<td>21</td>
</tr>
<tr>
<td>Measured HVL (mm Al)</td>
<td>0.77</td>
<td>0.89</td>
<td>7.1</td>
</tr>
<tr>
<td>$Q(\text{photons} \times \mu \text{Gy}^{-1} \times \text{mm}^{-2})$</td>
<td>5422</td>
<td>6161</td>
<td>21738</td>
</tr>
</tbody>
</table>

FIG. 6. Measured X-ray spectra for RQA 5 beam quality.
D.4. Signal Transfer Property (STP)

Figure 8 shows the detector response curve (STP) of the RadEye HR CMOS sensor under RQA 5 (70kVp) and mammographic (28kVp) beam qualities. The detector was found to have a linear response, covering the whole exposure range, with a pixel value offset of 144.879 for the RQA 5 beam quality and 328.912 and 296.796 for the W/Ag and W/Rh qualities, respectively. The linear no threshold fits gave correlation coefficients ($R^2$) greater than 0.9978, 0.9971 and 0.9989 for the RQA 5 and the Ag and Rh filtered beam qualities, respectively. The gain factor $G$ was determined as the slope of the characteristic curve, relating the mean pixel value to the incident exposure. Using flat-field images the gain factors were determined by linear regression to be $G=7.347$ digital units per $\mu$Gy for RQA 5 and 2.635 and 2.400 digital units per $\mu$Gy for Ag and Rh filtered beams, respectively.
D.5. Modulation Transfer Function (MTF)

Figure 9 shows the modulation transfer function (MTF) curves for three exposure levels under the RQA5 beam quality. In this figure the MTF of a CMOS sensor with 48 μm pixel pitch coupled to a Gd₂O₂S:Tb screen of 59.2 mg/cm² coating thickness (Cho et al 2008), is also shown for comparison purposes. This MTF has been obtained under RQA 5 beam quality at 43.5 μGy. As it can be depicted from figure 9, the MTF of the 48 μm CMOS sensor (Cho et al 2008) is 13% lower, on average, than the MTF of the 22.5 μm CMOS sensor (investigated in the present study) in the frequency range from 0 to 5 cycles/mm. In the higher spatial frequency range (5-10 cycles/mm), the MTF values differ by 3% on average. To explain these differences the combined effects of screen thickness/sensor pixel size, as well as the screen optical properties must be taken into account. The screen of our system is slightly thicker (65.1 mg/cm² instead of 59.2 mg/cm²) and, in addition, it has not been optically optimized (i.e. incorporation of special light absorbing dyes) like commercially available screens. However, taking into account our results, it could be
claimed that the red phosphor based system can show improved or acceptable signal transfer properties in the whole spatial frequency range.

Figure 9 shows the MTFs of the 65.1 mg/cm² Gd₂O₂S:Eu/CMOS detector with a 59.2 mg/cm² Gd₂O₂S:Tb/CMOS detector under RQA 5 beam quality.

Figure 10 shows MTF curves for mammographic beam qualities with two different anode/filter combinations (W/Ag and W/Rh). MTFs obtained with the Rhodium filters are higher than those obtained with the Silver filter by approximately 5, 6 and 5 %, for the three exposure levels respectively. Figure 11 shows a comparison between the MTFs of the present CMOS sensor combined with the 33.3 mg/cm² Gd₂O₂S:Eu screen and the, previously published, CMOS sensor coupled to a 33.91 mg/cm² Gd₂O₂S:Tb screen (Michail et al 2011). The MTFs of the present CMOS/33.3 mg/cm² Gd₂O₂S:Eu screen are comparable to those of the CMOS/33.91 mg/cm² Gd₂O₂S:Eu screen (Michail et al 2011) (deviations lower than 5, 2 and 1 % for the three exposure levels). However, a point by point comparison is not possible, since exposure levels and beam qualities were clearly different in the two studies, e.g. W/Rh and W/Ag.
anode/filters combinations in the present study instead of Mo/Mo (RQA M2) of the previous study.

![MTF curve for W/Rh and W/Ag beam qualities.](image1)

**FIG. 10.** MTF curve for W/Rh and W/Ag beam qualities.

![Comparison of the MTFs.](image2)

**FIG. 11.** Comparison of the MTFs for the CMOS/Gd$_2$O$_2$S:Eu sensor under investigation and a previously published CMOS/Gd$_2$O$_2$S:Tb sensor under W/Rh and RQA M2 beam qualities respectively.

Figures 12 and 13 show extracted 1D normalized noise-power spectra (NNPS), for the $u$ direction, obtained from uniformly exposed images under general radiography (RQA 5) and mammographic conditions with two different anode/filters combinations (Ag and Rh). NNPS shows variation over a large spatial frequency range. For example, a decrease from $3.54 \times 10^{-6}$ mm$^2$ down to $2.77 \times 10^{-6}$ mm$^2$ (i.e. 22%) was observed in NNPS values in the range from 2 to 5 cycles/mm (exposure level 11.2 $\mu$Gy for the RQA 5 beam quality). In the W/Ag and the W/Rh beam qualities (exposure levels 197.4 $\mu$Gy and 167.0 $\mu$Gy respectively) a decrease of 58% and 50% was observed. This is attributed to the fact that in the CMOS system, a 2D correction is applied reducing the structured noise in each image. In the medium to high frequency range, the noise levels of the image are related to the MTF of the system. The rather slow variation of the NNPS may be attributed to the fact that MTF remains high up and in the higher frequency range. For comparison purposes, NNPS of a CMOS sensor measured in a previous study is also shown in Figure 12. (Cho et al 2008) This curve was obtained from a 48 $\mu$m passive pixel sensor under the representative RQA 5 beam quality, at an exposure level of 43.5 $\mu$Gy. Figure shows that the noise levels of the CMOS sensor under investigation is higher than that of the previously published study (Cho et al 2008), due to the larger effective pixel size of the passive CMOS sensor (Cho et al 2008) leading to higher photon collection efficiency and to a subsequent increase in the signal to noise ratio (SNR) (Chen et al 2000, Farrell et al 2006). Furthermore in the fabrication process of the custom made Gd$_2$O$_2$S:Eu screens which has not been optically optimized (i.e. incorporation of
special light absorbing dyes) and are possible to be subjected to more structural non-uniformities than the commercial Gd₂O₂S: Tb screen used in the previous study.

![Graph](image)

**FIG. 12.** Comparison of the NNPS curves between the CMOS/Gd₂O₂S:Eu sensor under investigation and a previously published CMOS/Gd₂O₂S:Tb sensor under RQA 5 beam quality.

![Graph](image)

**FIG. 13.** NNPS curves for W/Rh and W/Ag beam qualities.

An additional comparison for the mammographic beam quality is shown in Fig. 14 in which the NNPS of a CMOS sensor measured in a previous publication of our group.
is also shown for comparison purposes (Michail et al 2011). This curve was obtained for the CMOS devise under investigation, coupled to a commercial Gd$_2$O$_2$S:Tb scintillation screen under the representative RQA M2 (Mo/Mo at 28KV) beam quality at exposure level of 40.1 $\mu$Gy (Michail et al 2011). NNPS of CMOS sensor/Gd$_2$O$_2$S:Eu screens combination shows higher noise in the whole spatial frequency range. The reduced NNPS values of Michail et al (2011) can also be attributed to the fabrication process of the Gd$_2$O$_2$S:Eu screen, as well as, in the lower energy components of the RQA-M2 used in that study. The latter may enhance the X-ray absorption properties of Gd$_2$O$_2$S, resulting in lower noise values.

![Graph](image)

**FIG. 14.** Comparison of the NNPS curves between the CMOS/Gd$_2$O$_2$S:Eu sensor under investigation and a previously published CMOS/Gd$_2$O$_2$S:Tb sensor under W/Rh and RQA M2 beam qualities respectively.

**D.7. Detective Quantum Efficiency (DQE)**

Figures 15 and 16 show DQE curves for the RQA 5, W/Ag and W/Rh qualities, obtained according to (5). DQE was investigated at various air Kerma settings at the detector surface. DQE decreases as the ESAK increases due to the influence of the
NNPS and MTF. All the DQE curves showed an increase in the spatial frequency from 0 to 1.8 cycles/mm for the RQA 5 and from 0 to 5 cycles/mm for the mammography beam qualities. This is due to the fact that NNPS falls off rapidly in these spatial frequency ranges. Thereafter the reduction rate of the NNPS falls off in contrast with the corresponding of the MTF curves, contributing to a decrease in the DQE. For comparison purposes, DQE of the 48 μm pixel size CMOS sensor (Cho et al 2008) (RQA 5) at an exposure level of 43.5 μGy is also shown in Figure 15. The DQE values of the previously published study (Cho et al 2008) were higher in the low spatial frequency range (up to 1.8 cycles/mm). This is due to the higher NNPS values of this spatial frequency range of the detector under investigation which contributes to lower DQE values. Thereafter DQE curve is comparable in the whole spatial frequency range due to the combined effects of MTF, NNPS and incident X-ray spectra, as depicted in the previous Figures. These results show that a further improvement on the DQE can be achieved with a commercialised Gd₂O₂S:Eu screen preparation which will lead to an additional noise reduction.

FIG. 15. Comparison of the DQE curves between the CMOS/Gd₂O₂S:Eu sensor under investigation and a previously published CMOS/Gd₂O₂S:Tb sensor under RQA 5 beam quality.
In Fig. 17, DQE curves, of the CMOS sensor under investigation, coupled to a commercial Gd$_2$O$_2$S:Tb screen, published in a previous study, is also shown for comparison purposes, under mammography conditions, at an exposure level of 40.1 $\mu$Gy (Michail et al 2011). DQE values of the currently investigated CMOS sensor/scintillating screen combination is lower than the corresponding of the CMOS/Gd$_2$O$_2$S:Tb screen combination (Michail et al 2011), due to the higher noise performance of the current CMOS/Gd$_2$O$_2$S:Eu for the reasons previously reported.
D.8. Noise Equivalent Quanta (NEQ)

Figures 18 and 19 show noise equivalent quanta (NEQ) of the RQA 5 and mammographic beam qualities, calculated from the MTF, NNPS and the measured X-ray photon distribution at the detector surface. The shapes of the NEQ curves are affected by both NNPS and MTF. The amplitude of NEQ is affected by the number of X-ray photons at the surface of the detector (photons/mm²). As shown in these Figures, NEQ reaches a maximum (73563 mm²) in the low frequency range (1.8 cycles/mm), under the RQA 5 (ESAK: 11.2 μGy) conditions and in the medium frequency range (183074 mm² at 3.6 cycles/mm) under mammographic conditions (ESAK: 167.0 μGy W/Rh).
In Fig. 20 NEQ values of a CMOS sensor is also shown for comparison purposes (Michail et al. 2011). NEQ values of this CMOS sensor is higher in the spatial frequency range up to 1 cycles/mm. NEQ curves of both CMOS sensors are comparable thereafter.
FIG. 20. Comparison of the NEQ curves between the CMOS/Gd$_2$O$_2$S:Eu sensor under investigation and a previously published CMOS/Gd$_2$O$_2$S:Tb sensor under W/Rh and RQA M2 beam qualities respectively.

D.9. Information Capacity (IC)

In Table II the information capacity values for the RQA 5 and mammographic beam qualities are shown. CMOS sensor/Gd$_2$O$_2$S:Eu screens combinations show higher IC values in the mammography energy range (2475-2821 bits/mm$^2$), in comparison to the radiography energy range (1730-1851 bits/mm$^2$) due to the higher MTF and incident air Kerma values. These data show that, for a given level of incident X-ray fluence, information capacity is mainly determined by the intrinsic phosphor material properties, while the role of the screen thickness is of low significance. Thin screens exhibit low levels of brightness (Michail et al 2010), which is compensated for by their higher spatial resolution (MTF), whereas thick screens show poor resolution properties but with high brightness gain (Michail et al 2010). The practical value of the IC is that it defines an imaging performance index that evaluates image information quantity by a single numerical value. Information capacity is not
expressed for specific frequency values since it is the outcome of integrating over the spatial frequency bandwidth (Michail 2010). IC may be considered as being roughly proportional to the area under the curve of the frequency-dependent image SNR. Thus, IC mainly reflects the quantity and not the quality of image information. Image quality is better described via the frequency-dependent signal parameters. It must be noted, however, that IC is significantly affected by the maximum frequency contained in the signal, i.e. the frequency bandwidth over which integration in (6) is performed. A practical problem is to properly select this maximum frequency, which may significantly affect the final IC result. In addition to this, information capacity is calculated using spatial frequency as a factor, which multiplies the logarithmic term containing SNR. Hence, the contribution of high spatial frequency SNR components to the final result may be relatively enhanced. This affects IC in a sense that it may be preferably emphasized by high frequency information content, i.e. the kind of information that is expressed by spatial resolution and sharpness, which is better displayed by thin layers. This may explain the higher IC values obtained for thin screens.

Table II. Information capacity and ESAK values for the mammographic and radiographic beam qualities.

<table>
<thead>
<tr>
<th>Beam Quality</th>
<th>Entrance Surface Air-Kerma (ESAK µGy)</th>
<th>Information capacity (bits/mm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>(W/Ag)</td>
<td>65.8±0.036</td>
<td>2475±26</td>
</tr>
<tr>
<td>(W/Ag)</td>
<td>197.4±0.071</td>
<td>2678±41</td>
</tr>
<tr>
<td>(W/Ag)</td>
<td>394.1±0.084</td>
<td>2733±39</td>
</tr>
<tr>
<td>(W/Rh)</td>
<td>55.7±0.031</td>
<td>2535±42</td>
</tr>
<tr>
<td>(W/Rh)</td>
<td>167.0±0.066</td>
<td>2821±34</td>
</tr>
<tr>
<td>(W/Rh)</td>
<td>334.0±0.073</td>
<td>2747±25</td>
</tr>
<tr>
<td>--------</td>
<td>-------------</td>
<td>---------</td>
</tr>
<tr>
<td>RQA 5</td>
<td>11.2±0.024</td>
<td>1730±31</td>
</tr>
<tr>
<td>RQA 5</td>
<td>34.1±0.032</td>
<td>1826±29</td>
</tr>
<tr>
<td>RQA 5</td>
<td>87.5±0.059</td>
<td>1851±34</td>
</tr>
</tbody>
</table>

Figures 21 show an initial image of a simple electronic device obtained from the CMOS/Gd2O2S:Eu sensor screen combination under the RQA 5 beam quality (obtained at 74 KVp with 21 mm Al filtration). This image demonstrates that the use of a CMOS/Gd2O2S:Eu sensor screen combination may be practically feasible.

FIG. 21. Image obtained from the CMOS/Gd₂O₂S:Eu combination at 74 KVp with 21 mm Al filtration.
CONCLUSIONS AND FUTURE WORK

E.1. Conclusions

In the present study the imaging performance of a high resolution CMOS based imaging sensor combined with custom made Europium (Eu\(^{3+}\)) activated Gd\(_2\)O\(_2\)S screens was investigated in terms of Modulation Transfer Function (MTF), Normalized Noise Power Spectrum (NNPS), Detective Quantum Efficiency (DQE), Noise equivalent Quanta (NEQ) and Information Capacity (IC) covering the mammography and general radiography energy ranges.

The spectral matching factor between the optical spectra emitted by the Gd\(_2\)O\(_2\)S:Eu and the Gd\(_2\)O\(_2\)S:Tb screens and the CMOS optical sensor, evaluated in the present study, was 1 and 0.95 respectively. The ESAK values ranged between 11.2-87.5 \(\mu\)Gy, for RQA-5, and between 65.8-334 \(\mu\)Gy, for W/Rh, W/Ag beam qualities.

It was found that the detector response function was linear for the exposure ranges under investigation. Under radiographic conditions the MTF of the present system was found higher than previously published MTF data for a 48 \(\mu\)m CMOS sensor, in the low up to medium frequency ranges. DQE was found comparable, while the NNPS appeared to be higher in the frequency range under investigation (0–10 cycles/mm). NEQ reached a maximum (73563 mm\(^{-2}\)) in the low frequency range (1.8 cycles/mm), under the RQA 5 (ESAK: 11.2 \(\mu\)Gy) conditions. IC values were found to range between 1730-1851 bits/mm\(^2\). Under mammographic conditions MTF, NNPS and NEQ were found comparable to data previously published for the 48 \(\mu\)m CMOS sensor while the DQE was found lower. The corresponding IC values were found ranging between 2475 and 2821 bits/mm\(^2\).
Results showed that the red emitting phosphor/CMOS sensor combination has comparable image quality parameters in terms of MTF, NNPS, DQE and NEQ compared to previously published data for Terbium activated Gd$_2$O$_2$S/CMOS sensor combinations. Since the imaging performance of Europium (Eu$^{3+}$) activated Gd$_2$O$_2$S screens, combined to CMOS sensors was found comparable to that of Gadolinium (Gd) activated Gd$_2$O$_2$S screens, red emitting phosphors could be used in digital imaging systems, where the Si based photodetectors are more sensitive to longer wavelength ranges, and particularly in the red wavelength range.

E.2. Future Work

The aim of the future work is the experimental and theoretical evaluation of the imaging performance and light emission efficiency of nanophosphor materials in the mammography and radiography energy ranges for application in a novel high resolution digital CMOS based X-ray sensor. The optimum CMOS/powder nanophosphor combination for digital imaging applications as well as for fast data acquisition (digital breast tomosynthesis, dual energy imaging, scanning slot mammography) techniques will be determined. Also the application of the information theory through the information capacity (IC) for the evaluation of the digital imaging CMOS based system will be used.
APPENDIX A: INFORMATION CAPACITY (IC)

According to Shannon’s information theory, if it is possible to distinguish $N_s$ different signal intensity levels of duration $T$ on a channel, we can say that the channel can transmit $\log_2 N_s$ bits in time $T$. The rate of transmission is then $\log_2 N_s / T$. More precisely, the image information capacity, per unit of image area, may be defined by (5) as follows (Shannon 1998):

$$IC = \lim_{T \to \infty} \log_2 N_s / T = n_p \log_2 N_s$$

(A.1)

Where $n_p$ is the number of image elements (pixels) per unit of area that can be registered in an image element. If the transmitter has a dynamic range ($D$) and the noise in the transmission channel is assumed to be Gaussian (Normal ($0, \sigma^2$) and independent of the signal amplitude, then an expression may be derived for $N_s$ in terms of $D$ and $\sigma$. For this case, the levels are equally spaced between zero and $D$ with a spacing determined by the acceptable error rate for signal transmission. If each level have an interval of $\pm k\sigma$, according to Wagner (Wagner et al 1979), then the spacing becomes $2k\sigma$, and one obtains: $N_s = 1 + D / 2k\sigma$. (A.1) may now be written as follows: $IC = n \log_2 (1 + D / 2k\sigma)$. Shannon showed that in the frequency domain there is a definition of information capacity that is consistent with the equations described above and which serves as a unique upper limit for the rate of information transmission (Shannon 1998). This upper limit for the number of distinguishable signals in communication channel of bandwidth $\Delta f$ is defined as:
\[ NS \leq \left( \frac{(SPS + NPS)}{NPS} \right)^{2TM} \]. Where SPS is the signal power spectrum which can defined as: \[ SPS = \left( MTF \cdot G \right)^2 \], where G is the gain factor (in digital units per \( \mu Gy \)) (Wagner et al 1979) and NPS is the noise power spectrum. Consequently, the channel capacity is limited by(Shannon 1998):

\[
IC = \log_2 \frac{NS}{T} \leq \Delta f \log_2 \left( \frac{(SPS + NPS)}{NPS} \right)
\] (A.2)

Where \( SPS \) and \( NPS \) are assumed constants over the frequency interval \( \Delta f \). Wagner has expressed this equation (Wagner et al 1979) under the following form:

\[
IC(\Delta u) = 2\Delta u \log_2 \left( \frac{(SPS + NPS)}{NPS} \right)^{1/2}
\] (A.3)

Where the exponent of \( \frac{1}{2} \) is explained since the ratio of amplitudes is replaced by a ratio of powers, and the factor \( 2\Delta u \) results directly from the sampling theorem. For any function having a bandwidth of \( \Delta u \), sampling at intervals \( 1/2\Delta u \) completely specifies the function. If the function is assumed to be limited to a time interval \( T \), then for a real function there are just \( 2\Delta uT \) independent values of \( 2\Delta u \) values per unit time. The dimensions of \( IC \) are bits per unit time (Wagner et al 1979). If (A.2) is integrated over the whole frequency range then information capacity is given by:

\[
IC = \int_0^1 \log_2 \left[ 1 + \frac{SPS(u)}{NPS(u)} \right] du
\] (A.4)
Where \( u \) is greater than or equal to the band limit of the signal power. The 2D form of (A.3) is given by

\[
IC = (2\Delta u_x)(2\Delta u_y)\log_2 \left( \frac{(SPS + NPS)}{NPS} \right)^{1/2},
\] written in terms of spatial frequency, leading similarly to an integral over the upper right-hand quadrant in \( u_x, u_y \) \((u = (u_x^2 + u_y^2)^{1/2})\) space or using polar frequency coordinates and assuming cylindrical symmetry of \( SPS \) and \( NPS \),

\[
IC = \frac{1}{2\pi} \int_0^{\infty} \log_2 \left[ \frac{1 + SPS(u)}{NPS(u)} \right] 2\pi u du
\]  

(A.5)

expressed in bits per unit area (Wagner et al 1979). For the detectors used in both conventional and digital X-ray imaging, power spectra can be represented as one-dimensional functions of spatial frequency \( u \), \( SPS(u) \) and \( NPS(u) \) due to rotational symmetry (Wagner et al 1979). (A.5) now becomes (Kanamori 1968, Brown et al 1979, Shaw 1979, Evans 1981, Kanamori and Matsuoto 1984, Shannon 1998, Kandarakis et al 2001c, Cavouras et al 2000)

\[
IC = \pi \int_0^{\infty} \log_2 \left[ 1 + \frac{SPS(u)}{NPS(u)} \right] u du
\]  

(A.6)

Where \( SPS(u)/NPS(u) = NEQ \), therefore (A.6) becomes (A.7):

\[
IC = \pi \int_0^{\infty} \log_2 \left[ 1 + NEQ(u) \right] u du
\]  

(A.7)
## APPENDIX B: ABBREVIATIONS

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>AMFPI</td>
<td>Active Matrix Flat Panel Imagers</td>
</tr>
<tr>
<td>a-Si:H</td>
<td>Amorphous Silicon</td>
</tr>
<tr>
<td>CCD</td>
<td>Charge-coupled device</td>
</tr>
<tr>
<td>CR</td>
<td>Computed Radiography</td>
</tr>
<tr>
<td>CMOS</td>
<td>Complementary metal Oxide semiconductor</td>
</tr>
<tr>
<td>Si</td>
<td>Crystalline Silicon</td>
</tr>
<tr>
<td>DQE</td>
<td>Detective Quantum Efficiency</td>
</tr>
<tr>
<td>ESAK</td>
<td>Entrance surface air-Kerma</td>
</tr>
<tr>
<td>FT</td>
<td>Fourier Transform</td>
</tr>
<tr>
<td>HVL</td>
<td>Half-value layer</td>
</tr>
<tr>
<td>IEC</td>
<td>International Electrotechnical Commission</td>
</tr>
<tr>
<td>IC</td>
<td>Information capacity</td>
</tr>
<tr>
<td>nI</td>
<td>Informational efficiency</td>
</tr>
<tr>
<td>( \Phi_o )</td>
<td>Incident X-ray photon fluence</td>
</tr>
<tr>
<td>FFT</td>
<td>Fast Fourier Transform</td>
</tr>
<tr>
<td>Gd2O2S:Tb</td>
<td>Gadolinium Oxysulphide</td>
</tr>
<tr>
<td>LSF</td>
<td>Line Spread Function</td>
</tr>
<tr>
<td>Lu2SiO5:Ce (LSO)</td>
<td>Lutetium Oxyorthosilicate</td>
</tr>
<tr>
<td>MPV</td>
<td>Mean pixel value</td>
</tr>
<tr>
<td>Mo</td>
<td>Molybdenum</td>
</tr>
<tr>
<td>MTF</td>
<td>Modulation transfer function</td>
</tr>
<tr>
<td>NEQ</td>
<td>Noise equivalent quanta</td>
</tr>
<tr>
<td>NPS</td>
<td>Noise power spectrum</td>
</tr>
<tr>
<td>NNPS</td>
<td>Normalized noise power spectra</td>
</tr>
<tr>
<td>Abbreviation</td>
<td>Description</td>
</tr>
<tr>
<td>--------------</td>
<td>-------------</td>
</tr>
<tr>
<td>PPS</td>
<td>Passive-pixel sensors</td>
</tr>
<tr>
<td>PSP</td>
<td>Photostimulable Storage Phosphor</td>
</tr>
<tr>
<td>Rh</td>
<td>Rhodium</td>
</tr>
<tr>
<td>SEM</td>
<td>Scanning electron microscope</td>
</tr>
<tr>
<td>SNR</td>
<td>Signal to noise ratio</td>
</tr>
<tr>
<td>Ag</td>
<td>Silver</td>
</tr>
<tr>
<td>SDD</td>
<td>Source to detector distance</td>
</tr>
<tr>
<td>as</td>
<td>Spectral matching factor</td>
</tr>
<tr>
<td>CsI:Tl</td>
<td>Thallium-doped Cesium Iodide</td>
</tr>
<tr>
<td>TFT</td>
<td>Thin film field-effect transistors</td>
</tr>
<tr>
<td>W</td>
<td>Tungsten</td>
</tr>
<tr>
<td>X</td>
<td>X-ray exposure</td>
</tr>
</tbody>
</table>
APPENDIX C: LIST OF FIGURES

FIG.1. Photostimulable storage phosphor system
FIG.2. Charged couple devise (CCD)
FIG.3. Active matrix flat panel imager (AMFPI)
FIG.4. Gd$_2$O$_2$S:Eu screen coupled to RadEye HR CMOS sensor.
FIG. 5. Comparison of the MTFs curves for both cases. MTF of the glass substrate is also shown.
FIG. 6. Measured X-ray spectra for RQA 5 beam quality.
FIG. 7. Measured X-ray spectra for W/Rh and W/Ag beam qualities.
FIG. 8. STP curves for the RQA 5, W/Rh and W/Ag beam qualities.
FIG. 9. Comparison of the MTFs of the CMOS /Gd$_2$O$_2$S:Eu sensor under investigation and a previously published CMOS /Gd$_2$O$_2$S:Tb sensor under RQA 5 beam quality.
FIG. 10. MTF curve for W/Rh and W/Ag beam qualities.
FIG. 11. Comparison of the MTFs for the CMOS/Gd$_2$O$_2$S:Eu sensor under investigation and a previously published CMOS/Gd$_2$O$_2$S:Tb sensor under W/Rh and RQA M2 beam qualities respectively.
FIG. 12. Comparison of the NNPS curves between the CMOS/Gd$_2$O$_2$S:Eu sensor under investigation and a previously published CMOS/Gd$_2$O$_2$S:Tb sensor under RQA 5 beam quality.
FIG. 13. NNPS curves for W/Rh and W/Ag beam qualities.
FIG. 14. Comparison of the NNPS curves between the CMOS/Gd$_2$O$_2$S:Eu sensor under investigation and a previously published CMOS/Gd$_2$O$_2$S:Tb sensor under W/Rh and RQA M2 beam qualities respectively.
FIG. 15. Comparison of the DQE curves between the CMOS/Gd$_2$O$_2$S:Eu sensor under investigation and a previously published CMOS/Gd$_2$O$_2$S:Tb sensor under RQA 5 beam quality.

FIG. 16. DQE curve for W/Rh and W/Ag beam qualities.

FIG. 17. Comparison of the DQE curves between the CMOS/Gd$_2$O$_2$S:Eu sensor under investigation and a previously published CMOS/Gd$_2$O$_2$S:Tb sensor under W/Rh and RQA M2 beam qualities respectively.

FIG. 18. NEQ curves for RQA 5 beam qualities.

FIG. 19. NEQ curves for W/Ag and W/Rh beam qualities.

FIG. 20. Comparison of the NEQ curves between the CMOS/Gd$_2$O$_2$S:Eu sensor under investigation and a previously published CMOS/Gd$_2$O$_2$S:Tb sensor under W/Rh and RQA M2 beam qualities respectively.

FIG. 21. Image obtained from the CMOS/Gd$_2$O$_2$S:Eu combination at 74 KVp with 21 mm Al filtration.
REFERENCES


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