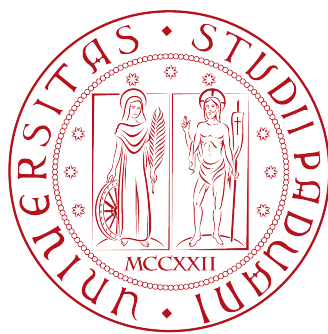




UNIVERSITÀ DEGLI STUDI DI PADOVA

Physics and Astronomy Department "Galileo Galilei"

Ph.D. course in Physics

Doctoral Thesis

X-ray monolithic detectors
for industrial CT scanning

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Abstract

Computed Tomography (CT) scanners have become indispensable in various industries due to their ability to perform non-invasive, real-time inspections. These systems utilize X-rays to generate detailed 3D reconstructions of scanned objects, enabling precise measurements and defects detection. Unlike traditional inspection methods that often require disassembly or destructive testing, CT scanners preserve the integrity of products, making them invaluable in fields such as manufacturing, food safety, wood processing, and even the preservation of cultural heritage. In particular, the context of this thesis is the *Food and Log Computed Tomography* (FLECT) project whose aim is to improve the performance of current industrial CT technology.

CT scanner systems and reconstruction algorithms are constantly evolving to improve efficiency, cost effectiveness, and market competitiveness. In particular, improvements in X-ray detectors are of prime importance, as detection efficiency and resolution ultimately define the CT system overall reconstruction performances. Traditional detection systems, which rely on scintillators and photodiodes, are constrained by slow signal acquisition, high-energy requirements, and susceptibility to signal distortion caused by photon pile-up. Furthermore, these systems often face challenges in maintaining cost-effectiveness while delivering high-resolution results, especially in competitive markets. In response to these challenges, the FLECT project, in collaboration with MiCROTEC company and the INFN ARCADIA initiative, seeks to integrate advanced Depleted Monolithic Active Pixel Sensors (D-MAPS) into industrial CT scanners. These sensors, originally developed for High-Energy Physics (HEP) experiments, offer low power consumption, fast readout speed and high hit rates capability that would enable the use of lower intensity X-ray source. Consequently, the X-ray intensity reduction leads to a lower amount of radiation interacting with the products scanned, as well as an increase in the productivity and competitiveness while reducing the waste at the same time.

This thesis initially gives an historical review of both the production and the detection mechanisms of X-ray particles, quantitatively modeling the phenomenon occurring when the inner structure of the substance is modified by the deposited energy. Then the sensors currently utilized in HEP are discussed, focusing on the chip designed by the INFN ARCADIA collaboration, which aims to develop a D-MAPS tailored towards large-area and low-power tracking applications. The GEANT4 simulation toolkit has been used to determine the feasibility of adapting such D-MAPS technology to industrial X-ray CT scanning applications. Moreover, scans of objects with simple geometries are simulated as in a tomography apparatus, and the image reconstruction is provided, showing the capability of distinguish materials of slightly different absorption coefficients. The risk for industrial employees in working with ionizing radiations are then considered, and geometries and materials that could be used as external shield of the apparatus to reduce the external dose rate to a safe level are explored. Finally, a prototype of the ARCADIA sensor is tested in two different environments: the Padova University laboratory with its X-ray tube source and the MiCROTEC company building with industrial CT scanner. The successful testing of D-MAPS sensors into existing MiCROTEC CT systems demonstrates their compatibility and potential for future industrial applications. By adapting this technology, the project envisions a new generation of faster, safer, and more cost-effective industrial CT scanners.

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Introduction

FLECT, *Food and Log Enhanced Computed Tomography*, is a project dedicated to improve the performance of industrial X-ray Computed Tomography (CT) scanners. CT scanners have a broad range of applications, from medical diagnosis and therapy to industrial uses.

In industrial settings, X-ray radiography remains the primary inspection method across a wide range of sectors, offering the ability to detect substances or components in complex products on a production line without the need to break them open for sample testing. This ability marks significant progress in industrial processes. Moreover, CT technology, an evolution of traditional radiography, enables the reconstruction of 3D images of objects, providing detailed information about the depth and dimensional structure of the materials within them. It serves as a powerful, non-destructive tool for identifying product defects, offering precise measurements with high dimensional resolution.

A typical CT system is composed of an X-ray source, a detection area, and a mechanical structure that allows synchronous rotation of the two around a moving sample. In some systems, the roles are reversed: the X-ray source and detection region remain stationary while the sample rotates and translates. In both configurations, the principle is the same: multiple views of cross-sectional slices of the sample are recorded, and a back-projection algorithm is applied to reconstruct the 3D volume. Therefore, in CT imaging modalities, the detector plays an essential role in determining image quality, making it the most critical element of a CT scanner for image formation, with a fundamental impact on the final 3D reconstruction.

The vast majority of CT scanners developed in recent years rely on indirect conversion. In this process, X-rays interact with a scintillating material to produce visible light, which is then converted into an electrical signal using an analogue photodiode or photomultiplier array. The analogue signal is then acquired, digitized, and parsed. A significant drawback of this approach is the time required to generate the final output image, due to the multiple steps that the signal undergoes during the acquisition, and the general system complexity, which encompasses several components and subsystems.

Within this context, the FLECT project aims to develop and refine a prototype Depleted Monolithic Active Pixel Sensor (D-MAPS) specifically designed for industrial X-ray CT applications. A fully operational D-MAPS characterized by low power consumption, high readout speed, and the capability to handle high hit rates would in fact enable using lower-intensity X-ray source, and consequently minimize radiation exposure to scanned products, enhance productivity and competitiveness, and contribute to waste reduction.

In recent years, the ARCADIA (Advanced Readout CMOS Architectures with Depleted Integrated Sensor Arrays) INFN collaboration developed a state-of-the-art pixel D-MAP sensor with integrated read-out electronics, specifically designed to cover large surfaces for High Energy Physics (HEP) and Space applications. One key advantage of these new sensors is their ability to easily scale the event-driven readout architecture, enabling coverage of detection areas up to several square centimeters [1]. This technology utilizes a high-resistivity substrate, allowing the creation of a fully depleted detection volume. In this setup, when a particle generates free charges, they quickly drift toward the collection electrode, optimizing detector efficiency and ensuring long-

term stability during use.

Comparison of this sensor type with other detectors highlights its significant advantages. In particular, a D-MAPS generates a digital signal directly from each pixel, eliminating the need for analog-to-digital conversion and further reducing signal acquisition time. Additionally, the detector design minimizes the pile-up effect: when a particle hits a pixel, the signal is promptly transferred to the end of the pixel column, ensuring efficient handling of high hit rates. This design also overcomes the inherent limitations of charge collection by diffusion found in standard MAPS designs, and improves the detector response speed.

As a member of the ARCADIA INFN group, I evaluate the sensor response to an X-ray spectrum up to 200 keV using a Monte Carlo simulation toolkit. The simulations are aimed at assessing the physical and technical challenges associated with this type of detector and determining how to address them to meet the needs of the industrial X-ray imaging sector. To do this, I collaborated with MiCROTEC, a global leader in non-invasive scanning technologies for the food and sawmill industry. This partnership seeks to optimize both the technical performance and cost-efficiency of these complex and expensive imaging systems.

During the year spent inside the MiCROTEC company, I mostly focused on the CT scanner used in the food industry, as the features of this machine made it the best candidate for implementing the new solid-state sensors that I propose. Thus, I characterize the scanner apparatus starting from the estimation of the dose acquired by the detector inside the gantry, considering the X-rays energy range used by the company, and also analyzed the ambient dose that exits from the gantry and how to block it. Inside the gantry, the X-ray tube continuously rotates with the generation of photons that interact with the conveyor belt and the products on it, scattering everywhere following the emission angular distribution according to the Klein-Nishina formula [2]. This is the reason why a part of the initial X-rays with energies higher than 20 keV exits from the gantry, posing a serious health threat. Therefore, to meet human safety dose limits, it is necessary to cover the gantry with an external shield to block these scattered X-rays.

For the estimation of external dose rate, the scanner system has been precisely recreated in GEANT using geometries directly imported from the original CAD drawings. The first interesting finding has been about the effects of Food Contact Materials (FCMs). FCMs are mandatory in a food industrial scanner for every part which may get in contact with the organic material; however, their interaction with X-ray photons produces a large amount of scattered particles due to their low density. A study of these materials shows the increase of scattering as a function of the material thickness. To reduce the amount of back-scattered X-rays, stacking of FCMs with substances that cannot be placed in direct contact with food has been evaluated. The optimum configuration resulted to be a sandwich of few millimeters of lead (not-FCM) and a few millimeters of stainless steel (FCM). Moreover, the geometry plays an important role in blocking stray X-rays. Two main elements have been considered: the variation of the conveyor belt direction and the insertion of internal layers to stop the particles. Both of these configurations have been examined with the inclusion of the so-called diaphragms.

The industrial partner requests a competitive solution for a detection system. Con-

sequently, my research must assess both the performance and cost of all detector types currently utilized in the industry, as well as the potential performance and projected costs of detectors that so far remain confined to research and academia. The simulations, refined during my time within the company, aim to answer these questions before starting to test the pixel sensor prototype developed by the INFN ARCADIA project with real-world industrial CT systems.

The initial image acquisition, conducted without any sample in a laboratory room, demonstrates the ability of the ARCADIA sensor to detect X-ray signals at varying tube voltages (photon energies) and tube current (photon flux) levels. The results confirm a linear correlation between the X-ray rate and the current, as expected. Several small objects made of different materials have been scanned by rotating them with a stepper motor, to simulate a real CT setup. Custom software has been developed to process and analyze the acquired data. Additionally, a separate package segments the scanned objects into slices, generates sinograms for each slice, and reconstructs both the 2D cross sections and the full 3D volume by stacking all the cross sections together.

Finally, I tested the ARCADIA sensor prototype under the X-ray flux of the Mi-CROTEC apparatus. The characterization of the sensor with a variable X-ray tube source voltage provides the possibility to evaluate the Signal-to-Noise Ratio (SNR) of the detection system, which is directly comparable with the once in use.

It is also important to consider the broader market potential of industrial CT applications. Although the wood and food processing industries are the original beneficiaries of the FLECT project, the development of fast and efficient marketable sensors has the potential to impact a variety of other fields. These include airport customs scanners, cultural heritage investigation tools, and even medical imaging systems.

This Thesis is structured as follows:

Chapter 1 gives an historical review of both the production and the detection mechanisms of X-ray particles. The interaction processes with matter, as a function of the X-ray energy, are then evaluated, with the quantification of the phenomenon occurring when the inner structure of the substance is modified by the deposited energy.

Chapter 2 provides an analysis of sensors currently utilized in HEP. This review establishes the feasibility of adapting D-MAPS technology to industrial X-ray CT scanning applications.

In **Chapter 3**, projections of simple geometries are sampled using the GEANT4 (*GEometry AND Tracking 4*) simulation toolkit [3][4][5] and a reconstruction algorithm is developed to provide the 3D image. Cylindrical geometries are taken into account because of their simple symmetry around the z-axis. The shapes of test samples of different materials and radii are considered to verify the detector capacity to distinguish them. A complete CT is provided by rotating the sample and reconstructing the image using a Java-based software called *Fiji* [6]. With the GEANT4 toolkit [3][4][5], I simulate the silicon pixel sensors to evaluate the ability to

distinguish two materials of slightly different absorption coefficients, such as water and glass. The results demonstrate the ability to distinguish up to 2 mm glass in water, a typical size of glass fragments that can be found in a broken jar.

Chapter 4 describes how to evaluate the dose produced by the CT scanner system developed by the MiCROTEC company for the food industry. The evaluation of the ambient dose rate from the simulations has been defined and comparisons between measurements from the real configuration set-up and the model are verified, geometries and materials that could be used as external shield of the gantry to reduce the ambient dose rate to a safety value are explored.

Chapter 5 reports on the results of experiments with the prototype. In particular it shows a linear dependence of the X-rays rate with respect to the current, as expected. The CT scanner is imitated in the laboratory by using a stepper motor to rotate the sample considered.

Chapter 6 summarizes the key results achieved during my Ph.D. research, highlighting the innovative ideas developed to enhance the competitiveness of the scanner system while reducing its emission dose. In addition, it includes a direct comparison between D-MAPS signal acquisition and the detection system currently employed by the company.

Finally in **Chapter 7** I conclude by addressing the multifaceted outcomes of the project, the challenges, and the broader implications for industries reliant on precise, non-invasive imaging solutions.

1 Electromagnetic Radiation

In this chapter we will briefly discuss *Radiation* but then quickly focus on the production and properties of X-rays and their modes of interaction with matter. Indeed Medical and Industrial Computed Tomography (CT) is an imaging technique that uses X-rays to produce images in a transmission modality: *transmission* refers to the passage of X-rays through an object (a body or an industrial product), with selective attenuation caused by the interaction of X-rays with different types of tissues or materials [7]; the variations in attenuation allow for a detailed visualization of internal structures.

1.1 *Radiation*

For practical purposes, radiation is generally categorized into three main groups:

- charged particles: electrons, positrons, protons, alphas, heavy ions ($A > 4$);
- neutrons;
- photons, the quanta of electromagnetic radiation ¹: gammas (γ), X-rays.

Radiation can further be classified as ionizing or non-ionizing. Ionization occurs when the radiation particles or photons have enough energy to remove tightly bound electrons from atoms, creating ions.

More specifically, organic and inorganic materials are generally composed by molecules with atoms bounded by the inter-atomic bonds, which are characterized by the electrons being shared between two or more atoms. Therefore, the interaction of a radiation with one orbital electron and the subsequent ionization leads to a break up of the chemical bonds between atoms and molecules.

This kind of interaction occurs in human tissues exposed to ionizing radiation during medical treatments. In such cases, the disruption of chemical bonds within living cells can be beneficial, as it helps inhibit cellular functions, particularly in the context of cancer treatment. By targeting and damaging the ability of cells to replicate, ionizing radiation aims to kill the malignant cells effectively.

Charged particles ionize directly (used for example in proton or hadron therapies ²), while neutrons and photons can only do it indirectly. Photons interact with charged particles, in particular bound atomic electrons: if a photon is energetic enough, it can create free an electron or even create an electron-positron pair. Neutrons instead interact with atomic nuclei to produce secondary radiation or charged particles that subsequently cause ionization.

¹The classical description of low-energy EM radiation involves a combination of perpendicular electric and magnetic waves traveling together at the speed of light in a vacuum characterized by parameters such as amplitude, wavelength, frequency. In *Quantum Field Theory*, a "particle" is quite different from a classical particle, namely a small localized object with a definite position and velocity; they are instead considered to be quanta of excitations of the underlying field. In particular, in *Quantum Electrodynamics* photons are discrete quanta of the EM field with no rest mass or electric charge. On the other hand, particles such as electrons, muons, protons, and neutrons, are quanta of matter fields distinct from the EM field.

²Proton and Hadron oncological therapies use protons and carbon ions to irradiate cancer cells: these charged particles ionize the most at a specific point inside the body called the Bragg peak. This type of therapy targets tumors more precisely, minimizing damage to surrounding healthy tissue.

1.2 Photons

The photon is a quantum of excitation of the EM field and as such has wave-particle duality: photons exhibit interference and diffraction patterns, consistent with wave behavior, but can interact with matter as individual particles, with energy ($E = h\nu = \hbar\omega$) and momentum given by the De Broglie-Einstein relations ($\vec{p} = \hbar\vec{k}$)³, such as in the Photoelectric and Compton effects, and in Pair Production (Figure 1.1) which may occur if the energy of the incoming photon is sufficiently high. In the Photoelectric and Compton effects, an atomic electron absorbs all or part of the energy of the photon, it is consequently freed and then proceeds to further ionize; in Pair Production a free electron-positron ionizing pair is created. These effects will be further discussed in Section 1.3.2.

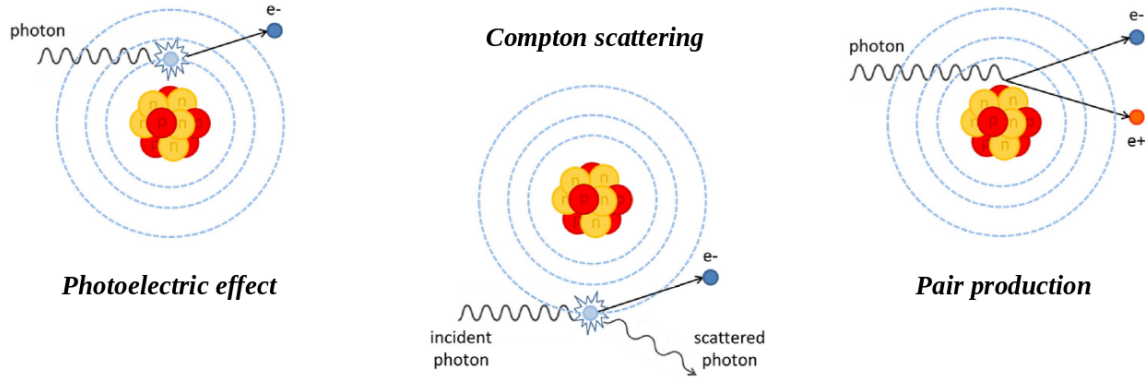


Figure 1.1: The particle behavior of EM radiation is exemplified in the ionizing processes: Photoelectric effect, Compton scattering and Pair production.

1.3 X-rays, γ -rays

Although the interaction of photons with matter depends solely on their energy, photons are not typically distinguished by energy alone. For practical purposes, they are classified instead according to their production process:

- photons emitted due to the deceleration of free charged particles (a process known as *Bremsstrahlung*, or braking radiation) are classified as X-rays. X-rays are also photons generated during the rearrangement of bound electrons between atomic shells in excited atomic states;
- photons produced during the radioactive decay of radionuclides, as a result of nucleons rearrangement within the atomic nucleus, are classified as γ -rays.

This thesis focuses on industrial Computed Tomography (CT) systems, specifically considering X-rays within an energy range of up to 200 keV.

³ $\hbar \approx 1.055 \times 10^{-34}$ J-s; the module of propagation vector $\vec{k}$ is $|\vec{k}| = \frac{2\pi}{\lambda}$

1.3.1 X-ray production

Bremsstrahlung X-rays are emitted when electrically charged particles undergo acceleration, such as when decelerating during stopping (breaking) or when curving (centripetal acceleration). A complete understanding of this process requires a quantum mechanical approach, as photons with energies comparable to that of the emitting particle can be produced. However, in certain regimes, a classical treatment provides an adequate description, with quantum effects appearing as corrections.

In classical physics ($v \ll c$), the intensity of the emitted radiation is proportional to the square of the particle's charge and acceleration. Considering a charge particle of charge ze and mass m moving through a material with atomic number Z , the coulomb force between the particle and an atomic nucleus of the material is approximately

$$F \sim zZe^2/r^2$$

where r is the distance between the two charges. The acceleration of the incident charged particle is

$$a = F/m \sim zZe^2/m$$

so the intensity of the breaking radiation is

$$I \propto a^2 \sim z^2 Z^2 e^4 / m^2$$

One concludes:

- comparing two equally charged particles traveling through the same material, the lighter particle will emit more bremsstrahlung radiation than the heavier one (other things being equal);
- more bremsstrahlung radiation is emitted when a given particle travels through a material with high atomic number Z compared to a material with a lower atomic number.

Since the electron is the lightest charged particle, it emits the most bremsstrahlung radiation under these conditions. This makes electrons a preferred choice for generating X-ray beams in various applications.

At a **synchrotron** accelerator, relativistic charged particles are guided by magnets to move continuously in a storage ring with somewhat circular trajectories (Figure 1.2); particles are given energy (*accelerated*) and bunched by radio-frequency accelerating cavities and are extracted at rates of up to 50 Hz [8]. Bremsstrahlung radiation is produced when the particles paths are bent by the magnetic fields and, in this context, is referred to as *synchrotron radiation*.

Although the emission of electromagnetic waves was theoretically described using Maxwell's equations in 1873, and the formula for energy loss of particles with relativistic velocities in circular motion was developed by Liénard in 1898, the first visual observation of *synchrotron radiation* occurred in 1947. This was achieved using the General Electric 70 MeV electron synchrotron in Schenectady, New York. The key advantage of *synchrotron radiation* lies in its spectral brightness and brilliance, which represent the flux density and the flux per unit area, solid angle, and spectral bandwidth. These characteristics make synchrotron radiation highly desirable for a wide range of sophisticated applications in fields such as Material Science, Biology, Chemistry, Environmental Science, and very Advance Medical Imaging such as phase-contrast imaging.

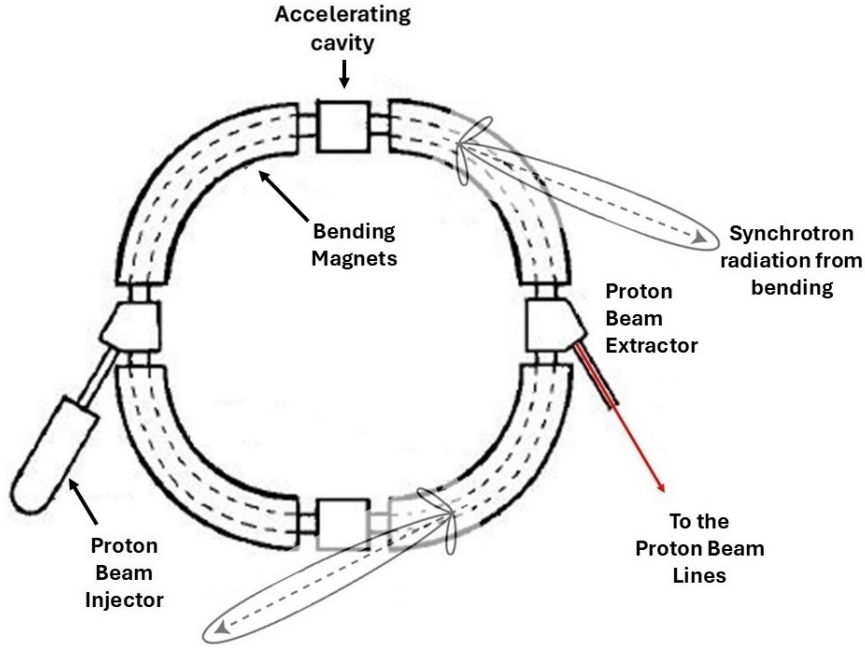


Figure 1.2: Schematic of a proton synchrotron accelerator. Synchrotron radiation is produced when the particle trajectories are bent by magnetic fields; overlaid at two points are typical radiation patterns.

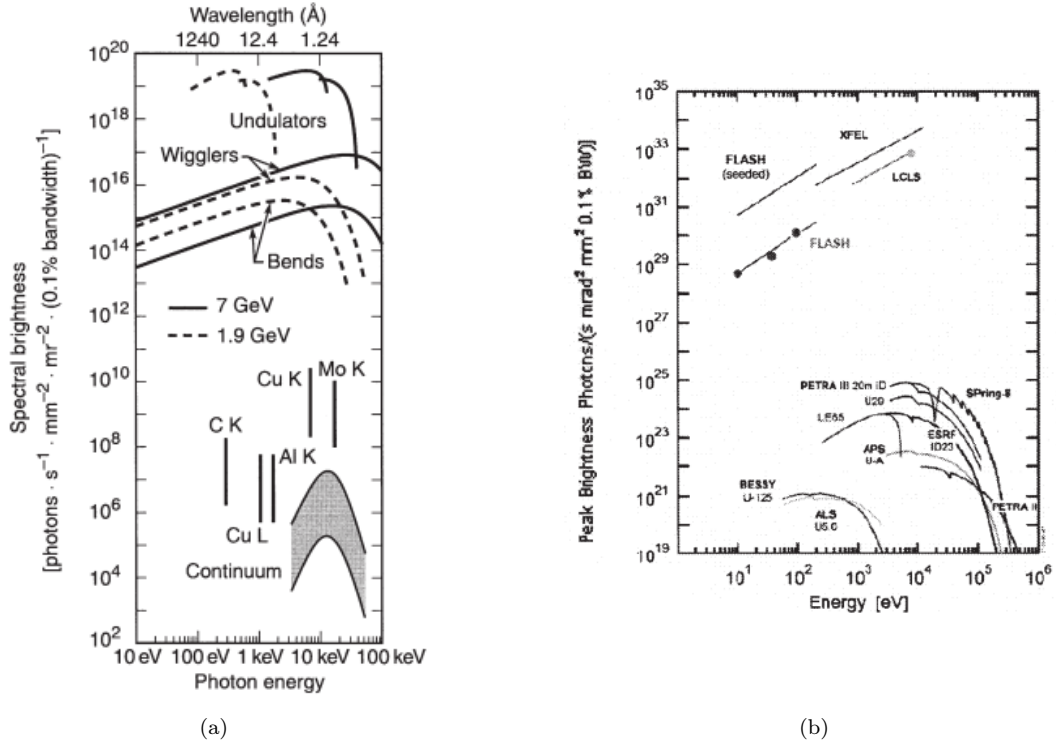


Figure 1.3: (a) Spectral brightness for several synchrotron radiation sources and conventional X-ray tube sources. The indicated two-order of magnitude ranges (the gray area) show the approximate variation that can be expected among stationary-anode tubes (lower end of range), rotating-anode tubes (middle) and rotating-anode tubes with micro-focusing (upper end of range). (b) FEL beams have very short pulse duration and full spatial coherence, with peak brightness many orders of magnitude higher than synchrotron facilities. Figures and captions taken from [8].

Over the following forty years, synchrotrons are redesigned to improve their performance. Devices such as *undulators*, which enhance intrinsic brightness, and *wigglers*, which shift the wavelength spectrum, are introduced to achieve this goal. The most significant advancement in synchrotron radiation sources has been the development of the Free Electron Laser (FEL) [9]. FEL technology enables a dramatic increase in the brightness of emitted radiation by several orders of magnitude, as illustrated in Figure 1.3b. It also provides ultra-short pulse lengths and spatially coherent beams. The wavelengths achieved by these pulse beams are on the order of a few angstroms, corresponding to X-rays with energies below 10 keV. However, X-rays in this energy range are entirely absorbed when passing through a patient or industrial product, rendering them unsuitable for such applications. Moreover, the high cost of constructing and operating a synchrotron further limits its feasibility as a source of X-rays in most medical or industrial environments.

For these reasons the X-ray tube remains the staple of practical applications as it does not suffer from the above mentioned limitations. The first stable X-ray tube, developed in 1913 by American engineer W. D. Coolidge, is named after its inventor. The *Coolidge X-ray tube* is notable for its high vacuum design and the use of a heated filament as the electron source. Unlike earlier cold cathode gas discharge tubes, which are unreliable in terms of both radiation hardness (measured in kV) and intensity (measured in mA), the *Coolidge tube* contains only a minimal amount of gas, which does not participate in the production of X-rays [10]. This innovation greatly improved the reproducibility and stability of X-ray generation, making it a critical advancement in X-ray technology.

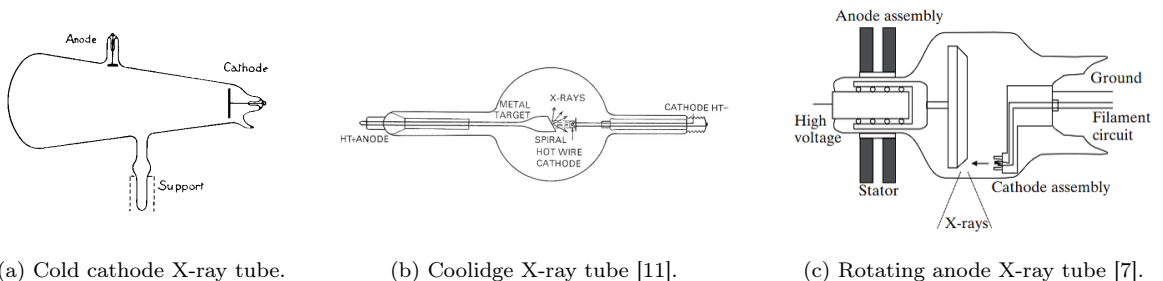


Figure 1.4: X-rays tubes used in the production of X-rays for radiography. (a) Type of gas filled tube with which Röntgen discovered X-rays. The cathode rays impinged on the broad end of the tube. This tube geometry was widely used for radiographs even if the large angular distribution of the emitted X-rays provides blurring effects in the image [12]. (b) Standard X-ray tube with near perfect vacuum and heated filament. Thanks to these features, the life of the tube was prolonged and the reproducibility of the results reached. (c) Modern X-ray tube with rotating anode. The rotation prevents melting the anode material thanks to the heat dissipation over a large area of the anode.

The X-ray production process involves heating a tungsten (W) filament to create a cloud of electrons, which are then accelerated toward an anode by applying a voltage ranging from a few kV to hundreds of kV. Thermionic emission occurs when the tungsten filament reaches a temperature between 2300 °C and 2700 °C [13]. To ensure the electron cloud is directed to the anode without dispersion, the geometry of the electrodes must be carefully designed to align with the shape of the filament. This design minimizes electron dispersion and helps define the form of the electron beam. A finer focus of the electron beam can be achieved by either reducing the focal spot size or increasing the voltage between the cathode head and the filament.

Once the electrons are accelerated by the electric field, they interact with the anode material. These interactions can be classified into two main processes: collisional and radiative transfers.

- **Collisional Transfer:** This is the dominant mechanism by which energetic electrons interact with the atomic electrons of the anode material. When an incident electron collides with an atomic electron, it transfers kinetic energy, which may result in the ejection of the atomic electron. If the atom returns to its original state, the excess energy is emitted as heat. The incident electron often transfers only part of its energy and deviates from its initial path undergoing multiple interactions. If the energy transfer is significant, an atomic electron is ejected from the outer part of the atomic shell with high kinetic energy. This ejected electron, known as a *delta ray*, will then create additional ionization paths, as shown in Figure 1.5a. On the other hand, an electron ejected from an inner atomic shell leaves a vacancy that must be filled by an electron from an outer shell. The energy lost by the electron filling the vacancy is emitted as EM radiation, known as *characteristic X-ray* (Figure 1.5b). The energy of this radiation is equal to the difference in binding energy between the two atomic shells.
- **Radiative Transfer (bremsstrahlung):** This process occurs when electrons interact with the electric fields of nuclei within the anode material (Figure 1.5c). The positive charge of the nucleus attracts the electrons, causing a deviation in their path and deceleration. This sudden loss of kinetic energy results in the emission of EM radiation. The outcome of the interaction depends on the impact parameter and the initial energy of the incident electron. If the electron approaches or directly collides with the nucleus, it may be annihilated, emitting an X-ray with energy equivalent to that of the electron.

Bremsstrahlung is the dominant mechanism of X-ray production in X-ray tubes, generating a spectrum of photons with energies ranging from zero to the maximum kinetic energy of the electrons, determined by the applied voltage. The resulting energy distribution follows an inverse linear relationship between the number of X-rays produced and their energy. However, this spectrum is inevitably modified by the glass or metal window of the X-ray tube, even by additional filters installed by the user, which attenuate low-energy particles. This filtering produces an absence of photons at very low energy values, as shown in Figure 1.6.

The average rate of energy loss through the collisional mechanism is well described by the stopping power formula, as presented in [15]. This loss primarily depends on $1/\beta^2$, with $\beta = v/c$ representing the electron velocity, when the incident energy is below 100 keV. In contrast, energy loss via radiative processes is strongly influenced by the atomic number and is directly proportional to the incident electron energy. Consequently, the probability of *bremsstrahlung* emission is very low in the energy range of a few keV.

The main requirements for the anode material include a high melting point, excellent thermal conductivity, high radiation yield, and low vapor pressure [10]. Therefore, the optimal choice is a tungsten-rhenium alloy, which resists cracking under thermo-

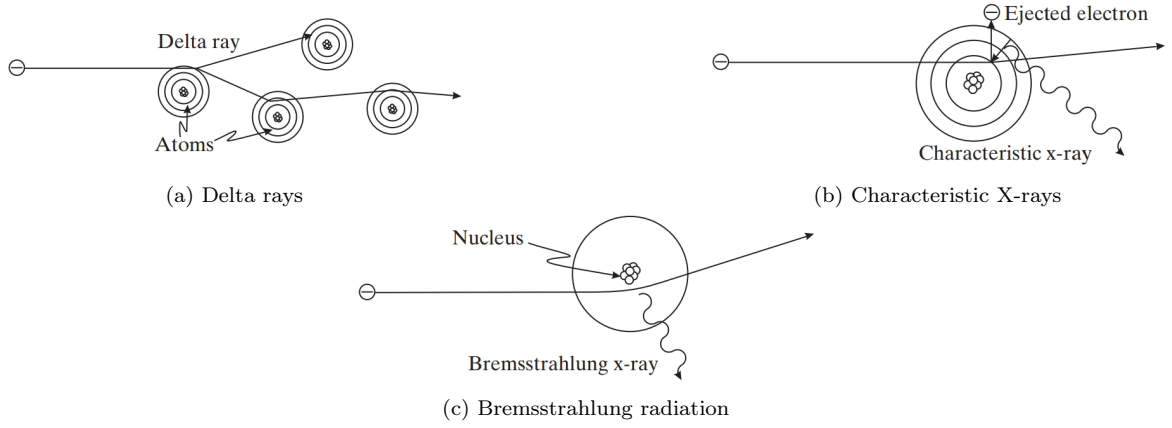


Figure 1.5: Interaction of energetic electrons with matter. (a) An electron interact via a collisional transfer with an external shell atomic electron. The path of the initial electron is deviated and the atomic electron acquires enough energy to be ejected from the atom, becoming a *delta ray*. (b) The impinging electron collides with an inner shell atomic electron, that is ejected from the atom. The vacancy is filled by an outer shell electron with the emission of an EM radiation called *characteristic X-ray*. (c) The fast and high energy electron is attracted by the nucleus of the material atom causing a bend and an immediately deceleration with the emission of EM radiation in an energy range between zero and the maximum energy value of the impinging electron. Figures taken from [7].

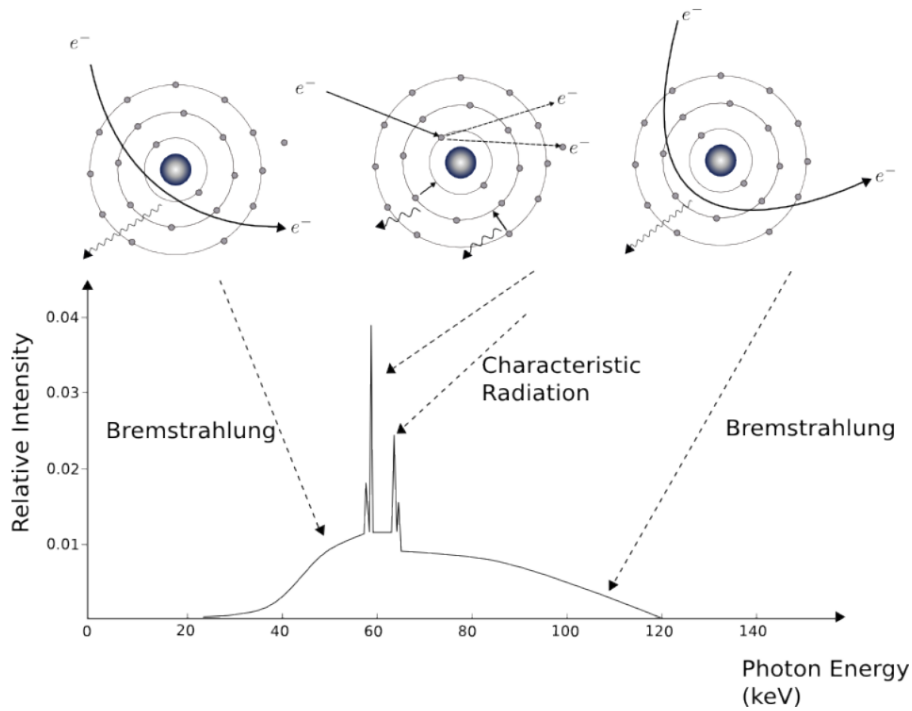


Figure 1.6: Example of an X-ray spectrum from an X-ray tube with tungsten rotating anode. The continuous part of the spectrum corresponds to the *bremsstrahlung* production mechanism, while the peaks represents the K_{α} and K_{β} *characteristic X-rays*. Image taken from [14]

mechanical stress and prolongs the lifespan of the tube, despite the efficiency of X-ray production being less than 1 %, as illustrated in Figure 1.7. Another candidate material for emitting radiation in the energy range of 25 keV-30 keV, with a primary peak at 17.4 keV, as used in mammography, is molybdenum.

The subsequent interactions of electrons with the anode material, the resulting en-

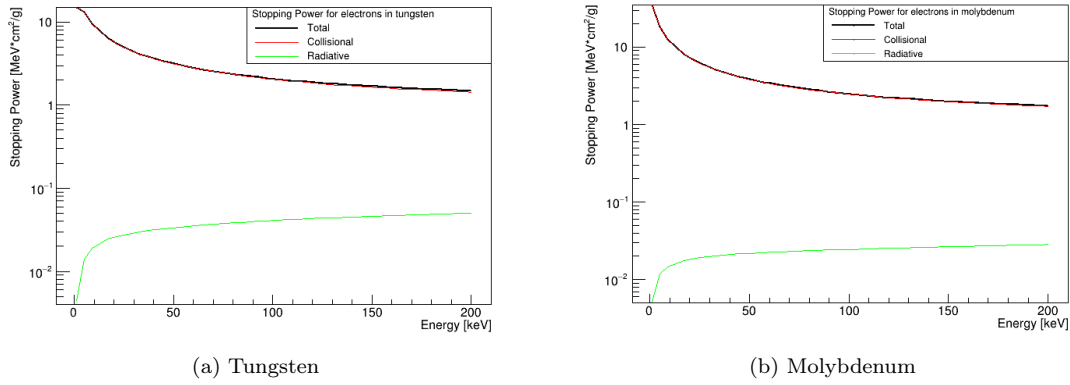


Figure 1.7: Stopping power of electrons in (a) Tungsten and (b) Molybdenum. Green and red curves represent the radiative and collisional stopping power while black curve is the sum of the two. Data taken from *ESTAR* computer program [16].

ergy deposition, and the continuous impact on the same spot significantly increase the likelihood of the anode melting, particularly at the focal spot. This leads to a reduction in the life span of the tube. To address this issue, a motorized rotating anode was introduced. In this design, the anode material is shaped into a disk that is set into rotation by applying an alternating current. The first successful commercial application of an X-ray tube with a rotating anode was developed in 1929 for clinical use by Philips. This innovation forms the basis of modern X-ray tubes, with enhanced heat dissipation that increases proportionally to the fourth power of the absolute temperature (Figure 1.4c).

1.3.2 X-ray interaction with matter

X-rays interact with matter in a manner fundamentally different from the interactions of charged particles. Photons, being neutral particles, do not experience electric fields of charged particles, such as the electrons and nuclei of the atoms of a material, and can only interact via collisions. Consequently, the probability of X-ray interaction is lower compared to that of charged particles.

In atomic, nuclear, and particle physics, the probability of an interaction is expressed as a cross-section (in area units): it quantifies the probability of an interaction with how "big" a target appears to incoming particles; larger cross-sections indicate a higher likelihood of interaction. A common unit of cross-sections is the barn (*b*) and its multiples or sub-multiples:

$$1 \text{ } b = 10^{-24} \text{ } cm^2$$

Cross-sections depend on the type of particles involved, on the interaction, and the energy of the incoming particles. In our case we are dealing with a beam of photons that can interact in various ways with the electrons and nuclei of a target and to each we associate cross-sections.

Photons primarily interact with the atomic electrons of matter through processes such as Rayleigh scattering, Photoelectric effect, Compton scattering, and pair production. There is also the photo-nuclear reaction, where a highly energetic photon interacts with an atomic nucleus. The cross-section of each interaction mechanism depends on the energy of the photon and is summarized in Figure 1.8, which compares these interactions for two different target materials: carbon and lead.

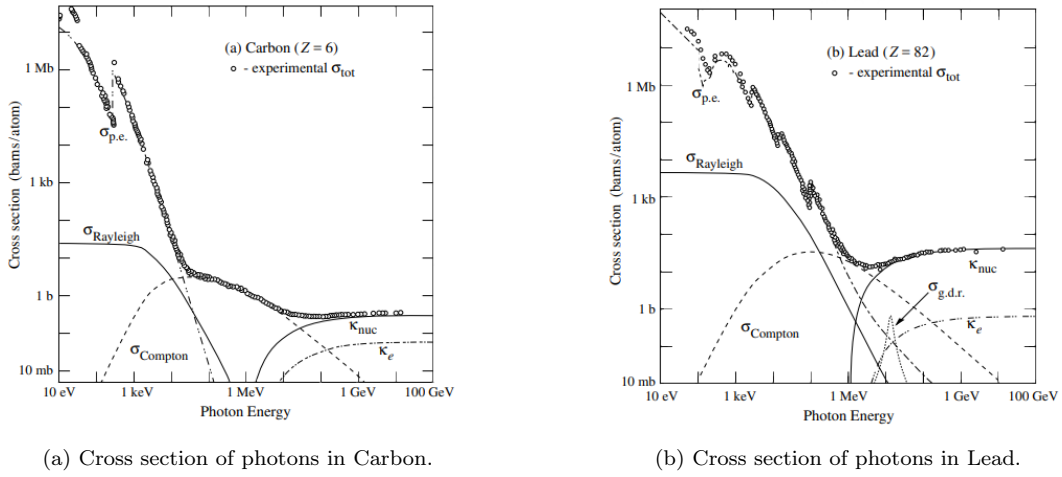


Figure 1.8: Total photon cross section (in units of $1\text{b} = 10^{-28}\text{m}^2$) in (a) light element Carbon and (b) heavy element Lead, as a function of photon energy, with the contributions of the processes listed in the text. These two materials are selected to demonstrate the dependence of the different interaction mechanisms as a function of the material density or atomic number (Z_t). The contributions to the total cross section are: photoelectric effect ($\sigma_{p.e.}$), Rayleigh scattering ($\sigma_{Rayleigh}$), Compton scattering ($\sigma_{Compton}$), pair production from nuclear field (κ_{nuc}), pair production from electron field (κ_e) and photonuclear interaction ($\sigma_{g.d.r.}$). Images taken from [15].

The total cross-section, the sum of all possible cross-sections, at a given photon beam energy

$$\sigma_{tot} = \sigma_{p.e.} + \sigma_{Compton} + \kappa_{nuc} + \kappa_e + \sigma_{Rayleigh} + \sigma_{g.d.f.} \quad (1.1)$$

provides the basis for an estimation of the survival probability of a photon beam after passing through matter. The total cross-section is directly related to the photon mass attenuation length (λ), also known as the mean free path, which depends on the material density (ρ) and the linear attenuation coefficient (μ) at a specific photon energy (E). Therefore, the linear attenuation coefficient is directly proportional to the total cross-section:

$$\mu = \frac{\rho N_A}{A_t} \sigma_{tot} \quad (1.2)$$

$$\lambda = \left(\frac{\rho}{\mu} \right)^{-1}$$

with ρ in units of kg m^{-3} , N_A the Avogadro number (mol^{-1}) and A_t the mass atomic number of the target material.

The attenuation of a beam of N_0 initial photons interacting with matter follows an exponential decay curve. This relationship arises from the direct proportionality between the number of photons lost within the target (dN) and both the number of photons remaining after interaction (N) and the target thickness (dx). Equation 1.3 demonstrates that this proportionality is governed by the linear attenuation coefficient (μ). Integrating this equation yields the number of photons that survive (N) after passing through the material:

$$dN = -\mu N dx \quad (1.3)$$

$$N = N_0 e^{-\mu x}$$

Equation 1.3 applies only to the case of a monoenergetic, narrow photon beam incident on a thin layer of homogeneous material. If the target is not homogeneous, the linear attenuation coefficient becomes position-dependent ($\mu(x)$). Similarly, for a polyenergetic photon beam, the energy dependence must also be included, resulting in a term expressed as $\mu(x; E)$.

The photon interaction mechanisms mentioned above are now detailed, with a focus on the energy ranges at which they are most likely to occur. This analysis helps determine their relevance for applications in medical and industrial tomography.

RAYLEIGH SCATTERING

It is a coherent scattering process that predominantly occurs for photons with energies below a few keV. During this interaction, the photon collides with an outer-shell atomic electron, which does not gain enough energy to be ejected from the atom but instead begins to oscillate. This oscillation leads to the emission of electromagnetic radiation with the same wavelength as the incident photon but in a different direction. Since no energy is transferred to kinetic energy, ionization does not occur. The Rayleigh cross-section is proportional to the square of the target atomic number (Z_t^2). Although Rayleigh scattering contributes to the total photon cross-section for X-rays used in tomographic scanners, its significance is generally surpassed by other interaction mechanisms, particularly the photoelectric effect.

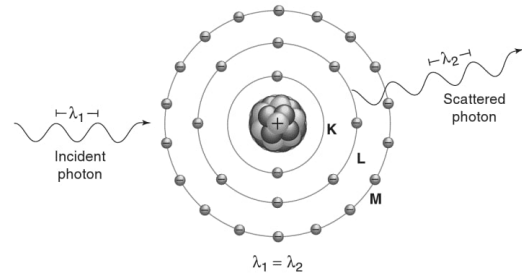


Figure 1.9: Rayleigh scattering [17]

PHOTOELECTRIC EFFECT

The trend of the experimental data in the first part of the cross-section, shown in both plots of Figure 1.8, closely follows the photoelectric cross-section curves. The probability of this process occurring is proportional to the fourth power of the target atomic number ($\propto Z_t^4$) and inversely proportional to the third power of the incident photon energy ($\propto 1/(h\nu)^3$). This is why the Rayleigh scattering mechanism is not considered in the description of X-ray interactions in medical and industrial fields. The Photoelectric effect occurs when an X-ray strikes an inner-shell atomic electron, and its energy is entirely absorbed by the electron. The kinetic energy gained by the electron is sufficient to eject it from the atom, and it is equal to the difference between the initial energy of the X-ray ($h\nu$) and the binding energy of the electron orbit (B). Thus, the kinetic energy (T) of the ejected *photoelectron* is given by:

$$T = h\nu - B \quad (1.4)$$

The vacancy left by the emission of the *photoelectron* can be filled by an outer electron from the atom, resulting in the production of a *characteristic X-ray*, as

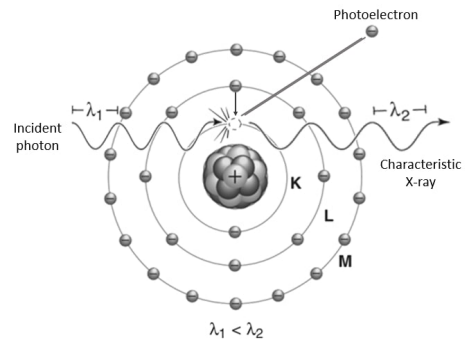


Figure 1.10: Photoelectric effect [17]

shown in Figure 1.10.

COMPTON SCATTERING

Considering the total photon cross-section in a light element, such as carbon, shown in Figure 1.8a, the central part is dominated by the Compton scattering process. This mechanism involves the inelastic scattering of a photon with a free electron at rest and can be accurately described using the relativistic expressions for energy and momentum conservation in a two-body system. In particular the photon has a zero rest-mass and $p = E/c$. Generally, the electrons affected by Compton scattering are the valence electrons, as they require less energy to be ejected from the atom. Considering Figure 1.11, let (x, z) represent the plane in which the interaction occurs. The initial four-momenta (p_x, p_y, p_z, E) of the two particles are given by:

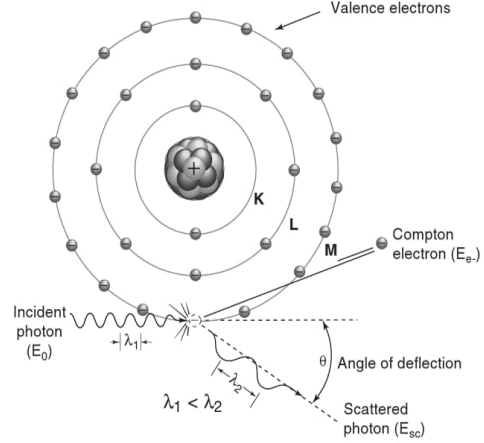


Figure 1.11: Compton scattering [17]

$$\begin{cases} p_\gamma = (0, 0, h\nu_0/c, h\nu_0) \\ p_e = (0, 0, 0, m_e c^2) \end{cases}$$

while the final four-momenta are:

$$\begin{cases} p'_\gamma = (-(h\nu/c)\sin\theta, 0, (h\nu/c)\cos\theta, h\nu) \\ p'_e = (p'_e\sin\phi, 0, p'_e\cos\phi, E'_e) \end{cases}$$

The energy and momentum conservation theorem provides the following equations:

$$\begin{aligned} h\nu_0 + m_e c^2 &= h\nu + E'_e \\ \frac{h\nu_0}{c} &= \frac{h\nu}{c}\cos\theta + p'_e\cos\phi \\ \frac{h\nu}{c}\sin\theta &= p'_e\sin\phi \end{aligned}$$

that adequately remodeled give the results:

$$\begin{cases} \left[\frac{h}{c}(\nu_0 - \nu) + m_e c \right]^2 = |\vec{p}'_e|^2 + m_e^2 c^2 \\ \left(\frac{h\nu_0}{c} - \frac{h\nu}{c}\cos\theta \right)^2 + \left(\frac{h\nu}{c}\sin\theta \right)^2 = |\vec{p}'_e|^2 \end{cases}$$

$$\begin{cases} \frac{h^2}{c^2}(\nu_0^2 + \nu^2 - 2\nu_0\nu\cos\theta) = \frac{h^2}{c^2}(\nu_0^2 + \nu^2 - 2\nu_0\nu) + 2\frac{h}{c}(\nu_0 - \nu)m_e c \\ \nu_0 - \nu = \frac{h}{m_e c^2}\nu_0\nu(1 - \cos\theta) \end{cases}$$

Comparing the two expressions and knowing that the recoil momentum of the scattering electron corresponds to the vector difference between the X-ray momentum before and after the collision [18], the resulting equation for the kinetic energy of the ejected electron (T_e) is:

$$T_e = h\nu_0 - h\nu = \frac{h^2\nu_0^2(1 - \cos\theta)}{m_e c^2(1 + h\nu_0/m_e c^2(1 - \cos\theta))} \quad (1.5)$$

The Compton scattering interaction mechanism primarily influences the photon cross-section in light elements, while it can be neglected for all energies in interactions with heavy elements such as lead. The probability of observing this process is highly dependent on the electron density of the target material ($\propto ED = (Z_t N_A)/W_m$), and generally decreases slowly with increasing photon energy, as described by the *Klein-Nishina* formula [2]. Therefore, in the energy range typically used in medical and industrial applications (up to several hundred keV), the Compton cross-section can be considered nearly constant, depending primarily on the properties of the target material. In this energy range, when comparing the probabilities of Compton scattering and the photoelectric effect, it is clear why Compton scattering is negligible in the photon cross-section for lead.

PAIR PRODUCTION

When photon energy exceeds a few MeV, the cross-section experiences a slight increase due to a new interaction process: *pair production*. The energy threshold for observing this interaction mechanism is twice the electron mass ($m_e = 0.511$ MeV), or 1.02 MeV. When the photon interacts with the nuclear or electric field of the atomic nucleus, its kinetic energy is converted into an electron-positron pair. The excess energy of the photon above the threshold value is transferred to the particles as kinetic energy. In the center-of-mass reference frame, the particle and antiparticle are generated with the same rest energy, moving in the same direction but with opposite velocities.

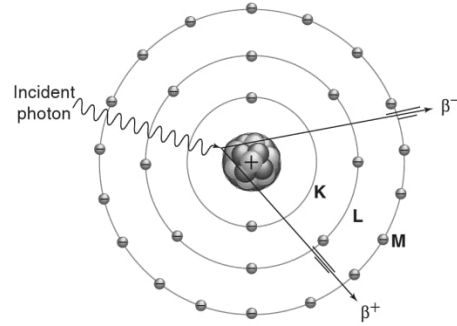


Figure 1.12: Pair production [17]

$$h\nu = 2m_e c^2 + T_{e-} + T_{e+} \quad (1.6)$$

The electron and positron, having kinetic energy greater than zero, can interact with the target material via excitation and ionization. The positron may annihilate with an atomic electron, resulting in the production of two oppositely directed photons, each with an energy of 0.511 MeV. The probability of the pair production interaction process is proportional to the square of the target atomic number ($\propto Z_t^2$), but it can be neglected in this context because the photon energy generated by medical and industrial apparatuses does not exceed 200 keV.

PHOTONUCLEAR REACTION

Another contribution to the total cross-section is the photonuclear reaction. This process occurs when a photon collides with the atomic nucleus and is completely absorbed, resulting in the ejection of a nucleon, either a proton or a neutron. Several MeV of energy are required to observe this phenomenon because the photon must have energy greater than the binding energy of the nucleons, making MeV γ -rays the most suitable candidates. As a result, this mechanism is not considered in the description of X-ray interactions for medical and industrial environments. The kinetic energy of the ejected nucleon is equal to the difference between the incident γ -ray energy and the nucleon's binding energy. The probability of detecting a nucleon

produced by this process is very small and is proportional to the target atomic number (Z_t).

1.3.3 Dose

The interaction of photons with matter releases energy within the material they strike, altering its internal structure. This phenomenon can be quantified using the concept of *dose*. Different methods are used to evaluate the dose, depending on the specific application. For example, the approach differs when analyzing a medical radiograph of a section of the human body versus scanning a commercial product on an industrial production line. In both scenarios, dose estimation begins by considering the amount of energy lost by X-rays within the target, a quantity referred to as *kerma* (K), or *kinetic energy released in matter*. *Kerma* is defined as the ratio of the mean total kinetic energy dE_{ptr} of all charged particles (typically atomic electrons) set in motion per unit mass dm of a material from the incident ionizing radiation (X-rays, gamma, beta, alpha,...) [19]:

$$K = \frac{dE_{ptr}}{dm} \quad (1.7)$$

The SI unit of *kerma* is J kg^{-1} , or *Gray* (Gy), which represents the total potential dose that the incident uncharged particles can impart to the target. However, *kerma* does not account for the possibility for scattered particles to escape from the material, leading to an overestimation of the actual dose within the target, especially when the incident particles possess high energies; i.e. it doesn't directly reflect the total energy absorbed by a material.

To address this limitation, another variable is used: the *absorbed dose* (D). This measures the energy truly absorbed per unit mass of the material due to ionizing radiation. The *absorbed dose* is defined as the ratio of the average energy $d\bar{\epsilon}$ imparted by ionizing radiation to a mass dm of material [19]. Its unit is also the Gy, and it is expressed as:

$$D = \frac{d\bar{\epsilon}}{dm} \quad (1.8)$$

From the definition of *absorbed dose*, it is possible to determine the appropriate dose metric for the specific environment under consideration:

- In the **Medical field**, the key variables are the *dose equivalent* (H) and the *effective dose* (E).

The *dose equivalent* H , expressed in sieverts (Sv), accounts for the biological effects of radiation in the tissue of a given organ. For a given tissue, H is the absorbed dose D multiplied by a dimensionless *radiation weighting factor* W_R that depends on the type of radiation

$$H = D \times W_R \quad (1.9)$$

$W_R = 1$ for X-rays, gammas, and beta particles (electrons), while alpha particles of typical energies have $W_R = 20$ because they are far more biologically damaging than beta particles of the same energy. Neutrons deliver most of their energy in the form of heavy charged ions, so their W_R is considerably greater than unity and varies significantly with energy.

The *effective dose* E is a further refinement: it is the weighted sum of absorbed doses across all specified organs and tissues of the body [20]:

$$E = \sum_T w_T \cdot H_T \quad (1.10)$$

While H_T is the dose equivalent absorbed by a given tissue, the dimensionless *tissue weighting factor* w_T reflects the relative biological effectiveness or radiation sensitivity of the specific tissue or organ. These values are determined by international bodies such as the International Commission on Radiological Protection (ICRP).

By taking into account the type of radiation (e.g., alpha, beta, gamma, X-rays) and the sensitivity of different tissues or organs to radiation, the *effective dose* is used to estimate the overall risk of exposure to ionizing radiation, not just the amount of energy deposited in the body.

- In **Industrial environments**, the relevant variable is the *ambient dose* (H^*). It is a measure of radiation exposure that takes into account the type, energy, and distribution of radiation in the environment. In a radiation field, at a given point in space, the *ambient dose* is defined as the product of the particle fluence at that point (Φ) and the conversion coefficient ($h_{E_{max}}^*$), which relates particle fluence to the maximum effective dose (E_{max}) [19]. The SI unit for H^* is J kg^{-1} , which is also expressed as Sievert (Sv). The *ambient dose equivalent* is a measure that accounts for the type and energy distribution of the radiation field at a point. It is generally calculated in terms of $H^*(10)$ which is the dose equivalent measured at a depth of material equal to 10 mm of tissue.

1.3.4 Ambient Dose at CT scanners

This thesis will focus exclusively on the *ambient dose*, as the study pertains to a CT scanner used in an industrial production line. This scenario involves estimating the dose absorbed by a material body in an environment where radiation sources are distributed around the object.

For an X-ray beam with kinetic energy E_X , the conversion coefficient $h_{E_{max}}^*(E_X) = E_{max}(E_X)/\Phi(E_X)$ is calculated for exposure of the whole-body ICRP/ICRU adult reference phantoms [20] assuming broad parallel beams of radiation incident under various irradiation geometries [19]. Considering a number dN of incident X-rays, passing through a sphere with a cross-sectional area da , located in a radiation field, the *ambient dose* can be expressed as:

$$\begin{aligned} H^* &= \Phi \cdot h_{E_{max}}^* \\ &= \frac{dN}{da} \cdot h_{E_{max}}^* \end{aligned} \quad (1.11)$$

The numerical values of ambient dose conversion coefficients from fluence for photons are provided in Table 1.1. These values are also visualized in Figure 1.13, where it can be easily observed that within the energy range of X-rays used in CT scanners, the

coefficients consistently remain below unity. Specifically, there is a gradual decrease across the photon energy range from 200 keV to 20 keV, followed by a sharp drop at energies down to 5 keV.

$E_p[\text{keV}]$	$h_{E_{max}}^* [\text{pSv} \cdot \text{cm}^2]$	$E_p[\text{keV}]$	$h_{E_{max}}^* [\text{pSv} \cdot \text{cm}^2]$	$E_p[\text{keV}]$	$h_{E_{max}}^* [\text{pSv} \cdot \text{cm}^2]$
5.0	1.34×10^{-2}	15.0	1.56×10^{-1}	70.0	4.11×10^{-1}
6.0	1.66×10^{-2}	17.0	1.81×10^{-1}	80.0	4.43×10^{-1}
7.0	2.25×10^{-2}	20.0	2.25×10^{-1}	100.0	5.18×10^{-1}
8.0	3.35×10^{-2}	25.0	2.75×10^{-1}	150.0	7.47×10^{-1}
9.0	4.90×10^{-2}	30.0	3.12×10^{-1}	200.0	1.00
10.0	6.85×10^{-2}	40.0	3.50×10^{-1}	300.0	1.51
12.0	1.05×10^{-1}	50.0	3.69×10^{-1}	400.0	2.00
13.0	1.22×10^{-1}	60.0	3.89×10^{-1}	500.0	2.47

Table 1.1: Tabulation of photon ambient dose conversion coefficients $h_{E_{max}}^*$ from fluence [19]. The energy tabulated here considers only the range useful in medical and industrial fields.

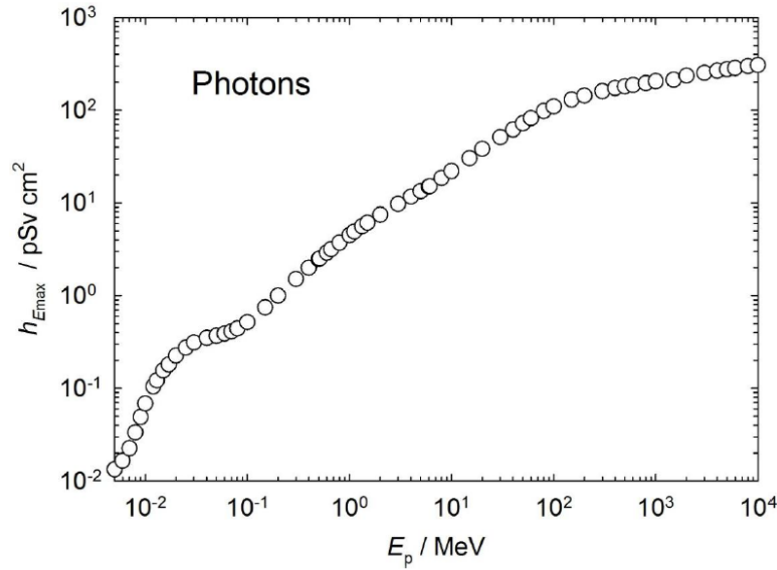


Figure 1.13: Ambient dose conversion coefficients versus photon energy [19].

The ambient dose equivalent $H^*(10)$ measured at a depth of 10 mm of tissue, or equivalent, is a standard value provided by many dose measurement devices, including the dosimeter used to collect the data discussed in Chapter 3. Consequently, comparisons must consider the X-ray conversion coefficients from ambient dose to ambient dose equivalent at 10 mm, as listed in Table 1.2. As depicted in Figure 1.14, X-rays within the energy range of 18 keV to 60 keV are particularly significant due to their conversion coefficients exceeding unity, indicating their enhanced contribution to the ambient dose equivalent at this depth.

$E_p[\text{keV}]$	$h^*(10)/h_{E_{max}}^*$	$E_p[\text{keV}]$	$h^*(10)/h_{E_{max}}^*$	$E_p[\text{keV}]$	$h^*(10)/h_{E_{max}}^*$
10.0	8.70×10^{-1}	50.0	1.46	200.0	1.17
15.0	5.23×10^{-1}	60.0	1.29	300.0	1.17
20.0	4.65	80.0	1.18	400.0	1.17
30.0	2.58	100.0	1.15	500.0	1.17
40.0	1.79	150.0	1.15		

Table 1.2: Tabulation of photon conversion coefficients from ambient dose to ambient dose equivalent at 10 mm [19]. The energy tabulated here considers only the range useful in medical and industrial fields.

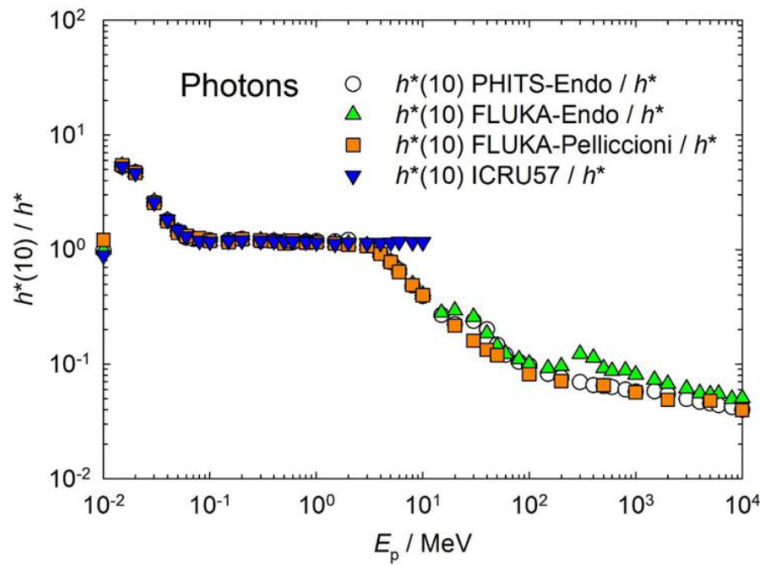


Figure 1.14: Ratio between ambient dose equivalent at 10 mm and ambient dose conversion coefficients as a function of photon energy. Comparisons of conversion coefficients for photons of $h^*(10)$ PHITS-Endo and $h^*(10)$ FLUKA-Endo (Endo, 2016b), $h^*(10)$ FLUKA-Pelliccioni (Pelliccioni, 2000), and $h^*(10)$ (ICRU, 1998) with kerma approximation; with those for h^* [19].

1.3.5 Detection

In **this work** we focus on evaluating and developing technologies specifically for detecting X-rays. In this section we will make an extensive overview of various detection systems, after which we will finally focus on semiconductor sensors for their suitability in direct conversion technology and their efficiency in detecting particle interactions. Among semiconductor materials, silicon is selected due to its advantageous properties. We will then evaluate two different pixel technologies, hybrid pixel sensors and MAPS. The properties of silicon semiconductors and pixel sensors will be addressed in detail in Chapter 2.

X-rays were first discovered in 1895 by Wilhelm Conrad Röntgen, who observed that a cathode ray tube covered by paper foil produced a detectable signal on a photographic plate, even though the electrons emitted could not penetrate the shielding paper. This led to the first detection of X-ray radiation.

Photographic screens became the earliest detection devices for X-rays, but they were highly inefficient for radiography, as less than 2 % of the incident X-rays were captured by the plate. Consequently, the resulting images were extremely dark, even after prolonged exposure to radiation. The first radiograph, shown in Figure 1.15, took approximately 15 min to produce. Photographic screens remained in use for a significant period, thanks to the development of specialized emulsions that increased efficiency. In the 1930s, a new type of nuclear emulsion was created, enabling the identification of nuclear tracks. These improved screens were widely employed in cosmic ray research [22].



Figure 1.15: The bones of a hand with a ring on one finger, viewed through X-ray. Photoprint from radiograph by W.C. Röntgen in 1895 [21]

From the 1940s onward, the next major advancement in X-ray detection involved the indirect conversion of X-ray energy into an electrical signal and its subsequent amplification. It consists of a scintillator coupled with a Photomultiplier Tube (PMT) method, where photons interact with a scintillating material, depositing energy that excites the atoms or molecules of the material. As these atoms or molecules return to their ground state, they emit luminescent flashes of visible light. This light is then guided to the photo-cathode of the PMT, where it is converted into photoelectrons. The resulting electrical signal is amplified through a system of dynodes and ultimately collected at the anode. The stages of this process are illustrated in Figure 1.16.

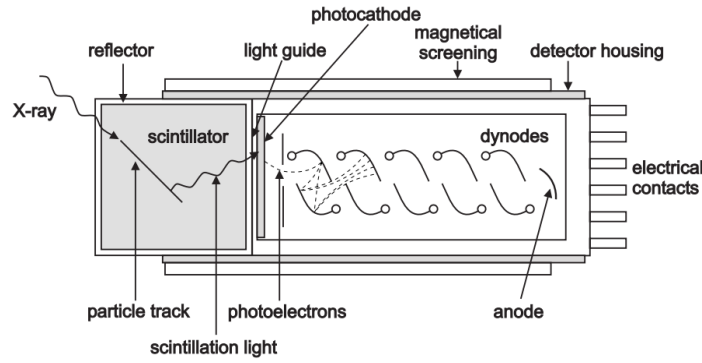


Figure 1.16: Schematic of a scintillation detector with a photomultiplier tube to convert the X-rays and amplify the light signal [23].

The operational mode of a scintillator can be described using the following equations. When a photon interacts with the scintillation material, transferring part or all of its kinetic energy E_{abs} to the material, the number of visible photons (N_{vis}) generated by the scintillation process is given by:

$$N_{vis} = \eta_S \frac{E_{abs}}{\langle E \rangle} \quad (1.12)$$

where η_S is the scintillation response of the material and $\langle E \rangle$ is the average energy required to produce a single visible photon within the scintillation medium. The visible light emitted by the scintillation material is isotropic, meaning it radiates uniformly in

all directions. As a result, not all the generated light can reach the PMT photo-cathode. The fraction of light that reaches the photo-cathode depends on the internal reflections within the scintillation material and the efficiency of the optical coupling, described by the light transfer factor k_l . The number of emitted photoelectrons ($N_{e^-}^{cathode}$) is directly proportional to the number of visible photons that reach the PMT photo-cathode ($k_l \cdot N_{vis}$) and the quantum efficiency of the photo-cathode ($\eta_{\gamma e}$):

$$N_{e^-}^{cathode} = \eta_{\gamma e} k_l N_{vis} \quad (1.13)$$

The photoelectrons are then directed to the first dynode, where their survival fraction is determined by the electron transfer factor k_e . After this, the photoelectrons are accelerated through the subsequent n dynodes, each of which produces Δ secondary electrons. Once the electrons reach the PMT anode, the total amount of charge collected (Q) is given by the following expression [23]:

$$\begin{aligned} Q &= e N_{e^-} \\ &= e N_{e^-}^{cathode} k_e \Delta^n \\ &= e k_e \Delta^n \eta_{\gamma e} k_l \eta_S \frac{E_{abs}}{\langle E \rangle} \end{aligned} \quad (1.14)$$

The scintillator coupled with a photomultiplier tube (PMT) remains in use in academic laboratories today. However, it is not practical for medical and industrial applications due to the high voltage required, limited tube lifespan, high noise levels, and low quantum efficiency. These challenges are, in part, addressed through the development of Silicon Photomultipliers (SiPMs), which are single-photon-sensitive devices with smaller dimensions, and photodiodes (PDs), semiconductor devices that convert light into electrical current. The structure of avalanche photodiodes (APDs) is designed to support high reverse bias operation, enabling avalanche breakdown. This process multiplies each photo-generated carrier, creating internal gain in the photodiode and enhancing its effective responsiveness. One notable application of this technology is in flat-panel detectors (FPDs), a projection radiography technology that collects light using photodiodes, as shown in Figure 1.17b. This system consists of a photodiode matrix attached to the scintillation layer, where the photodiodes convert light into electric charge. The accumulated charge is stored in capacitors within each pixel. The readout of the collected charge is controlled by a thin-film transistor (TFT) switch enclosed in each pixel [14].

Other methods for collecting light can also be integrated with scintillation materials. For example, Charge-Coupled Device (CCD) detectors combine light-sensitive capacitors with a CCD, which stores charge proportional to the incident light. The charge is then read-out through a linear charge transfer process, as shown in Figure 1.17c. A major limitation of CCD detectors, however, is the small pixel size, which is inadequate for matching the field of view required in projection radiography. Scaling up the pixel size is not feasible, as the charge transfer process over a large detection area introduces significant noise, leading to very low resolution in the final image. To address this, a light demagnification process is employed, which reduces the signal acquired by the capacitors but resolves the size issue. This makes CCD detectors a cost-effective solution for medical and industrial X-ray systems [7].

The introduction of *Complementary Metal-Oxide-Semiconductors* (CMOS) in the late 1990s solved the issue of sensor size. CMOS technology uses complementary and

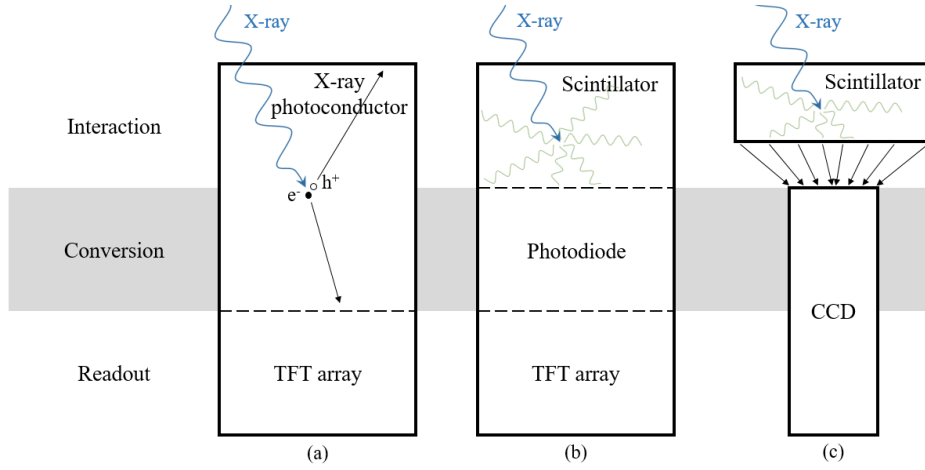


Figure 1.17: Readout electronic X-ray detectors using direct (a) and indirect conversion (b,c) techniques to convert X-rays into electric signal. (a) X-rays interact with a photoconductor that transform the energy released into electric charge; while during indirect conversion X-rays interact with a scintillation material where they are converted into visible light and then collected (b) by a photodiode array, like in the FPD or (c) by a CCD, optical coupled with the scintillator. TFT arrays can be used both in direct and indirect conversion.

symmetrical pairs of p-type and n-type MOSFETs (Metal–Oxide–Semiconductor Field-Effect Transistors) to provide logic functions. This technology enables the production of larger pixels compared to CCD detectors, while offering advantages such as low cost, low power consumption, high X-ray radiation tolerance, and fast pixel readout. CMOS sensors with Active Pixel Sensors (APS) capture the light produced by a scintillator within individual pixels. These pixels are activated and provide an amplified electrical signal along with a set of coordinates, which indicate the row and column of the pixel that was hit by the radiation.

X-ray detection relies not only on the production of visible light but also on the ability of the material with which the photons interact to ionize. An example of a detector based on ionization and signal acquisition by a TFT array is shown in Figure 1.17a. The ionization mechanism, along with the ability to move the charges produced by applying an electric field, forms the basis of *ionization chambers*. These devices were developed before the discovery of X-rays and were initially used to measure particles generated in radioactive decay. The first generation of ionization chambers used gas as the target material. These were immediately applied for measuring high-intensity radiation sources. However, the low density of gases prevents charge multiplication, making it difficult to detect low-energy radiation efficiently.

In 1928, the first *Geiger-Müller counter* was developed by the German physicists of the same name. The key advantage of this new gas-filled counter was its ability to detect individual radiation events and amplify them significantly. However, it lacked the ability to discriminate between different types of radiation and did not provide energy resolution. Due to its simplicity and low cost, the Geiger-Müller counter became widely used and further improved over time. Notably, research explored using different target materials, eventually leading to the realization that solid materials offered significant advantages due to their higher density and the possibility of creating smaller detectors [22].

Semiconductor detectors emerged in the late 1950s. These detectors are made of single crystals with a specific energy band structure: the valence and conduction bands are separated by a band gap that is not excessively large, allowing for the possibility of electrons being excited from the valence to the conduction band. At temperatures above 0 K, the thermal energy of electrons can excite them into the conduction band, where they behave as free electrons.

The transition of an electron from the valence band to the conduction band can occur in two different ways, depending on the initial and final momentum of the charged particle:

- **direct interband transition:** the minimum of the conduction band and the maximum of the valence band occur at the same momentum in the energy band diagram so an electron transition from the valence band to the conduction band can occur without any change in momentum. This process is typical in materials like Gallium Arsenide (GaAs), which has an energy gap of $E_g^{GaAs}(300\text{ K}) = 1.42\text{ eV}$.
- **indirect interband transition:** the minimum of the conduction band and the maximum of the valence band occur at different momentum values in the energy band diagram so the transition of an electron from the valence band to the conduction band requires the involvement of an external particle, such as a phonon (a quantum of lattice vibration): the excess momentum of the electron is transferred to the crystal lattice. This means that the electron does not directly jump to the conduction band but must pass through an intermediate state, such as crystal imperfections or impurities. Indirect interband transitions are observed in materials like Silicon (Si) and Germanium (Ge), which have energy gaps of $E_g^{Si}(300\text{ K}) = 1.12\text{ eV}$ and $E_g^{Ge}(300\text{ K}) = 0.67\text{ eV}$, respectively.

The main challenge associated with using Silicon and Germanium semiconductors as detectors, is the need to produce them with high purity. This difficulty was largely overcome in the early 1970s, leading to the development of high-purity Germanium (HPGe) detectors. HPGe detectors are known for their high resolution, thanks to the large number of charge carriers produced by the narrow band gap. However, this high resolution also presents a challenge: thermal noise at room temperature can interfere with the measurements. To reduce this noise, HPGe detectors must be cooled. To avoid the need for cooling, detectors can be made from materials with larger band gaps, such as Silicon or Cadmium Telluride (CdTe), with energy gaps of $E_g^{CdTe}(300\text{ K}) = 1.5\text{ eV}$. However, these materials have lower resolution and efficiency compared to Germanium.

When an X-ray interacts with a semiconductor through the photoelectric effect, the photon energy is completely absorbed by an electron, which is then excited into the conduction band. This can happen in one of three ways:

- If the energy of the X-ray is greater than the band gap energy, the electron reaches an empty level within the conduction band, leaving behind a hole in the valence band. Both the electron and the hole then move toward the boundaries of the band gap, releasing energy in the process.
- If the X-ray energy is equal to the band gap energy, an external electron from the valence band moves to the conduction band, filling the lower level

of the conduction band.

- If the photon energy is lower than the band gap and an "indirect interband transition" occurs, a hole is created in the valence band, and an intermediate state is ionized.

Once the charges are separated between the valence and conduction bands, applying an external voltage, by means of an inhomogeneous charge carrier distribution or doping concentration, accelerates the free particles in the direction defined by the electric field. This makes it possible to easily detect the signal generated by the interaction of an external particle with the semiconductor. Since the 1970s, the development of Silicon detectors has been integrated with microelectronic technologies, leading to the creation of low-noise, low-power analog read-out electronics for semiconductor detectors. Furthermore, the introduction of the drift mechanism in these sensors has paved the way for the production of new semiconductor structures [24].

Semiconductors have been widely exploited in various fields over the last 30 years, leading to the development of numerous systems and devices. In the medical field, for example, they have been integrated as photon-counting detectors for CT scans and radiography, thanks to their excellent time resolution and adaptability in size. Similarly, in particle physics, semiconductors play a critical role due to their high spatial resolution and fast acquisition rates, making them indispensable in laboratories such as the Large Hadron Collider (LHC) at CERN. The continued advancement of this technology also offers opportunities for enhancing industrial apparatus, which is a key focus of this thesis. Therefore, the following chapters will explore different types of semiconductor-based sensors and their advancements for industrial applications.

2 Silicon Detectors

The continuous advancement of semiconductor sensors to address emerging challenges has led to significant innovations in industrial X-ray CT scanning. In the field of High Energy Particle Physics, the gold standard among semiconductor detectors is the ALPIDE (ALICE Pixel Detector) chip. This is a Silicon Monolithic Active Pixel Sensor (Si-MAPS) known for its high integration density, compact structure, great detection efficiency and excellent spatial resolution.

Among the available semiconductor materials, silicon is widely used for fabricating new detectors due to its advantageous properties. The energy required to create an electron-hole pair is only 3.6 eV, which is much lower than the energy required for gases used in ionization chambers or proportional counters, and the band gap is moderate, equal to 1.2 eV. Silicon is also favored for its high radiation tolerance, low material cost, and abundant availability. Furthermore, while it is less efficient than germanium, silicon operates at room temperature, eliminating the need for cooling. This is particularly advantageous in environments such as medical settings.

Resistivity (ρ) is an important parameter that characterizes the ohmic electrical property of the semiconductor materials used in a detector. For silicon, the intrinsic resistivity is $\rho_{Si} \approx 330 \text{ k}\Omega \cdot \text{cm}$ at 293 K (room temperature). Conductivity (σ), the inverse of resistivity, is a direct measure of the ability of free charges to move within the material in response to an electric field: the formula for conductivity is related to charge mobility (μ), which is the ratio between the applied electric field and the drift velocity, as well as the charge densities and is expressed as $\sigma = e(\mu_e n + \mu_p p)$ where e is the electron charge, and n and p represent the densities of electrons and holes, respectively. At room temperature the intrinsic conductivity of silicon is $3 \times 10^{-6} \text{ S/cm}$.

To achieve a desired resistivity in a silicon layer, the material properties must be altered by adding impurities to the crystal lattice, a process known as *doping*. Through doping the resistivity can be reduced by even orders of magnitude. However, in practice *high-resistivity* doped silicon, with values typically in the range of a tens of $\text{k}\Omega \cdot \text{cm}$, is used for detectors as it minimizes leakage currents, enhances the efficiency of charge collection, improves signal quality: a high-resistivity detector performs with greater precision in situations that require low noise and high sensitivity.

When a material is doped, if the added substances contribute an excess of electrons to the conduction band, the material becomes an *n-type* semiconductor. Conversely, if the impurities introduce holes into the valence band, the material becomes a *p-type* semiconductor, as shown in Figure 2.1. In an *intrinsic* semiconductor (i.e., a pure silicon wafer), the energy at which exactly half of the available levels are occupied, known as the *Fermi level*, is positioned in the middle of the forbidden energy gap. However, when the material is doped, the *Fermi level* shifts toward the conduction or valence bands depending on whether the doping results in an *n-type* or a *p-type* semiconductor, respectively, as illustrated in Figure 2.2a.

After doping a semiconductor, the subsequent implantation of a semiconductor of the opposite type forms a *p-n junction*, a simple diode structure. This junction produces a sharp gradient in the density of conduction electrons. As a result, free charge carriers tend to move from the region of higher concentration to the region of lower concentration, where they recombine with opposite charges, restoring equilibrium. At the interface of the two differently doped semiconductors, a net diffusion of electrons

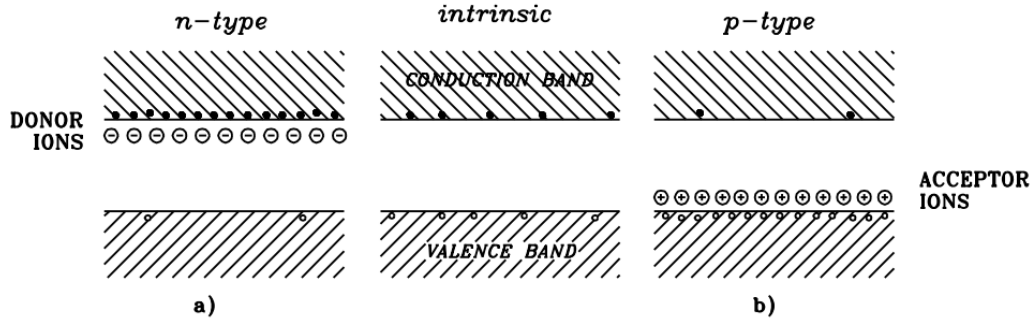


Figure 2.1: Representation of the energy band structure of (a) a *n-type* semiconductor, (b) a *p-type* semiconductor and a pure material semiconductor (*intrinsic*) [24].

from the *n-type* to the *p-type* semiconductor creates a region with a potential difference, known as the *built-in potential* (V_{bi}). While the diffusion process tends to expand this region, the electric field generated by the migration of charge carriers acts in the opposite direction. As a result, a *depletion region* forms at equilibrium, as shown in Figure 2.2b. The extent of this region depends on the concentrations of donors and acceptors, denoted N_D and N_A , respectively. If the concentrations are similar, the width of the depletion region will be roughly the same in both the *n-type* and *p-type* materials. However, if there is an excess of donor concentration, for example, the depletion region will be wider in the *p-type* material, as it works to recombine all the free conduction electrons with holes.

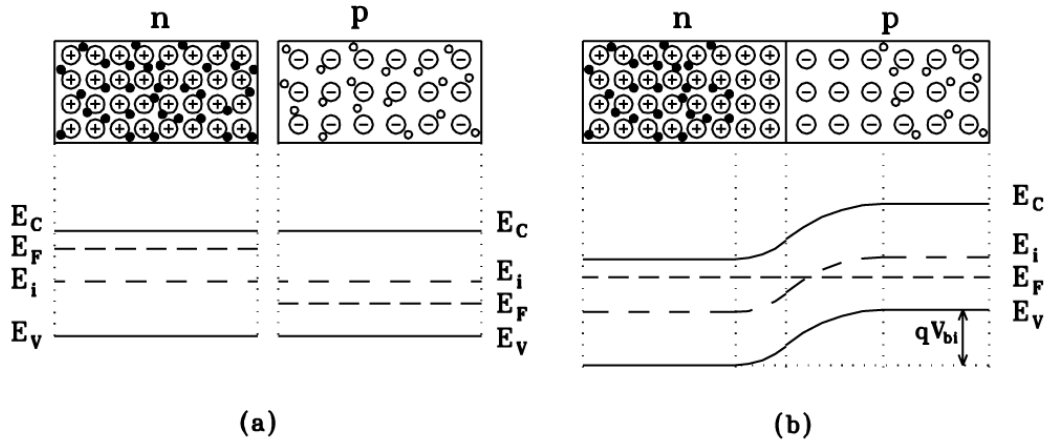


Figure 2.2: Schematic representation of a *p-n junction* in thermal equilibrium, with E_C the conduction band, E_F the Fermi level of the doped semiconductor material, E_i the Fermi level associated to the non-doped material, corresponding to the middle of the band gap, and E_V the valence band. (a) *n-type* and *p-type* semiconductors separated. (b) The same semiconductors brought together: a depletion zone is formed at the surface between the two substances [24].

Even though a *p-n junction* is formed, it is not ideal for radiation detection due to the limited performance of the diode at thermal equilibrium. As is, the device generates a narrow *depletion region* due to the movement of a few electrons from the *n-type* material into the holes of the *p-type* material, creating a *built-in potential* of about 1 V. This potential difference is too small to cause a rapid movement of any charge carriers generated during irradiation: the slow electrons and holes may recombine during their diffusion within the semiconductor, causing the signal to be

lost.

The width of the *depletion region* can be controlled by applying an external voltage to the diode. For a one-dimensional junction, the charge density ρ within the two materials can be derived from the Poisson equation:

$$\nabla^2 V = -\frac{\rho}{\epsilon_{Si}\epsilon_0} \quad (2.1)$$

where $\epsilon_{Si} = 11.7$ is the dielectric constant of silicon relative to the vacuum dielectric constant $\epsilon_0 = 8.854... \times 10^{-12}$ F/m.

The result shows that in the *n-type* region, the charge density is positive ($n = eN_D$), while in the *p-type* region, it is negative ($p = -eN_A$). By integrating these charge densities and applying boundary conditions that specify the electric field is zero at the junction boundaries, we can calculate the electric field $E(x) = -dV/dx$ in the two doped regions. The result is:

$$\begin{aligned} E_n(x) &= -\frac{eN_D}{\epsilon_{Si}\epsilon_0}(x+a), & -a < x < 0 \\ E_p(x) &= +\frac{eN_A}{\epsilon_{Si}\epsilon_0}(x-b), & 0 < x < b \end{aligned}$$

with $a+b = W_{dep}$ corresponding to the total width of the region depleted of free charge carriers, where a and b are the widths of the depletion regions in the *n-type* and *p-type* materials, respectively ($a, b > 0$). The electric field must be continuous at the junction, which gives the condition $aN_D = bN_A$, ensuring that the total charge on both sides of the junction is balanced. The component of electric field in any direction is the negative of rate of change of the potential in that direction, therefore, another integration provides the potential values across the depletion region:

$$\begin{aligned} V_n(x) &= \frac{eN_D}{2\epsilon_{Si}\epsilon_0}(x+a)^2, & -a < x < 0 \\ V_p(x) &= \frac{eN_A}{2\epsilon_{Si}\epsilon_0}(x+b)^2, & 0 < x < b \end{aligned}$$

The *built-in potential* results the voltage across the junction, equal to:

$$\begin{aligned} V_{bi} &= -\int_{-a}^b E(x)dx = \frac{e}{\epsilon_{Si}\epsilon_0} \left[\int_{-a}^0 N_D(x+a)dx - \int_0^b N_A(x-b)dx \right] \\ &= \frac{e}{2\epsilon_{Si}\epsilon_0}(N_D a^2 + N_A b^2) \\ &= \frac{e}{2\epsilon_{Si}\epsilon_0}(N_A + N_D) \frac{N_D}{N_A} a^2 \\ &= \frac{e}{2\epsilon_{Si}\epsilon_0}(N_A + N_D) \frac{N_A}{N_D} b^2 \end{aligned}$$

Now it is possible to evaluate the width of the natural depletion region W_{dep} , that depends to the doping levels of the acceptor N_A (*p-type* material) and donor N_D (*n-type* material) semiconductors and the *built-in potential*:

$$\begin{aligned} W_{dep} &= a + b \\ &= \sqrt{\frac{2\epsilon_{Si}\epsilon_0 V_{bi}}{e(N_A + N_D)}} \left[\sqrt{\frac{N_A}{N_D}} + \sqrt{\frac{N_D}{N_A}} \right] \\ &= \sqrt{\frac{2\epsilon_{Si}\epsilon_0}{e} \frac{(N_A + N_D)}{N_A N_D} V_{bi}} \end{aligned}$$

When an external bias V_{ext} is applied to the junction, the potential across the junction deviates from the neutral *built-in potential* (V_{bi}), and the width of the depletion region (W_{dep}) is consequently adjusted. The relationship governing this behavior can be expressed by the general formula:

$$W_{dep} = \sqrt{\frac{2\epsilon_{Si}\epsilon_0}{e} \frac{(N_A + N_D)}{N_A N_D} (V_{bi} - V_{ext})} \quad (2.2)$$

When a *forward bias* is applied to the junction, connecting the *p-type* region to the positive terminal and the *n-type* region to the negative terminal of the voltage supply, the width of the depletion region decreases. Conversely, applying a *reverse bias* to the diode, where the *p-type* region is connected to the negative terminal and the *n-type* region to the positive terminal, results in an increase in the depletion region width. As shown in Equation 2.2, under a *reverse bias* condition ($V_{ext} < 0$), the term under the square root remains positive, leading to an expansion of the depletion region.

In the *reverse bias* condition, the pre-existing *built-in potential* is further enhanced. The associated reverse-bias current remains very small because the concentration of charge carriers attracted to the junction (holes in the *n-type* region and electrons in the *p-type* region) is minimal. Due to this property, a reverse-biased *p-n junction* acts as a rectifier, a system that allows current to flow predominantly in one direction, blocking significant current flow in the reverse direction.

2.1 Pixel detectors

2.1.1 Introduction

The diode structures described above can be fabricated in various sizes and with different doping substances. The development of the detector, including the choice of materials and doping levels, must be carefully aligned with the associated read-out electronics: the electronics must be capable of acquiring and analyzing the signal output efficiently while accounting for the required acquisition rate.

In the domain of HEP, particularly in large-scale experiments such as those conducted at the LHC at CERN, the increasing rate of particle collisions sets the foundation for stringent electronics requirements. These include high acquisition rates, the ability to store and process large amounts of data, and robust radiation tolerance. To meet these demands, it is crucial to have greater logic density close to the diode

structure and to minimize the sensitive area where the signal output is collected. This necessity has led to the adoption of pixel detectors. Small pixels are essential for several reasons: they increase detection efficiency by reducing the risk of signal pile-up, enhance spatial resolution, and minimize leakage current in each pixel. The latter contributes to improve radiation tolerance, making pixel detectors ideal for high-intensity environments [25].

Presently, the gold standard in the HEP field is the ALPIDE chip, a *Monolithic Active Pixel Sensor* (MAPS) matrix with 512×1024 pixels, each with an area of approximately $27 \mu\text{m} \times 29 \mu\text{m}$. The MAPS technology is described below in Section 2.1.3. This ALPIDE MAPS chip, a cross section of which is shown in Figure 2.3, forms the foundation of the second-generation Inner Tracking System (ITS2) of the ALICE experiment at LHC. Its use significantly enhances the read-out rate to approximately 100 kHz for Pb-Pb collisions and 200 kHz for proton-proton collisions. In comparison, the first-generation Inner Tracking System (ITS) was developed using hybrid pixel sensors, a technology described below in Section 2.1.2 that continues to find widespread application in other fields. The hybrid pixel cells had an area of about $50 \mu\text{m} \times 425 \mu\text{m}$ and a read-out rate of just 1 kHz. The shift from hybrid to MAPS technology offers a significant advantage, also in terms of cost efficiency, as it reduces the material budget for the ALICE experiment by a factor of seven.

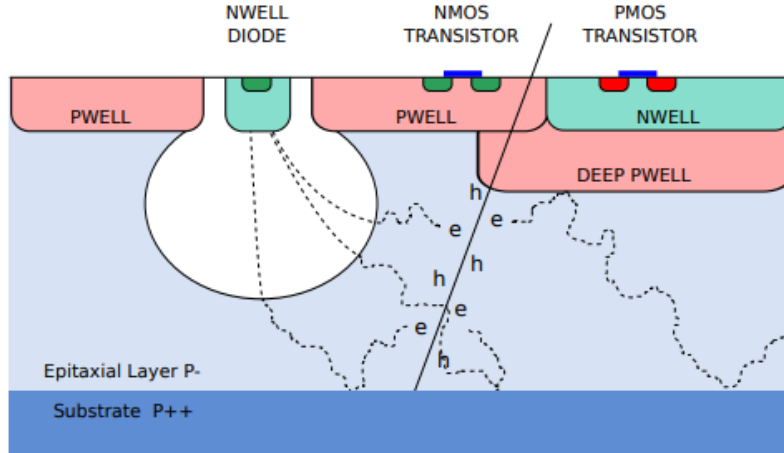


Figure 2.3: Cross section of the ALPIDE chip. It is a MAPS pixel in the 180 nm imaging CMOS with the deep p-well feature [26]. The full thickness of the sensor is $50 \mu\text{m}$ with a reverse biased epitaxial layer of about $25 \mu\text{m}$. The application of the external voltage provides a partial depletion region (white area) starting from the *n-well diode*, so that the negative charge collection happens firstly via diffusion and then drift.

In ITS2, each pixel is a single silicon die that includes a high-resistivity silicon epitaxial layer, which constitutes the sensor’s active volume and serves as the fundamental MAPS element of the ALPIDE chip. The charge collection diode (pixel) amplifies and digitizes the signal, while also performing zero-suppression using integrated electronics. The signal output from each pixel conveys critical information about whether or not a particle has interacted with that specific pixel [26].

The aim of the ITS2 is to significantly improve tracking accuracy over a wider range of particle momenta and enhance the vertexing performance, the ability to precisely identify the point where particles originate, both primary vertexes from main beam-

beam interactions, and secondary ones created by decaying particles. The strategy is to move the innermost detection layer closer to the beam pipe, increasing granularity, and reducing the material budget, all achievable by using MAPS.

2.1.2 Hybrid pixel sensors

Hybrid pixel detectors consist of a pixel matrix sensor integrated with an electronic circuit for each pixel, enabling signal processing at the pixel level. The electronic circuit's area must match the pixel size to ensure seamless alignment between the pixel matrix and the electronic board, as illustrated in Figure 2.4. The interconnections between the n^+ electrodes in the sensitive region and the electronic chip areas can be established using one of two methods: wire bonding or a more advanced technique called *bump bonding*. The *bump bonding* process involves placing a solder ball on each pixel [27]. However, using two separate components, one for the sensor and one for the electronics, that must be connected later increases production costs and raises the scattering probability of the particles being detected. This scattering degrades spatial resolution. Consequently, in applications where spatial resolution is a top priority, hybrid technology has been largely supplanted by MAPS, which provide superior integration and performance.

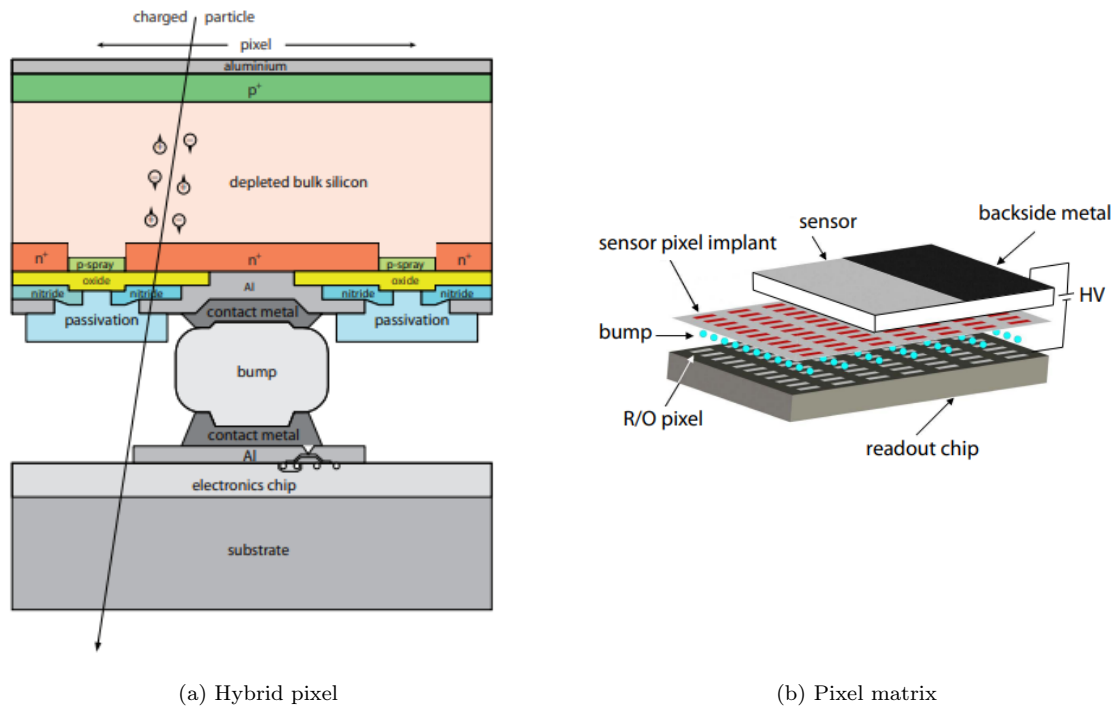


Figure 2.4: Picture of an hybrid pixel detector. (a) Single pixel cell showing the correspondence between the sensor and the electronics. (b) The entire pixel matrix: sensor and electronics chip have pixels of the same size, bonded to each other by means of bump contacts [25].

Hybrid pixel structures have been extensively tested and implemented in nearly all experiments at CERN since 1995, when challenges related to track localization and radiation hardness became increasingly critical. The subsequent advancements in hybrid pixel technology, alongside its integration into the four primary LHC experiments, have spurred its evolution and adaptation for applications beyond HEP. Hybrid

pixel detectors offer several key advantages: high detection efficiency, nearly zero noise, fast acquisition speeds, and the ability to perform energy selection by setting multiple thresholds in the electronic chip. Compared to CCD cameras and other X-ray detection systems, hybrid pixel technology has demonstrated clear benefits, leading to its adoption for applications like detecting X-rays generated by synchrotron light sources. Notable examples of circuits developed from this technology for X-ray detection include *Medipix*, *Xpad*, *Pilatus* and *Pixie*. *Medipix*, in particular, was the first collaborative effort with CERN to create a sensor tailored for medical imaging. Hybrid pixel sensors have also been tested in clinical CT scanners. For instance, the *Siemens Healthcare* group developed a prototype scanner capable of photon counting with dual energy thresholds for each pixel. The results have been promising, showing reduced radiation doses for patients during contrast-enhanced diagnostic procedures [27]. However, this technology has yet to be fully industrialized, and certain materials used for the sensitive area, such as Cadmium Telluride (*CdTe*), remain prohibitively expensive.

2.1.3 Monolithic Active Pixel Sensors (MAPS)

The limitations of hybrid pixel sensors can be addressed by adopting a single-element detector in which the sensor and the read-out circuitry are integrated on the same silicon chip. The *Complementary Metal-Oxide-Semiconductor Monolithic Active Pixel Sensor* (CMOS MAPS) is a compact silicon-based device (*Monolithic*) organized into a pixel matrix, where each pixel incorporates an internal analog circuit (*Active*) for signal acquisition. While MAPSs share conceptual similarities with hybrid detectors, such as features like guard rings and channel stops, they operate with a significantly lower external voltage compared to hybrid structures. Additionally, the monolithic design minimizes the sensitive region occupied by the circuitry, providing two key advantages: an improved signal-to-noise ratio and the possibility to rely commercial fabrication processes. This approach enables cost-effective production and coverage of large areas.

A typical MAPS consists of an epitaxial layer grown on top of a substrate of the same type. The thickness of these layers varies, with the epitaxial one typically ranging from 1 μm to 20 μm , while the overall sensor thickness falls within 50 μm to 400 μm . The CMOS front-end and read-out electronics are located on top of the epitaxial layer, where the external bias is generally applied. The NMOS and PMOS transistors in the CMOS are enclosed within p-wells and n-wells, respectively.

As previously discussed, the gold standard for this technology is represented by the MAPS sensors installed in the ITS2 of the ALICE experiment, as illustrated in Figure 2.3. These sensors are fabricated using standard 180 nm CMOS technology designed for imaging sensors, with an additional processing step that introduces a high-energy "deep p-well" implant to shield the sensing nodes from the epitaxial layer. This configuration provides a foundation for the n-well region where the CMOS front-end electronics are integrated and resolves a key challenge: it prevents the n-well layer of the CMOS circuitry from interfering with the signal acquisition by the n-well sensing diode. This solution achieves a sufficient signal-to-noise ratio without increasing power consumption, making it a highly effective design for radiation detection. The total sensor thickness is 50 μm , which includes a high-resistivity ($> 1 \text{ kW} \cdot \text{cm}$) p-type epitaxial layer of 25 μm on a p-type substrate.

The advanced technologies employed in the ALPIDE sensor, have been leveraged for the development and enhancement of systems in various fields beyond HEP. The 180 nm transistor feature size, for instance, is instrumental in increasing the density of radiation-sensitive areas while simultaneously reducing the power consumption of digital circuits. This design facilitates placing the digital circuitry at the periphery of the pixel matrix, thereby minimizing the insensitive area. Additionally, the thick, high-resistivity epitaxial layer of the ALPIDE sensor, plays a critical role in extending the depletion region. This feature not only improves the efficiency of the detector but also enhances the signal-to-noise ratio, making the system more effective for particle detection. The most notable innovation, however, is the use of a "deep p-well" layer.

Anyway, the epitaxial layer typically has lower resistivity compared to the substrate, and the implanted electronics limit the external bias to a relatively low voltage. Consequently, this structure creates a partially depleted region starting from the collection nodes. As a result, the negative charges generated by the interaction of particles with the detector primarily migrate to the collection pad through diffusion, rather than drift.

The full depletion condition can be achieved by modifying CMOS technology while ensuring the pixel capacitance remains as low as possible to enable low power consumption and a high signal-to-noise ratio. Two main approaches are typically considered: increasing the bias voltage through the adoption of specialized high-voltage technologies, or utilizing a mid-to-high resistivity substrate.

The first approach introduces a "deep n-well" surrounding the CMOS electronics, which serves as a large collection node. This configuration enhances the sensor's radiation tolerance but also increases pixel capacitance. A key advantage of this method is that the electric field generated by the external bias is nearly uniform due to the extensive area covered by the n-well layer.

The second approach employs a smaller collection electrode positioned outside the CMOS electronic area, as shown in Figure 2.3 for the ALPIDE design. In this configuration, the electrode is still enclosed by a "deep n-well" layer, but the layer is significantly smaller compared to the first approach. This creates a planar junction between the n-well and the epitaxial layer, enabling full depletion of the entire epitaxial thickness with a relatively low external bias. However, in this case, the electric field produced by the external bias is non-uniform due to the small size of the electrode relative to the substrate, resulting in regions with zero electric field, particularly at the sensor corners.

2.2 MAPS for X-rays detection

The ideal X-ray imaging detector must meet several key requirements. Foremost, the signal-to-noise ratio (SNR) should remain uncompromised. This issue is addressed by transitioning from an energy integration system, where the energy information of individual photons is lost due to signal integration over the exposure period, including noise sources, to a quantum imaging system. In the latter, the read-out architecture can detect individual X-rays. This requires dividing the detector into a pixel matrix, with each pixel set to a threshold above the noise baseline. This approach must also be supported by high-speed signal read-out with minimal dead time, the interval between

consecutive event acquisitions within the same pixel. This is crucial for achieving precise and efficient X-ray detection and imaging [28].

The primary difference between hybrid and monolithic technologies lies in the simplicity of fabrication for the latter, along with differences in spatial resolution and the applied high voltage. The objective of this thesis is to evaluate the most suitable detection system for industrial applications. This requires not only considering technological advancements and upgrades but also focusing on reducing the material budget and optimizing sensor fabrication to scale production for industrial purposes. Based on these considerations, silicon MAPS have been chosen as the foundational technology for X-ray detection.

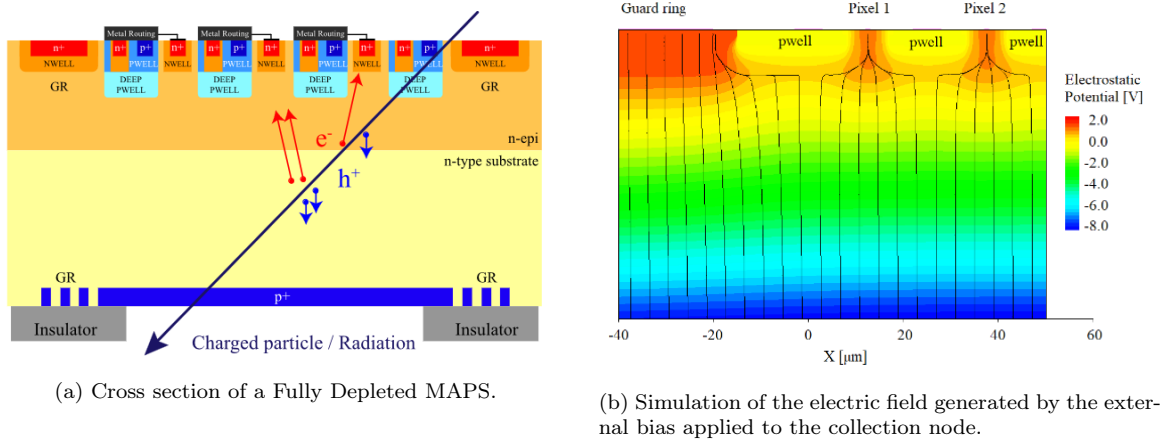


Figure 2.5: Example of a Fully Depleted MAPS considering the modified high-resistivity approach solution [29]. 2.5a Schematic of the cross section of a FD-MAPS with the passage of a particle. The interaction with silicon material produces electron-hole couples that drift to the electrodes. 2.5b TCAD simulation of the electrostatic potential across the sensor. The black lines represent the electric field lines.

Figure 2.5a illustrates an example of a Fully Depleted MAPS featuring an n-type epitaxial layer grown on a high-resistivity n-type substrate, with a deep p-well shielding the electrodes. The right-hand image (Figure 2.5b) depicts the electrostatic potential distribution across the sensor when an external bias is applied to the n-wells.

Building on the CMOS architecture shown in Figure 2.5, two collaborations funded by the *Italian Nuclear Physics National Institute* (INFN), SEED and ARCADIA, have developed a novel solution to address the electric field challenges in the high-resistivity approach for producing sensors with a depletion thickness exceeding 100 μm. Their design employs a standard 110 nm CMOS technology provided by *LFoundry*, combined with a high-resistivity n-type substrate and a p⁺ layer at the bottom. The external bias is applied to this p⁺ layer, enabling the depletion region to expand from the sensor backside, thereby eliminating the need for high voltages on the collection nodes. Applying an external negative bias to the p⁺ region facilitates the growth of the depletion region. This region captures the charges generated by external particle interactions, allowing them to drift directly to the collection nodes within each pixel. When the depletion region extends to the front-end metalization, the device is fully depleted. The voltage required to achieve this state is termed the *full depletion voltage* and can

be calculated as:

$$V_{FD} = \frac{e}{2\epsilon_{Si}\epsilon_0} \frac{(N_A + N_D)}{N_A N_D} D^2 \quad (2.3)$$

with D the distance between the p^+ layer and the metal contact.

In this customized configuration, the maximum voltage permissible at the n-well node is less than one order of magnitude compared to that in a standard CMOS configuration, allowing compatibility with embedded transistors. As a result, the bias applied to the sensing electrodes must remain below 1.2 V. The use of an n-type epitaxial layer enhances the punch-through voltage, facilitating full sensor depletion even when utilizing low-voltage CMOS processes. When the difference between the bias applied to the p^+ backside region and the punch-through voltage is maximized, the electric field within the detector reaches its highest possible value. This results in the fastest charge collection, optimizing the detector performance.

2.2.1 MATISSE

The SEED (*Sensor with Embedded Electronics Development*) collaboration has developed a chip called **MATISSE** (*Monolithic ActIve pixel SenSor Electronics*), a fully depleted monolithic pixel sensor designed to offer excellent timing resolution, on the order of ns, and low power consumption. Prototypes of this technology have already been fabricated, featuring a total thickness of 300 μm and a pixel capacitance of approximately 20 fF. The device covers an area of 2 mm $\times$ 2 mm, which is divided into two regions: four sectors, each containing 6 $\times$ 24 pixels, and an orthogonal area dedicated to hosting the *End-of-Column* (EoC) logic, as depicted in Figure 2.6a. The four sectors vary in the geometry of the implemented collection nodes, with the objective of identifying the optimal configuration to achieve the highest signal-to-noise ratio.

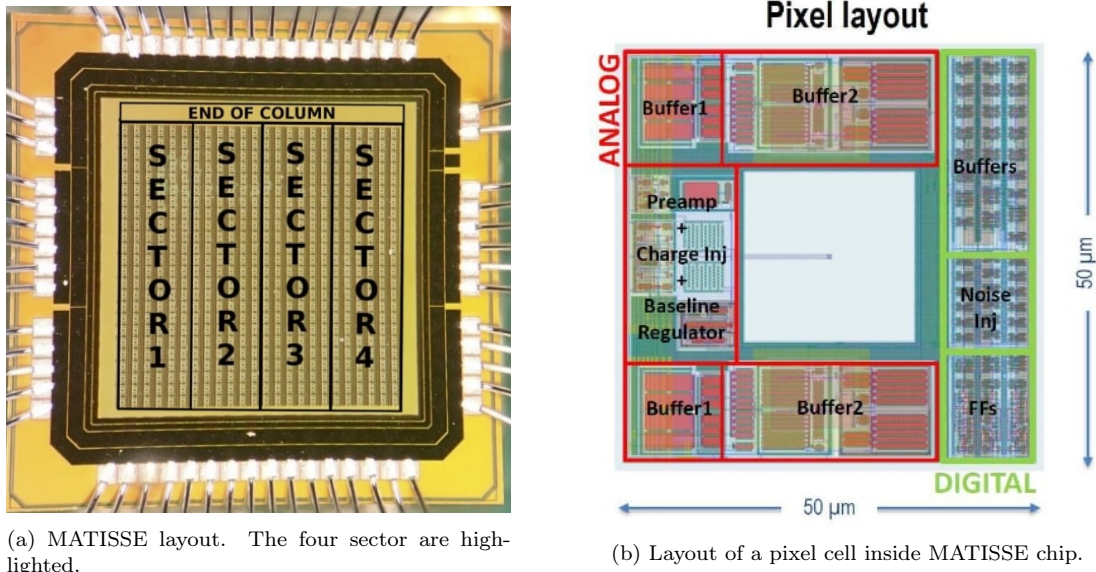


Figure 2.6: Layouts of the MATISSE chip [30]. (a) Photo of a MATISSE demonstrator divided in sectors and *End-of-Column* region. (b) Subdivision of each single pixel. The central part represents the diode of 20 μm side; in red the region that hosts the analog read-out process with the amplification signal stage; in green the region that hosts the digital part with the registers.

Each pixel cell in the MATISSE chip occupies an area of $50\,\mu\text{m} \times 50\,\mu\text{m}$, with the central $20\,\mu\text{m} \times 20\,\mu\text{m}$ designated as the sensitive region. The remaining external surface houses the read-out electronics, as illustrated in Figure 2.6b. The analog read-out includes a charge-sensitive amplifier with a conversion gain of $130\,\text{mV/fC}$, two memory units, and four low-power buffers for transferring the analog signal to the digital processing region. In the digital readout section, three programmable flip-flops are implemented alongside additional functionalities such as pulse generation and pixel masking.

The readout of the full pixel matrix is managed by the EoC logic, where the four sectors transmit signal information, including the initial and final values of the integration time and the addresses of the activated pixels. The reading process is completed in just $40\,\mu\text{s}$, operating at a clock frequency of $50\,\text{MHz}$. This ensures rapid signal transfer, with the sensor consuming $3.6\,\text{mW}$ during integration and $14.4\,\text{mW}$ during data transmission. These results demonstrate the feasibility of implementing a MAPS device with pixel sizes of $50\,\mu\text{m} \times 50\,\mu\text{m}$ or larger, incorporating in-pixel read-out electronics occupying an area of $30\,\mu\text{m} \times 30\,\mu\text{m}$ [30][31].

2.2.2 ARCADIA

Building on the technology developed and tested within the SEED project, a new initiative has emerged in collaboration with a silicon foundry to create scalable MAPS with high signal-to-noise ratio and low power consumption. The **ARCADIA** (*Advanced Readout CMOS Architectures with Depleted Integrated sensor Arrays*) project aims to develop innovative CMOS sensors with the following key features:

- Sensitive detector thickness range of $50\,\mu\text{m}$ to $400\,\mu\text{m}$
- Full depletion of the sensor thickness with signal collection only via drift mechanism
- Small collection electrodes for the optimization of the signal-to-noise ratio
- Scalable event driven read-out architecture with low power consumption to enable air-cooled operation
- Compatibility of the fabrication processes with the standard CMOS production.

To ensure compatibility with various commercial fabrication techniques, the primary design constraint for the ARCADIA sensor is to minimize alterations to the standard fabrication process. The deep p-well is the only custom implant on the upper side of the sensor. Additionally, the target power consumption is below $20\,\text{mW}/\text{cm}^2$ for a signal rate of $100\,\text{MHz}/\text{cm}^2$ and a time resolution on the order of $10\,\text{ns}$ [1]. Figure 2.7 presents the conceptual design of the ARCADIA sensor. When a particle interacts with the detector via ionization, electron-hole pairs are generated and immediately drift toward the collection electrodes. On the sensor's bottom side, a p^+ boron-doped layer is used to apply the external high voltage, creating a reverse-bias *p-n junction* that fully depletes the sensor from the backside region. This method effectively separates the high voltage required for depletion from the voltage needed for the read-out electronics, which enhances radiation resistance and reduces electronic noise.

The ARCADIA sensor is designed for a broad range of applications. It meets the needs of HEP experiments, which require fast timing, low power consumption, high

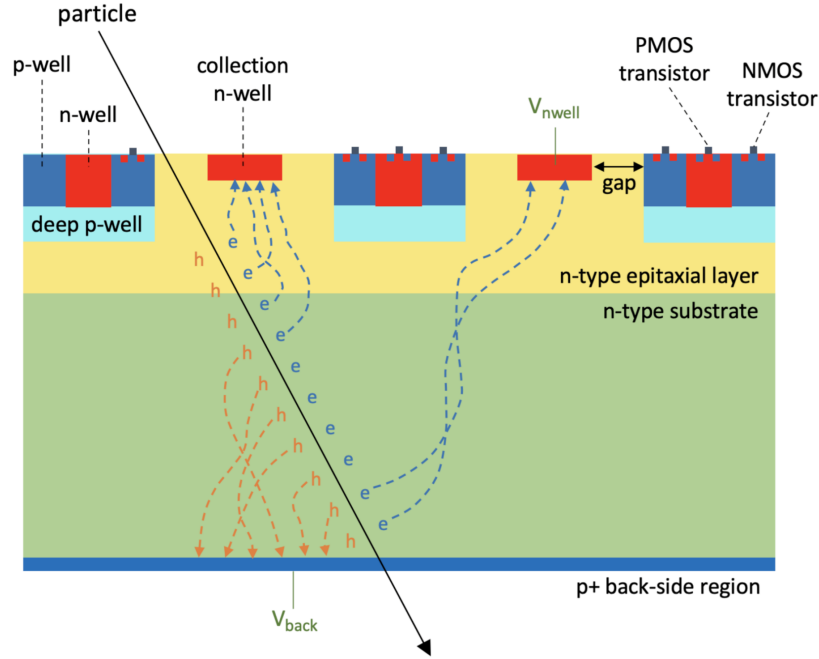


Figure 2.7: ARCADIA chip cross section [1]. It is a FD-MAPS implemented in a 110nm CMOS technology with deep p-well feature.

radiation hardness, high spatial resolution, and scalability. It is also suitable for space experiments, which prioritize ultra-low power consumption and scalable matrix sizes, as well as for medical environments, where fast timing, low voltage, and low material budget are essential.

The collaboration has developed large-area prototypes with an active surface of $1.28 \text{ cm} \times 1.28 \text{ cm}$, consisting of 512×512 pixels with a $25 \mu\text{m}$ pitch. This architecture allows scalability up to a matrix size of 2048×2048 pixels, covering a sensitive area of about 26.2 cm^2 with a single device. The demonstrators have a total thickness ranging from $50 \mu\text{m}$ to $300 \mu\text{m}$. For detectors with a depleted thickness greater than $100 \mu\text{m}$, the fabrication process uses a high-resistivity n-type substrate with a backside process to form the p^+ layer. For thinner devices, the approach involves a low-doped $n\text{-on-}p^+$ epitaxial substrate [32].

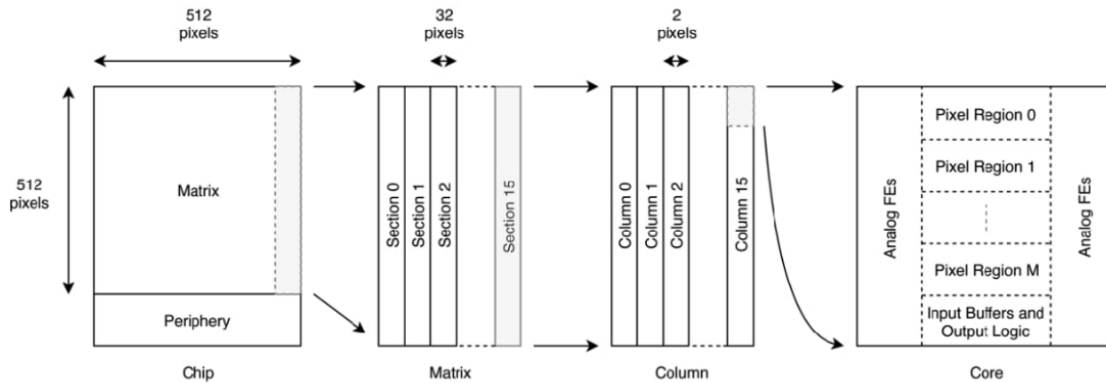


Figure 2.8: A schematic representation of the first ARCADIA demonstrator (MD1) [33]. Various engineering runs are done from 2020 to now but the structure is preserved.

The signal information flows from the clockless pixels within the matrix to the clocked periphery, where all the data processing is carried out. Figure 2.8 illustrates how the sensitive area is organized. The matrix is divided into 16 vertical sectors, each consisting of 16 columns, with each column containing a 512×2 pixel array. These columns are further divided into *cores*, each consisting of a 4×2 pixel array, which is read as a single entity, referred to as a "hitmap". The 16 sectors transmit their data to the periphery simultaneously, facilitated by the 16 serializers that are associated with each sector.

The two collaborations described above have initiated their projects to develop new technologies for academic experiments. Specifically, the INFN ARCADIA project aims to provide an innovative solution for satellite-based space sensors and future collider experiments in HEP [1]. However, the tests performed on the chips have primarily relied on radioactive sources that produce low-energy X-rays, demonstrating their applicability in the industrial and medical fields where X-ray sources are commonly used. Therefore, it is now essential to evaluate the performance of the latest chip version using a photon particle source with an energy range from 0 keV to 200 keV.

2.3 Simulation of X-rays interaction with ARCADIA

The ARCADIA chip, which builds upon the MATISSE project, is designed with enhanced scalability and ultra-low power consumption, making it an ideal candidate for chip characterization under X-ray sources. The evaluation of this sensor for radiography and CT scanning begins with simulations that assess the efficiency of silicon sensors, ranging from a few hundred μm in thickness, with and without samples to be imaged. Additionally, the simulations estimate the preferential direction of the scattered X-rays.

These simulations are conducted using the GEANT4 toolkit, which models the passage of particles through matter in various fields such as high-energy physics, nuclear and accelerator physics, as well as medical and space science [3][4][5]. GEANT4 allows the definition of key parameters to accurately describe the interactions of X-rays with silicon sensors. For this evaluation, the FD-MAPS ARCADIA prototype is modeled as a silicon layer with a thickness of 200 μm , capable of detecting electronic signals generated within it through the photoelectric effect and Compton scattering. Other interaction methods, as discussed in Chapter 1, are excluded for two main reasons: pair production and photonuclear reactions are not relevant within the energy range considered (0 – 160 keV), while Rayleigh scattering does not ionize silicon atoms.

When a sample is placed in front of the silicon layer, the interaction probability of the incident X-rays with the object is considered, accounting for the photoelectric effect and Compton scattering. Coherent scattering can also occur inside the sample, but its probability is significantly lower compared to the photoelectric effect, by 3-2 orders of magnitude as the X-ray energy increases, as shown in Figure 2.9. Therefore, Rayleigh scattering is not considered in the simulations.

The experimental setup with the real prototype chip involves positioning the sensor so that the metal read-out region is on the opposite side of the X-ray source, thus allowing the sensor to be back-illuminated. This configuration avoids any interaction of X-rays with the metal components before they reach the conversion layer and eliminates the need to consider the CMOS read-out region in the simulation. The advantage of this approach is a significant increase in the speed of the simulation, enabling the

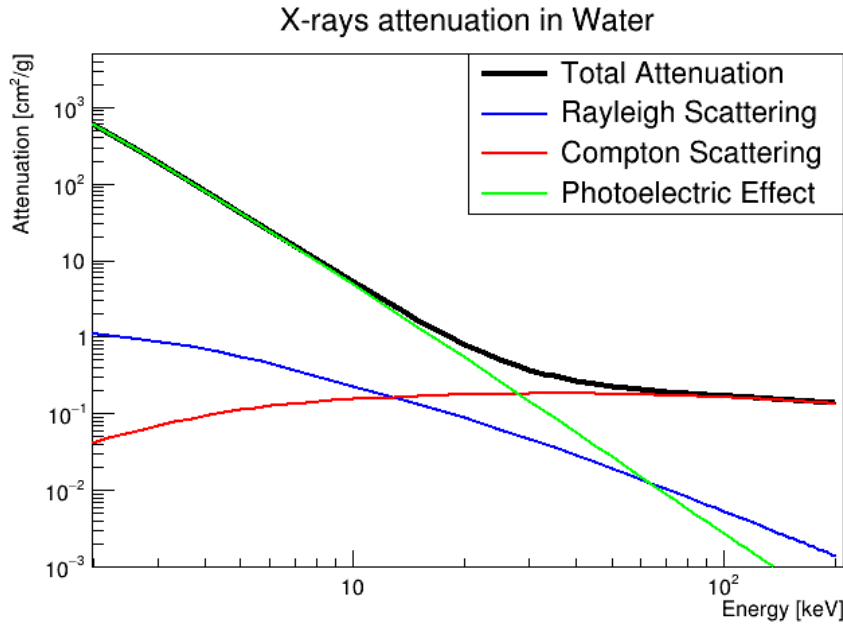


Figure 2.9: Cross section of photons in Water. The energy range is the same used in medical imaging, from 1 keV to 200 keV [34].

acquisition of high-statistics results within a few simulation days. However, the trade-off is that charge sharing between neighboring pixels is not accounted for. This is not a problem for the current evaluation, as the goal is to assess the efficiency of an ideal fully depleted silicon sensor. This approach helps establish an upper boundary for the performance that can be expected from the ARCADIA technology.

2.3.1 Efficiency

To detect the amount of X-rays interacting with the silicon sensor, I define simulations with the following features:

- **Silicon Layer and Pixel Matrix:** A square silicon layer of various thicknesses, divided into a matrix of 512×512 pixels, each with a pitch of $25 \mu\text{m}$, matching the real dimensions of the detector. The matrix is sensitive to photon interactions, recording their energy in a histogram.
- **Photon Source:** A photon source with an energy range from 0 keV to 160 keV. The beam source is a cone, originating from a point 610 mm away from the sensor. This distance is selected to facilitate a direct comparison of simulation results with the features observed in the experimental apparatus used by the company I collaborate with. The energy distribution simulates a typical beam from an X-ray tube with a tungsten anode.
- **Water Box:** A box containing water, with a $50 \text{ mm} \times 50 \text{ mm}$ area and a variable thickness ranging from 0 cm to 10 cm. The box fully covers the sensor, ensuring that only photons which either pass through or scatter from the material reach the sensor surface.

The above setup defines the physical parameters required to develop the simulation. The standard GEANT4 electromagnetic physics package is used, with the Monash University model for Compton scattering and the Livermore model for the photoelectric

effect.

The first characterization of the sensor involves using a silicon layer with a thickness of $200\text{ }\mu\text{m}$ to evaluate the number of X-rays hitting the material and those interacting inside it. Five simulations are produced, each with a different initial seed and 2×10^6 X-rays fired per simulation. This approach is used to assess statistical fluctuations in the results when the initial seed for event production is changed. The simulations are then repeated by adding one sensor at a time, with each sensor placed 1 cm apart from the previous one. The efficiency is calculated as the ratio between the number of X-rays that interact with the silicon layer and the total number of X-rays that hit the surface of the first sensor. The results are displayed in Figure 2.10, which shows the increase in efficiency as a function of the number of layers.

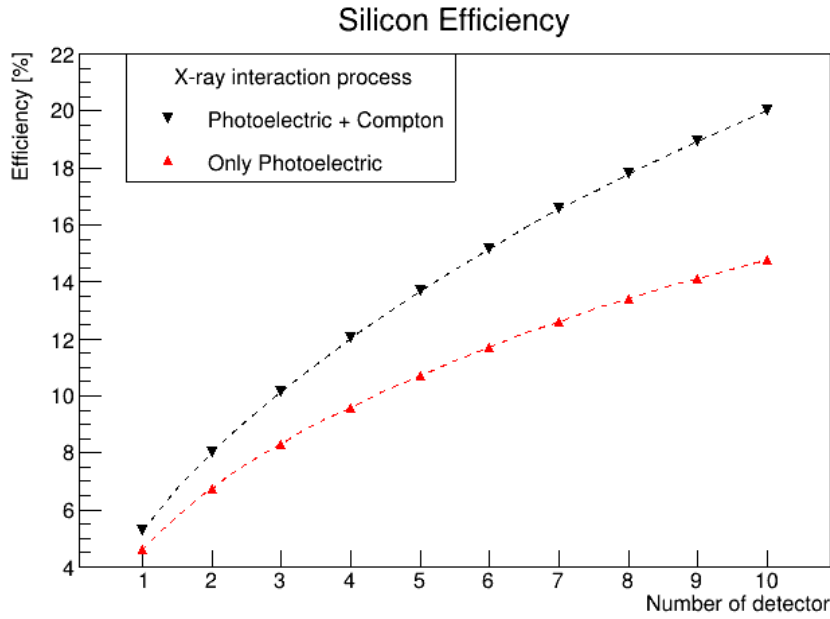


Figure 2.10: Efficiency of silicon layer as a function of the number of chip. The black triangles display the total efficiency, which considers both the photoelectric effect and Compton scattering interaction processes. If a particle scatter inside a sensor layer and change its direction, there is also the probability to observe the same event in the following layers. The red markers show the efficiency associated to X-rays that loose their entire energy in the first interaction (photoelectric effect).

The plot compares the efficiency in two cases: one considering only X-rays that release all their energy inside the silicon via the photoelectric effect, and the other considering all ionizing processes inside the silicon sensor, including Compton scattering. When a particle undergoes Compton scattering, its direction changes, and there is a certain probability that the scattered particle will interact with the same detector or another one. This causes the signal acquired by the detector to increase as the number of stacked detectors increases, reaching about 0.25 % efficiency for 10 layers, starting from 0.03 % for a single layer. The latter value represents the probability of observing two signals generated in a single sensor due to a single scattered event. The increase in total efficiency as a function of the number of detectors is not linear. For a single sensor, the efficiency is 5.3 %, but adding another layer at 1 cm distance does not double the efficiency. The efficiency increase is about 50 % when adding the second layer, and this gain diminishes to about 20 % adding the tenth layer. As the number

of detectors increases, the probability of scattering interactions also rises, since the electron density of the material increases with each additional layer. However, after an X-ray interacts with one detector, the number of surviving X-rays decreases due to the material thickness, which limits the efficiency gain from subsequent layers. Therefore, the efficiency of each additional layer becomes smaller than the first.

This simulation study provides the foundation for future simulations. It suggests that using a single device may be more efficient in terms of material usage, as the total efficiency increase with multiple layers is not linear. Furthermore, adding more layers increases the complexity of the system, requiring more read-out electronics and calibration efforts for precise positioning of the layers, which impacts the material budget.

To further evaluate the efficiency of the sensor, simulations are performed considering larger sensor widths and various sample thicknesses. Although the prototype tested so far has a total thickness of 200 μm , future implementations may increase this thickness up to 800 μm to enhance efficiency with a single detector. The goal is to assess whether the efficiency increases proportionally to the sensor width and to investigate how the efficiency is affected when a sample with varying thicknesses is placed in front of the sensor. For each condition, five simulations are conducted, each with a different initial seed, resulting in a total of 1×10^7 X-rays considered for each case of sensor width and sample thickness. The multiple acquisitions are designed to evaluate the statistical fluctuations resulting from different initial seeds used for event production. This approach will help determine the relationship between the sensor's size and its efficiency and provide insights into the impact of sample thickness on the sensor's performance. The results will guide decisions on whether increasing the sensor thickness is a beneficial strategy for enhancing detection efficiency in practical applications.

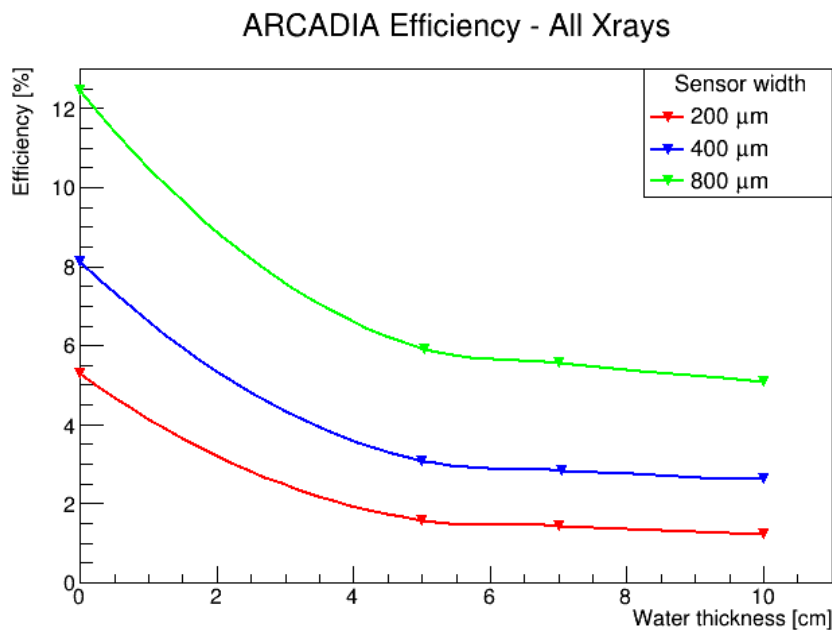


Figure 2.11: Simulation of the total signal acquired by a pixelated silicon detector when a sample is placed in front of it. Three sensor thicknesses are considered: 200 μm , 400 μm and 800 μm . The efficiency increases as a function of sensor width and decreases as a function of water object thickness with the tendency to reach a constant value after 6 cm.

The simulations were conducted for three different silicon thicknesses (200 μm , 400 μm and 800 μm) and four different water object depths (0 mm, 50 mm, 70 mm and 100 mm). The results are shown in Figure 2.11. The key findings from these simulations include:

- Non-linear efficiency increase: Similar to previous simulations involving stacked layers, the increase in silicon width does not result in a linear increase in efficiency. However, in this case, the efficiency is evaluated for a single device, which hosts all necessary electronics.
- Decreasing efficiency with water object thickness: As expected, the efficiency decreases as the water object depth increases. This is due to the attenuation of the X-rays as they pass through the material. While the attenuation curve is expected to follow an exponential trend based on the material's width and the X-ray energy, the data are not explicitly fitted to an exponential model. This is because the simulation tracks the interaction of X-rays with the detector rather than recording the precise energy of the incoming photons. The plotted points are connected to visually highlight the comparison between the three silicon thicknesses.

In the context of medical imaging, an image is reconstructed using a back-projection algorithm based on the X-ray beam attenuation process. In this process, the amount of X-rays that interact with the material and are detected depends on the material's width, density, and the energy of the incoming X-rays. This mathematical relation considers only photoelectric interaction mechanism and follows an exponential attenuation curve. However, in real experiments, photons with energies up to 160 keV also interact via Compton scattering with materials like water. This scattering process contributes to the detection signal, and Figure 2.12 presents the percentage of X-rays that scatter inside the water object before being detected by the silicon sensor.

The simulations for different silicon sensor thicknesses and water object depths reveal that the amount of scattered photons remains nearly constant for each sensor thickness. However, when the sensor width is increased, as in the previous case of multiple layers stacked together, the number of scattered photons detected by the sensor increases. In these simulations, considering a 12.8 mm $\times$ 12.8 mm sensor area, for each sensor width, up to 5 % of the total detected X-rays result from scattering within the water object.

As previously mentioned, scattering X-rays pose a challenge in imaging reconstruction. This is because the photons that scatter inside the sample and are detected by the sensor do not provide accurate spatial information. The position where these scattered photons are detected deviates from the initial incoming direction, which distorts the image. To mitigate this issue, I decided to explore whether it is possible to reduce the amount of scattered X-rays detected after a Compton scattering event inside the sample. The goal is to evaluate potential geometrical solutions that do not require the addition of specific equipment such as anti-scattering grids. Anti-scattering grids are often used in medical imaging to absorb scattered photons before they reach the detector, but this approach may not always be feasible or cost-effective in every scenario.

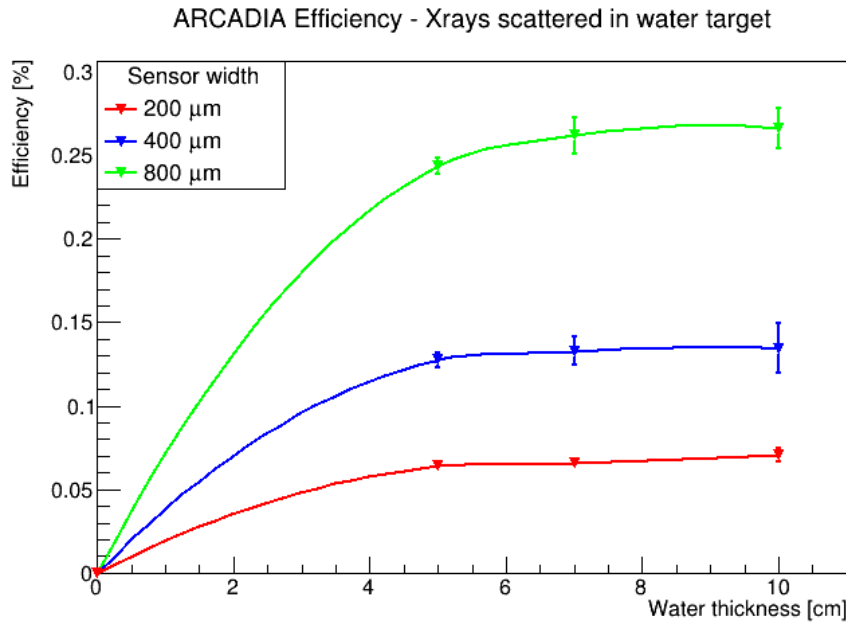


Figure 2.12: Simulation of the amount of signal acquired when X-rays scatter from a sample placed in front of the pixelated silicon detector. Three sensor thicknesses are considered: 200 μm , 400 μm and 800 μm . Considering one sensor width at a time, the amount of scattered X-rays remains almost constant for the three different water thicknesses considered.

2.3.2 Compton scattering angular distribution

The photoelectric effect and Compton scattering are the primary interaction processes within the X-ray energy range. Two distinct energy regions can be identified: one where a single of these processes is dominant and another where both overlap. These energy ranges vary depending on the material involved. For example, when interacting with lead, the Compton effect is negligible, while in the case of carbon, the probability of observing the Compton effect is approximately 100 % within the energy range from 100 keV to 10 MeV.

In X-ray imaging, scattering is an undesirable interaction because it introduces artifacts into the image reconstruction. The study of scattering dynamics began after the discovery of the electron, when scattering experiments on atoms revealed significant deviations from the cross-section predictions made by J.J. Thomson. Consequently, further research into X-ray radiation scattering, incorporating relativistic and quantum mechanical effects and assuming target electrons to be initially free and at rest, led to the development of an analytical expression for the differential cross-section. This expression was derived by Klein and Nishina in 1928 [2] using the formalism of Quantum Electro-Dynamics (QED)⁴; it is one of the first successes of QED in explaining experimental observations of high-energy photon interactions.

The Klein-Nishina formula takes into account the initial and final wavelengths, denoted as λ and λ' , as well as the scattering angle θ of the X-ray interacting with an

⁴QED is built on the principles of special relativity and quantum field theory: the foundation lies in the Dirac equation (describing relativistic electrons) and Maxwell's equations, combined in a quantum field framework.

incoherent free electron:

$$\left(\frac{d\sigma}{d\Omega}\right)_{KN} = \frac{r_0^2}{2} \left(\frac{\lambda}{\lambda'}\right)^2 \left\{ \frac{\lambda}{\lambda'} + \frac{\lambda'}{\lambda} - \sin^2\theta \right\} \quad (2.4)$$

where r_0 is the classical electron *radius* ⁵

$$r_0 = \frac{e^2}{4\pi\epsilon_0 m_e c^2} = 2.82 \text{ fm}$$

The scattering wavelength calculated from the Compton scattering formula results:

$$\Delta\lambda = \lambda' - \lambda = \frac{h}{m_e c} (1 - \cos\theta) \quad (2.5)$$

The result of the Klein-Nishina formula is shown in Figure 2.13. The differential cross-sections for four different energies are presented, demonstrating a trend where the angular range in which the photon can be scattered decreases as the energy increases. For X-ray energies in the range of a few keV, the angular distribution is symmetric relative to the direction perpendicular to the incoming particle. As the energy increases, however, this symmetry begins to break, leading to an increasing imbalance towards the forward direction when the energy exceeds a few hundred keV.

In real scenarios, X-rays interact with electrons bound to the atomic nucleus of the selected material. Laboratory experiments yield different results compared to the theoretical Klein-Nishina formula, as shown in Figure 2.14. This variation in shape is attributed to the effects of electron binding. To account for these effects, the Klein-Nishina differential cross-section formula can be corrected by including the Incoherent Scattering Functions $S(x, Z)$, where Z is the atomic number of the target material, x is the *momentum transfer parameter* ⁶:

$$x = \frac{\sin(\theta/2)}{\lambda(\text{\AA})}$$

and $\lambda(\text{\AA})$ is the photon wavelength in angstroms

$$\lambda(\text{\AA}) = \frac{hc}{E} = 12398.52/E(\text{eV})$$

Thus, the expression becomes:

$$\frac{d\sigma}{d\Omega} = \left(\frac{d\sigma}{d\Omega}\right)_{KN} \times S(x, Z) \quad (2.6)$$

Figure 2.15 illustrates the Incoherent Scattering Function $S(v)$ for silicon ($Z = 14$), using a new variable $v(x, Z) = 4\pi x(a_0/3)Z^{-2/3}$, where $a_0 = 0.53\text{\AA}$ is the Bohr

⁵In QED, the electron is a point particle, a point charge with no spatial extent. However r_0 is a combination of fundamental physical quantities that defines a length scale that characterizes electron interactions in atomic-scale problems.

⁶The module of the momentum transfer vector is $|\mathbf{q}| = 2\hbar |\mathbf{k}| \sqrt{\frac{1-\cos(\theta)}{2}} = 4\pi\hbar \frac{\sin(\theta/2)}{\lambda}$

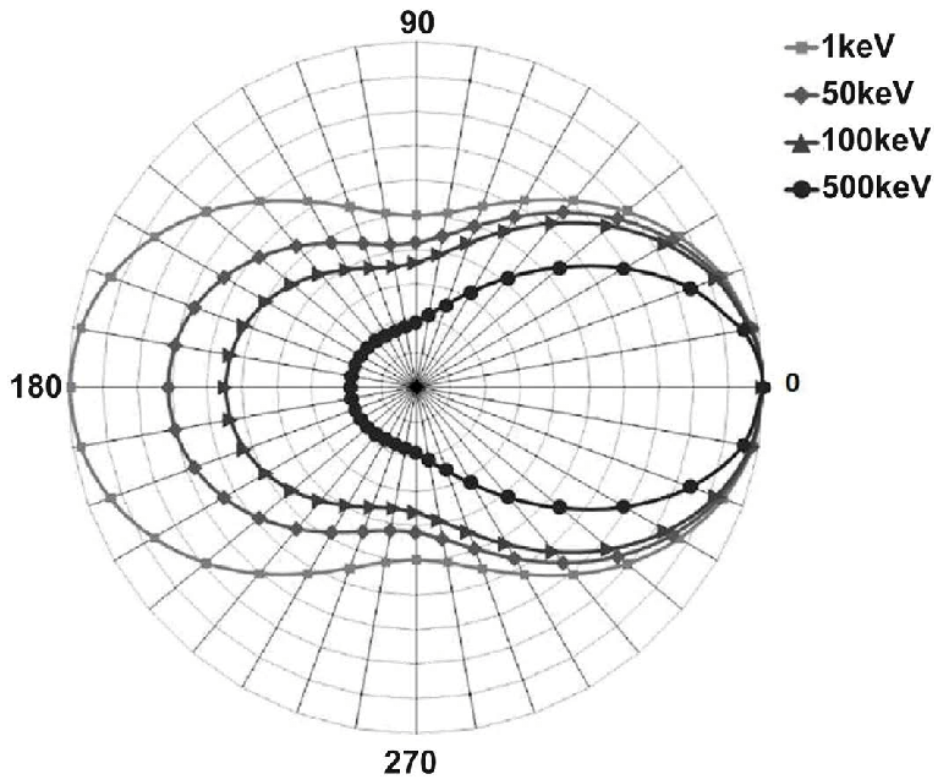


Figure 2.13: Differential cross section as a function of scattering angle and incoming photon energy, obtained with the free electron Klein-Nishina formula. Four energies are considered, in order to cover the range used for interventional medical procedures [35].

radius⁷. By examining the differential cross-section, it is evident that for high values of the momentum transfer v , the Incoherent Scattering Function approaches the number of electrons in the atom, causing the modified Klein-Nishina formula to approximate the version assuming free and at-rest electrons. The significant difference occurs at small scattering angles, where the Incoherent Scattering Function causes the differential cross-section to decrease toward zero as the momentum transfer diminishes.

In crystallographic and chemical physics literature, tables of the Incoherent Scattering Functions $S(x, Z)$ are constructed from theoretical data derived by combining both relativistic and non-relativistic expressions for the atomic form factors $F(x, Z)$.

The calculations of the atomic form factor $F(x, Z)$ and the Incoherent Scattering Functions $S(x, Z)$ reported in Table 2.1 are approximations of the full theoretical treatments. These approximations do not account for the effects of electron pre-collision motion and are primarily valid for small scattering angles [36]. However, the energy range considered extends up to 200 keV, which excludes the consideration of additional scattering processes, nuclear resonances, and proton Compton scattering, which are not described in the tables. Table 2.1 presents the values for three elements:

- Hydrogen
- Oxygen
- Silicon

To estimate the modified Klein-Nishina expression for electron binding effects when an

⁷In terms of fundamental constants, the Bohr radius is expressed as $a_0 = \frac{\hbar^2}{m_e e^2} = 0.52917706 \text{ \AA}$.

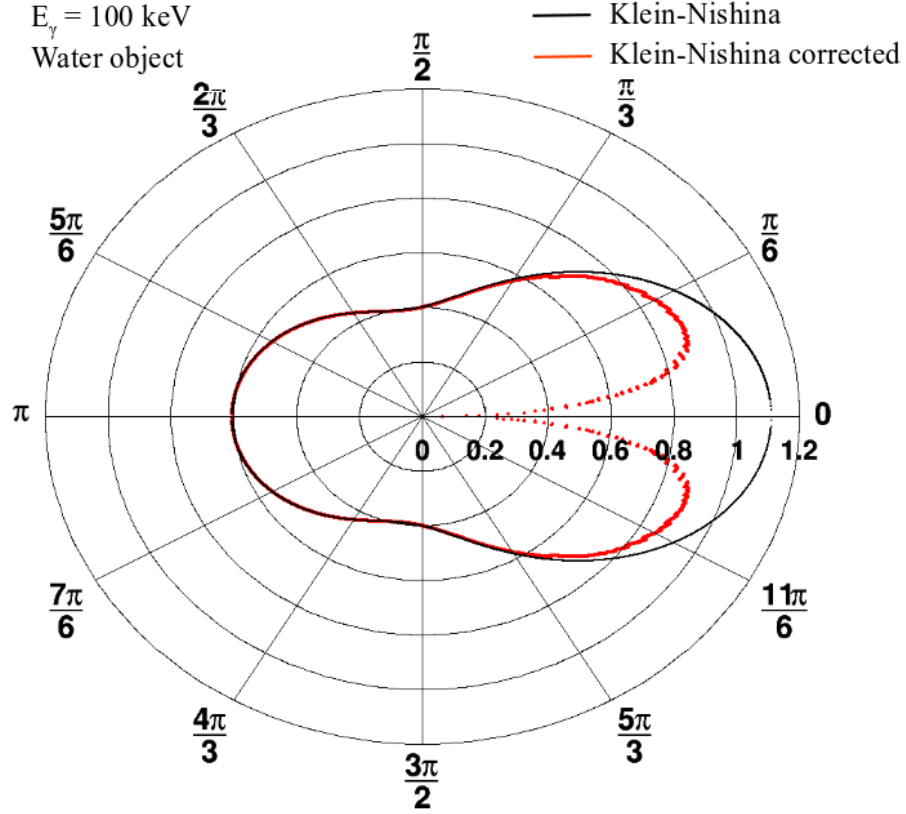


Figure 2.14: Comparison of the differential cross section as a function of scattering angle obtained with the free electron Klein-Nishina formula (black) and the Klein-Nishina corrected one with the incoherent scattering factors (red) for an X-ray beam of 100 keV on a sphere of water. The rectilinear axis of the polar graph represents the differential cross section in barn units.

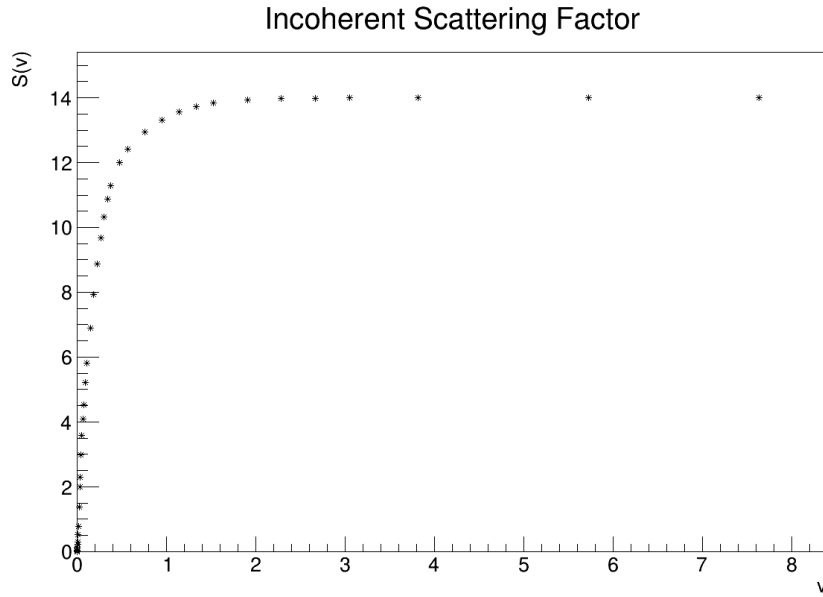


Figure 2.15: Incoherent scattering function $S(x)$ for Silicon ($Z = 14$) as a function $v(x)$. Values taken from [36].

x	HYDROGEN		OXYGEN		SILICON	
	$F(x, Z)$	$S(x, Z)$	$F(x, Z)$	$S(x, Z)$	$F(x, Z)$	$S(x, Z)$
0.00	1.0	0.0	8.0	0.0	14.0	0.0
$5.0 \cdot 10^{-3}$	$9.99 \cdot 10^{-1}$	$1.10 \cdot 10^{-3}$	7.99	$3.00 \cdot 10^{-3}$	13.99	$9.00 \cdot 10^{-3}$
$1.0 \cdot 10^{-2}$	$9.98 \cdot 10^{-1}$	$4.41 \cdot 10^{-3}$	7.98	$1.10 \cdot 10^{-2}$	13.97	$3.50 \cdot 10^{-2}$
$1.5 \cdot 10^{-2}$	$9.95 \cdot 10^{-1}$	$9.89 \cdot 10^{-3}$	7.97	$2.53 \cdot 10^{-2}$	13.95	$7.90 \cdot 10^{-1}$
$2.0 \cdot 10^{-2}$	$9.91 \cdot 10^{-1}$	$1.75 \cdot 10^{-2}$	7.96	$4.48 \cdot 10^{-2}$	13.91	$1.39 \cdot 10^{-1}$
$2.5 \cdot 10^{-2}$	$9.86 \cdot 10^{-1}$	$2.72 \cdot 10^{-2}$	7.95	$6.98 \cdot 10^{-2}$	13.85	$2.15 \cdot 10^{-1}$
$3.0 \cdot 10^{-2}$	$9.80 \cdot 10^{-1}$	$3.88 \cdot 10^{-2}$	7.93	$1.00 \cdot 10^{-1}$	13.79	$3.05 \cdot 10^{-1}$
$4.0 \cdot 10^{-2}$	$9.66 \cdot 10^{-1}$	$6.77 \cdot 10^{-2}$	7.87	$1.76 \cdot 10^{-1}$	13.63	$5.23 \cdot 10^{-1}$
$5.0 \cdot 10^{-2}$	$9.47 \cdot 10^{-1}$	$1.03 \cdot 10^{-1}$	7.80	$2.71 \cdot 10^{-1}$	13.44	$7.82 \cdot 10^{-1}$
$7.0 \cdot 10^{-2}$	$9.00 \cdot 10^{-1}$	$1.90 \cdot 10^{-1}$	7.61	$5.14 \cdot 10^{-1}$	12.97	1.37
$9.0 \cdot 10^{-2}$	$8.42 \cdot 10^{-1}$	$2.90 \cdot 10^{-1}$	7.38	$8.12 \cdot 10^{-1}$	12.42	1.99
$1.0 \cdot 10^{-1}$	$8.11 \cdot 10^{-1}$	$3.43 \cdot 10^{-1}$	7.24	$9.77 \cdot 10^{-1}$	12.14	2.29
$1.5 \cdot 10^{-1}$	$6.41 \cdot 10^{-1}$	$5.89 \cdot 10^{-1}$	6.47	1.89	10.77	3.59
$2.0 \cdot 10^{-1}$	$4.81 \cdot 10^{-1}$	$7.69 \cdot 10^{-1}$	5.62	2.80	9.67	4.52
$3.0 \cdot 10^{-1}$	$2.51 \cdot 10^{-1}$	$9.37 \cdot 10^{-1}$	4.09	4.29	8.22	5.81
$5.0 \cdot 10^{-1}$	$7.06 \cdot 10^{-2}$	$9.95 \cdot 10^{-1}$	2.34	5.83	6.23	7.94
$7.0 \cdot 10^{-1}$	$2.43 \cdot 10^{-2}$	$9.99 \cdot 10^{-1}$	1.71	6.41	4.46	9.67
1.0	$6.88 \cdot 10^{-3}$	$10.0 \cdot 10^{-1}$	1.38	6.90	2.69	11.29
2.0	$4.89 \cdot 10^{-4}$	1.0	$6.72 \cdot 10^{-1}$	7.76	1.26	12.94
5.0	$1.30 \cdot 10^{-5}$	1.0	$6.69 \cdot 10^{-2}$	7.99	$3.45 \cdot 10^{-1}$	13.94
15.0	$1.62 \cdot 10^{-7}$	1.0	$1.29 \cdot 10^{-3}$	8.0	$1.11 \cdot 10^{-2}$	14.0

Table 2.1: Tabulation of Atomic Form Factor $F(x, Z)$ and Incoherent Scattering Functions $S(x, Z)$ for materials useful for my studies (Hydrogen, Oxygen and Silicon)

X-ray beam interacts with a material, the differential cross-sections of the individual elements can be combined, weighted by the number of atoms that make up the compound. The atomic form factors and Incoherent Scattering Functions for hydrogen and oxygen are presented here, as their combination is useful for determining values for the water compound. Silicon is also included for completeness, as these three elements are the most commonly used in simulations.

The processes and equations described above are of primary interest in the fields of medical and industrial imaging. The resolution of a reconstructed image depends not only on the efficiency of the detectors but also on the amount of X-rays that scatter inside the sample and are subsequently detected. To determine this value, a GEANT4 simulation is implemented. The geometry of the simulation consists of:

- A mono-energetic pencil beam of X-rays.
- A sphere of 1 mm radius. A single element, silicon, is selected in order both to simplify the results, avoiding the calculation of the atomic form factor and the Incoherent Scattering Factor as a combination of two elements, like for water, and to increase the scattering probability, thanks to the higher atomic number.
- A spherical shell of 1 mm air around the silicon sphere, placed at a distance of 10 mm from the center.

Figure 2.16 illustrates the geometrical setup of the simulation. The air shell that

surrounds the silicon sphere is used to detect X-rays scattered from the sphere. This configuration allows for the detection of scattered X-rays across all possible emission angles.

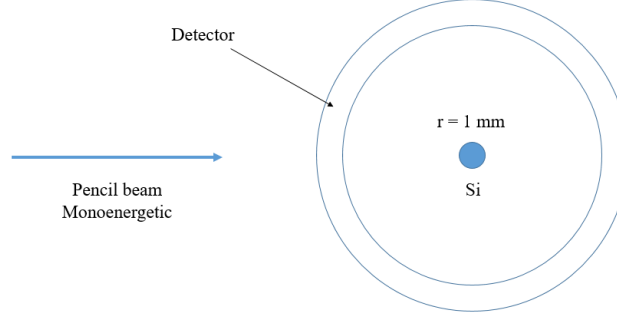


Figure 2.16: Scheme of the simulation geometry to evaluate the angular distribution of the X-rays that do Compton scattering.

The silicon sphere is chosen to be small enough to ensure that only a single interaction occurs within it. The interaction point inside the silicon sphere and the corresponding point in the air shell are recorded to calculate the scattering angle relative to the direction of the incoming particle. The scattered angles of X-rays that change direction and reach the 4π detector are shown in Figure 2.17a and Figure 2.17b for two different energies: 100 keV and 511 keV. The data (black points connected by a segmented line) are fitted using the modified Klein-Nishina formula, demonstrating excellent agreement.

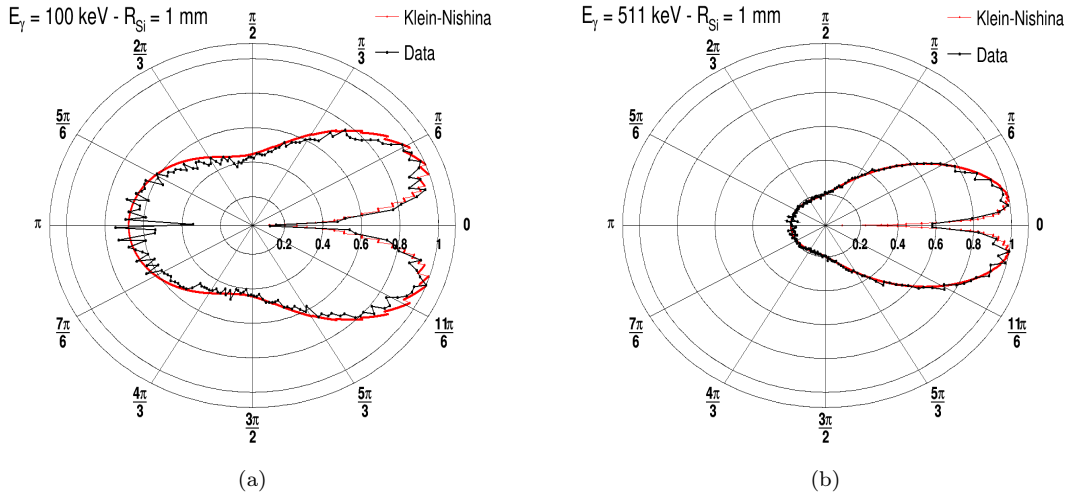


Figure 2.17: Comparison of the differential cross sections obtained with the Klein-Nishina formula corrected for the Incoherent Scattering Factor (red) and the simulation result (black) for X-rays with incident energy of (a) 100 keV and (b) 511 keV.

The 3D results obtained from the simulations are represented in Figure 2.17 as 2D plots. These plots are generated by considering only the arc-tangent of the two orthogonal dimensions relative to the incoming photon direction. To visualize the complete 3D distribution, the 2D plot can be rotated by 180° around the central axis of the image, which spans from 0 to π rad.

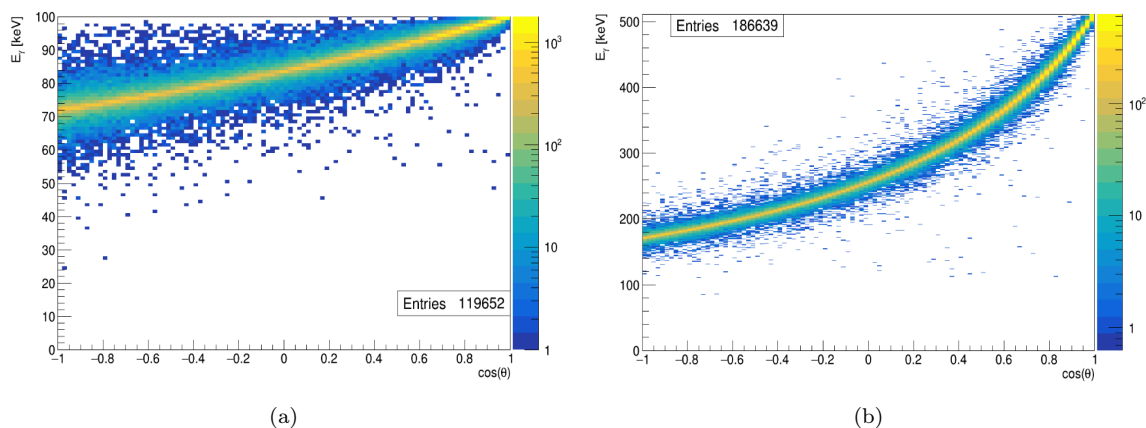


Figure 2.18: Energy of the X-rays scattered from the silicon sphere starting from a beam of (a) 100 keV and (b) 511 keV as a function of the cosine of the scattered angle.

The plots in Figure 2.18 depict the energy of scattered X-rays as a function of the scattering angle, represented by its cosine. These plots reveal a non-linear relationship between the scattered energy and the angle for both initial energies considered. The behavior aligns with the Compton formula presented in Equation 2.5, which shows a direct proportionality between the wavelength difference, corresponding to the subtraction of the reciprocals of the energies, and the cosine of the scattering angle.

The analysis yields two key findings: first, the angular deviation caused by Compton scattering is inversely proportional to the incident energy of the X-rays; second, the minimum amount of scattered X-rays occurs at an angle of 90° relative to the incoming photon direction, while no photon re-emission is observed along the incident path. These insights are crucial when designing a new apparatus. Careful consideration should be given to the placement of detectors in regions where scattered X-rays are almost absent. Moreover, it might be possible to reduce the material budget by omitting the purchase of anti-scatter grids, provided the design of the system can compensate for their absence.

Simulation studies on efficiency and scattering probability demonstrate the feasibility of using Fully Depleted MAPS for X-ray imaging applications.

MiCROTEC offers various CT scanner models for two key industries: sawmills and food processing. Sawmill scanners require a large field of view to accommodate entire logs passing through the system, while food industry scanners deal with smaller samples, necessitating reduced field of view, lower X-ray energy, and thinner sensors. The efficiency of the silicon sensor under investigation poses a challenge for both industries. For food industry applications, where an efficiency of at least 15% is required to produce CT images suitable for quality control, this necessitates stacking 4–5 silicon sensors of $200\text{ }\mu\text{m}$ thickness each, or fewer sensors of greater thickness. Additionally, current industrial systems are nearly insensitive to X-rays below 20 keV. In contrast, the FD-MAPS photon counter can discriminate photons releasing as little as 3 keV energy in a pixel by setting an appropriate threshold. This advantage, combined with the reduced material budget of the FD-MAPS technology, enables the acquisition of multiple sensor layers without necessitating the development of an expensive, fully integrated system. Consequently, silicon detectors emerge as a promising solution for future scanner systems.

To further explore their potential, characterization experiments with various X-ray sources are underway. These include radioactive sources from the Padova University laboratory and X-ray beams generated by X-ray tubes at different voltages from the industrial environment. The prototype under study, an ARCADIA FD-MAPS chip, features a thickness of 200 μm and operates with a reverse bias of 90 V.

3 Industrial Applications

In the industrial sector, CT scanners are widely used as automated, non-contact inspection systems during both the initial and final stages of production. These systems, along with their reconstruction algorithms, are continually evolving to enhance performance, cost efficiency, and market competitiveness. Among these advancements, improvements in X-ray detectors play a pivotal role, as their detection efficiency and resolution are critical factors that determine the overall reconstruction performance of a CT system.

Today, scanner systems utilizing X-ray beams are employed across various industries. Applications include custom scanners at airports, equipment for investigating cultural heritage artifacts, detection of foreign objects in food processing, and even medical imaging.

MiCROTEC, the company I collaborate with, specializes in scanner solutions for the sawmill and food industries, providing non-destructive testing for products. The company defines itself as:

"the global technology and market leader in intelligent wood characteristics detection to maximize recovery for the wood processing industry" which "brings excellent service and world-leading scanning solutions to the complete wood processing industry."⁸

A spin-off of this group, BIOMETiC, was founded in 2012 to apply MiCROTEC's innovative technology in the field of food scanning. It offers:

"innovative tools and solutions based on advanced technologies, such as tomography and X-ray, for the automation and optimization of production processes in the food industry that increase the added value of the final product" in order "to guarantee to the final consumer a safe and high-quality product."⁹

3.1 Computed Tomography

In industrial environments, the decision to purchase a new, useful system is not solely based on process optimization but also takes other factors into account. For example, the new technology must integrate seamlessly into the production line, maintaining the same speed as other processes to ensure consistent productivity. As a result, the adoption of new technology in industrial settings is not immediate. It depends on continuous advancements over the years, considering factors such as budget constraints, ease of component fabrication, and the durability of the final product. This explains why CT scanners were introduced to the industrial sector only about 30 years after their initial use in the medical field. Today, this technology is applied in various industries, ranging from non-destructive defect analysis (such as detecting cracks, porosity, and inclusions) to evaluating 3D sample geometry and inspecting the

⁸<https://www.microtec.eu>

⁹<https://www.biometric.com>

assembly of components to prevent critical issues.

Since the late 1990s, industrial CT has been primarily used to identify defective areas in a 3D virtual environment, aiming to improve and optimize component production. The technology used in industrial CT closely followed the advancements made in medical diagnostic imaging, beginning with the first generation of medical CT scanners. The earliest industrial CT scanners featured a simple configuration, where a single detector was positioned opposite the X-ray source. The scanning system would first translate to cover the entire width of the sample, and then rotate. This process was repeated until all necessary angles were captured to reconstruct the 3D image. The use of a single detector and a pencil beam source had the advantage of completely eliminating scattering X-rays, but it resulted in very long scan times. In an industrial setting, while this long duration does not pose health risks as it does in medical applications, it slows down the production line, reducing its efficiency.

The need to speed up the scanning process led to the adoption of newer scanner technologies. In medical CT, technological advancements allowed for the simultaneous irradiation of multiple detectors with a cone beam source, followed by fast data acquisition and analysis. The latest generation of CT scanners, known as Multiple-row Detector CT (MDCT), can detect data from multiple parallel planes at once. It uses a single X-ray tube that emits photons in a cone-shaped beam. Multiple arrays of detectors are read simultaneously while the X-ray tube continuously rotates, and the sample moves along a plane perpendicular to the direction of the X-ray emission. This setup enables the rapid production of 3D images, although the scanners themselves are significantly more expensive.

As scanner technology has evolved over the years, the fundamental detection process has remained largely unchanged, focusing on evaluating the amount of X-rays that reach the detector. By analyzing the energy distributions of X-rays before and after interacting with the sample, it is possible to assess the average density the photons pass through, which helps distinguish different materials.

A quantitative description of this phenomenon begins with the concept of interaction cross-section, as explained in Chapter 1. In the case of a complete CT system, where a poly-energetic cone beam source interacts with a non-homogeneous sample, the formula for calculating the X-ray beam intensity when photons interact with the detector depends on both the initial energy distribution of the beam and the densities of the materials that the photons encounter:

$$I_d = \int_0^{E_{max}} S_0(E) E \exp \left[- \int_0^d \mu(s; E) ds \right] dE \quad (3.1)$$

with I_d the intensity of the beam at the detection point, E_{max} the maximum energy of the incoming photon beam, $S_0(E)$ the initial energy distribution, s the distance between the X-ray tube and the detection point and $\mu(s; E)$ the linear attenuation coefficients of the materials crossed along the line segment s between the source and the detector for a specific energy E .

The energy integration cannot be solved directly with the information obtained from the detector, so a simplification regarding the energy must be applied. The integral over multiple energies is approximated by a single effective energy value (E_{eff}), which corresponds to the amount of energy in a substance that produces the same

observed intensity as when measured with a mono-energetic source. Equation 3.1 can be rearranged to:

$$I_d = I_0 \exp \left[- \int_0^d \mu_{eff}(s; E_{eff}) ds \right]$$

so that it is possible to evaluate the basic measurement g_d of a CT scanner, equal to the line integral of the linear attenuation coefficient at the effective energy value:

$$g_d(s) = -\ln \left(\frac{I_d}{I_0} \right) = \int_0^d \mu_{eff}(s; E_{eff}) ds \quad (3.2)$$

The effective energy value aids in evaluating the total attenuation coefficient along the ray path, but it also introduces artifacts into the results. Attenuation generally decreases with higher X-ray energy, while it increases with the amount of material crossed. Using a single effective energy value, instead of a full energy distribution, leads to the opposite effect, resulting in a reduction of the effective attenuation coefficient when a large volume of material is traversed. Common artifacts in the final image produced by this phenomenon include *cupping* and *scattering*. *Cupping* is a beam hardening effect caused by the absorption of low-energy X-rays as they pass through thick substances, resulting in an effective energy that is higher than the simple mean of the initial energy distribution. In radiography, the *cupping* artifact appears as stripes around the edges of high attenuation areas, along with an overestimation of the effective attenuation values in the centroids of the continuous phase. On the other hand, the *scattering* effect is related to the Compton scattering interaction process and leads to an offset in the intensity values detected at certain points.

The 2D image of a sample's cross-section can be mathematically reconstructed if the beam intensity at the detection point is measured and the effective attenuation coefficient can be inverted from I_d . Considering a parallel beam source originating from a line source and the linear attenuation coefficient along an X-ray path, the primary coordinates involved are: the distance of the rays from the line passing through the detector's central position (l), the orthogonal axis (s), and the rotation angle (θ). Referring to Figure 3.1, a line parallel to the s -axis can be defined by:

$$L(l, \theta) = \{(x, y) | x \cos \theta + y \sin \theta = l\}$$

with the rectilinear coordinates defined as a rotation of the (l, s) plane:

$$\begin{cases} x(s) = l \cos \theta - s \sin \theta \\ y(s) = l \sin \theta + s \cos \theta \end{cases} \quad (3.3)$$

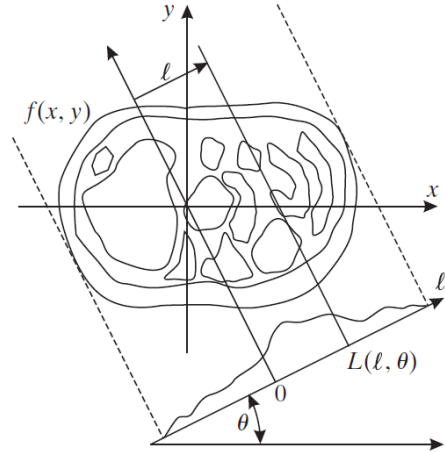


Figure 3.1: Example of a projection of a cross section. The natural reference system corresponds to (l, s) plane with l orthogonal to the rays emission direction (s). The system is rotated of θ angle w.r.t. (x, y) plane, following the source-detector rotation. The 2D image $f(x, y)$ in rectilinear coordinates is transformed into a line integral distribution along the l -axis [7].

The distribution of the attenuation coefficients of the sample along the l -axis represents the values of $f(x, y)$ at the points crossed by the line $L(l, \theta)$, multiplied by the 2D impulse function along the same line. Consequently, the function $g_d(s)$ can be expressed as:

$$\begin{aligned} g(l, \theta) &= \int_{-\infty}^{+\infty} f(x(s), y(s)) ds \\ &= \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} f(x, y) \delta(x \cos \theta + y \sin \theta - l) dx dy \end{aligned} \quad (3.4)$$

The line integral for each angle θ is known as a *projection*, and $g(l, \theta)$ represents the 2D Radon Transform of $f(x, y)$.

The concept of the 2D Radon Transform for image reconstruction was introduced in 1917 by the mathematician Johann Radon and further advanced in 1956 by R. N. Bracewell [37]. The Radon space represents the line integral values of the object's distribution and is characterized by the following properties:

- Periodicity in θ : $g(l, \theta) = g(l, \theta + 2n\pi)$ with $n \in \mathbb{N}$
- Symmetry: $g(l, \theta) = g(-l, \theta + \pi)$
- A point in Radon space represents a line in object space
- A point in the object space is described by a sinusoidal curve in Radon space

Therefore, tomography requires a uniformly sampled Radon space of the object's distribution [37]. Based on this concept, the Radon Transform can be directly visualized as a *sinogram*, as shown in Figure 3.2b.

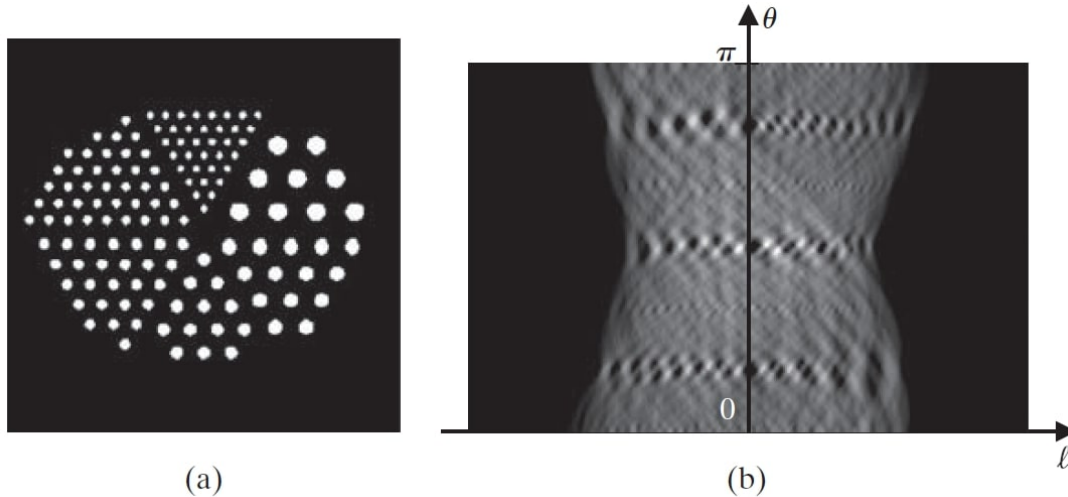


Figure 3.2: (a) a 2D image, resembling a cross section, and (b) its 2D Radon Transform, visualized as a sinogram [7].

The name *sinogram* derives from the sinusoidal path a point in the 2D image follows when plotted in the (l, θ) plane. In Figure 3.2b, in fact, it is possible to observe oblique lines that run through the sinogram from the bottom row equal to a 0 rad to a π rad angle projections. The scanning rotation stops at 180° in order to avoid redundant information.

To advance the mathematical analysis of the image reconstruction problem, the functionality of the CT scanner system must satisfy the following three conditions:

1. Attenuation: The incident particles must propagate in straight paths through the sample to ensure the system deals exclusively with transmission tomography
2. Linearity: The data must represent the integral values of the sample function along straight lines
3. Dense and Homogeneous Radon Space Population: The line integral measurements must be sufficiently dense and uniformly spaced to enable the reconstruction of high-resolution images

The effective attenuation coefficient of a pixel can be calculated as the sum of the attenuation coefficients obtained from the *projections* at different angles, considering only the ray passing through that pixel. However, since the contribution of each pixel along a ray is unknown, the initial assumption is that all pixels contribute equally. This approach is known as *direct back-projection*. Despite its simplicity, it is rarely used in image reconstruction because the integral approximation leads to a loss of structural detail and introduces blurriness in the resulting image.

To address these issues, a refined approach builds upon the concept of *direct back-projection* but incorporates filtering to mitigate blurriness. This improved method is derived through a mathematical analysis of the problem, starting with the inversion of the 2D Radon Transform. A key principle in this analysis is that the 1D Fourier Transform of the Radon Transform at a given angle θ corresponds to the 2D Fourier Transform of the effective attenuation coefficients in rectilinear coordinates.

$$\begin{aligned} {}^1D \mathcal{F}_l[g(l, \theta)](u = \omega, \theta) &= {}^2D \mathcal{F}_{(x,y)}[\mu_{eff}(x, y)](u = \omega \cos \theta, v = \omega \sin \theta) \\ G(\omega, \theta) &= M(\omega \cos \theta, \omega \sin \theta) \end{aligned}$$

This equation, known as the *Fourier Slice Theorem* or *Projection Slice Theorem*, is highly valuable for image reconstruction. Specifically, the 2D Fourier Transform of an image can be determined through an inversion process if the Radon Transform is measured for all angles θ within the range 0° to 180° .

The effective attenuation coefficient values in the object space are then computed using the inverse Fourier Transform of $M(u, v)$. By substituting the general (u, v) coordinates in Fourier space with $u = \omega \cos \theta$ and $v = \omega \sin \theta$, the Jacobian for the change of variables becomes $|\omega|$. Consequently, the integral over the variable θ is restricted to the interval $[0, \pi]$:

$$\begin{aligned} \mu_{eff}(x, y) &= \frac{1}{(2\pi)^2} \int_0^\pi \int_{-\infty}^{+\infty} M(\omega \cos \theta, \omega \sin \theta) \exp\{j\omega(x \cos \theta + y \sin \theta)\} |\omega| d\omega d\theta \\ &= \frac{1}{(2\pi)^2} \int_0^\pi \int_{-\infty}^{+\infty} G(\omega, \theta) \exp\{j\omega(x \cos \theta + y \sin \theta)\} |\omega| d\omega d\theta \\ &= \frac{1}{2\pi} \int_0^\pi \left[\frac{1}{2\pi} \int_{-\infty}^{+\infty} G(\omega, \theta) |\omega| \exp\{j\omega l\} d\omega \right] d\theta \\ &= \frac{1}{2\pi} \int_0^\pi {}^1D \mathcal{F}_\omega^{-1} [G(\omega, \theta) |\omega|] (l, \theta) d\theta \end{aligned}$$

The result represents a filtered version of the *direct back-projection*, which effectively eliminates the blurriness found in the unfiltered approach. In practical implementation, a high-pass filter is applied during the inverse Fourier Transform of the Fourier Transform of the Radon Transform. This filter enhances the reconstruction by preserving high-frequency components, ensuring that the edges of the object are reconstructed with high resolution.

Thus, this provides the mathematical framework for image reconstruction of a cross-section using parallel rays under the properties of Radon space. From an algorithmic perspective, the effective attenuation coefficient can be evaluated as a dense set of point values, each assigned a constant value given by $\mu_{eff}(x, y)$. The convolution of the point function with the effective attenuation coefficient in Fourier space corresponds to simple multiplication. In this case, the Fourier transform $M(\omega \cos \theta, \omega \sin \theta)$ represents the ratio between the Fourier transform of the full data and the Fourier transform of the point function. This straightforward description poses a challenge, however, because when the denominator is zero, the inverse solution does not exist. The solution to the problem is provided by the *Central Slice Theorem*. To compute the Fourier transform of the effective attenuation coefficient $\mu_{eff}(l, s)$ for a given θ , the process is carried out in two steps: first, a 1D transformation along one dimension ($l \rightarrow \omega_l$), while keeping the other fixed, and then, conversely, a transformation along the second dimension ($s \rightarrow \omega_s$) with the first fixed. This approach is feasible because the 2D Fourier transform of $\mu_{eff}(l, s)$ is invariant under changes in the order of integration and rotation. The rotational invariance implies that, for any angle θ , the ray segment s in Fourier space crosses the origin and is perpendicular to the direction of the projection, thus the following equivalence holds:

$$M(\omega_l, \omega_s = 0) = \int_{-\infty}^{+\infty} M(\omega_l, s) ds = \int_{-\infty}^{+\infty} g(l, \theta) \exp\{j\omega_l l\} dl = G_\theta(\omega_l)$$

This equation corresponds to the *Central Slice Theorem*.

Other algorithms have been developed for evaluating the 2D Fourier transform of the effective attenuation coefficient and performing the final back-projection. These algorithms differ in the specific steps they use, which involve various combinations of back-projection, corrections, and inverse Fourier transformations in different coordinate spaces. The choice of algorithm for image reconstruction primarily depends on the speed of the procedure and the ability to analyze the entire set of projection frames simultaneously, rather than processing them one by one.

After applying the algorithms to obtain a *filtered back-projection*, the result of a CT scan is a 2D map of the linear attenuation coefficients traversed by the cone-beam, considering a half rotation of the source around the sample. However, even if the X-ray sources are made of the same materials, they can emit slightly different beams. As a result, different CT scanners may yield slightly varying results when scanning the same sample.

The map of effective linear attenuation coefficients obtained from two different CT scanner systems could vary because each system provides a different effective energy value. Therefore, to enable a meaningful comparison of the data, the effective linear attenuation coefficients must be transformed into absolute values, which are represented

as *CT numbers* expressed in Hounsfield Units (HU).

$$CT\ number = 1000 \times \frac{\mu_{eff} - \mu_{H_2O}}{\mu_{H_2O}}$$

with μ_{H_2O} the linear attenuation coefficient in water at the effective energy value of the apparatus considered. CT numbers range from -1000 HU for air to about 3000 HU for metal substances, with a value of zero representing water. This standardization allows for more consistent comparisons between scans from different systems.

3.2 Photon Counters for industrial CT

The image reconstruction description provided in the previous section applies to energy-integrated systems. These systems typically use detectors such as scintillators read by photomultipliers or photodiodes, which are commonly employed in industrial settings.

An example of this is the MiCROTEC company, which manufactures large-scale scanners for the sawmill industry. One of their products is the CT LOG, designed to scan logs with a field of view of 90 cm, suitable for logs with diameters up to 70 cm and lengths of a few meters. The field of view is larger than the maximum diameter to account for the probability of scanning logs that are not perfectly straight. The detection system developed by MiCROTEC addresses the need for a large field of view while maintaining high transportation speed. The solution is a half cylindrical array detector composed of small-area crystal scintillators with a long extension, providing high resolution along the transversal direction (with a pitch of 2.2 mm) [38]. Longitudinal resolution is less critical, as the features and defects of the log primarily develop along the radial direction from the central axis.

Similarly, in the food scanning industry, energy-integrated systems are also used. The BIOMETiC group has developed a CT technology based on a scintillator plate and a set of CCD cameras for signal acquisition. Their MITO apparatus is the first X-ray inspection system with 3D reconstruction technology on the market. This system is designed to optimize food industry processes, covering various aspects such as ingredient uniformity control, ensuring the correct products are in each package, and detecting glass fragments and foreign objects. Food inspection requires very high spatial resolution, typically on the order of 1 mm.

In contrast, D-MAPS technology does not provide a map of the linear attenuation coefficient values while scanning an object with tomography. As a result, the mathematical framework described in the previous section cannot be directly applied. In a digital architecture like the ARCADIA chip, the system only detects the ability of the object to stop an X-ray, indicated by an on-off variation of a sensor pixel. This approach bypasses the issues associated with the energy resolution of scintillators and the inefficiency introduced by photon-to-electric signal conversion via external devices, such as photodiodes, which can reduce time resolution. Moreover, recent technological advancements have significantly enhanced spatial resolution. A single pixel with a pitch of $25\ \mu\text{m}$ can now house the necessary electronic circuitry for read-out. Another advantage is the ability to discriminate between photons, allowing the system to evaluate the interaction position of each individual particle.

The image obtained from a single beam shot resembles a radiograph, with the output information related to the material density encountered by the X-rays. Consequently, image reconstruction can be performed using similar steps to those in energy-integrated systems, but without the complications associated with defining an effective energy.

The analysis begins with evaluating the probability of X-ray interaction at the detector position, but unlike previous systems, the intensity cannot be directly acquired. As described in Chapter 1, the number of photons (N) that survive after interacting with matter depends on the initial number of photons (N_0), the linear attenuation coefficient (μ), and the material thickness (x):

$$N = N_0 e^{-\mu x} \quad (3.5)$$

From a statistical perspective, given the initial energy distribution emitted by the X-ray tube and the amount of incident X-rays, the expected number of photons reaching the detection point can be modeled by considering the X-ray production process within the tube.

The amount of X-rays generated by an X-ray tube depends on the number of electrons accelerated from the cathode to the anode (N_e). This value is determined by the product of the number of electrons accelerated and the probability of their conversion into X-rays (p_{ex}), which can be modeled using a Poisson distribution. The Poisson distribution describes the probability of a given number of events (in this case, the number of X-ray photons generated) occurring within a fixed interval, assuming the events occur independently and with a constant average rate. The expected number of X-ray photons generated is then given by:

$$Prob(N = n_x) = \frac{(N_e \cdot p_{ex})^{n_x}}{n_x!} \exp[-(N_e \cdot p_{ex})]$$

with n_x the number of X-rays generated in a fixed interval of time and N a discrete random variable with expectation value and variance equal to $N_e \cdot p_{ex}$.

Thus, the X-ray beam interacts with the object to be scanned, and the survival probability of the X-rays after interacting with the object (Equation 3.5) can be described by the binomial distribution. The survival probability is given by:

$$\begin{aligned} Prob(N = n_s) &= \sum_{n_x=n_s}^{\infty} Prob(n_x) Prob(n_s|n_x) \\ &= \sum_{n_x=n_s}^{\infty} \frac{(N_e \cdot p_{ex})^{n_x}}{n_x!} e^{-(N_e \cdot p_{ex})} \cdot \binom{n_x}{n_s} (e^{-\mu_{obj} x_{obj}})^{n_s} (1 - e^{-\mu_{obj} x_{obj}})^{n_x - n_s} \\ &= \frac{(N_e \cdot p_{ex} e^{-\mu_{obj} x_{obj}})^{n_s}}{n_s!} \exp[-(N_e \cdot p_{ex} e^{-\mu_{obj} x_{obj}})] \end{aligned}$$

with n_s the number of photons traversing the object and μ_{obj} the linear attenuation across the object, which depends on the energy of each single X-ray. Finally, the probability of observing an X-ray at the detection point, considering the binomial

distribution for the number of photons absorbed by the detector, can be expressed as follows:

$$\begin{aligned}
 Prob(N = n_d) &= \sum_{n_s=n_d}^{\infty} Prob(n_s) Prob(n_d|n_s) \\
 &= \sum_{n_s=n_d}^{\infty} \frac{(N_e \cdot p_{ex} e^{-\mu_{obj}x_{obj}})^{n_d}}{n_d!} \exp \left[- (N_e \cdot p_{ex} e^{-\mu_{obj}x_{obj}}) \right] \cdot \\
 &\quad \cdot \binom{n_s}{n_d} (1 - e^{-\mu_{det}x_{det}})^{n_d} (e^{-\mu_{det}x_{det}})^{n_s-n_d} \\
 &= \frac{(N_e \cdot p_{ex} e^{-\mu_{obj}x_{obj}} (1 - e^{-\mu_{det}x_{det}}))^{n_d}}{n_d!} \cdot \exp \left[- (N_e \cdot p_{ex} e^{-\mu_{obj}x_{obj}} (1 - e^{-\mu_{det}x_{det}})) \right]
 \end{aligned}$$

with n_d the number of detected X-rays. The final distribution is again a Poisson distribution with expected value and variance equal to:

$$E[n_d] = Var[n_d] = N_e \cdot p_{ex} e^{-\mu_{obj}x_{obj}} (1 - e^{-\mu_{det}x_{det}}) \quad (3.6)$$

The statistical description provides a method to evaluate the Signal-to-Noise Ratio (SNR) in order to compare it with the results of the apparatuses used by the company. The SNR results:

$$\begin{aligned}
 SNR &= \frac{E[n_d]}{\sqrt{Var[n_d]}} = \sqrt{N_e \cdot p_{ex} e^{-\mu_{obj}x_{obj}} (1 - e^{-\mu_{det}x_{det}})} \\
 &= \sqrt{N_x e^{-\mu_{obj}x_{obj}} (1 - e^{-\mu_{det}x_{det}})}
 \end{aligned} \quad (3.7)$$

with N_x representing the number of X-rays evaluated experimentally from the X-ray tube by placing a photon counter in front of the beam source.

The statistical evaluation of the number of X-rays acquired by the detector after passing through the matter is crucial for determining whether the image reconstruction method described in the previous section can be applied to this case. As discussed earlier, the single image acquired represents a radiograph of the object before any rotation by an angle θ is performed. These radiographs correspond to a matrix of the same size as the D-MAPS, filled with integer values representing the number of photons detected. From each radiograph at a specific angle, the Radon transformation is applied, converting the 2D image into a linear distribution along the axis orthogonal to the ray emission direction in the (l, s) plane. The assumption here is that the X-ray beam consists of parallel rays. The accumulation of these line distributions for each acquired angle forms the *sinogram*, which can later be converted back into a 3D image.

Under these conditions, it is possible to evaluate the tomography of an object, determining the ability of a D-MAPS to distinguish between different materials inside the object. The preliminary results regarding the discrimination of various substance densities from a CT with a photon counter sensor are presented in the final paragraph of this chapter, which includes a complete simulation of the image reconstruction process.

3.2.1 The MiCROTEC case study

After defining a method to compare the results obtained from two different detector systems, specifically using the Signal-to-Noise Ratio (SNR), it is essential to describe the apparatus employed in the industrial field. The collaboration with MiCROTEC provides an opportunity to directly test the performance of a new detector system developed under the ARCADIA research program. This partnership enables the university and research group to assess the sensor performance in a real-world environment, offering the chance to explore its potential applications beyond research fields like HEP, space exploration, and medical imaging. This collaboration can benefit both parties: the company gains access to new technology for the market, while the research team secures the necessary funding from the company to further improve and develop the technology.

Therefore, it is essential to define the apparatus system used for the chip characterization. The customers of MiCROTEC can be divided into two main categories: sawmills and food industries. The application of silicon D-MAPS technology, as discussed in Chapter 2, depends on the X-ray energy distribution involved. The efficiency of silicon detectors begins to decrease for X-rays above approximately 20 keV, making them unsuitable for highly penetrating X-rays. As a result, the CT LOG apparatus cannot be considered an appropriate option for silicon D-MAPS use. On the other hand, the BIOMETiC group works with thinner objects, where the X-ray energy can be lowered to find a balance between detector efficiency and the required X-ray energy for CT scanning.

In every case, the CT scanners developed by MiCROTEC for quality control in the food industry are tailored to each customer, with each system being slightly different from the others, primarily in terms of the reconstruction algorithm and analysis methods used. The geometry of all the MITO systems, on the other hand, remains consistent. The X-ray source consists of a tungsten anode X-ray tube that emits photons in a cone beam shape, with a maximum energy of 160 keV and a current of 3.12 mA. The emission angle is designed so that a single scan can cover the entire sample region, eliminating the need for translation of the X-ray source.

The detection system used is a Gadolinium Oxysulfide (Gd_2S_2O) scintillator plate, also known as Gadox, with dimensions of 350 mm $\times$ 400 mm. This material is activated with rare earth ions such as Praseodymium (Pr^{3+}) or Terbium (Tb^{3+}), and is selected for its high density (7.32 g/cm³) and efficient X-ray to light conversion. The luminous efficiency of Gadox is approximately 15 % [39]. Additionally, the light emitted by Gadox is well aligned with the spectral response of common silicon light detection devices, such as CCD and CMOS sensors, allowing them to operate at high quantum efficiency. However, a drawback of this ceramic scintillator is its strong optical scattering and slow signal decay, which leads to a reduction in energy resolution. Furthermore, it has very low efficiency for X-rays at 50.24 keV, since this energy corresponds to the binding energy of the K-shell electrons in the Gadolinium substance. This results in increased X-ray absorption at this energy, causing a loss of information when using a beam source with characteristic X-ray emission lines at 57.98 keV and 59.32 keV, like the one used by the company. Despite the high efficiency in X-ray to light conversion, with an estimated light yield of $(7.1 \pm 0.5) \times 10^4$ photons/MeV [40], Gadox is not the ideal solution as a CT detector. The coupling between the scintillator and the CCD

cameras inside the MITO apparatus is influenced by the distance between the object and the cameras, as well as the number of cameras used. A larger distance reduces the resolution of the camera, as the same CCD must capture more spatial information. Additionally, another important parameter for 3D image reconstruction is affected: the acquisition time. The decay time of this material is $458 \pm 3 \mu\text{s}$ [40], which is quite long compared to other scintillators on the market, which typically range from a few ns to a few μs .

Finally, both the source and detector rotate synchronously to obtain a 3D image. A scanner of this type typically records around 1000 projections in the range of 1 s to 20 s when using a fan-beam source with an angle of 30° to 60° interacting with a pixelated detector [7].

The replacement of the MITO detection system, which currently uses a ceramic scintillator read by CCD cameras, with layers of Silicon FD-MAPS could significantly enhance the performance of the apparatus. As previously discussed, energy resolution will no longer be an issue due to the different detection behavior of the silicon-based sensors. Additionally, the ability to distinguish materials can be achieved in a similar manner as with the current system. To evaluate the feasibility of using these sensors for complete tomography, a simulation is created using the GEANT4 toolkit, and the results are analyzed with the image processing software *Fiji* [6].

3.2.2 Simulation of a CT scan

The GEANT4 simulation toolkit provides an effective method for evaluating the suitability of using a silicon digital pixel detector for a complete CT scan.

The main challenge lies in simulating the rotation of the entire system, which consists of both the source and the sensor. To address this, I begin with a simple description of the X-ray emission. Instead of modeling the source as a point-like emitter, I simulate it as an area source, the size of which matches that of the silicon layer. This ensures that only parallel rays are emitted, striking the entire detector area (as shown in Figure 3.3a). The X-ray energy distribution used in the simulation ranges from 0 keV to 160 keV, consistent with the energy spectrum used in previous simulations. For the simulation, the detector size is adjusted and does not match the exact dimensions of the silicon layer proposed by the ARCADIA project. This adjustment is made for qualitative purposes. The silicon layer in this simulation has a thickness of $300 \mu\text{m}$ and an area of $100 \text{ mm} \times 100 \text{ mm}$, which is divided into a matrix of 1000×1000 pixels, each with a $100 \mu\text{m}$ pitch. These values are selected to reflect potential industrial applications, where spatial resolution is a critical factor. A pitch of $100 \mu\text{m}$ strikes a balance between achieving high resolution and the need to incorporate the electrical circuit and memory into each pixel.

The estimation of silicon efficiency for an X-ray beam generated by a Tungsten anode X-ray tube is previously presented in Chapter 2. In that section, I calculated the number of $200 \mu\text{m}$ layers required to achieve an efficiency comparable to the threshold set by the company. This calculation indicated that five sensors would be needed. For this simulation, however, I opted to use four detectors, each with a thickness of $300 \mu\text{m}$, placed at a distance of 5 mm apart, in order to account for the thickness difference.

The object to be scanned is modeled as a water-filled cylinder with a height of 70 mm and a radius of 30 mm. This cylindrical shape is chosen for its simple symmetry

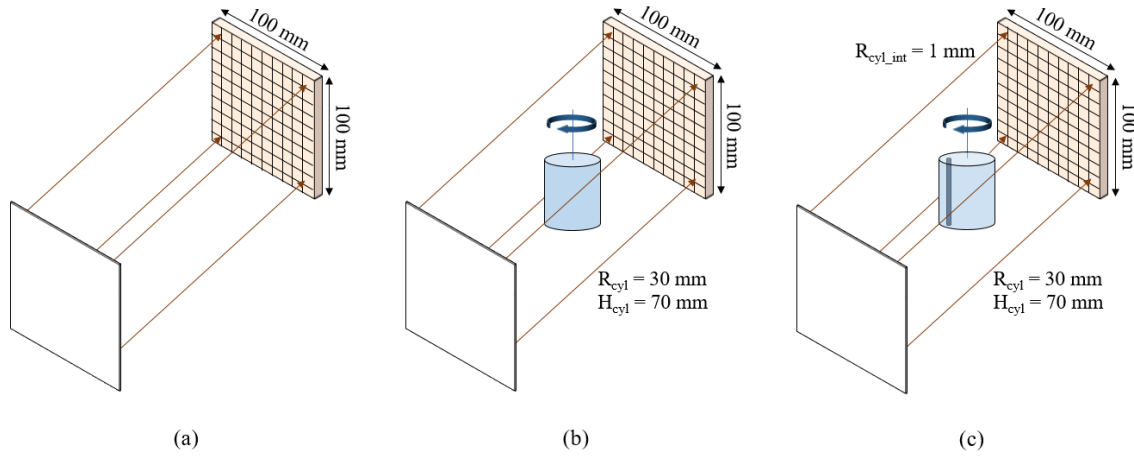


Figure 3.3: Configuration of the simulation with a water cylinder. The three representations show the progressive implementation inside the GEANT4 simulation, starting from (a) a source of parallel X-rays (white square) distant 610 mm from the pixelated Silicon detector, (b) the positioning of the water cylinder at a distance of 310 mm from the source and (c) the addition of the glass cylinder of 1 mm radius inside the water object. The detection system consists of four Silicon layers but for clarity reasons, only one slab is depicted here.

around the central axis, which simplifies the modeling and is also similar to the shape of a jam jar. Figure 3.3 illustrates two different configurations of the same object: (b) represents a perfect jar, while (c) shows the jar with a glass foreign body inside. The glass fragment is modeled as a cylinder of the same height but with a radius of 1 mm.

The imaging process begins by capturing a single radiography using the four sensor layers. Afterward, a 1° rotation is applied to the object, providing a complete CT scan after 180 projections are collected. For the reconstruction of an horizontal slice of the full sample, the process goes through the sinogram, as described earlier. To achieve this, I have written a software tool to collect the sinogram data. The same slice of each projection is considered to evaluate the profile distribution, which is then folded over the previous one. The final product, as displayed in Figure 3.4a, represents the sum of the images acquired by all four detectors used to scan the object. This process results in a sinogram that captures the projections of the object cross section from various angles, ready to be reconstructed into a 2D image.

3D image reconstruction is performed using a Java-based software called *Fiji* [6]. However, the sinogram presented in Figure 3.4a represents only a single cross-section of the entire object, making it insufficient to reconstruct the full sample volume. The 2D image is reconstructed using a square filter, and the resulting reconstruction is shown in Figure 3.4b.

In the reconstructed image, the water cylinder and the glass fragment within it are clearly visible and distinguishable from the background. An estimate of the position of the glass fragment aligns well with its placement in the simulation geometry, confirming the precision of the reconstruction. However, a closer analysis of the image reveals the presence of a third cylinder outside the water cylinder. This is identified as an artifact arising from the reconstruction process. The artifact is caused by the non-precise calibration of the black background between the radiographs of the object and those of the reference (empty) background. As a result, the software mistakenly interprets the lateral areas of the sinogram, where the background intensity differs from the

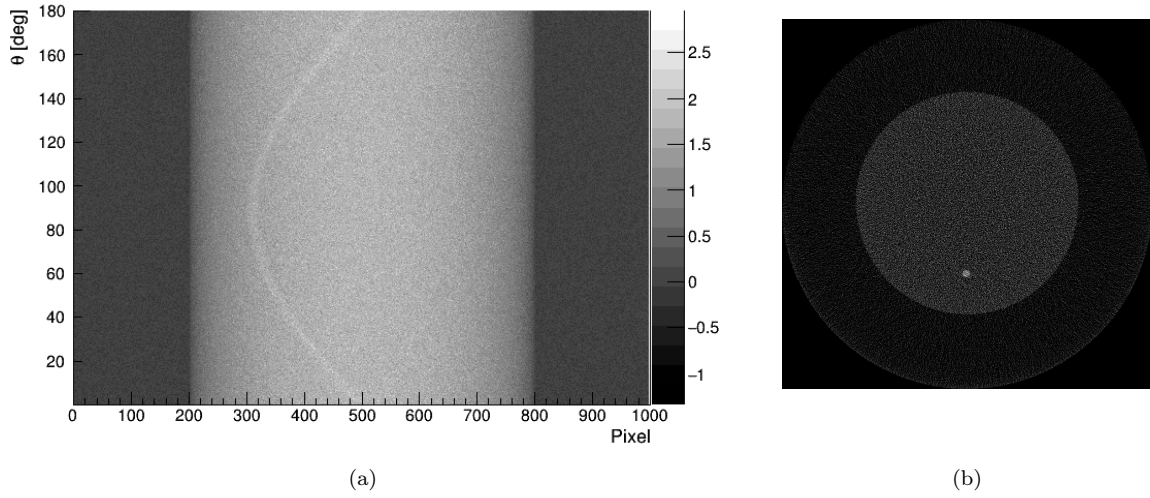


Figure 3.4: (a) Sinogram of the object scanned via GEANT4 simulation. The projections of the object on the four detector layers at the same rotation angle are summed up and then the profile distribution is calculated. The result provide a single sinogram considering a Silicon efficiency of 15.6 %. (b) 2D image back-projection of the water cylinder with a glass fragment.

reference, as another cylinder with low material density. This artifact is particularly evident when the profiles of different positions in the reconstruction are plotted, as shown in Figure 3.5.

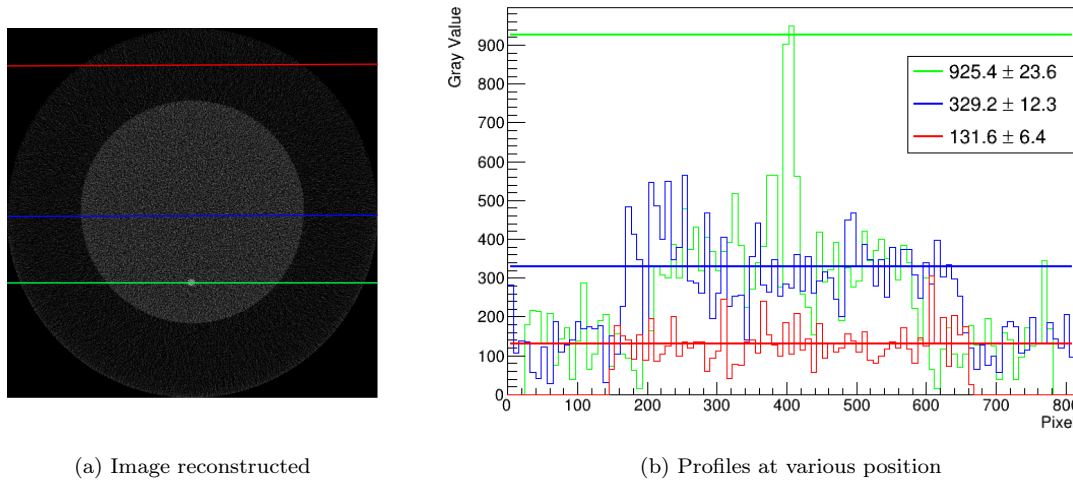


Figure 3.5: Histogram profiles of the reconstructed image (a) at different positions. (b) Overlap of the gray values for three profiles. The red fit represents the level of the image background, due to a problem in the calibration analysis (see text). The blue one is the value associated to water density, while the red one to the glass.

In Figure 3.5b, the red constant fit represents the gray value of the background, while the blue and green fits correspond to the values associated with the water and glass materials, respectively. These values are not universal reference points for all image reconstructions because they depend on several factors, such as the calibration of the background value and the type of filter used during the *back-projection* step. Nevertheless, the ability to distinguish between two materials with relatively similar densities, water ($\rho_{H_2O} = 1 \text{ g/cm}^3$) and glass ($\rho_{glass} = 2.23 \text{ g/cm}^3$), highlights the potential of this sensor type for X-ray detection in radiography and CT scanning

applications. This capability is crucial because it demonstrates the effectiveness of the sensor in differentiating materials that are not significantly different in terms of their physical properties.

The next step is the reconstruction of a complete 3D image. The method follows exactly the same procedure as described earlier, but instead of capturing a single radiography of the sample at each angle, now the object is further scanned along another direction. This allows for the collection of multiple cross sections, which are necessary to reconstruct a full 3D image. The simulations follow the same approach as before, with the primary difference being the geometry of the X-ray source. In this case, a point-like X-ray source is used to more accurately simulate the configuration of a real CT scanner system. This change in source geometry allows for a more realistic representation of how X-rays interact with the object during scanning, which is crucial for achieving accurate information about the 3D reconstructions.

For the reconstruction validation, a different geometry shape (glass cone) is chosen. This shape maintains the central axis symmetry, as is done in the cylinder case, but avoids the replication of the same cross-section along the length of the object. The cone rotates along a segment perpendicular to its axis, passing through the center of gravity. A rotation of 5° is applied after every scan of the volume, starting from 0° to 180° . Each projection is divided into a set of horizontal slices, each 1 mm thick, which are separated during the analysis to recover 100 cross sections for volume reconstruction. The projections are generated from the emission of 10^7 X-rays, ranging in energy from 20 keV to 160 keV. Few of them are shown in Figure 3.6.

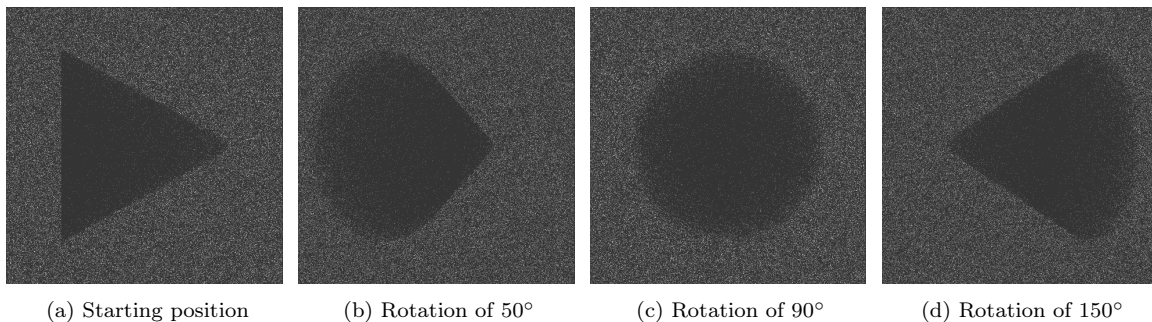


Figure 3.6: Examples of the cone projections obtained via GEANT4 simulation. The X-ray beam is represented by a point-like source at the same distances described for the cylinder cross section analysis. The detector corresponds to four slabs of $100\text{ mm} \times 100\text{ mm} \times 300\text{ }\mu\text{m}$ pixelated Silicon sensors.

Sinograms for every cross section of the cone are generated by filling a 2D histogram with the negative value of the logarithm of the pixel hits, normalized for the background of the selected slice. The x-axis represents the position of each pixel along the slice, while the y-axis corresponds to the rotation angle. These sinograms are shown in Figure 3.7, which displays the sinusoidal paths for three different slices highlighted in Figure 3.7a. The sinograms exhibit a non-homogeneous distribution in the background area due to the probability that some pixels acquire no signal. In the background region of the projection images, in fact, the mean value of pixels that are switched on is 0.45, indicating the sparsity of the signal in these areas.

The 3D reconstruction of the cone volume is performed using the *Fiji* software [6], which can back-project the 2D Radon Transform and apply various algorithms

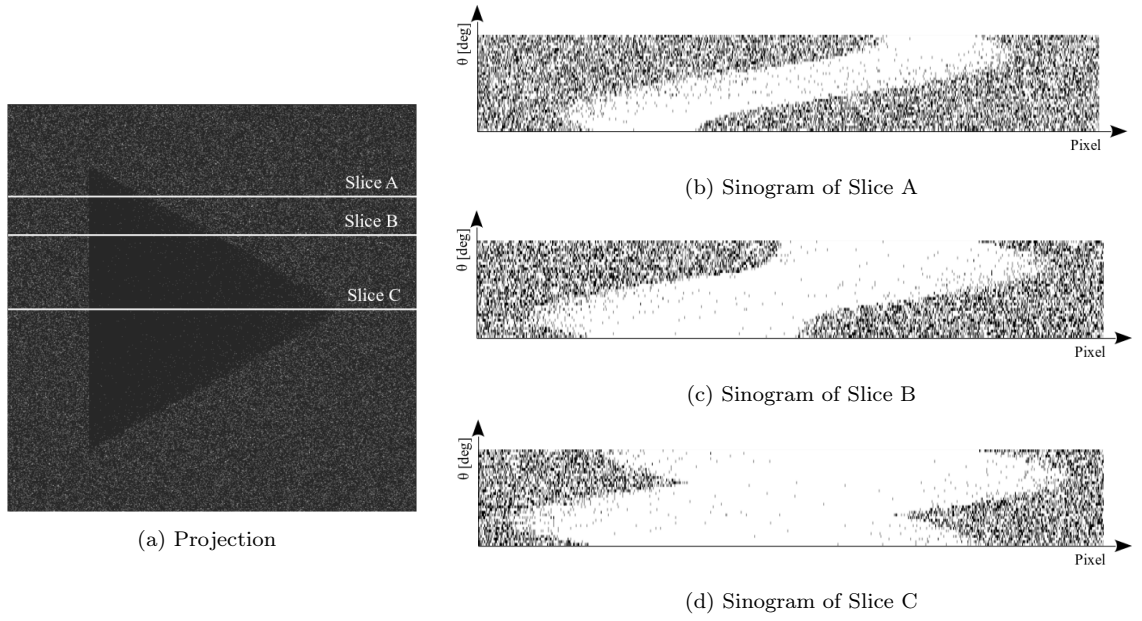


Figure 3.7: Sinograms created from the simulated projections of a cone geometry. (b), (c) and (d) are those associated to the slices A, B and C, respectively. The number of pixels in the x-axis is equal to the one in projection radiography of (a) while the y-axis corresponds to the rotation angles in degrees.

and filtration processes. In this case, the *filtered back-projection* method with a ramp filter is used. After adjusting the necessary parameters, the 3D reconstruction of the back-projected data reveals the scanned volume, as shown in Figure 3.8.

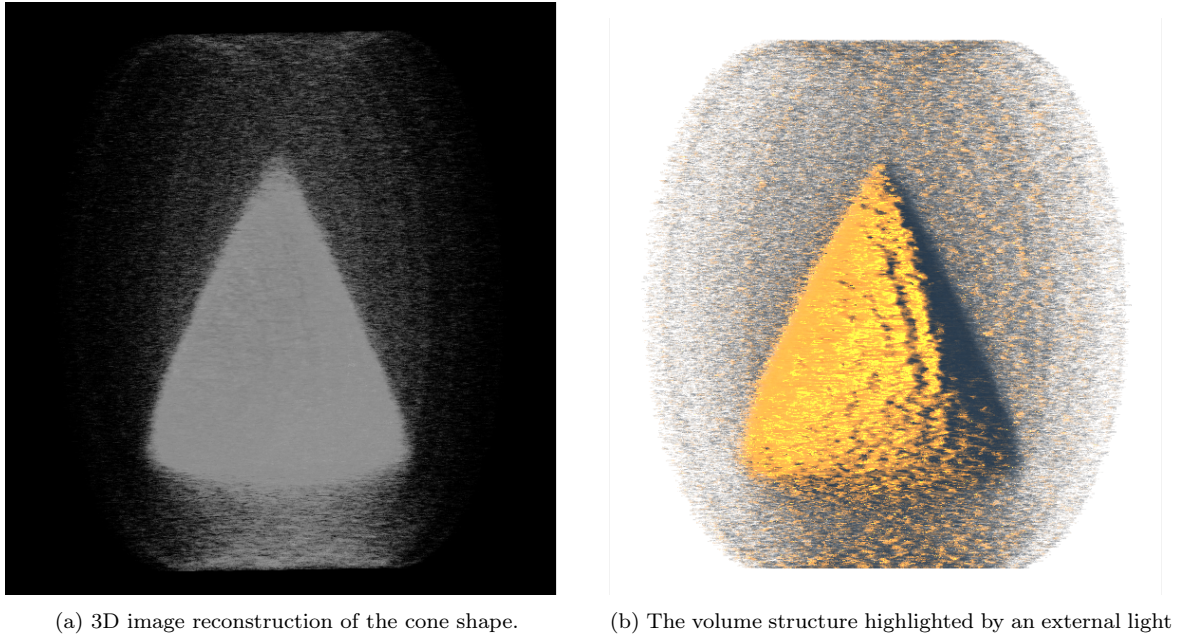


Figure 3.8: 3D volume reconstruction of a cone. (a) shows a photo of the reconstructed volume. (b) represents the same volume adding color and an external light source in order to emphasize the tridimensional structure. The gray ring around the cone, present in both images, is an artifact due to the non-homogeneity of the sinograms background.

By selecting the appropriate threshold for the gray value scale, it is possible to

clearly differentiate between the air and glass materials, allowing the cone volume to be visualized as in Figure 3.8a. Adding color and an external light source enhances the visualization of the 3D structure, as demonstrated in Figure 3.8b. Figure 3.8b also facilitates a qualitative analysis of the reconstructed image, highlighting the sliced structure of the 3D volume. This slicing is due to the way the projections are divided into individual slices during the reconstruction process. Each slice represents a thickness of 10 pixels along the y-axis, resulting in a thick slice during reconstruction. To improve the quality of the reconstruction and reduce the visible slicing effect, it is necessary to increase the number of slices, ideally matching the number of pixels in the sensor to achieve finer resolution.

In conclusion, the use of the GEANT4 toolkit [3] has enabled the simulation of silicon pixel sensors to assess their ability to distinguish materials with slightly different absorption coefficients, such as water and glass. Various test sample shapes and material combinations have been considered to evaluate the detector capacity to differentiate between materials. The results indicate that the silicon pixel sensors can effectively distinguish glass fragments as small as 2 mm in water, which is a typical size for glass fragments found in broken jars. Additionally, the 3D image reconstruction of a cone-shaped object demonstrates the feasibility of using silicon D-MAPS sensors in CT scanner systems, even with a non-optimized software setup. These findings suggest that silicon D-MAPS technology has the potential to be integrated into advanced CT systems, offering enhanced material discrimination and imaging capabilities.

4 Dose evaluation in MiCROTEC systems

The determination of the effectiveness of an industrial CT scanner extends beyond the optimization of the detection system, for which a basic simulation tool is implemented in Chapter 3. While the optimization is crucial, it is only one part of a comprehensive evaluation process. This process must also consider other key aspects such as system design, operational safety, and compliance with radiological protection standards.

As outlined in Chapter 1, in fact, the interaction of X-rays with matter releases energy that can alter the internal structure of materials. While the scanned materials, such as logs, wood, food containers, and scanner system boundaries, can tolerate higher energies than those used in medical imaging, and for prolonged periods, the possibility of X-rays escaping through scanner openings must be addressed. This is particularly crucial in ensuring the safety of industrial employees working nearby.

To comprehensively address the safety and efficiency of the X-ray scanner system, I analyze the dose generated by X-rays with an energy of up to 180 keV. The study has two primary objectives:

1. **Estimate the dose acquired by the detector inside the gantry:** this involves evaluating the amount of radiation absorbed by the detector during normal operation. The analysis considers the energy range employed by the company X-ray sources, ensuring that the detector system efficiency and durability align with the operational requirements.
2. **Evaluate and control the ambient dose outside the gantry:** here, I assess the potential for X-rays to escape from the scanner into the surrounding environment, which could pose a risk to industrial employees. For this purpose, I have to quantify the ambient dose equivalent (H^*), the operational quantity used to measure external radiation exposure. Strategies for minimizing radiation leakage, such as shielding and structural modifications, are evaluated.

To implement radiological security measures, the *International Commission on Radiological Protection* (ICRP) provides guidelines and defines safety quantities to limit radiation exposure and mitigate risks. These guidelines, in conjunction with recommendations from the *International Commission on Radiation Units and Measurements* (ICRU), offer conversion coefficients for both internal and external radiation exposure scenarios.

Moreover, during my research period, the company developed a new CT scanner, the MITO500. This open, large-dimension apparatus differs from previous models as it is not positioned along the production line for single object scanning. Instead, it is designed to handle large-sized or multi-package products and is placed at the end of the industrial line. Therefore, the development of this machine requires a comprehensive evaluation of the dose generated by its operations.

The ambient dose equivalent (H^*) is selected as the primary operational quantity for external exposure evaluation. The objective is to create a tool capable of reproducing real-world results, taking into account the CT system developed by MiCROTEC. The results are expressed as dose rates, which require detailed analysis of the data

generated by the simulation tools. To this end, a simplified geometry of the MITO500 is created, exported as STL files from CAD software, and imported into the GEANT4 toolkit for simulations. Simplifying the geometry is necessary to reduce file sizes and computational time, enabling efficient simulation while maintaining the accuracy of the results. This approach aims to ensure a practical balance between computational efficiency and the fidelity of the dose evaluation, aiding in both machine development and compliance with radiological safety standards.

4.1 Ambient Dose Simulation

The GEANT4 simulation toolkit requires a precise definition of geometries, materials, and sources, all accurately positioned within the simulated space. Characterizing the scanner system with high accuracy is a complex task. Therefore, an efficient approach for simulation development involves directly exporting solid geometries from CAD drawings.

The quality of the designed simulation can be assessed by comparing its results with measurements obtained using a dosimeter at a bunker site operated by the company in Brixen (Bressanone, Italy). The instrument used to measure radiation doses is the *RaySafe 452 Radiation Survey Meter*, a highly capable device for detecting ionizing radiation across various applications, including the measurement of scattered radiation from X-ray machines. The *RaySafe 452* utilizes a combination of a silicon sensor cluster and a Geiger-Müller pancake detector, which activate dynamically based on the particle rate [41]. The instrument offers an ambient dose range of 0 Sv/h to 1 Sv/h with a resolution of 0.01 μ Sv/h, ensuring precise and reliable measurements.

The GEANT4 simulation I developed incorporates an external particle source, described using an energy spectrum histogram. The energy distribution of the X-ray source is measured with a Cadmium Telluride (CdTe) pixel sensor, the *X-Card ME3*, a standalone multi-energy detector board designed for linear X-ray imaging systems [42]. The sensor consists of a single array of 128 pixels, each with an intrinsic pitch of 0.8 mm and a thickness of 2 mm, offering an energy resolution of 7.7 keV at 60 keV, corresponding to 105 counts/pixel/s.

The spectrum shown in black in Figure 4.1 represents the measured energy distribution of the X-ray tube at a nominal voltage of 160 kV, used in the company CT systems. This spectrum has been corrected for the CdTe sensor efficiency. The operational energy range extends from 20 keV to 160 keV. Below 20 keV, the spectrum is represented by a connecting curve between the origin and the first data point in the plot.

A comparison with a spectrum generated using a free toolkit for modeling X-ray tube spectra, *SpekPy* [43], highlights the limitations of the CdTe sensor, including its poor energy resolution and the absence of data below 20 keV. While the reduced energy resolution is a disadvantage for the scanning system, the missing information below 20 keV has minimal impact on object radiography. This is because more than 90 % of X-rays in this energy range are absorbed within 3 cm of water-equivalent material.

The simulation output comprises energy distribution histograms of particles that survive after interacting with the sample. The dose rate is calculated from the simulation results, taking into account the detector area, the energy distribution, and the efficiency of the sensor material used. The reproducibility of the measurements relies

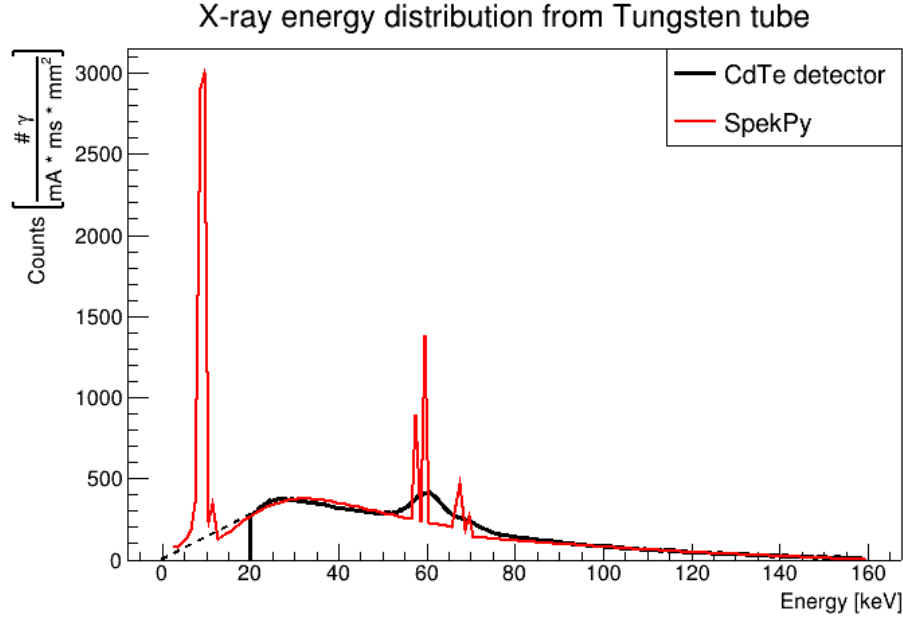


Figure 4.1: Energy spectrum of the X-rays source generated by the tungsten tube of 160 kV nominal voltage. The black distribution represents the measurement with the pixel sensor *X-Card M3* corrected for the CdTe efficiency. The red curve corresponds to the attended spectrum calculated using the *SpekPy* software without any filtration [43]. The values are normalized in order to allow a direct comparison between the two distributions.

on how accurately the CT system is modeled in the simulation. Therefore, the initial validation involved comparing the dose values obtained by considering only the source.

4.1.1 X-ray source

The first set of measurements is taken using an X-ray tube operating at a nominal voltage of 160 kV, positioned 5.3 m away from an array of CdTe pixels. This sensor operates in two distinct modes: it can function as either a photon counter or a spectroscopic detector. This flexibility allows the definition of the X-ray source within an energy range starting at 20 keV, enabling the measurement of the number of particles observed and their energy as a function of the subtended angle and the integration time. These measurements are then used as inputs for subsequent simulations.

To evaluate the compatibility between the simulation tool and the experimental data, I compare the calculated dose values. For this comparison, measurements are taken in a bunker, placing the *RaySafe 452* dosimeter in the same position as the CdTe pixel sensor. The simulation outputs consist of two histograms: the initial X-ray energy distribution and the kinetic energy carried by photons interacting with the detector. In the simulation, the detector is modeled as a simple air layer with a thickness of a few millimeters and a variable area, corresponding to a 100 % efficient sensor. The dose rate calculation begins with converting the fluence to ambient dose using Equation 1.11. The resulting integral value, expressed in pSv, is then scaled based on the number of photons detected by the CdTe sensor, corrected for the efficiency of the sensor material. The measured and simulated dose values, as a function of the current of the X-ray tube, are presented in Table 4.1.

Current [mA]	Dose Measured [mSv/h]	Error [mSv/h]	Dose Simulated [mSv/h]	Standard Error [mSv/h]
0.0	1.4×10^{-4}	1.0×10^{-5}	0.0	0.0
0.1	4.49	1.0×10^{-5}	5.92	6.3×10^{-5}
0.2	10.99	1.0×10^{-5}	11.85	1.2×10^{-4}
0.3	18.05	1.0×10^{-5}	17.77	2.0×10^{-4}
0.4	25.01	1.0×10^{-5}	23.70	2.4×10^{-4}
0.5	31.46	1.0×10^{-5}	29.62	3.1×10^{-4}
0.6	38.46	1.0×10^{-5}	35.55	3.7×10^{-4}
0.7	45.50	1.0×10^{-5}	41.47	4.3×10^{-4}
0.8	51.92	1.0×10^{-5}	47.40	4.8×10^{-4}
0.9	58.94	1.0×10^{-5}	53.32	5.6×10^{-4}
1.0	65.91	1.0×10^{-5}	59.25	6.1×10^{-4}

Table 4.1: Dose rates as a function of the X-ray tube current from 0 mA to 1 mA. Both data measured with the *RaySafe 452* dosimeter and results from the simulations are presented.

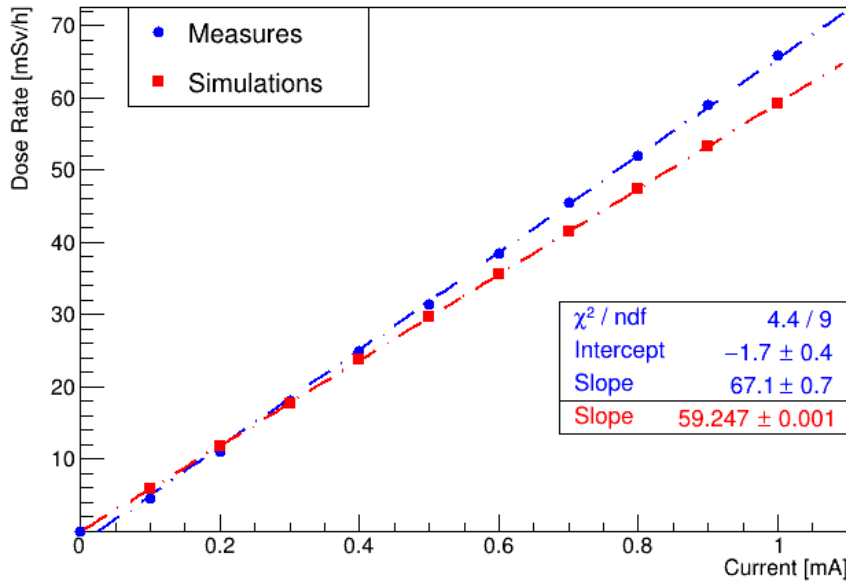


Figure 4.2: Direct measurements of the ambient dose rate provided by the X-ray tube with nominal voltage of 160 kV (blue circle) compared with the simulation results (red square) obtained for the same set-up. The measured values are fitted with a line (blue parameters) while the simulated numbers are interpolated with a line constrained to pass through the origin (the mathematical calculation of the dose rate for 0 mA must be zero).

The data presented in Table 4.1 are plotted in Figure 4.2, demonstrating a compatibility of within 10 % between the measured and simulated values. The measurement errors correspond to the systematic error of the instrument. Each measurement is obtained from a single acquisition for each X-ray tube current, so no statistical error can be assigned. Additionally, a non-zero value is observed when no current is applied. This can be attributed to the acquisition method, as the measurements are taken after the X-ray tube is powered on. During the warm-up phase of the system, an intrinsic offset is introduced, which contributes to this baseline value. Conversely, the simula-

tions include standard error estimates for the ambient dose values. This is achieved by running five identical simulations, each initialized with a different random seed for the X-ray production energy distribution. As expected, at zero current, no X-rays are generated by the tube, so the ambient dose curve derived from the simulations originates at zero.

The data are fitted using linear interpolation, as shown in Figure 4.2. Although data errors are considered, they are too small to be distinguishable from the markers. Despite the positive offset present in the experimental measurements, the parameters of the blue fit indicate a steeper slope compared to the simulated values, along with a negative intercept. The result shows that the two curves intersect at approximately 2.5 mA, with a maximum relative difference between the measured and simulated values of 10 %. The company uses operating currents of 3.12 mA and 5.5 mA for the in-line and end-of-line scanner systems, respectively. Based on the fit parameters obtained, the relative difference between the measured and simulated ambient dose rate values remains below 10 % for these operating currents as well.

4.2 External shielding structure for Mito500

After establishing a reliable method for evaluating the ambient dose rate through simulation and validating it against measurements obtained from the same configuration setup, it becomes possible to analyze the geometries and materials that can be employed as external shielding for the scanner apparatus developed by the company (Mito500). The aim is to reduce the ambient dose rate to a safe level for industrial workers.

In the food industry, the conveyor belt serves as the transport system for food products, connecting the industrial line to the Mito500 gantry and vice versa. The X-rays generated by the rotating tube continuously interact with the conveyor belt and the products being transported and scatter in all directions, without any specific trajectory. As a result, some of the initial X-rays, primarily those with energies exceeding 20 keV, escape the gantry. Therefore, an external shield covering the Mito500 gantry is essential to block scattered X-rays and ensure that the ambient dose rate remains within safe limits for human exposure [44].

Experimentally, it is only possible to compare various external shield configurations when they are fully assembled. However, producing these configurations is extremely costly, especially given the size and complexity of the entire CT system. To address this challenge, my role involves developing a simulation tool to evaluate the efficiency of different shielding geometries and materials that could be used to cover the Mito500 framework.

The first simulation I developed included a simple configuration comprising a plastic belt, a carbon pipe, and an external lead box with holes for the pipe, while accounting for the correct distances between the source and the belt. However, the results of this simulation are not consistent with the experimental measurements obtained using the *RaySafe 452* device at the exit point of the pipe. To achieve more accurate results, a detailed model of the entire CT system is required. Consequently, I utilize geometries directly imported from CAD drawings to describe the system. Although this approach significantly increases both the simulation loading times and durations due to the large number of particle interactions the tool has to process, it produces results that are

more consistent with experimental data. Ultimately, a balance is achieved between accurately representing the Mito500 and optimizing the simulation runtime to deliver reliable outcomes.

<i>SpekPy Web</i> toolkit options	
Physics	<i>kqp</i>
Attenuation data	Penelope
Target material	Tungsten
Anode angle	20°
Nominal voltage	180 kV
Filtration material	Polycarbonate
Filtration thick	1 mm

Table 4.2: Options used in *SpekpyWeb* toolkit [43] to reproduce the X-ray tube source spectra of 180 kV nominal voltage installed in Mito500 gantry.

Additionally, a customized source model is necessary, as this CT scanner operates with an X-ray tube that has a nominal voltage of 180 kV and a filtration layer of 1 mm of polycarbonate. The source model is generated using the open-source toolkit *SpekpyWeb*, which is specifically designed for modeling X-ray tube spectra. The *kqp* physics model option is selected to achieve the desired high accuracy, particularly for modeling the anode heel effect, as it comprehensively accounts for bremsstrahlung anisotropy effects [43]. At photon energies of a few hundred keV, discrepancies between the PENELOPE and NIST datasets are minimal, primarily limited to data on the photoelectric effect. In PENELOPE, these are renormalized subshell cross-sections, whereas in the NIST dataset (based on the XCOM code), they remain unrenormalized [45]. The PENELOPE Monte Carlo code system is employed for attenuation data due to its reliability and precision. The specific options selected for generating the X-ray tube spectra are summarized in Table 4.2, while Figure 4.3 illustrates the resulting energy spectrum.

A comparison of the shape distributions in Figure 4.3 and Figure 4.1 reveals that the filtering material layer effectively blocks nearly all X-rays with energies below 10 keV. However, the rest of the distribution remains unchanged. The simulation tool can now be enhanced by incorporating the precise energy distribution of the X-ray source, as well as the external shielding configurations intended for evaluation. Before simulating the entire scanner setup, it is essential to carefully analyze the materials and geometries currently in use and consider potential alternatives for future implementation.

4.2.1 Materials

Food, before it is ingested, comes into contact with many materials and items during its manufacture, processing, storage, preparation, and final serving. Examples of such items include food processing equipment, kitchenware, and dinnerware.

This also applies to industrial CT scanners, such as the Mito500, which are used as automatic inspection systems during various stages of food production. Scanner systems must be designed using materials that can safely come into contact with food. These materials are referred to as *Food Contact Materials* (FCMs). The European

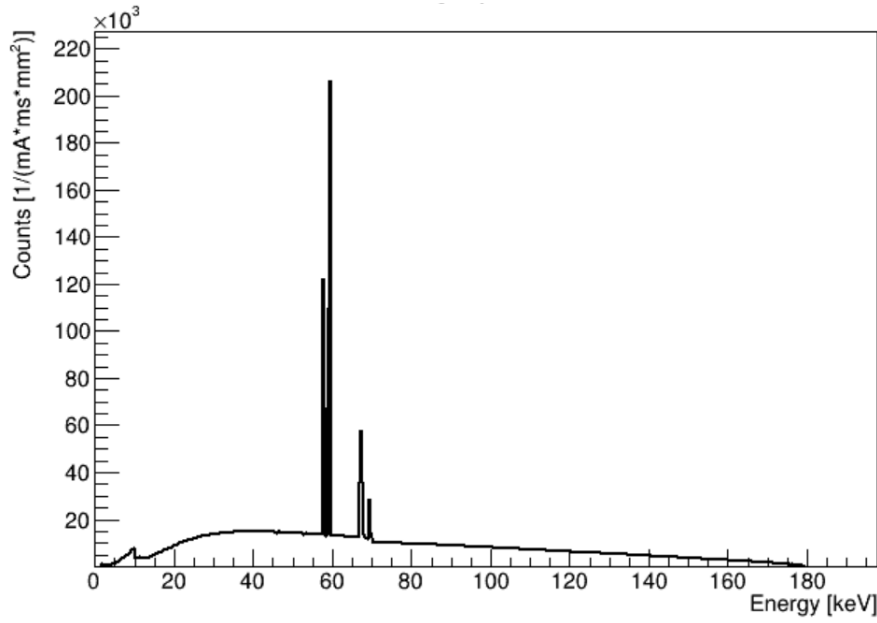


Figure 4.3: Spectrum of X-rays energy emitted from a tube of 180 kV nominal voltage. The plot is created using the *SpekpyWeb* toolkit and aims to reproduce the real spectra used by MiCROTEC company.

Commission regulates these materials to ensure "the effective functioning of the internal market in relation to the placing on the market in the Community of materials and articles intended to come into contact directly or indirectly with food, whilst providing the basis for securing a high level of protection of human health and the interests of consumers", as stated in the purpose of the document [46]. FCMs typically consist of low-density materials like plastic, glass, and aluminum, which are not effective at blocking high-energy X-rays. When X-rays interact with FCMs, they scatter away from their original path due to the Compton effect. The scattering direction is non-preferential and depends on the initial and final energy of the photon, as explained in Chapter 1. Therefore, selecting materials that minimize particle backscattering is advantageous.

To evaluate this, a simulation is conducted using a 1 mm thick layer of single elements or materials. The aim is to determine the amount of X-rays within the energy range 20 keV to 160 keV that are backscattered and their corresponding energy. The simulation involves firing a beam of X-rays, with energies up to 160 keV, at the material layer positioned 420 mm away. The initial X-ray energy distribution is based on tests conducted with a CdTe sensor, starting from 20 keV. The space between the source and the layer, instead, mirrors the distance between the X-ray tube and the conveyor belt in the Mito250 in-line CT scanner apparatus. The Mito250 scanner, which has been produced and sold since the inception of the BIOMETiC spin-off group, is designed to be placed along production lines, scanning individual products immediately after their assembly. Its compact size, about half that of the new Mito500 system, makes it suitable for this purpose. Despite the size difference, the gantry and detection system of both scanners are comparable. This similarity allows the results obtained from the Mito250 configuration to be applied to the Mito500, as the attenuation of X-rays in air is negligible.

This study investigates both Food Contact Materials and non-FCMs. The list of tested substances is presented in Table 4.3. For each material, the study evaluates the spectra of X-rays backscattered in a hemisphere on the same side as the source (Figure 4.4), the spectra of X-rays transmitted through the layer (Figure 4.6), and the energy deposition distribution within the layer (Figure 4.5).

Substances tested	
FCMs	Non FCMs
HDPE	
Mylar	
Polycarbonate	
Plexiglass	
Teflon	Lead
Carbon	Bismuth trioxide
Stainless steel	
Aluminium	
Tin	

Table 4.3: List of the substances considered for the X-ray backscattering evaluation. The materials are divided between Food Contact and NON-Food Contact Materials, following the European document [47].

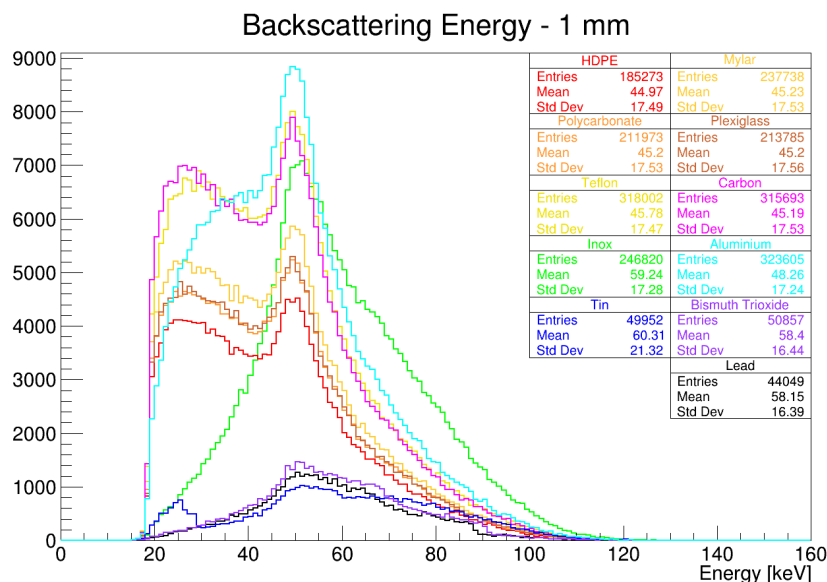


Figure 4.4: Energy spectra of backscattered X-rays. All the particles that interact with a semi-spherical detection area placed between the source point and the 1 mm layer of material are considered.

The results indicate that high-density materials, such as tin, bismuth trioxide, and lead, effectively block nearly all the initial X-rays. In contrast, low-density materials like carbon, plastic, and aluminum exhibit minimal photon interaction. Stainless steel falls between these two categories, making it a potential candidate for use in food industry apparatuses. A balance among material costs, their X-ray attenuation capabilities, and compliance with food contact regulations identifies stainless steel as

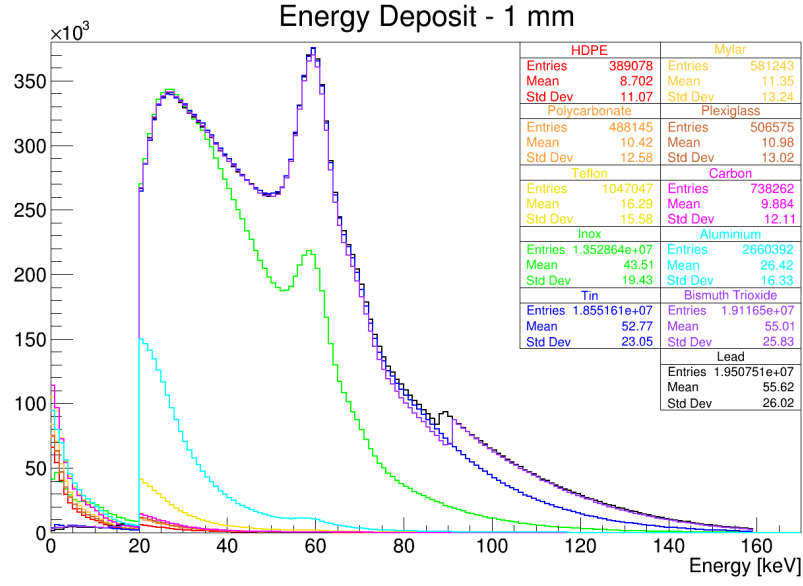


Figure 4.5: Spectra of the energy deposit by the X-rays when interact with the 1 mm layer material.

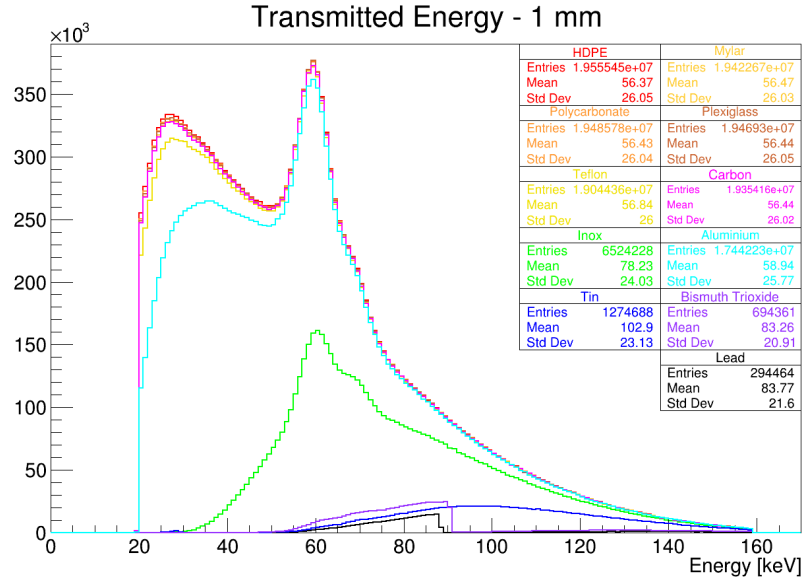


Figure 4.6: Energy spectra of X-rays that are transmitted over the 1 mm layer material. All the particles which cross the slab but do not interact or are deviated less than 90° with respect to the incident photon direction are considered.

the most suitable choice. A summary of the energy deposition, backscattering, and transmission properties of the tested materials is provided in Table 4.4.

When the layer thickness increases from 1 mm to 2 mm, the results depend significantly on the material density and electron shell configuration. For high-density substances, the number of backscattered X-rays remains nearly constant, whereas the number of transmitted photons decreases by factors of 1.9, 7.7, and 9.4 for tin, bismuth trioxide, and lead, respectively. The enhanced attenuation in bismuth trioxide and lead is attributed to their electron shell configurations. As shown in Figure 4.6, the absorption increase due to the K-shell electrons (K-edge) in these materials results in a sharp

Substance	Density [g/cm ³]	Percentage of X-rays detected		
		Interaction	Backscattering	Transmission
HDPE	0.93 to 0.97	1.9 %	0.9 %	97.8 %
Mylar	1.38	2.9 %	1.2 %	97.1 %
Polycarbonate	1.20	2.4 %	1.1 %	97.4 %
Plexiglass	1.18	2.5 %	1.1 %	97.3 %
Teflon	2.20	5.2 %	1.6 %	95.2 %
Carbon	2.20	3.7 %	1.6 %	96.8 %
Aluminium	2.71	13.3 %	1.6 %	87.2 %
Stainless steel	7.50 to 8.00	67.6 %	1.2 %	32.6 %
Tin	7.31	92.8 %	0.2 %	6.4 %
Bismuth trioxide	8.90	95.6 %	0.3 %	3.5 %
Lead	11.33	97.5 %	0.2 %	1.5 %

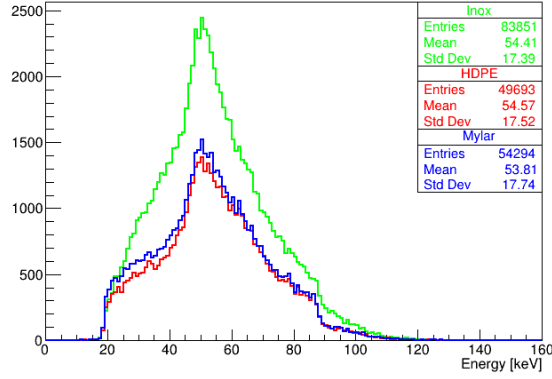
Table 4.4: Percentages of the X-rays that interact, are backscattered or transmitted considering 1 mm of listed substances.

drop in transmission around 88 keV-90 keV. In contrast, for low-density materials, the amount of backscattered X-rays nearly doubles, while the number of transmitted photons remains largely unchanged. Stainless steel, with a density in the range of approximately 7.5 g/cm³ to 8.0 g/cm³, exhibits behavior closer to that of low-density materials in terms of backscattering as a function of layer thickness. Specifically, the number of backscattered X-rays increases by a factor of 1.3 when the thickness grows from 1 mm to 2 mm. Meanwhile, the transmitted X-rays decrease by approximately 25 %.

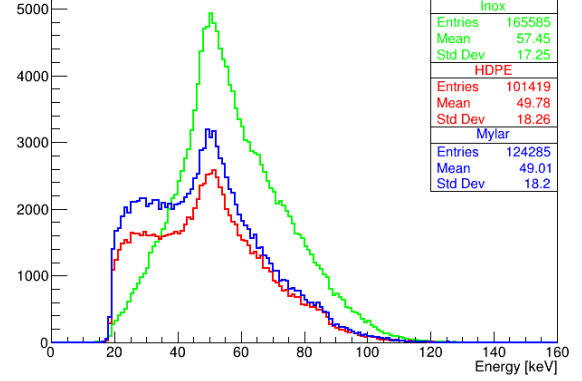
Low- and high-density substances exhibit distinct behaviors regarding the amount of backscattered and transmitted X-rays. While a 2 mm layer of lead would be the optimal X-rays shielding solution, it cannot be placed in direct contact with food products. To address this limitation, a layered "sandwich" configuration is proposed, incorporating both NON-FCM and FCM materials. The objective of this design is to leverage the benefits of high-density materials while ensuring food safety. The ideal configuration consists of a thick lead core wrapped with a thin layer of a material with a low X-ray scattering coefficient. Lead is chosen over tin or bismuth trioxide due to its cost-effectiveness and market availability. Three wrapping materials commonly used in the company, stainless steel, HDPE, and Mylar, are tested in configurations with a central 2 mm lead layer.

Comparisons between these four simulations reveal an increase in the number of backscattered X-rays for all configurations as a function of the wrapping thickness. Notably, while the number of particles detected increases almost linearly for configurations using HDPE and Mylar, it tends to saturate for stainless steel. As shown in Figure 4.7e, plastic wrapping is the best option for thicknesses below 1 mm; however,

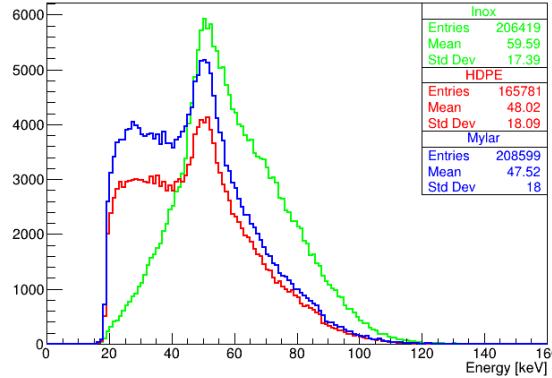
at these thicknesses, their durability over prolonged use needs to be verified. For this reason, stainless steel is again selected as the preferred material for the external shielding of the Mito500. In contrast, the transmitted X-rays remain nearly constant across all configurations, thanks to the shielding effectiveness of the lead core.



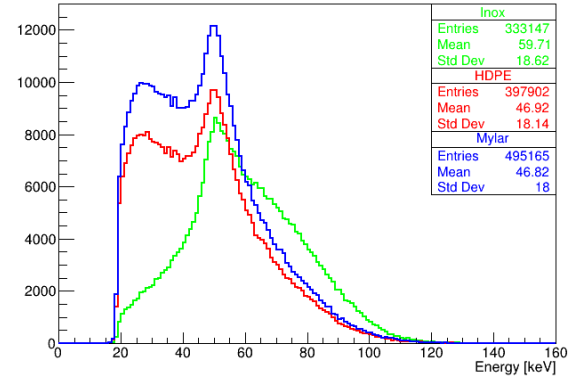
(a) 2 mm of Lead wrapped with 100 μm of lower density material.



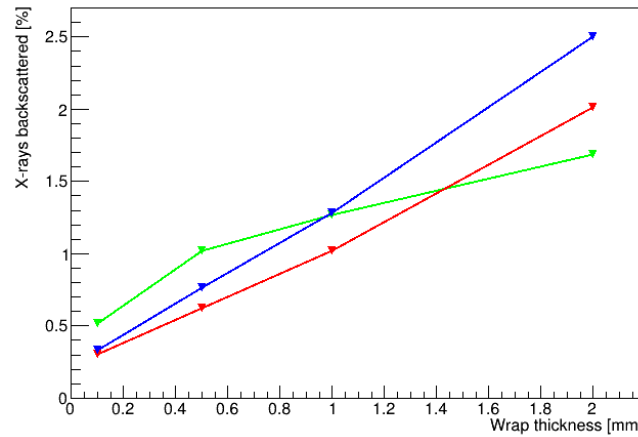
(b) 2 mm of Lead wrapped with 500 μm of lower density material.



(c) 2 mm of Lead wrapped with 1 mm of lower density material.



(d) 2 mm of Lead wrapped with 2 mm of lower density material.



(e)

Figure 4.7: Energy spectra of X-rays that backscattered from a sandwich of materials. The centre is 2 mm of Lead while the thickness of the wrap increases from (a) 0.1 mm to (d) 2 mm, passing through (b) 0.5 mm and (c) 1 mm. (e) The amount of X-rays backscattered as a function of the wrap thickness.

4.2.2 Geometries

To reduce the ambient dose rate detected in the external area of the Mito500, it is essential to combine external and internal shielding geometries and materials. External shielding refers to all visible elements surrounding the gantry, such as pipes and enclosures, while internal shielding includes elements placed inside the system to reduce particle scattering in specific directions. The Mito250 scanner employs a shielding structure comprising approximately 1 m of stainless steel and lead pipes on both sides of the gantry openings as external shielding, complemented by lead-free curtains as internal shielding. This configuration effectively ensures the ambient dose rate remains within safe levels for humans.

However, the Mito500 scanner cannot adopt this approach for two primary reasons: the large field of view of the gantry and the nature of the scanned samples. In fact, the Mito500 is designed to handle large multi-packs, which would keep the lead-free curtains perpetually open, rendering them ineffective. As a result, a different external shielding configuration must be devised, considering the placement of the scanner at the end of the industrial production line and its space constraints, as the entire system cannot exceed 10 m in length. The first priority is to block X-rays that exit the gantry openings after only a single interaction with the scanned material. The proposed solution involves introducing an angular deviation to the conveyor belt, which can be implemented along two distinct planes.

Considering the deviation along the vertical plane, two configurations are evaluated, as shown in Figure 4.8. Several simulations are run for three different inclination angles, with the addition of a pipe either before or after the slipway. The selection criteria for the evaluated sizes are twofold: the conveyor belt cannot be tilted more than 20° , and the exit area must be positioned below the lowest point of the gantry opening. The ambient dose rates from these simulations are presented in Table 4.5. Between the two configurations that involve placing the horizontal pipe before or after the slipway, the geometry shown in Figure 4.8b yields lower dose results, with a reduction in the ambient dose rate ranging from 25 % to approximately 40 %, depending on the configuration sizes.

Configuration	Z length [mm]	Y length [mm]	Slope [°]	Pipe length [mm]	Dose simulated [μSv/h]	Std deviation [μSv/h]
Figure 4.8a	3000	600	11.3	1000	178.6	9.0
	3000	600	11.3	1500	127.2	5.1
	3000	1000	18.4	1000	55.2	8.0
	3000	1000	18.4	1500	29.6	4.0
	4000	1000	14.0	500	90.6	11.0
	4000	1000	14.0	1000	58.1	13.5
Figure 4.8b	3000	600	11.3	1000	73.5	5.8
	3000	600	11.3	1500	34.2	6.5
	3000	1000	18.4	1000	20.0	3.0
	3000	1000	18.4	1500	5.9	2.3

Table 4.5: Ambient dose evaluated at the exit area of the pipes for the two configurations of Figure 4.8. The results are the average of 5 simulations starting with different seeds. The central columns gives the sizes of the structure simulated.

However, neither of the two structures provides a configuration in which the external

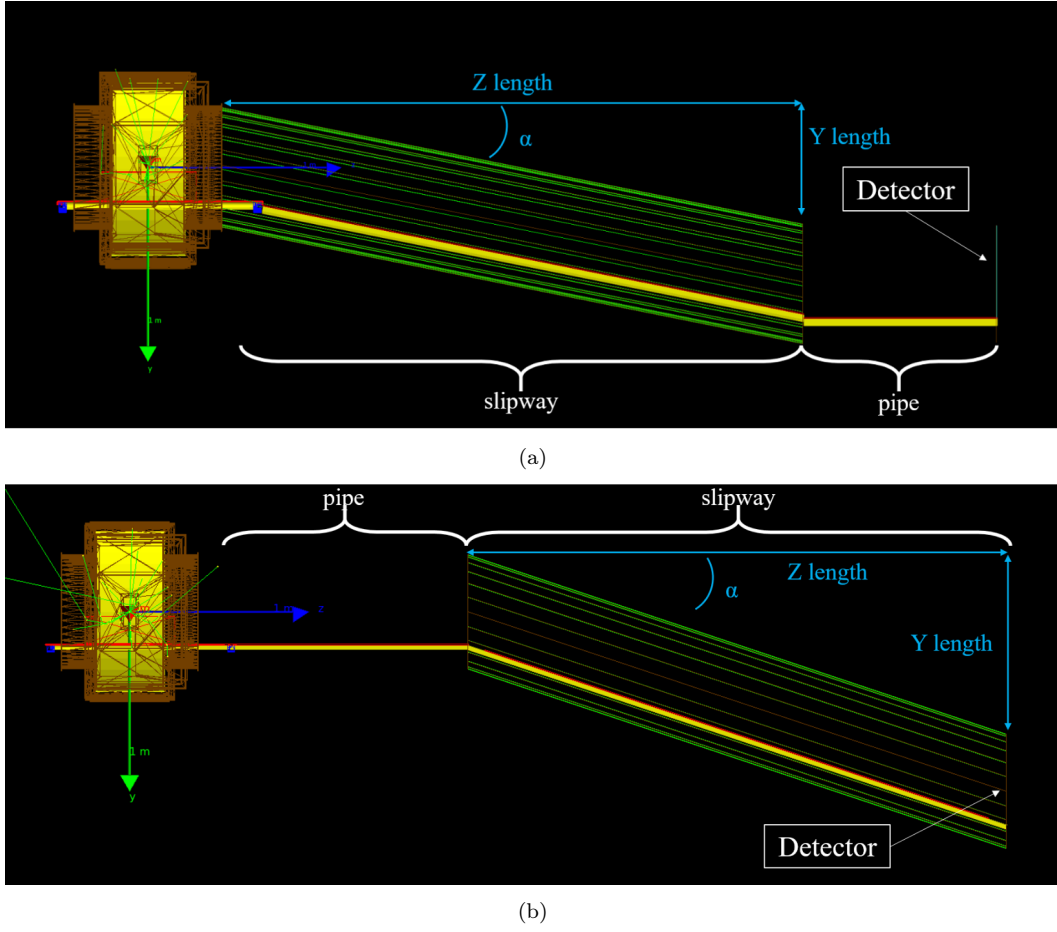


Figure 4.8: Pictures of the structures simulated for the external shielding of Mito500 considering a deviation along the vertical plane. The conveyor belt (yellow) is tilted up to 20° to avoid the roll and drop of the sample scanned. Figure 4.8a shows an inclination immediately after the gantry, while in Figure 4.8b the tilted plane is placed after an horizontal plane.

ambient dose rate falls below the safety threshold of $0.5 \mu\text{Sv/h}$. To achieve such a reduction, the size of the slipway and horizontal pipe would need to be significantly increased, resulting in a very long apparatus. Additionally, the proper elevation of the conveyor belt would have to be maintained, further increasing the overall size of the structure, potentially exceeding 10 m. For these reasons, deviation along the vertical plane is not considered a viable solution.

The second option involves deviating the conveyor belt without altering its elevation, by introducing a 90° or two 60° curves. This approach eliminates issues with object rolling and avoids the need to adjust the height of the conveyor belt. Three simulations are conducted: one to evaluate the ambient dose rate exiting from the gantry and detected after approximately 2 m at *pos 1* (Figure 4.9b), and two others after the conveyor belt curves at 90° angle (*pos 2*, Figure 4.9c) and two 60° angles (*pos 3*, Figure 4.9d). These results are compared to the first simulation to determine the dose rate reduction.

An improvement to this basic configuration involves the addition of elements that can block the continuous scattering of particles within the large internal stainless steel panels. The selected shape for this purpose is a diaphragm, and a study is conducted to evaluate the residual ambient dose rate after placing one or two diaphragms inside

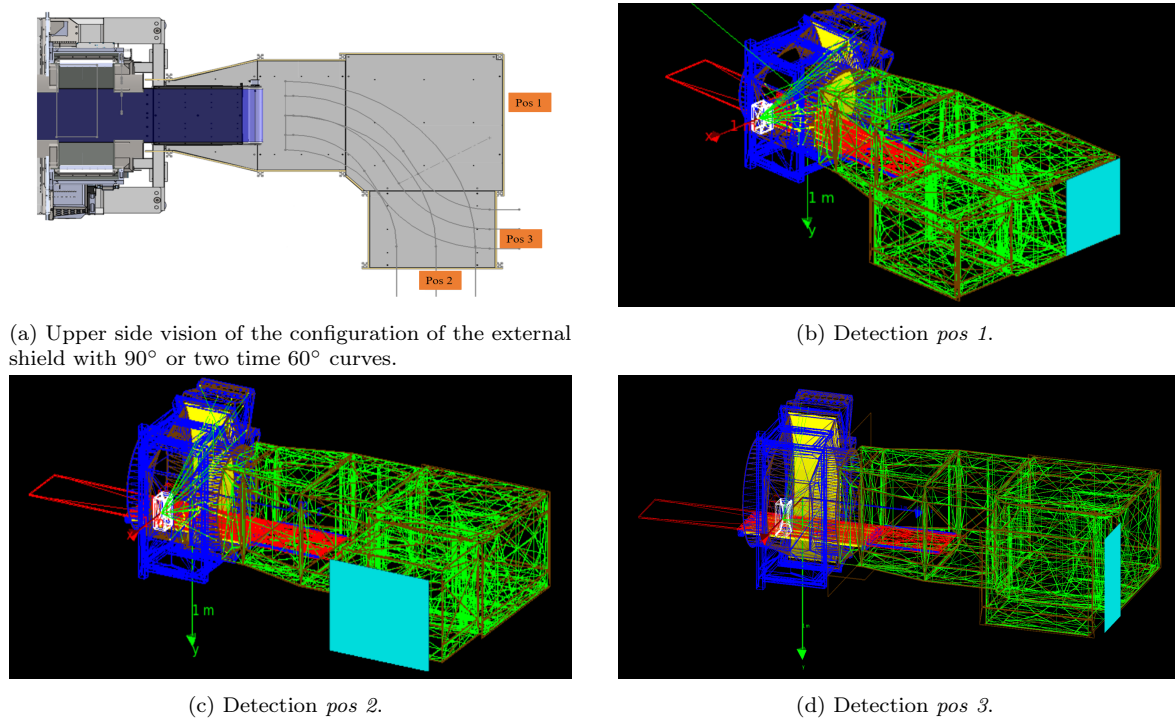


Figure 4.9: Pictures of the external shielding apparatus for Mito500 simulated with the GEANT4 simulation toolkit. These figures show a deviation along the same plane of the conveyor belt of two different angles. Detection areas of the three configurations are highlighted by a cyan squared surface.

the box. The X-ray source spectrum used for these simulations is based on the energy distribution measured with a CdTe solid-state sensor array, which was positioned vertically 18 cm away from the gantry opening of the Mito250. To simulate the energy distribution after interaction with a product inside the gantry, a polyethylene block with dimensions $150 \text{ mm} \times 150 \text{ mm} \times 500 \text{ mm}$ is placed over the conveyor belt, with its center of gravity aligned with the projection of the source position. Although the sizes and energy spectrum of the X-ray tube in the Mito250 are not identical to those in the Mito500, the large amount of X-rays detected are primarily in the energy range of 20 keV to 80 keV, which is also expected for the Mito500 configuration. Therefore, I can approximate this spectrum as representative of the real one.

A circular shape with a radius equal to that of the gantry opening represents the source area for these simulations, with the X-ray beam following the energy distribution shown in Figure 4.10 and an emission angle of 60° . The three images in Figure 4.11 illustrate the positions of the diaphragms (white) used in the simulations. The amount of ambient dose and the ambient dose rate detected at the exit area (cyan rectangle) are presented in the plots of Figure 4.12.

Two types of diaphragm components are considered: stainless steel only, and lead wrapped with 0.5 mm of stainless steel, in order to compare the results. Additionally, for the configuration with lead, three tests are conducted with increasing material thicknesses, while maintaining the same 0.5 mm stainless steel wrap. This allows for an evaluation of the effects of inserting the diaphragms, compared to the configuration without any internal shielding structure.

The use of lead in the configurations with one or two diaphragms (Figure 4.12b and Figure 4.12d) reduces the ambient dose rate by about 60 % for all the thicknesses

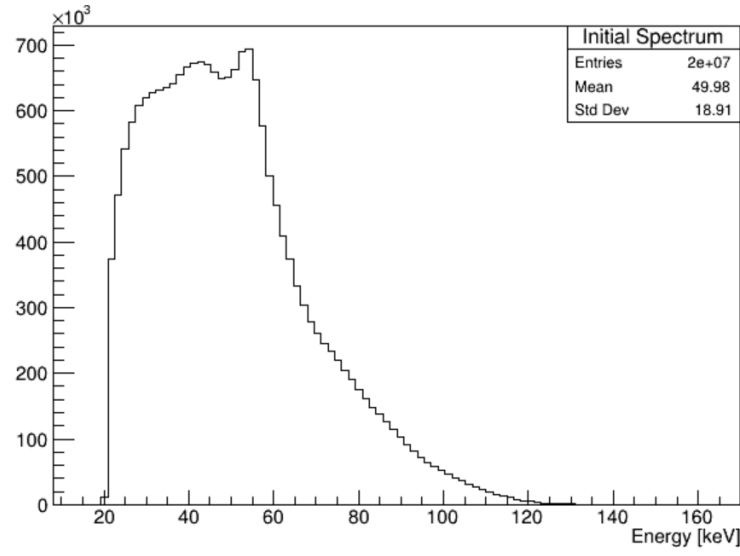


Figure 4.10: The energy spectrum of X-rays exiting the Mito250 gantry after the interaction with a $150\text{ mm} \times 150\text{ mm} \times 500\text{ mm}$ polyethylene block.

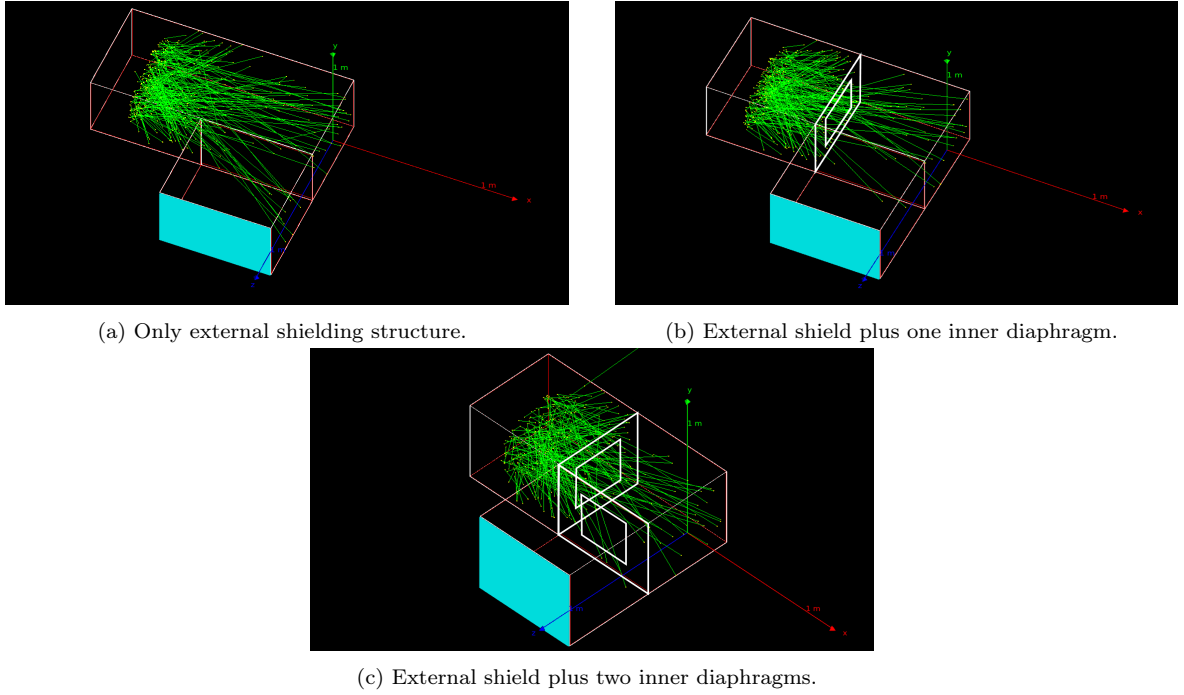


Figure 4.11: Images of the simulations with diaphragms of various materials for ambient dose rate detection after 90° curve. Diaphragm structures are highlighted in white.

tested. In contrast, using only stainless steel (Figure 4.12a and Figure 4.12c) results in reductions of about 40 %, 57 %, and 59 % for thicknesses of 1 mm, 3 mm, and 5 mm, respectively. This demonstrates that stainless steel diaphragms, just a few millimeters thick, can provide blocking capability similar to that of lead. Moreover, stainless steel is cheaper than lead and is an acceptable *FCM*. Therefore, the decision is made to use 3 mm diaphragms of stainless steel only.

Thanks to the simple geometry of the external apparatus shown in Figure 4.9, it is possible to build and place it in contact with the gantry of the Mito500, inside the

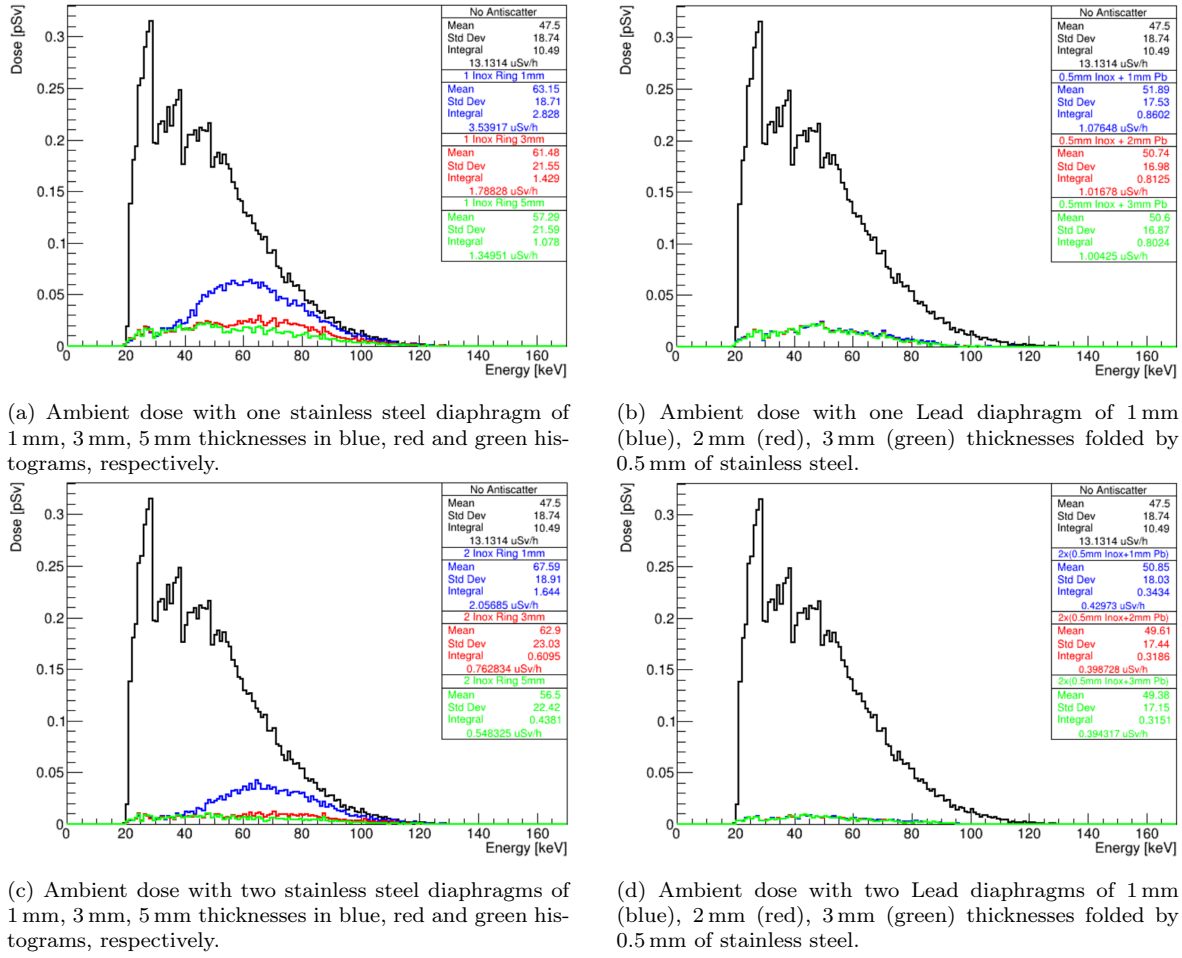


Figure 4.12: Ambient dose distributions evaluated after the 90° curve, considering the addition of diaphragms of different materials. The plots show the comparison for the same substance of different thickness, with the amount of ambient dose rate inserted in each panel.

company bunker. This setup allows for direct ambient dose rate measurements. Table 4.6 presents a comparison between the measured and simulated ambient dose rates. The main difference between the real and simulated environments is the rotation of the X-ray tube and detector inside the gantry. The simulation cannot replicate the movement, so three orthogonal positions of the source-detector are considered: TOP, BOTTOM, and LATERAL. These names refer to the relative positions of the X-ray tube with respect to the conveyor belt. Both in the simulated and real apparatuses, several internal elements resembling the diaphragms discussed earlier are inserted. These elements cover three out of four internal sides of the box, as the fourth side is occupied by the conveyor belt support. Figure 4.13 shows images of the external shield prototype that is developed and studied. The *RaySafe 452* dosimeter is placed at the three positions defined in the simulation, as illustrated in Figure 4.9, though the dosimeter detection area is smaller.

In the simulations, large rectangular areas are used instead of the actual dosimeter surface size to increase the statistics of X-rays reaching the detection region, without increasing the number of initial photons fired. The advantage of this approach is that it allows for a reasonable simulation time. However, this method may result in higher ambient dose rate values in the simulations. This is indeed the case when comparing the



Figure 4.13: Photographs of the external shielding apparatus for Mito500 built in the bunker corresponding to the three positions presented in Figure 4.9. The measures are taken with the *RaySafe 452* dosimeter.

measured and simulated ambient dose rates at the two positions after the conveyor belt curve (*pos 2* and *pos 3*). For these two positions, Table 4.6 shows that the simulated ambient dose rates are nearly double the measured values for all configurations of *pos 2* and for the TOP and BOTTOM configurations of *pos 3*. The LATERAL configuration of *pos 3*, however, yields an anomalous result, with the simulated value being more than three times higher than the measured ambient dose rate. The cause of this deviation is currently under investigation.

A different consideration must be made for the ambient dose rate evaluated in *pos 1*. In this case, the simulated values for the different source positions are always smaller than the measured values, showing a relative difference ranging from about 30 % to 50 %. This behavior can be explained by the differences in simulation geometry compared to the real measurement setup. As shown in Figure 4.13a, the dosimeter is placed directly in front of the gantry overture. However, the shielding structure has a hole due to the absence of one of the lateral sides of the box. This gap allows X-rays to escape from the shielding structure and scatter off the bunker walls and other materials around the gantry, reaching the detection area via a different path. Unfortunately, this feature cannot be accurately reproduced in the simulation because it is impossible to recreate the exact environment inside the bunker during the test.

Detection Position	Dose Measured [$\mu\text{Sv/h}$]	Source placement	Dose Simulated [$\mu\text{Sv/h}$]	Std. Deviation [$\mu\text{Sv/h}$]	Difference [%]
Pos 1	3800	top	2211.7	28.5	41.8
		bottom	2495.5	40.5	34.3
		lateral	1963.7	55.0	48.3
Pos 2	4	top	8.3	2.5	107.5
		bottom	8.9	3.0	122.5
		lateral	7.6	1.8	90.0
Pos 3	20	top	35.5	7.9	77.5
		bottom	36.6	7.5	83.0
		lateral	65.2	7.7	226.0

Table 4.6: Ambient dose rate evaluated in the three different positions of Figure 4.9. The dose measured and the one simulated via the GEANT4 toolkit are compared, showing large differences when it drops under few mSv/h, as the last column highlights.

The validity of this theory is supported by another ambient dose rate measurement campaign. In this campaign, various material thicknesses placed 48 cm from the gantry exit region are tested, as shown in Table 4.7. Two ambient dose rate values were measured with the *RaySafe 452* dosimeter at the same distance for each material. The third column shows the ambient dose rate acquired with an unshielded dosimeter, while the final column shows the values when the dosimeter is wrapped in 1 mm of lead, except for the region in front of the gantry overture. The results show a significant reduction in the ambient dose rate when the scattered X-rays are blocked by the lead foil. Furthermore, the measurements with 2 mm and 4 mm thick lead material reveal that the dose increases as the material thickness increases when measured with an unshielded dosimeter, which is undesirable. This confirms the theory related to the large signal acquisition observed in *pos 1*. Therefore, an accurate measurement of the ambient dose rate would likely result in a reduction of the measured value in *pos 1*, making it comparable to or even smaller than the simulated value, as observed in the other cases.

Material	Thickness [mm]	Ambient dose rate	
		No shielding [$\mu\text{Sv/h}$]	Lead foil [$\mu\text{Sv/h}$]
-	-	85500	67000
Iron	15	900	340
Iron	30	600	160
Lead	2	500	41.2
Lead	4	530	3.2

Table 4.7: Ambient dose rate detected considering various shielding materials placed in front of the dosimeter. A comparison of the dose acquired considering a naked dosimeter placed over the conveyor belt (third column) and the one measured if the dosimeter is wrapped with a 1 mm of lead foil except the detection area in front of the gantry (fourth column).

Following the study of materials and geometries that can be used for shielding

structures in the Mito500 scanner system, and their influence on the final ambient dose rate results, the company has decided to develop a shielding structure similar to the prototype shown in Figure 4.13. The design and production of this external shielding box ensures that the ambient dose rate measured is compliant with the limits set by the International Commission on Radiological Protection.

Thus, the shielding structure described above provides optimal results for the new CT scanner system Mito500 while the smaller Mito250 scanner is equipped with a different suitable apparatus to block X-ray emissions from the gantry openings. However, the main issue with both shielding systems is their length. The size for only one side of the gantry opening in Mito250 is about 85 cm; in Mito500 about 220 cm. This limits the sale of MiCROTEC products in the food industry, as transportation lines are generally designed to fit store dimensions. Therefore, adding a large module of 2.7 m or about 6 m is not always feasible. Moreover, the shielding system for the Mito500 CT scanner is not suitable for creating a competitive product on the market due to the high cost of the coating process. To break away from these limitations a new shielding scheme is considered.

4.3 A Screw Conveyor

A novel approach to shielding the emission of direct and scattered X-rays from the scanner system to the external environment is the implementation of a *screw conveyor*.

The helical design of screw conveyors is widely utilized in the industrial sector, particularly for transporting granular materials like gravel or seeds. Its design allows for movement in any direction, including from ground level to elevated regions, owing to its unique shape and functionality. In this project, the screw conveyor will serve not for product transport but as a mechanism to block photon emissions in areas where employees work. This innovation eliminates the need for the conveyor belt path deviations described in previous geometries and ensures the industrial production line remains compact. A key advantage of this system is its ability to reduce the overall size of the CT scanner by at least one meter, maintaining operational efficiency without compromising safety.

The feasibility of this shielding system is evaluated using GEANT4 simulations. The X-ray source is planar with a circular shape and a radius of 130 mm, matching the gantry overture of the Mito250. The Mito500 apparatus is not considered in this study due to its operational constraints: the helical design separates individual products on the conveyor belt, a process that becomes challenging when handling large items or multi-packs.

The X-rays emitted from the gantry are isotropic, with no preferred direction, and their energy distribution corresponds to the spectrum shown in Figure 4.10. The corkscrew I designed, is created using INVENTOR software, incorporating adjustable parameters for both material thickness and screw pitch. The design is then exported to the GEANT4 simulation toolkit, where it is integrated with the conveyor belt of the Mito250 apparatus. Additionally, a shielding box consisting of 2 mm of lead encased in 1 mm of stainless steel is added to enclose the screw system. Images of the corkscrew are shown in Figure 4.14.

The helical structure is built around a tubular stainless steel support with a radius of 2.5 cm. During operation, the helix must fully cover the gantry aperture, extending

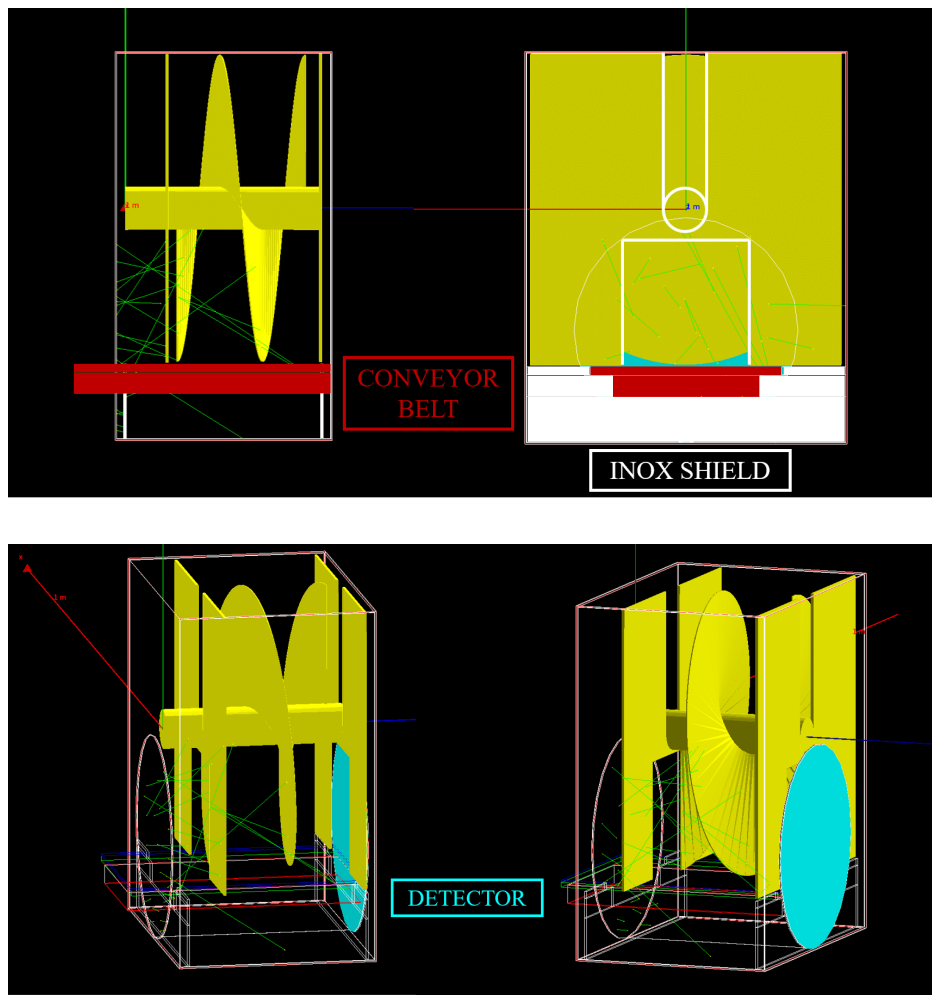


Figure 4.14: Pictures of a screw conveyor system used as shielding system for a CT scanner. The images display the helical structure supported by two layers of 2 mm Lead wrapped with 1 mm stainless steel each with a squared bottom hole where products pass. In red the conveyor belt, while in white an addition of 3 mm stainless steel materials under the belt for blocking the X-rays passing on the bottom region. The cyan surface corresponds to the detection region.

down to the conveyor belt. Consequently, the radius of the helical system must be at least equal to the distance between the highest point of the gantry exit region and the top surface of the belt. In the simulations, a helix with a radius of 155 mm is used. However, the circular design of the helix results in material gaps at the vertex zones within the external shielding box, composed of lead and stainless steel. These gaps allow excess photons to escape the shielding structure without undergoing any interactions. This issue is resolved by adding additional layers of shielding, each consisting of 2 mm of lead encased in 1 mm of stainless steel. These layers are mounted on the front and rear sides of the helix and supported by the external box. In the front view of the system shown in Figure 4.14, the shapes of these panels are emphasized with thicker white lines. The panels feature a square opening at the bottom to allow product passage and a cutout at the top to accommodate the rotating helical structure. The entire mechanism must make contact with the conveyor belt, depicted in red in the figure. Additionally, a white region below the belt represents an extra shielding panel

made of 3 mm thick stainless steel. This panel is positioned both immediately after the gantry aperture and before the detection area to block X-rays escaping beneath the system. Finally, the number of turns in the helix warrants consideration. While theoretically, a single turn would suffice to cover the entire region, the probability of X-rays scattering within the shielding box is not negligible. To address this, the design incorporates one and a half turns.

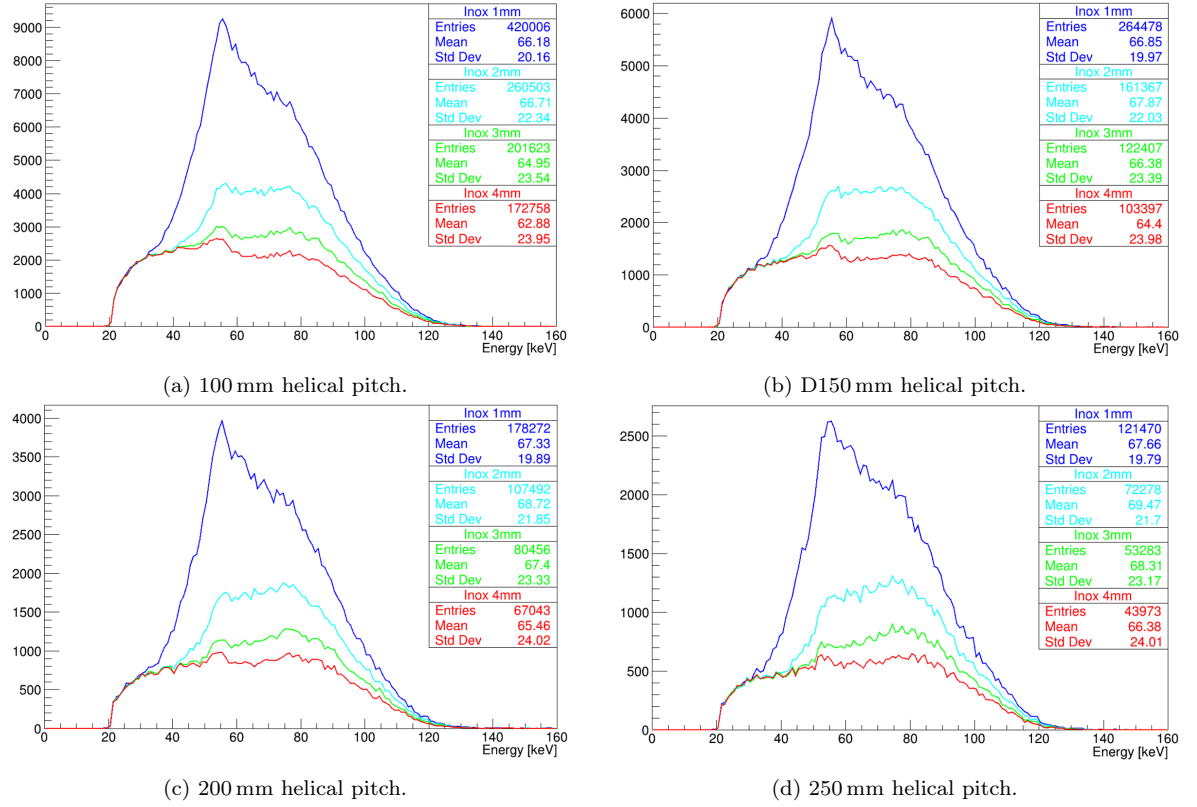


Figure 4.15: Energy distributions of the X-rays traversing the screw conveyor. The four plots display the reduction of the integral values of the energy spectra when the material thickness of the helix increases.

Various simulations of the system are conducted, varying both the thickness of the helix and its pitch. This flexibility is necessary as the company handles a wide range of industrial product sizes, including jam jars, loose olives, and meat boxes. The pitches evaluated ranged from 100 mm to 250 mm, increasing in increments of 50 mm. Figure 4.15 presents the plots of the X-ray energy distribution detected at the exit surface, grouped according to pitch size.

For all pitch configurations, the energy spectra exhibit a sharp reduction in the 50 keV to 100 keV range when the thickness increases from 1 mm to 2 mm and from 2 mm to 3 mm. However, the variation becomes less pronounced when a thickness of 4 mm is considered. The ambient dose rate, calculated when the screw conveyor is used as the external shielding structure for the CT scanner gantry, is evaluated and presented in Table 4.8.

With the screw conveyor as the shielding system, the CT scanner of the company will adopt the structure depicted in Figure 4.16. The external circular pipes of the Mito250, previously equipped with lead-free curtains, will be replaced by a screw con-

Thickness [mm]	Ambient dose rate with helical pitch of			
	100 mm [$\mu\text{Sv/h}$]	150 mm [$\mu\text{Sv/h}$]	200 mm [$\mu\text{Sv/h}$]	250 mm [$\mu\text{Sv/h}$]
1	1127	800	561	434
2	504	372	260	197
3	286	209	144	105
4	198	146	96	69

Table 4.8: Ambient dose rate evaluated for the various screw conveyor configurations described in text. The screw is of stainless steel material, with variation on the thickness of the helix and the pitch.

veyor with dimensions ranging from approximately 200 mm to 400 mm, depending on the size of the products being processed. The modification will reduce the overall dimensions of the scanner by at least 40 cm on each side of the gantry, enhancing its suitability for integration into existing production lines, thereby increasing its appeal to industrial customers.

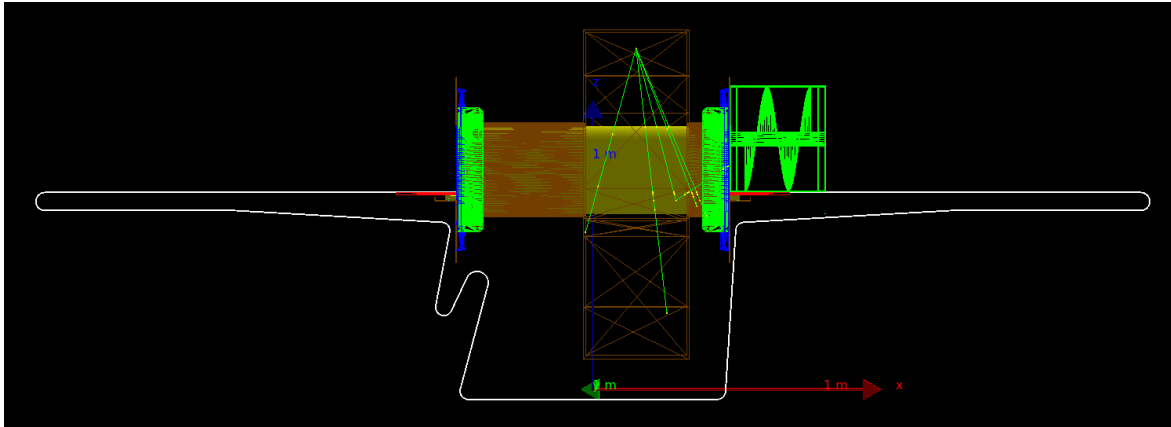


Figure 4.16: Representation of the full CT scanner of the MiCROTEC company with the addition of the screw conveyor as the only shielding structure at the gantry overtures. The 800 mm long circular tubes at the right side of the gantry is here replaced by the longer screw conveyor shielding system of about 400 mm.

The simulated ambient dose rate for the original scanner apparatus, without the external pipes and lead-free curtains, is approximately $3563 \mu\text{Sv/h}$. This value is comparable to $8958 \mu\text{Sv/h}$, the dose recorded by the dosimeter when it is not shielded with lead foil. A comparison with the simulated dose rates listed in Table 4.8 shows that this ambient dose rate is the highest among all the screw configurations considered. The International Commission on Radiological Protection (ICRP) defines $0.5 \mu\text{Sv/h}$ as the maximum permissible ambient dose rate for daily scanner operation.

While the simulated results for the screw conveyor system in the present configuration do not achieve this stringent limit, the shielding structure, if carefully designed and optimized, could offer an alternative solution for scanners with fields of view of a few hundred millimeters in radius.

5 ARCADIA experimental tests

The aim of the study on the best shielding geometry for the CT scanner is both to reduce the emitted radiation and to decrease the length of the entire apparatus. In fact, the size of the scanner could be a problem when it must be installed in the industrial production line. However these modifications must not change the CT acquisition features nor the necessity to use a specific amount of X-rays in a certain time in order to scan and reconstruct an object.

In order to generate and emit the lowest amount of radiation while maintaining the same productivity, and to reduce both the power consumption and the manufacturing costs, a radical customization of the inner parts of the scanner must be considered. The aim of the project pursued during the Ph.D. years wants to prove the possibility of implementing an already existing CT scanner system with a different detection system.

The INFN ARCADIA collaboration designs a FD-MAPS for HEP and space environments but it could be used also for X-rays imaging with few adjustments. The simulations, described in the previous sections, guarantee the possibility of the detector to reconstruct a 3D image and to distinguish different materials of few millimeters thickness.

In this section, experimental tests with the ARCADIA prototype sensor are managed in order to characterize it and investigate its capability in the X-ray imaging field. The FD-MAPS chip corresponds to the third engineering run main demonstrator (MD3).

5.1 Set-up

The sensor prototype described in Section 2.2 provides a digital output, which does not allow for the evaluation of the energy spectra of the acquired data. This acquisition method improves the time resolution of the system, as only the hit information needs to be recorded and transmitted from the pixel matrix to the periphery, where a clock input is included. However, this approach prevents a direct comparison with the simulation studies discussed in the previous sections. Nonetheless, experimental tests are conducted using various X-ray sources. Beams of X-rays with energies up to 160 keV are generated in two distinct ways:

- Radioactive sources
- X-ray tubes of various nominal voltages

5.1.1 Source

Radioactive sources, such as Fe-55, are ideal for the initial characterization of the sensor features, allowing for the evaluation of the signal response and calibration of the chip matrix. Fe-55 primarily emits at 5.89 keV and 6.49 keV, corresponding to two very narrow characteristic lines. The separation between these lines is clearly visible when using a thermo-electrically cooled Si-PIN X-ray detector, as shown in Figure 5.1. However, this detection system is suitable only for applications requiring moderate energy resolution and moderate count rates [48]. The spectrum is acquired with a 6 mm² area, 500 μ m thick silicon detector, and the energy resolution is calculated to be 204 eV FWHM.

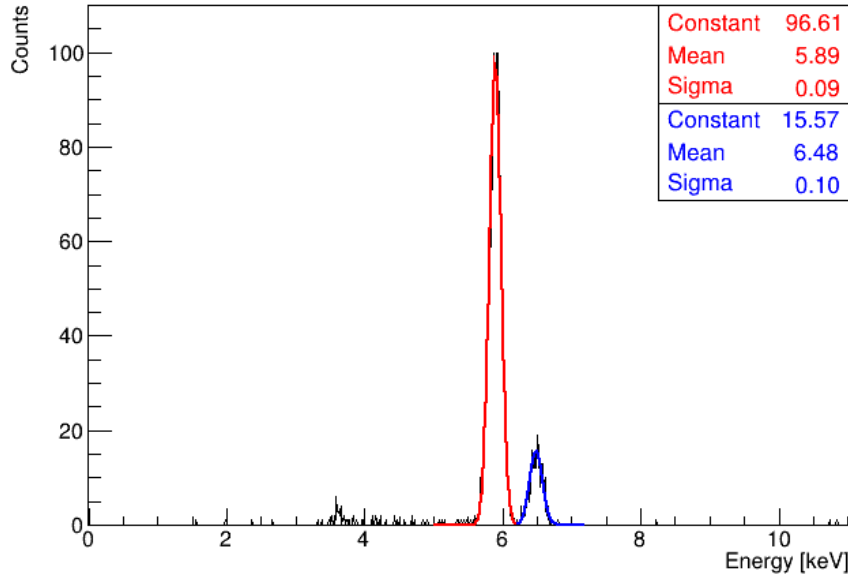


Figure 5.1: Energy distribution of Fe-55 radioactive source taken with the Amptek Silicon detector XR-100CR. The energy resolution is calculated from the gaussian interpolation of the X-ray characteristic line at 5.98 keV with a FWHM value of 204 eV.

Alternatively, when using a sensor from the same family as the ARCADIA prototype, but with an analog readout, such as MATISSE, it is not possible to distinguish the two characteristic lines of Fe-55. As a result, the radioactive source is treated as a monochromatic beam. This implies that the energy resolution of these detectors cannot achieve the same level as the XR-100CR Si-PIN detector, and it is estimated to be an order of magnitude worse [49]. The worst energy resolution associated with the FD-MAPS is expected due to the varying characteristics of the detectors, further related to their operating temperatures: the Si-PIN detector is cooled, while the MATISSE and ARCADIA sensors operate at room temperature.

The second type of source is described in the first paragraph of Chapter 1. The model used for the tests at the Padova University laboratory is a SEIFERT X-ray tube, model DX-W12X04-S (Long Fine Focus W), with a maximum current of 50 mA and a nominal voltage of 40 kV. The X-ray spectrum emitted from the tube source, with the current and voltage set to 2 mA and 40 kV, respectively, and acquired by the XR-100CR Si-PIN diode, is shown in Figure 5.2. Only three of the five characteristic lines of tungsten are visible at 8.4 keV, 9.7 keV, and 11.3 keV, due to the limited energy resolution of the acquisition system. To resolve the lines resulting from atomic electronic transitions that originate from the same energy level, a detection apparatus with an energy resolution better than 100 eV is required. The peaks at energies below 8 keV are associated with characteristic X-ray lines produced by other substances within the tube, which can be effectively blocked using a filter.

However, the energy spectrum acquired by the D-MAP sensor MATISSE does not resolve the individual tungsten peaks, but produces instead a broad peak in the 5 keV to 13 keV range [49]. Therefore, it is not expected that the ARCADIA D-MAPS silicon prototype will achieve high resolution either. The X-ray tube setup used in laboratory is shown in Figure 5.3:

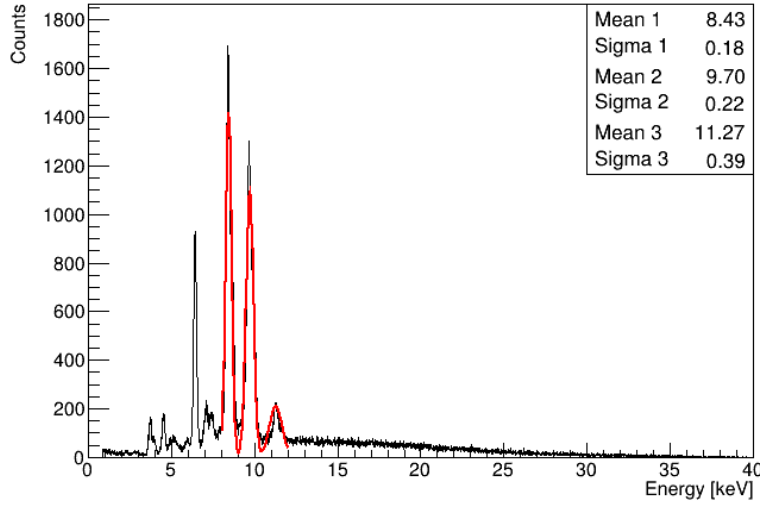


Figure 5.2: X-ray spectrum from the SEIFERT X-ray tube at 2 mA current and 40 kV nominal voltage. Under 40 keV, the Tungsten anode produces 6 emission lines L_{α_1} , L_{α_2} , L_{β_1} , L_{β_2} , L_{γ_1} and M_{α_1} at 8.40 keV, 8.34 keV, 9.67 keV, 9.96 keV, 11.29 keV and 1.78 keV, respectively. The M_{α_1} emission line is not visible, while among the others only the 3 peaks highlighted in red can be distinguished, grouping together the lines produced by the same initial level. The other peaks relate to the X-rays soft components from different substances present inside the tube.

5.1.2 Detector

The ARCADIA MD3 chip is mounted on a Front-End Board (FEB) and installed on a custom-made support made of rigid plastic to prevent any interaction with other substances. A plastic slab with a circular hole at the center is placed over the detector, leaving only its sensitive area exposed. The side of the detector facing the source is referred to as the back-side, corresponding to the surface without the metalization, where the transistors are located. This side is preferred when using an X-ray beam, as the metal layer on the front-side tends to attenuate or completely block low-energy photons. The sensor on its board is shown in Figure 5.4.

Each FEB in Figure 5.4b is connected on one side to the power supplies and on the other to the data acquisition system. Each chip requires two power supplies: one low voltage supply (3 V and 4 V) for the electronics on the board and to power the transistors on the chip, and one high voltage supply (90 V) for the sensor depletion.

The sensitive area of the MD3 chip acquires only spatial information, known as the "hitmap", while time information is captured when the "hitmap" reaches the peripheral region. At the periphery, 16 serializers simultaneously send the data to the data acquisition system (DAQ). The data read by the DAQ consist of 32-bit words, which combine both time and spatial information. Starting from the least significant bit, the data are organized as described in [50] and shown in Figure 5.5:

- 1 bit dummy
- 8 bits for the *one-hot encoded* "hitmap" of the pixel region (HITMAP)
- 7 bits for the address of the pixel region inside the double column (PR address)
- 4 bits for the address of the double column inside the section (COL address)
- 4 bits for the section address inside the array (SCT address)
- 8 bits for the time (TIMESTAMP)

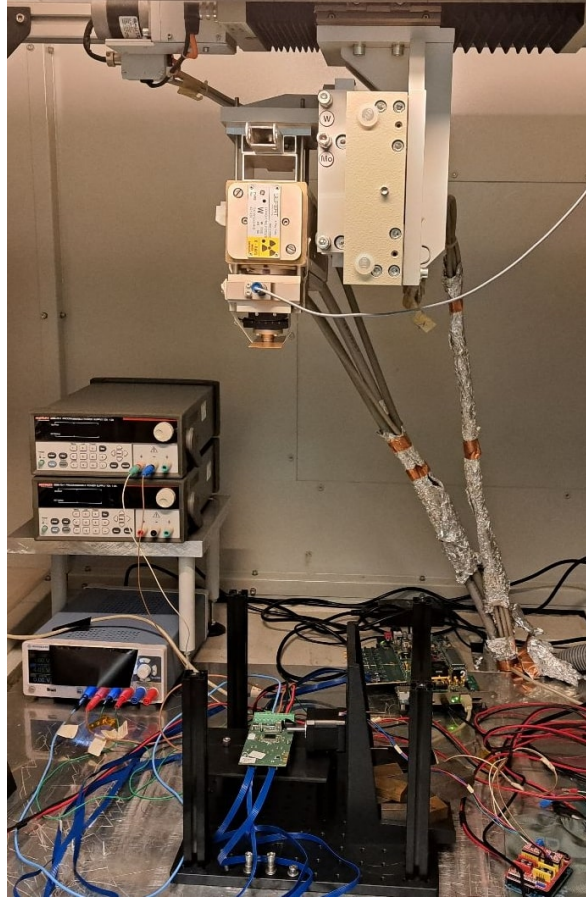
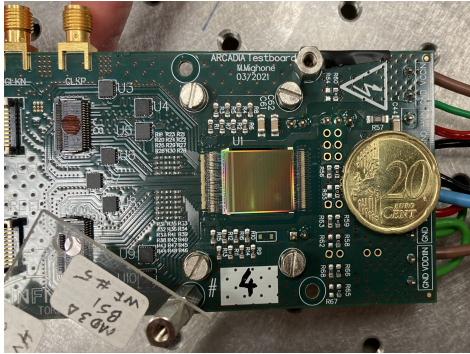
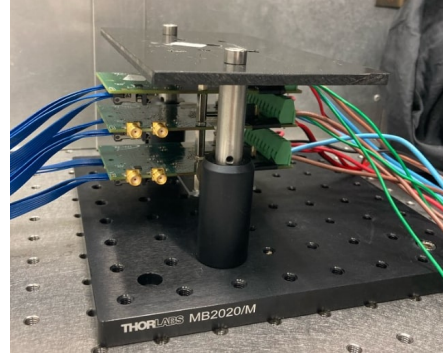


Figure 5.3: Setup of the X-ray tube source from the Physics department at Padova University. The X-ray tube, positioned at the top of the image, features a tungsten anode. A 2 mm aluminum layer is visible at the X-ray emission point. The ARCADIA MD3 chip is aligned with the tube for detection. The sensor electronics is powered by the power supplies visible on the left side of the image.



(a)



(b)

Figure 5.4: Photos of the ARCADIA MD3 sensor. (a) The chip mounted on a FEB. Its sensitive area is $12.8\text{mm} \times 12.8\text{mm}$. (b) Three detectors stacked together at 1 cm distance and installed on a plastic support. Samtec cables communicate with the FPGA, while the cables on the opposite side are connected to the power supplies.

The 32-bit words are transmitted to the Xilinx evaluation board (KC705), which hosts a Kintex-7 FPGA [51]. An additional 24 bits are appended to each event on the FPGA to include the timestamp indicating when the data is read. The augmented data is then stored in a FIFO (First-In First-Out) memory within the FPGA and sent

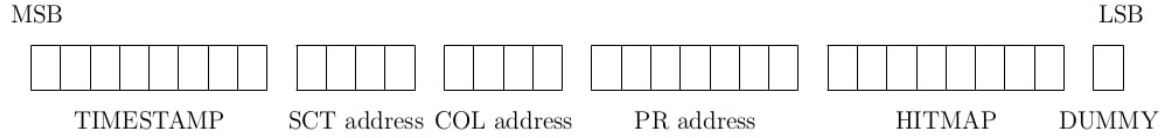


Figure 5.5: Representation of an event acquired by the ARCADIA MD3 chip. The 32 bits give information about time and space of the event [50].

to the target computer via an Ethernet connection. The final data consists of 64-bit words, whose type can be determined by examining the four most significant bits.

A software tool developed by the ARCADIA collaborators is used to acquire the data and save them in a ".csv" file for offline analysis. The timestamp associated with each event must account for both the time provided by the chip periphery and the time recorded by the FPGA. While both times are based on the same clock with a step size of 200 ns, they are not inherently synchronized, necessitating a synchronization process. The *timestamp synchronization* mechanism involves sending a test pulse to the sensor matrix prior to the start of experimental data acquisition. The difference between the two timestamp values is recorded and subsequently subtracted from the FPGA timestamp for each event.

At this stage, the last 8 bits of the 24 bits added by the FPGA align with the 8 bits representing the chip timestamp. The software combines these values into a variable referred to as the *extended timestamp*, augmenting it with an arbitrarily long time maintained by the software. This operation is essential when the 24-bit FPGA timestamp overflows. Specifically, the 24-bit counter resets every 3.35 s, corresponding to $2^{24} \times 200$ ns, but the software timestamp counter continues to increment, ensuring a consistent temporal reference is always available.

The *extended timestamp* operation is a routine consistently performed by the software during experimental acquisition. However, other actions, such as the real-time visualization of acquired events, immediate cluster analysis, or applying a mask for noisy pixels, can be optionally disabled and deferred to offline analysis. This flexibility allows the acquisition mode to be tailored to the operating environment, accommodating varying event acquisition rates, from low to high. Conversely, certain parameters configured before the acquisition run cannot be altered during offline analysis. Among these, two are particularly relevant to the experiments and are described below.

One critical parameter is the VCASN value, an integer ranging from 0 to 63, which sets the threshold for each pixel in the matrix. The relationship between the selected VCASN value and the minimum number of electrons detectable by a pixel is calibrated during experiments using a test pulse. This calibration involves injecting a known amount of energy into the electronic circuitry of the pixel to determine its response. It is essential to choose an appropriate VCASN value for the acquisition run. For reference, a VCASN value of 1 corresponds to a threshold of approximately $835e^-$, while a value of 37 corresponds to about $145e^-$.

Another key parameter is the BACKGROUND REFERENCE (BGR) value, which defines the voltage reference for the analog front-end of each sector. This value depends on two parameters: MEAN and SLOPE, both of which range from 0 to 15. When MEAN is fixed, the voltage tends to decrease non-linearly with changes in SLOPE. Conversely, when SLOPE is fixed, the voltage increases linearly with the MEAN value. The default

setting for MEAN is 7, as this provides the chip with a reference voltage of approximately 600 mV, based on simulations of the analog front-end performed by the ARCADIA collaboration. However, the default BGR settings often fail to deliver the precise reference voltage due to variations in the responses of different sectors. Therefore, an equalization process is crucial to achieve consistent performance across the matrix.

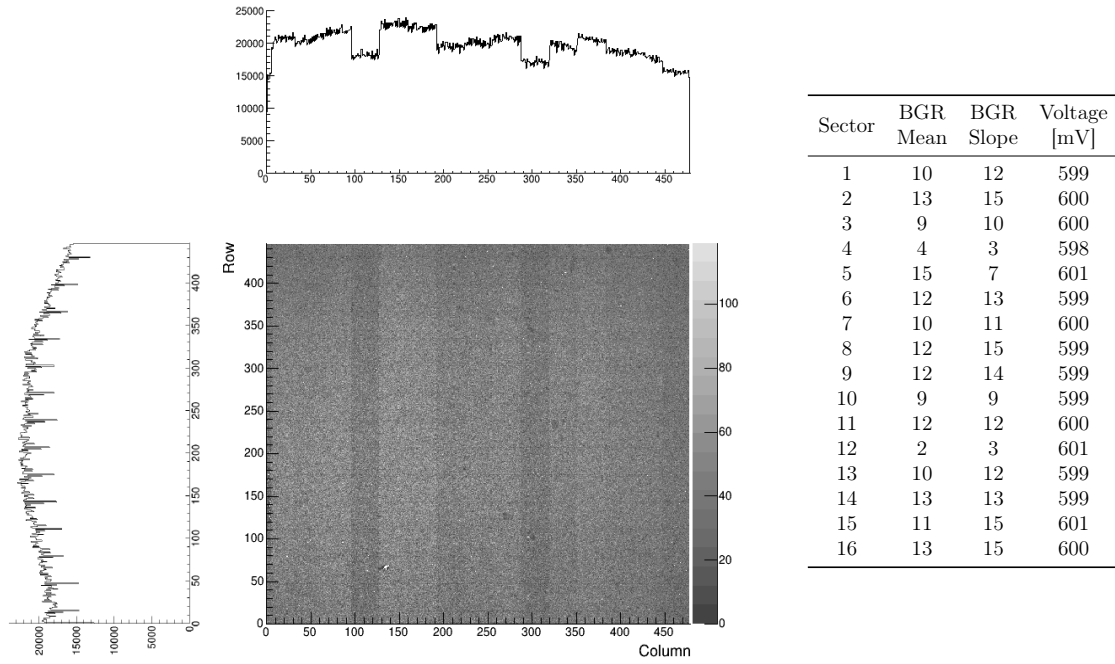
5.1.3 System calibration

A Fe-55 radioactive source is used for pixel signal calibration. In this context, calibration does not refer to associating a channel with the energy released by a particle in the sensor. Instead, it focuses on ensuring the homogeneous detection of signals across all sectors of the matrix. To assess this homogeneity, a two-dimensional image of the chip sensitive area is generated, providing a visual representation of the matrix uniformity. Additionally, projections along the axes are analyzed to quantitatively evaluate the uniformity of the signal response across the matrix.

For calibration, the MD3 chip is positioned with its front side facing the radioactive Fe-55 source. The parameter used for sensor calibration is the BACK-GROUND REFERENCE (BGR), as described earlier. Among the various combinations of MEAN and SLOPE values, the BGR can be adjusted to achieve the target voltage of approximately 600 mV, consistent with the front-end simulations conducted by the collaboration. Table 5.6b lists the parameter combinations and the resulting voltage measured in a test point for each sector, while Figure 5.6a presents the 2D map obtained using these settings.

The 2D map illustrates a matrix of 480×448 pixels instead of the full 512×512 , due to two factors: the last 32 columns (corresponding to the 16th sector) are unresponsive to signal input, and the top 64 rows are obscured by the mezzanine board. The histogram of the row count projections reveals a Gaussian distribution with a mean of 210 pixels and a sigma of 296 pixels, alongside peaks of lower counts spaced at 32-pixel intervals. The Gaussian shape with a large sigma is consistent with the isotropic emission characteristics of the Fe-55 source, which is encapsulated in a dense plastic container with a circular hole for X-ray emission. The plastic thickness blocks some X-rays, which carry an energy of approximately 6 keV, contributing to the observed distribution. The regularly spaced lower-count peaks have been extensively discussed within the collaboration, though no definitive explanation has been reached. One hypothesis suggests that horizontal circuit lines supplying current to the pixels might interfere with the signal acquisition process along the sectors, causing hits at these specific positions to be discarded.

In contrast, the projection along the fifteen active sectors reveals an incomplete equalization among them. While a Gaussian distribution with a large sigma value is also expected along this axis, certain sectors register significantly fewer hits than others. As shown in Table 5.6b, the voltage values differ slightly between sectors because it is not possible to achieve exactly 600 mV using the available combinations of MEAN and SLOPE parameters. This discrepancy causes variations in the hit counts per sector, although it does not fully explain the pronounced differences observed in sectors 4 and 10, for instance. One potential explanation is the variability in the internal electronic circuitry of the pixels, which is not perfectly uniform across the matrix, resulting in a threshold fluctuation for signal detection. Adjusting the VCASN parameter could po-



(a) 2D map of the ARCADIA MD3 matrix after the sector calibration and the relative projections along the axis. (b) Analog front-end voltage values after calibration of the sensor.

Figure 5.6: ARCADIA MD3 matrix is irradiated with a Fe-55 source after the calibration of the sectors. The signal acquisition homogeneity is not perfect even if all sectors are equalized around 600 mV.

tentially equalize the sectors more precisely. However, this adjustment would limit the ability to modify the parameter for different X-ray emission rates during experiments. Consequently, we have chosen to retain the calibration shown in Figure 5.6 as the optimal configuration for this MD3 chip, balancing the trade-offs between equalization and operational flexibility.

5.2 X-ray signal acquisition

Once the primary operations are completed, the chip is ready to be used. Two measurement campaigns are considered: one with the X-ray tube of the Padova University, at the Physics and Astronomy department "Galileo Galilei"; the other with the industrial source of the MiCROTEC company apparatus named Mito250.

The tests at the University deal with the capability of the sensor to acquire a radiography and a set of images of real objects, distinguishing the materials. In Chapter 3, Monte Carlo simulations demonstrate the possibility to discriminate substances of slightly different absorption coefficient when a CT with 90 projections and 4 silicon layers is performed. Here only one sensor is considered, and the analysis of materials is achieved looking at a simple radiography. Then, a full CT of a single element is completed.

The experiments with the industrial CT scanner of MiCROTEC, instead, are aimed at a more specific characterization of the chip, in order to obtain a comparison with the detection system already in use inside Mito250. In this case, a further analysis of data is required, taking into account the clusterization process.

The energy spectra of the two sources have the same shape displayed in Figure 4.1 because the anode is of the same material. The difference is the energy range and the maximum power reached, and thus the lifetime of the system. The laboratory source, in fact, reaches a power emission of 2 kW with 40 kV and 50 mA, while the company X-rays tube has a maximum limit of 500 W with 160 kV and 3.125 mA.

The main issue for tube lifetime is filament breakage. In fact, the application of a high current increases the number of atomic ions evaporating which consumes the filament; thinning at a hot spot then leads to a thermal runaway. Therefore, the tube lifetime depends on the maximum current applied and follows an exponential curve. In a laboratory environment, with proper heating during operation and keeping the whole system cool, tube lifetime can exceed 10 years. For industrial use, a tube must operate continuously 7 days at week, 24 hours a day, thus the current limit must be decreased by magnitude orders. This is the reason for the low current value which characterizes the tube selected by the MiCROTEC company.

The current reduction also represents an important variable regards radiological protection. In fact, the current parameter sets the amount of X-rays emitted by the tube: lowering it, the number of photons decreases and so the dose generated.

On the other hand, the possibility of producing X-rays in the high region of the energy distribution is not null because the voltage is not modified. Therefore, the probability of generating photons which provide high ambient dose is still important. The solution is a tube with lower nominal voltage, but two factors have to be considered: the voltage lower limit and the sensibility of the detector. The low voltage limit is determined by the material density and thickness to be crossed, because the single projections in a CT must show variations of attenuation due to different substances. Sensitivity depends instead on the characteristics of the detector: for the D-MAPS MD3 chip, tests with X-rays show that the lower limit to distinguish a signal from the background noise in a pixel is about $145e^-$, corresponding to a minimum detectable X-ray energy equal to $145e^- \cdot 3.6 \text{ eV}/e^- = 522 \text{ eV}$; therefore the chip shows good behavior for low energy photons detection.

5.2.1 Laboratory experiments

RADIOGRAPHY

Substance discrimination is evaluated using radiography of various objects, two of which are presented here to demonstrate the capabilities of the ARCADIA chip. The samples, each approximately the same size as the sensor, are placed in contact with the back-side of the chip. To ensure safety and prevent potential damage to the chip during placement, the samples are encapsulated in a kapton layer. This set-up also helps minimize geometric distortions in the resulting radiography. Each sample is scanned using a 40 kV X-ray tube operating at a current of 2 mA for a few seconds.

Figure 5.7a shows a radiograph of a light bulb with a broken filament, featuring a surface area of $8 \text{ mm} \times 23 \text{ mm}$ and a width of 8 mm. The bulb consists of a cylindrical glass shell with a thickness of 1.2 mm, filled with a noble gas, and containing a metal filament approximately 100 μm in diameter. The glass thickness and filament diameter are determined directly from the radiography by counting the number of pixels with the same shade of gray and multiplying by the pixel pitch value.

The image on the right in Figure 5.7 depicts a hexagonal plastic nut. Its measured

dimensions include a flat-to-flat diameter of 5.7 mm, a distance of 1.47 mm between the external and internal borders, and a thickness of 3 mm. The dimensions calculated from the radiography are summarized in Table 5.1. While the measurements from the radiography are slightly larger, this discrepancy is attributed both to the fixed pixel pitch of the detector and the possibility of placing the nut not exactly parallel to the sensor area, due to the thickness of kapton layer. Overall, the agreement between the calculated and actual measurements is satisfactory.

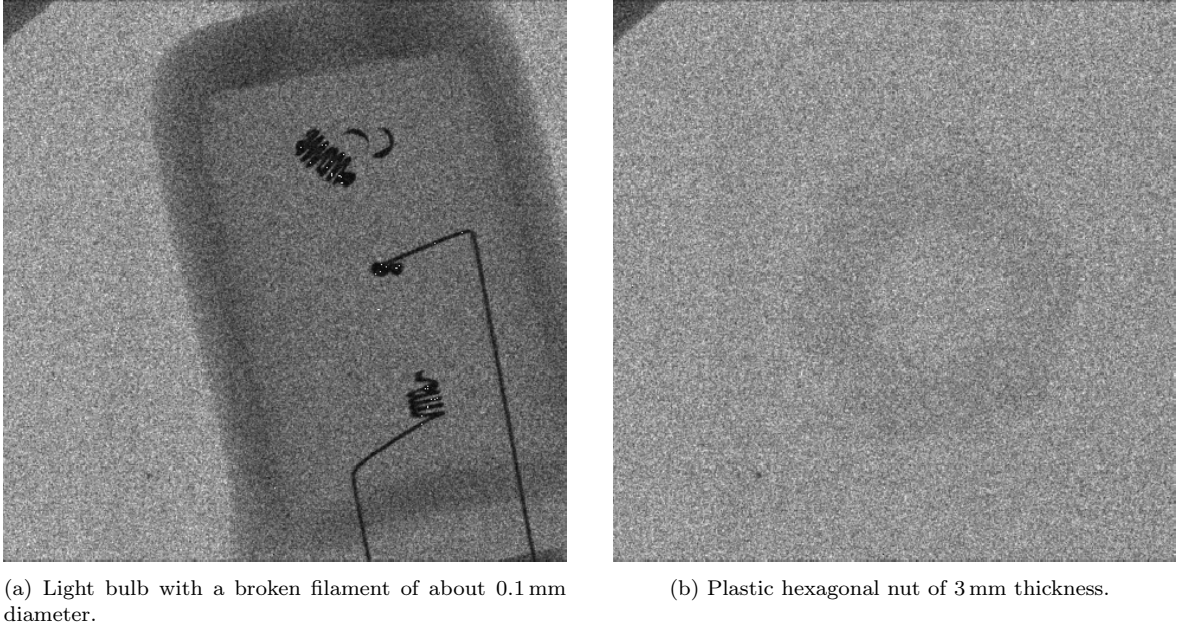


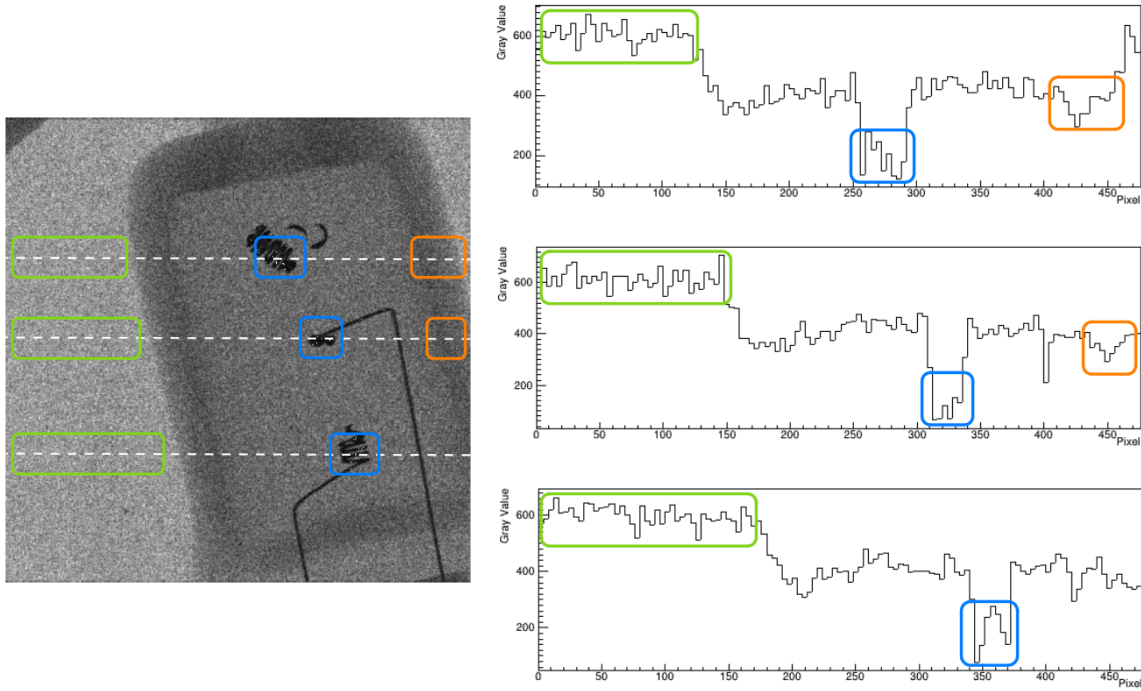
Figure 5.7: Radiography of two small objects. The samples are placed under a 40 kV X-ray tube with a current of 2 mA for one second. X-rays signals are detected by the ARCADIA D-MAPS Silicon prototype.

Measure of nut size		
	Gauge [mm]	Image [mm]
Across flat	5.70	5.75
Between borders	1.47	1.5
Thickness	3	-

Table 5.1: Hexagonal plastic nut sizes. The first column reports the real measure with a digital gauge, while the second column the dimension achieved from the radiography analysis with *Fiji* software.

Using the data decoding structure originally implemented in Python software [49], I developed a C++ version, achieving a significant improvement in the speed of analysis. The radiographs are generated by normalizing the counts to the acquisition time under consistent X-ray tube parameters: 40 kV, 2 mA, and a 2 mm aluminum filter. This standardization allows for direct comparison of the images. The 512×512 pixel matrix is not fully examined due to two constraints: the 16th sector is non-responsive, and the upper portion of the matrix is obscured by the mezzanine board. Additionally, the circular hole in the plastic support does not entirely align with the chip active area, resulting in a small black triangle visible in the top-left corner of both radiographs.

Despite these limitations, the objects in the radiographs are visually distinguishable. Further analysis reveals variations in gray levels corresponding to differences in material density. Figure 5.8 and Figure 5.9 present plot profiles for selected rows of the radiographs. These profiles correspond to the dotted lines marked on the left 2D maps, with colored regions highlighting zones composed of different materials.



(a) Radiography of the light bulb with a broken filament.

(b) Profiles at 3 different positions.

Figure 5.8: Three profile plots of the radiography on the left are presented, highlighting with colors the different substances encountered by the X-rays. Green corresponds to air (the background), blue the metal filament, and orange the external glass of the bulb.

For the light bulb, the profile plots clearly distinguish three regions:

- Background: around a gray scale value of 600
- Glass: falling within the range of 300 to 450
- Metal: with values below 200

All three plots show spikes in gray values at the positions of the metal coil, highlighted with blue squares. Notably, the first and third plots reveal two distinct peaks at these locations, indicating that the detector can differentiate between two neighboring spires of the same coil. In contrast, the orange zones in the first and second plots demonstrate the detector response to the same glass material with differing thicknesses. This variation occurs because the cylindrical structure of the bulb (1.2 mm thick glass) appears thicker at its borders than at the center. This effect is likely due to beam hardening, where low energy X-rays are preferentially attenuated along longer path lengths in a polychromatic X-ray spectrum.

The comparison of background gray values in both radiographs shows consistent results. The mean values for the five profiles (Figure 5.8b and Figure 5.9b) are around

600. Consequently, with a standardized setup, the gray values associated with specific materials can help identify substances in unknown objects.

For the plastic hexagonal nut (3 mm thick), Figure 5.9a shows minimal contrast from the background. However, the material is clearly identifiable in the profile plots, with an approximate difference of 10 % relative to the background. In Figure 5.9b, averaging at least 20–25 pixels along the columns is necessary to approximate the true gray value due to substantial fluctuations. Alternatively, contrast enhancement can be achieved by smoothing the 2D map. This involves replacing each pixel value with the average count of its neighbors in a 5×5 pixel region, as described in the master thesis of a group collaborator [49].

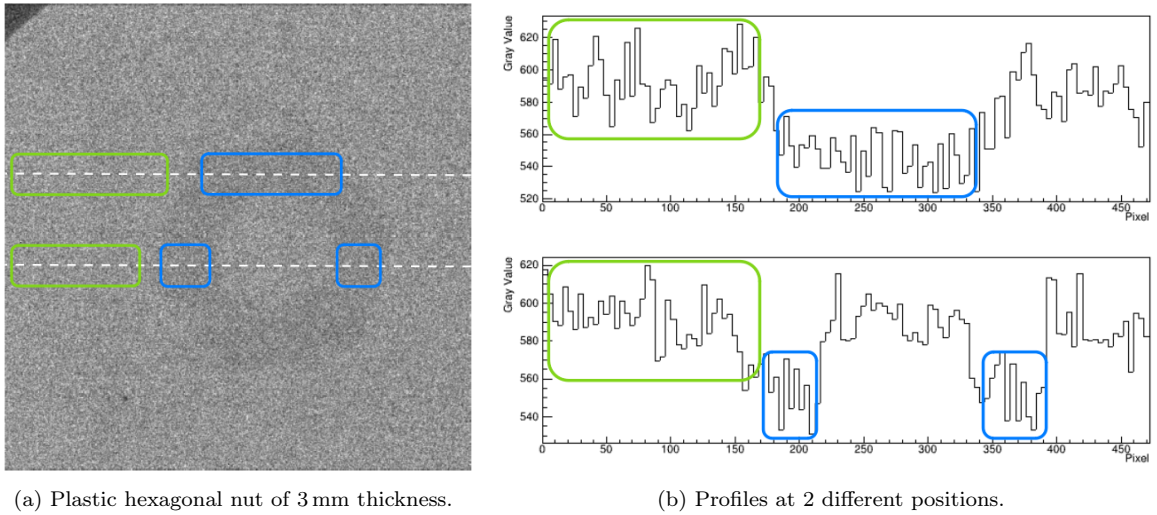


Figure 5.9: Two profile plots of the radiography on the left are presented, highlighting with colors the different substances encountered by the X-rays. Green corresponds to the background, thus air, and blue the plastic material.

COMPUTED TOMOGRAPHY

The next phase of the study focuses on reconstructing a 3D representation of an object. A tomographic image is produced using a CT scanner, where the X-ray source and detection system rotate around the sample. In the laboratory setup, however, the source remains stationary and the object is rotated instead. This arrangement, shown in Figure 5.10, utilizes a stepper motor to which the sample is attached, allowing for a minimum rotational step of 1.8° . The sample selected is an iron screw of 8 mm length, 5.8 mm diameter at the head position and a major diameter of the thread length equal to 3 mm, with a screw pitch of 0.5 mm. To ensure stability and reduce precession motion and parallax effects, an aluminum optical breadboard secures both the detector support and the stepper motor. The screw is connected to the motor shaft using a heat shrink tube made of polyolefin material, with a thickness of 2 mm. The material density, ranging from 0.97 g/cm^3 to 1.33 g/cm^3 , is comparable to air, ensuring it does not interfere with the captured projections. To minimize image distortion and maintain accurate size measurements, the distance between the object and the detector chip is kept as small as possible.

The rotation of the screw is managed using an *Arduino* micro-controller, which ensures precise control over the motion. Projections for the reconstruction are acquired each time the screw is stopped. Although this method differs from a conventional CT

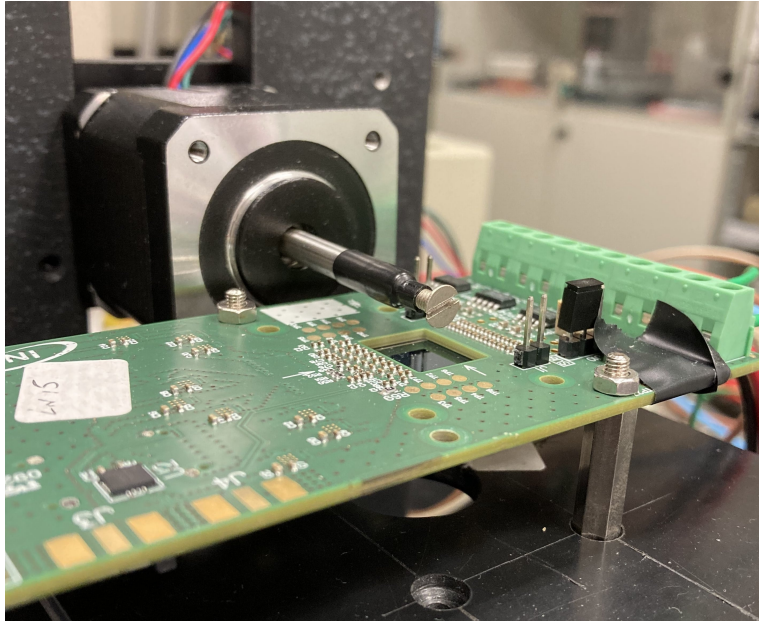


Figure 5.10: Set up for the 3D image reconstruction of an iron screw of $5.8 \text{ mm} \times 5.8 \text{ mm} \times 8 \text{ mm}$ [49]. Projections of the complete CT are acquired using an X-ray tube of 40 kV nominal voltage and the ARCADIA prototype D-MAPS sensor. The screw is connected to a step motor, so that it is rotated at steps of 1.8° .

acquisition, it simplifies the 3D reconstruction process. To generate consistent radiographs, each projection is normalized to the acquisition time and the mean background value, resulting in radiographs that represent 1 s of exposure and are already corrected for the initial intensity I_0 .

As demonstrated in the radiographs of the hexagonal plastic nut, dimensions can be calculated from individual projections. Two examples of such projections are shown in Figure 5.11, with the measured and calculated dimensions summarized in Table 5.2. This comparison validates the accuracy of the dimensional analysis based on the acquired projections.

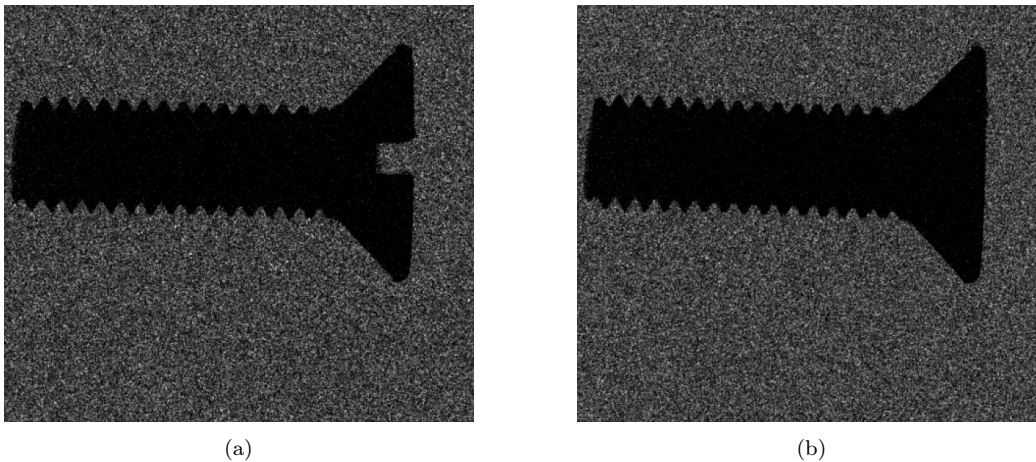


Figure 5.11: Two projections of the iron screw. The threads are perfectly distinguishable. (a) Starting position with the screw drive horizontal indentation completely visible. (b) Projection after a rotation of 21.6° .

Measure of screw size		
	Gauge [mm]	Image [mm]
Head	5.80	5.95
Minor diameter	3.0	3.0
Length	8.0	9.5
Pitch	0.5	0.5
Recess	-	1.0

Table 5.2: Iron screw dimensions. The first column reports the real measure with a digital gauge, while the second column the dimension achieved from the radiography analysis with *Fiji* software.

The dimensions evaluated from the projections are highly precise and align well with manual measurements taken using a digital gauge. However, for the length measurement, the calculated value shows slightly more deviation compared to other dimensions. This discrepancy may be due to magnification effects caused by the object distance from the detector and the use of a cone beam source. As a result, the elongated parts of the projection appear slightly distorted.

A total of 100 projections, spanning a 180° rotation around the screw central axis, are analyzed to generate sinograms for the 3D reconstruction. Each projection is divided into vertical slices of 5 pixels, equivalent to $125\text{ }\mu\text{m}$, resulting in 96 regions. For each region, the mean hit values along the rows are computed to provide the distribution representing the line integrals of the slice at a given angle. This distribution is transformed using the negative logarithm of the values, already normalized to the initial intensity. By overlapping the 2D Radon transforms across all angles, a sinogram for each slice is generated. Ultimately, 96 sinograms are processed for the reconstruction. The 3D image reconstruction is performed using the filtered back-projection method in the *Fiji* software, applying a rectangular filter. A more detailed description of the mechanism for this specific case is presented in [49].

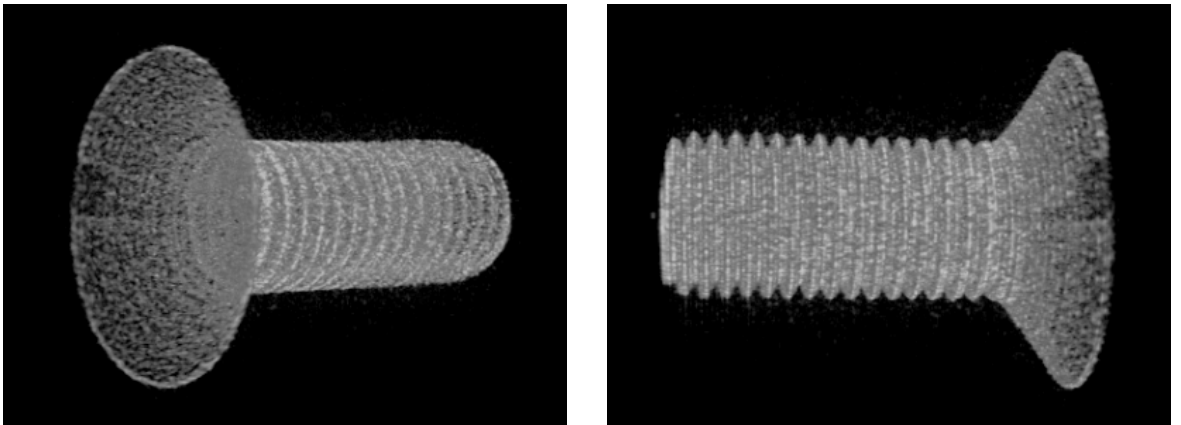


Figure 5.12: 3D reconstruction of an iron screw using the ARCADIA D-MAPS prototype as detector.

The resulting 3D image, shown in Figure 5.12, faithfully reproduces the threads of the screw. However, the horizontal indentation of the slot screw drive is not clearly visible. Only part of the cavity on one edge of the screw head is discernible, but

the reconstruction software could not differentiate the horizontal indentation from the background or the screw body. Several factors may have contributed to the discrepancies in the reconstructed shape: precession motion, observed in the projections, that could introduce artifacts; the rectangular filter used in the back-projection, that may not have been optimal for preserving fine details; the threshold applied to differentiate the screw from the background in the 3D volume, which renders the screw head region excessively transparent, obscuring finer details.

Despite these limitations, the reconstructed 3D volume demonstrates the capability to accurately resolve the screw threads, exhibiting the system potential for detailed tomographic imaging.

RATE MEASUREMENT

The characterization of the ARCADIA chip proceeds with an analysis of the particle acquisition rate. For this purpose, the bare chip is positioned under the 40 kV X-ray source of the university laboratory, using a 2 mm aluminum filter. The current is gradually adjusted, starting from the minimum stable value, to determine the acquisition rate curve.

The VCASN parameter plays a crucial role in this process. It sets the acquisition threshold for each pixel in the matrix, meaning that increasing this value raises the probability of detecting a signal, thereby enhancing the acquisition rate. In this study, the parameter is set to 4, corresponding to a threshold of $625e^-$, which is equivalent to 2.25 keV.

An issue is immediately identified during data acquisition: when the entire chip matrix is exposed to the tube source for more than a few milliseconds, events are lost. Under these operating conditions, the rate is estimated to exceed 18 MHz/cm^2 , which surpasses the capacity of the acquisition system. Figure 5.13 illustrates the 2D map obtained from the acquisition. Aside from three sectors that exhibit different equalization compared to the others, resulting from the use of a different MD3 chip than the one calibrated earlier, a recognizable pattern emerges across the sectors. Data from the top side of the matrix appears to be properly acquired and processed by the hardware, whereas a significant loss of information is evident on the bottom side.

This discrepancy is attributed to the limitations in the acquisition software of the DAQ system, which provides slow communication of all the signals acquired at such high rates across the entire matrix area. Consequently, a significant portion of the data is discarded, preventing the determination of the total acquisition rate. The observed lack of information in the bottom region of the matrix aligns with the read-out mechanism. Specifically, data must propagate to the periphery before spatial information is acquired and a timestamp is assigned. The system prioritizes the earliest detected events at the top of the matrix. Therefore, when the event rate approaches or exceeds the periphery read-out rate, data from the bottom region of each sector is inevitably discarded.

To address the issue, the entire 512×512 matrix is masked, leaving only a 16×16 pixel region in one sector of the chip active for subsequent testing. In this way the lack of information observed in Figure 5.13 does not occur.

The acquisition rate is subsequently evaluated by setting the X-ray tube current to 2 mA and incrementing it in steps of 2 mA up to 40 mA. The collected data are then analyzed using C++ offline software equipped with a cluster analysis function. This

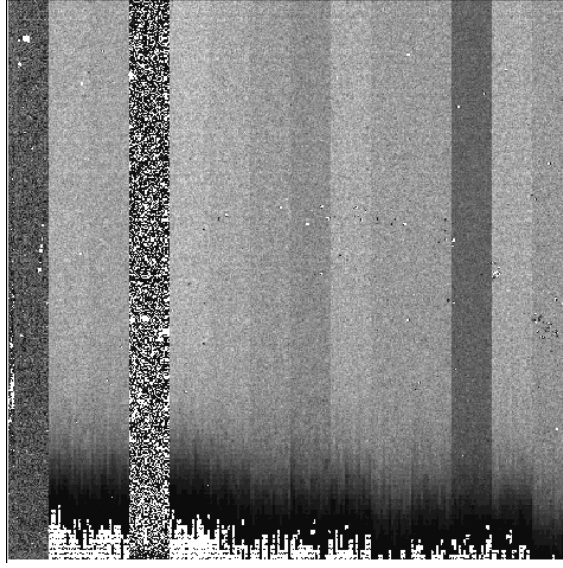


Figure 5.13: 2D maps of another MD3 ARCADIA sensor when an X-ray beam of 40kV voltage and 2mA impinges on the pixel matrix. The VCASN parameter is set to 4. The emission rate of 18 MHz/cm^2 is high enough to avoid the reading of all the hits acquired instantaneously along the same column, so that events on the bottom side are discarded.

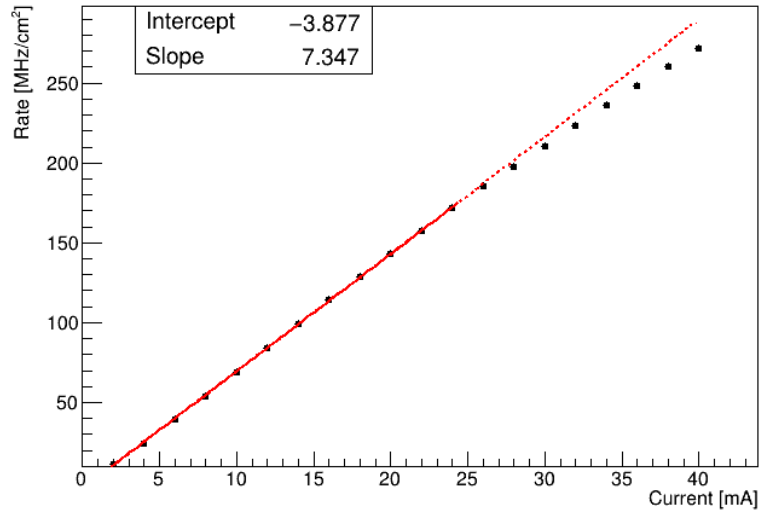


Figure 5.14: Plot of the cluster rate considering the X-ray tube of 40kV nominal voltage. Only a sub-matrix region is considered for the experiment, a 16×16 pixels square along one central sector of the ARCADIA MD3 chip. The reason lays in the low capacity of the acquisition system of storing all the information about the 512×512 matrix at high rates. VCASN sets to 4 and 2mm Aluminium filter at the exit of the tube.

function identifies neighboring pixels activated within a $5 \mu\text{s}$ time window as belonging to the same event hitting the detector, grouping them into a single cluster.

The number of clusters acquired per second by the MD3 chip is normalized to the active area, yielding the acquisition rate values presented in Figure 5.14. The error bars are so small that they are obscured by the markers. The data points in the plot are fitted with a linear curve, resulting in a slope of $7.35 \text{ MHz}/(\text{cm}^2 \cdot \text{mA})$. Linearity is maintained up to 24 mA, corresponding to approximately 180 MHz/cm^2 . This rate is notably high, nearly double the expected rate based on the chip design, but this is due

to the small area selected for the test. Beyond 24 mA, the data points begin to deviate from the linear fit, indicating a decline and a tendency toward saturation.

The acquisition rate study demonstrates the ability of the ARCADIA chip to handle high-rate data detection. However, the DAQ system needs to be redefined to support the storage of all the matrix information.

5.2.2 Industrial tests

Industrial tests for food inspection are conducted in collaboration with the MICROTEC company. Within the company, I have the opportunity to directly investigate the MD3 chip using industrial CT scanners. The X-ray tube, with a nominal voltage of 160 kV and a current set to 3.125 mA, is not suitable for the detection system I proposed. For the test, the current is initially reduced to its minimum stable value of 100 μ A and then gradually increased until the DAQ system encounters issues. The chip area is limited to a 32×32 pixel region, corresponding to the width of a single sector, and no objects or filters are placed between the source exit and the chip.

The Mito250 CT scanner is designed with a rotational gantry. For the upcoming tests, the gantry is fixed in a vertical position, with the X-ray tube exit point directed towards the upper side of the conveyor belt. The MD3 chip, placed in its plastic support, is positioned on the belt with the sensitive area nearly directly beneath the tube exit point projection. The distance between the tube and the chip is calculated to be 445 mm.

Two tests are conducted to evaluate the acquisition rate, varying the VCASN parameter. Figure 5.15 shows the 2D maps obtained with a low pixel threshold, while Figure 5.16 displays the maps with the highest possible threshold.

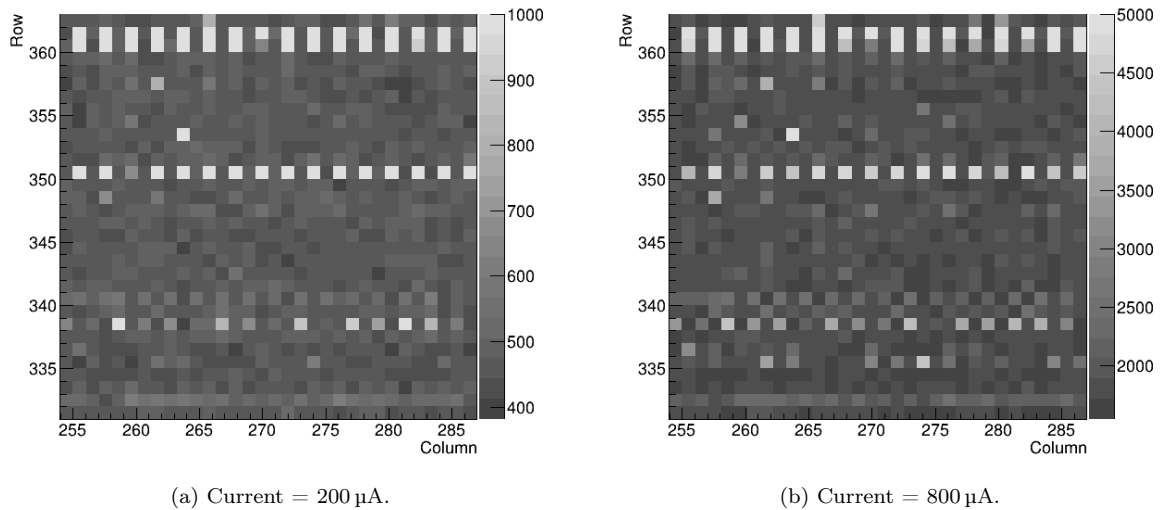


Figure 5.15: 2D map of the 32×32 sub-matrix of the ARCADIA MD3 chip placed into Mito250 under a source of 160 kV and variable current. The threshold is the lower value at which hits can be distinguished from noise, thus VCASN is set to 30. A cut in the acquired event per pixel is inserted for a visualization purpose: it is set to 1000 in (a) and to 5000 in (b).

All the plots are normalized for acquisition time, showing an increase in the number of events observed as the current rises. The 32×32 pixel matrices in Figure 5.15 show

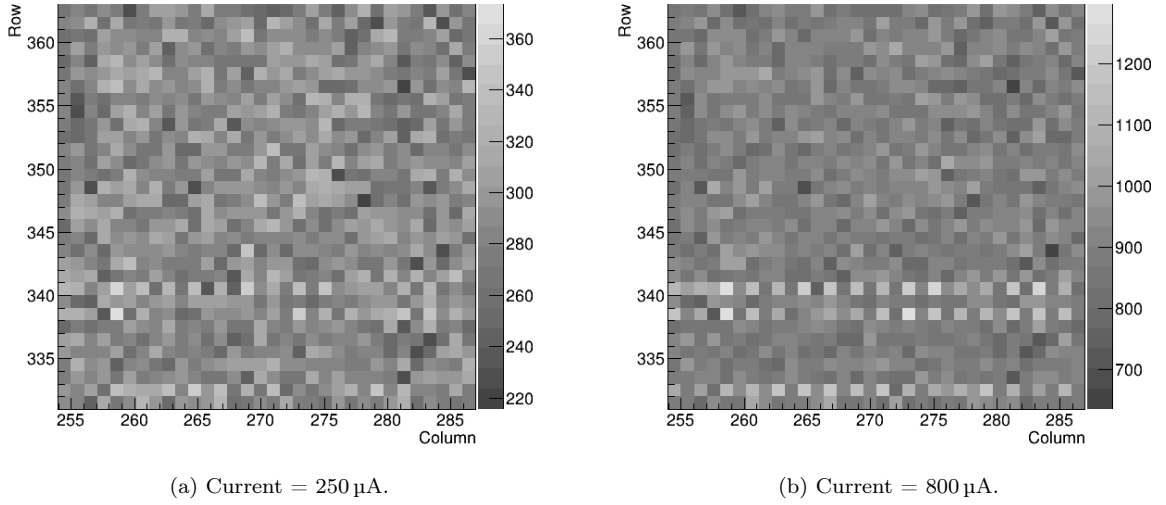


Figure 5.16: 2D map of the 32×32 sub-matrix of the ARCADIA MD3 chip placed into Mito250 under a source of 160 kV and variable current. The threshold is the highest possible, in fact VCASN is set to 1.

a few blank pixels, mostly in the top area, corresponding to pixels that collect a high number of X-rays in one second. A pattern is clearly visible at this threshold, where only odd pixels tend to exhibit a higher acquisition rate. These pixels extend the z-axis to high values, reaching over 3000 counts/s/pixel at 200 μ A and 12 000 counts/s/pixel at 800 μ A, which causes a lack of visual differentiation among pixels with lower acquisition rates. To improve visualization, a cut-off is applied to the number of events collected per pixel. For this threshold, the optimal acquisition condition involves masking all pixels with an acquisition rate more than 3 sigma away from the mean value.

The plots in Figure 5.16, on the other hand, do not show any blank pixels. Instead, they exhibit a nearly homogeneous collection of events. The tube conditions are similar to those of the matrices described earlier, highlighting the importance of the VCASN parameter. No cuts are applied in these plots, and the z-axis values decrease by approximately one order of magnitude for both selected currents.

A distinct pattern is observed across all acquisitions: two rows on the bottom side of the plots tend to acquire more events than the rest of the matrix. Notably, the even pixels of rows 338 and 340 in all four figures appear lighter than the others, suggesting that this pattern is not related to the acquisition threshold set for the pixels. A similar trend is seen in Figure 5.6a, but with the opposite effect: a repetitive pattern of lower event counts is observed along the row projection. Therefore, an explanation for this phenomenon may lie in the electronic circuitry design of the chip.

The low pixel threshold (VCASN set to 30) is used to evaluate the cluster acquisition rate at the voltage values of the CT scan apparatus. Points ranging from 100 μ A to 1 mA are considered for voltages of 80 kV and 120 kV, while at 160 kV, the study starts at 20 μ A and goes up to 1.1 mA. The last case, shown in Figure 5.17c, highlights the instability of the X-ray tube emission under 100 μ A, which is observed as a constant emission rate, as well as the onset of DAQ issues at around 700 μ A, corresponding to approximately 110 MHz/cm². A few points are missing between 200 μ A and 400 μ A, but acquisition at the other voltage values shows a linear increase in the cluster rate,

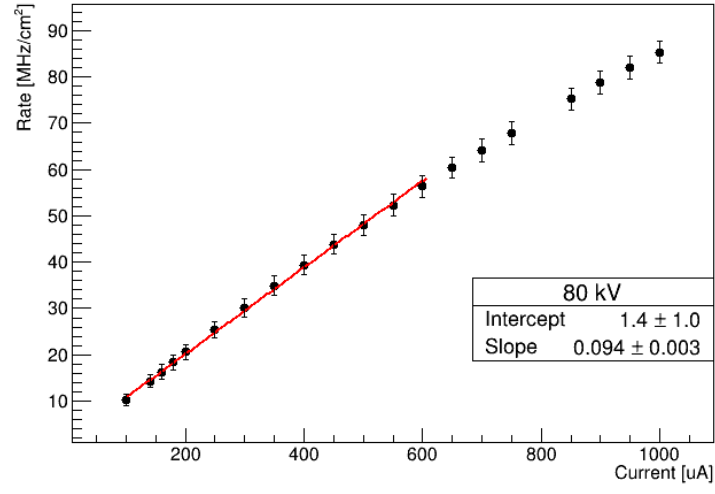
allowing the data to be linearly fitted for this case as well.

The three plots show a linear increase in the rate up to about 110 MHz/cm², followed by a decrease in the signal rate. This trend is particularly evident with a 160 kV source, while at 80 kV, only the increasing slope is observed within the current range considered. At the 120 kV source, the maximum rate is reached at 1 mA. For comparison, all the plots are fitted in the range from 100 μ A to 600 μ A. The linear interpolation reveals different slope values, with the gain varying as a function of the voltage. However, the gain is not directly proportional to the voltage: the slope at higher voltages is smaller than expected. This behavior may be due to the high VCASN value, which increases the number of events acquired per pixel, thereby slowing down the DAQ system.

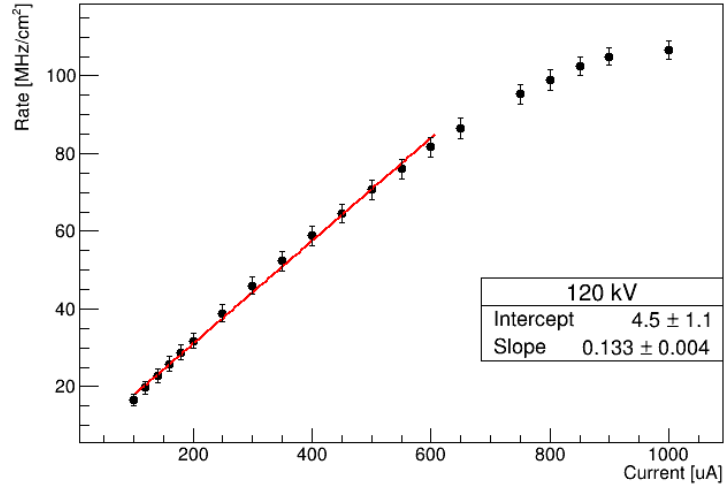
The X-ray emission and detection processes are defined in Chapter 3 as a combined Poisson distribution, with the error corresponding to the square root of the detected signal. Consequently, the error bars represent the square root of the expected value. These error bars tend to increase after reaching the cluster rate maximum value, as the expectation is a reduction in the number of events acquired due to limitations of the DAQ system, loss of information in the bottom region of the matrix, and increased uncertainty during the clusterization process.

A similar test is conducted with a higher threshold value (VCASN set to 1) and the addition of a sample between the X-ray source and the detector. The objective of this test is to directly compare the performance of the MD3 chip with the Mito250 detection system under the same conditions. Data are acquired at 250 μ A, 400 μ A, 600 μ A, and 800 μ A for the three working voltages mentioned above. A solid polymethyl methacrylate (PMMA) cylinder is inserted with thicknesses of 0 mm, 40 mm, and 83 mm. The current range is chosen to avoid both the instability in the X-ray tube emission at lower values and the reaching of the acquisition rate limit at higher currents. The PMMA cylinder thicknesses, with a density of 1.19 g/cm³, are selected to represent objects passing through the scanner in a real industrial line.

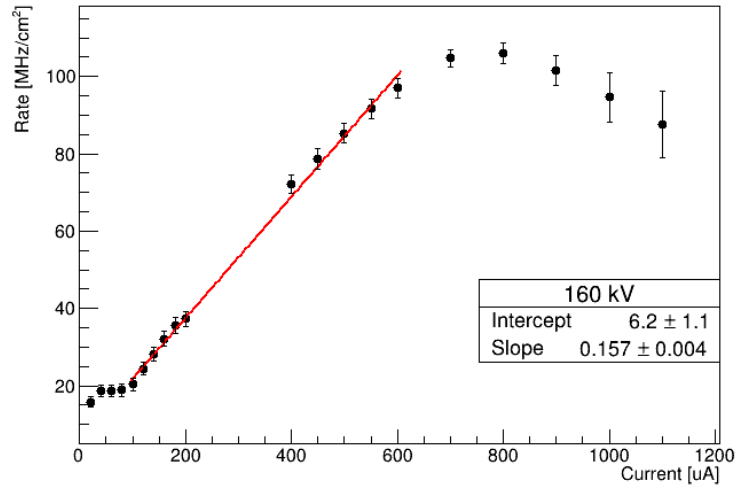
Figure 5.18 presents three plots of the cluster acquisition rates for various PMMA thicknesses. When the thickness is non-zero, two types of events are acquired: X-rays that pass through the material without losing kinetic energy and those that interact with the PMMA, deviating from their initial path via Compton scattering. Additional measurements are taken by inserting a block of lead material to cover the hole in the plastic support of the sensor, with the PMMA cylinder placed nearby. This setup allows for the observation of only those photons that interact with the PMMA electrons and are scattered into the sensitive area of the chip. These measurements are then subtracted from the direct measurements, with the non-scattering variance calculated as the difference between the total variance and the variance from the scattered events. All acquired data under each condition are normalized for acquisition time, and the clusterization algorithm is applied before evaluating the rate. The plot for data acquired at 120 kV differs from the others for two reasons: experimental data with the 40 mm PMMA cylinder are missing, and information about scattering events is not recorded. As a result, the comparison of these values with the others is excluded from the following analysis.



(a) 80 kV.

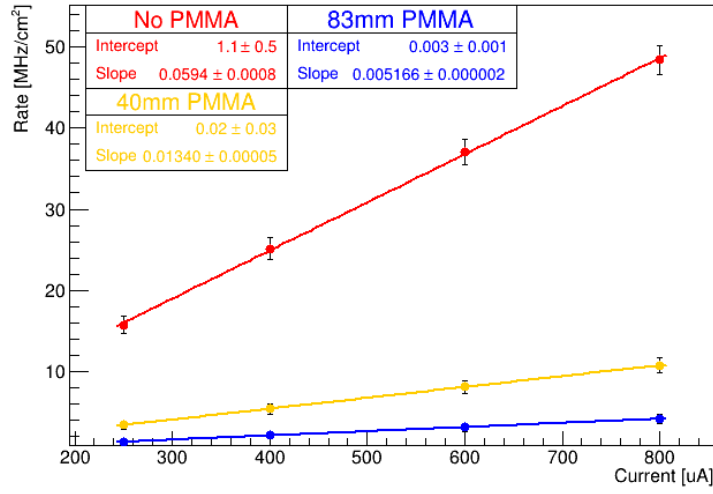


(b) 120 kV.

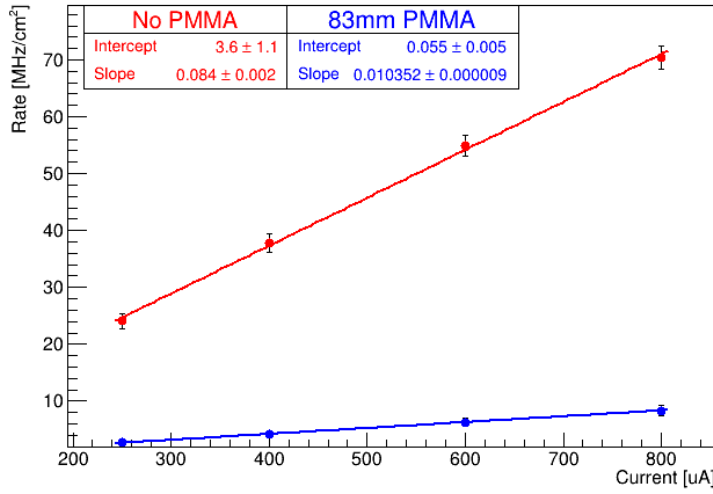


(c) 160 kV.

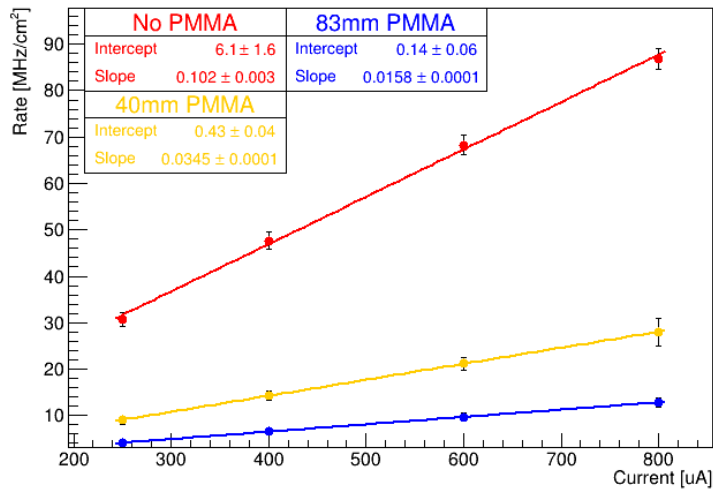
Figure 5.17: Cluster rate plots for three different voltages of Mito250. The signal threshold is set to the lower possible value in order to acquire all the signals, including noise. VCASN equal to 30 and area of 32×32 pixels.



(a) 80 kV.



(b) 120 kV.



(c) 160 kV.

Figure 5.18: Cluster rate plots for three different voltages of Mito250 and three PMMA thicknesses: 0 mm in red, 40 mm in yellow and 83 mm in blue. The signal threshold is set to the higher value in order to avoid electronic noise. VCASN equal to 1 and area of 32×32 pixels.

Under all conditions, the cluster rate increases linearly with current without reaching the upper limit observed in Figure 5.17 due to the pixel threshold setting. At 800 μA , the maximum rate detected with the highest voltage considered is below 90 MHz/cm^2 , whereas in previous plots with the VCASN parameter set to 30, the rate limit is reached. The error bars correspond to the square root of the variance difference between the total and scattered data. Unlike the case illustrated above, at VCASN set to 1, the linear interpolations show a gain that is proportional to the voltage.

The information obtained with the ARCADIA MD3 chip tested at 80 kV and 160 kV is particularly valuable for the company. In fact, the use of a digital D-MAPS detector could be supported by lowering the X-ray tube voltage, as 200 μm of silicon is nearly transparent to energies higher than 20 keV. Reducing the X-ray tube nominal voltage also helps lower the material cost for the external shielding structure, making the final product more competitive in the market.

The signal to noise ratio (SNR) as a function of material thickness is then evaluated for the two voltage values. The data are processed to calculate the SNR for an area of 0.8 mm $\times$ 0.8 mm over a time frame of 2.5 ms, with a distance from the X-ray tube corresponding to the setup used by MiCROTEC for their detection system, and a current that delivers 500 W of power. The mean number of clusters per pixel acquired by the 32 $\times$ 32 pixel matrix during the experimental tests at 80 kV and 160 kV for each time frame is calculated along with the standard deviation. To reduce fluctuations between consecutive frames, the mean for all frames is calculated, and the variance is determined through error propagation. The SNR is then computed as the ratio of the signal to the square root of the variance. The results are presented in Table 5.3.

PMMA thickness [mm]	Voltage [kV]	SNR	Err SNR
0	80	58.9	0.3
	160	58.2	0.9
40	80	24.2	0.2
	160	27.2	0.2
83	80	12.4	0.1
	160	15.4	0.1

Table 5.3: Signal to Noise Ratio calculated from data acquired at VCASN equal to 1 and voltages of 80 kV and 160 kV.

The values are consistent for a 0 mm PMMA thickness, but they tend to deviate as the PMMA cylinder is introduced, with the number increasing as a function of the selected voltage. The inconsistency in the numbers could stem from several factors. From a mechanical perspective, the central point of the Mito250 gantry is not easily accessible, so the positioning of the detection system might not be perfectly aligned with the tube exit point during each test repetition, and the various materials placed on top of the chip could shift slightly. From a physical standpoint, the efficiency of a 200 μm silicon sensor in detecting X-rays decreases for energies above approximately 20 keV. Even though the ARCADIA chip is fully depleted, which increases the signal acquisition probability, only 2 % of X-rays interacting with the detector at energies

greater than 80 keV are detected. Simulations indicate that, regardless of the material thickness, about 30 % of the total X-rays reaching the chip carry energy greater than 80 keV. However, the efficiency of the sensor material creates a difference in signal acquisition. The number of detected photons relative to those reaching the sensor with energy greater than 80 keV increases with object thickness for two primary reasons: on one hand, the object acts as a filter, blocking low-energy particles and reducing the total number of acquired particles; on the other hand, the probability of detecting a photon with energy greater than 80 keV remains constant. Consequently, the SNR associated with an X-ray tube of 160 kV is slightly higher.

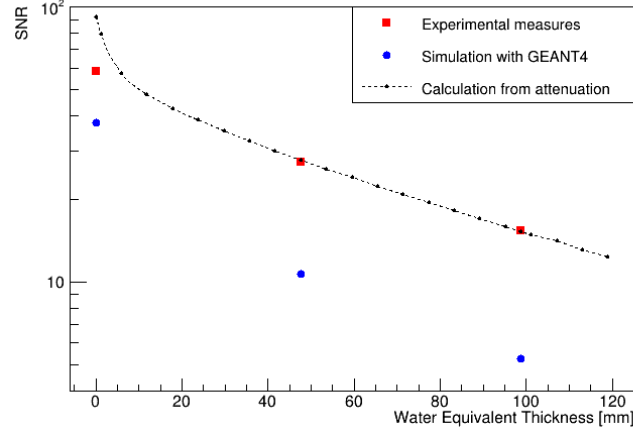
Figure 5.19 shows the SNR plots for both voltage values, alongside the values calculated from GEANT4 simulations and the attenuation data from NIST tables. The initial energy distribution for the literature and simulated values is derived from the web version of the software toolkit SpekPy [43], which models X-ray tube spectra emission. For the simulations, the tube potential is set to 160 kV and 80 kV, with the tungsten anode angle set to 12° . The physics model used is *kqp*, and attenuation data is sourced from *nist*. No filtration material is included, and both Bremsstrahlung and characteristic X-ray components are considered in the spectrum.

For the SNR evaluation under the specified time, distance, and current conditions, a normalization value is calculated based on the real energy spectrum acquired with an array of CdTe sensors, which have an area of $0.8\text{ mm} \times 0.8\text{ mm}$ and a thickness of 2 mm [42]. The tube is set to 160 kV and 80 kV at a distance of 610 mm with a current of 0.1 mA and an acquisition time of 10 ms. The energy distribution measured by the CdTe detector starts from 20 keV. This energy limit does not present an issue when scanning a PMMA object of 40 mm thickness, as almost all X-rays with energy lower than 20 keV are stopped by the material. However, the problem arises when no sample is scanned. In this case, the tungsten characteristic X-rays, which range from 8.3 keV to 11.3 keV, are not blocked by the object, and there is a 50 % to 70 % probability of these photons reaching the detection area. Consequently, only photons with energy higher than 20 keV are considered in the simulated spectra for the normalization of the signal.

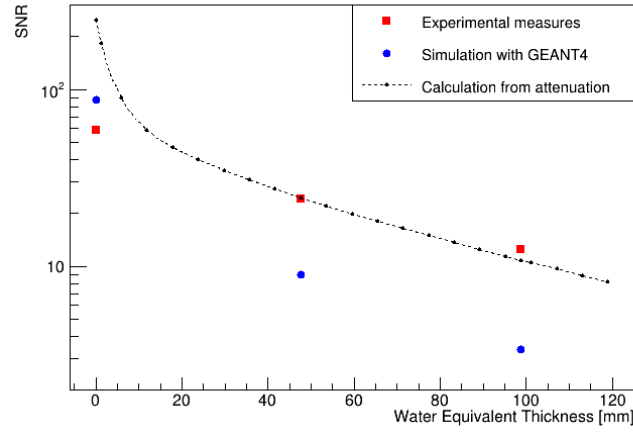
To calculate the normalization value for the analysis, the following steps are performed:

- Spectrum integration: the integral of the energy spectrum, corrected for the CdTe detector efficiency, is computed. This represents the total number of photons in the relevant energy range detected by the CdTe sensor.
- Scaling by area: the integral value is scaled by the area selected, ensuring the calculation is representative of the spatial resolution of the detector system.
- Incorporating acquisition time: the product is multiplied by the acquisition time set by the scanner working mode, accounting for the temporal duration of the measurement.
- Adjusting for current and power: the current required to deliver a power of 500 W is factored into the calculation to normalize the photon flux based on the tube operating conditions.
- Accounting for distance: the product is divided by the square of the distance from the X-ray source to the detector. This step corrects for the inverse-

square law governing photon flux as it propagates away from the source. This normalization value provides a standardized measure of the signal under the specified experimental and operating conditions, enabling comparison between measurements and simulations.



(a) 160 kV.



(b) 80 kV.

Figure 5.19: Signal to Noise Ratio (SNR) comparison between data acquired during the Noale tests with the ARCADIA MD3 chip, the GEANT4 simulations and the attenuation from NIST tables.

Starting with the plot in Figure 5.19a, where the tube voltage is set to 160 kV, the experimental data closely aligns with the attenuation values in PMMA material derived from the NIST table. In contrast, the GEANT4 simulations yield lower SNR values. This discrepancy arises from the sensor description in the simulations: the detector is modeled as a silicon layer without incorporating the depletion mechanism, and only photoelectric and Compton scattering interactions are accounted for, to maintain a reasonable simulation runtime. Conversely, the NIST tables offer reliable results for a silicon layer by also considering the radiative losses experienced during the X-ray passage through the material.

A similar trend is observed in Figure 5.19b, where the tube voltage is set to 80 kV. In this case as well, the SNR curve derived from the NIST tables closely matches the experimental data.

The significant discrepancy observed when no sample is placed between the source and the detector is attributed to the normalization issue described earlier. Generally, silicon is efficient at detecting particles with energies up to approximately 20 keV. However, the spectrum used as a reference for both simulations and theoretical values is effectively blind to energies below this threshold. As a result, data acquired without a sample cannot be directly compared with literature values or simulations, leading to noticeable differences in the behavior of the curves. Nevertheless, the experimental data show consistent alignment, following a single exponential function. This consistency enables the determination of SNR values for various sample thicknesses.

In conclusion, I tested the ARCADIA sensor prototype under the X-ray flux of the MiCROTEC apparatus. These experiments offered a direct evaluation of the signal response of the ARCADIA MD3 chip using the X-ray tube of the Mito250 scanner. The results demonstrated the sensor ability to acquire signals with a linear increase in the X-ray rate as a function of current. However, saturation occurred due to the inherent limitations of the data acquisition (DAQ) system. This behavior is linked to bottlenecks in the acquisition software and the overall DAQ system, which are not fully optimized for this application.

5.2.3 Comparison with the state-of-the-art

Tests conducted with the ARCADIA MD3 chip under an industrial X-ray source, both with and without objects, demonstrate its potential as a detection system for imaging applications. Performance could be further enhanced by optimizing the data acquisition system. Notably, this is not the first instance of a single-photon counting detector being used for commercial applications across various fields but still it is the first test assessing the potential use of MAPS. As referenced in Section 2.1.2, various technologies developed over the past 30 years have been employed for applications with X-ray sources. Examples of pixel sensors for X-ray imaging include hybrid detectors like *Medipix*, *Xpad*, *Pilatus* and *Pixie*. These detectors are widely used in various applications, such as medical imaging, synchrotron experiments and HEP, due to their high spatial resolution, energy discrimination capabilities, and photon-counting technology.

The most recent Medipix collaboration, established in 2005, developed the *Medipix 3* chip using 130 nm CMOS technology. This design integrates two counters within a single small pixel, allowing one image to be captured while the previous one is being read out. The chip features a pixel size of $55\text{ }\mu\text{m} \times 55\text{ }\mu\text{m}$ in a 256×256 matrix, connected to a $300\text{ }\mu\text{m}$ thick silicon layer. The largest commercially available Medipix-based detector comprises 6.5×10^6 pixels, arranged in a 10×10 array of single *Timepix* chip assemblies. Its roof-tile mechanics enable an almost seamless $14\text{ cm} \times 14\text{ cm}$ coverage. *Medipix 3* technology achieves a hit rate ranging from 2.8 GHz/cm^2 to 82.6 GHz/cm^2 , depending on the configuration, and provides color-coded images distinguishing up to three different photon energy levels [52]. X-ray CT imaging applications are being explored through a collaborative project involving Nikhef's neighboring institute, CWI for Mathematics and Informatics, the Nikhef spin-off company Amsterdam Scientific Instruments, and the Belgian company XRE [53]. Although the hit rate of *Medipix 3* is significantly higher than that of the D-MAPS chip presented in this thesis, its commercial use is limited by certain drawbacks. The primary limitation is its total power consumption: at 100 MHz, the chip consumes 440 mW for the analog circuitry

and 450 mW for the digital part. In contrast, D-MAPS has a total power consumption of less than 20 mW/cm² for a signal rate of 100 MHz/cm².

The *XPAD 3* (*X-ray Pixel chip with Adaptable Dynamics*) is designed for integration with a pixel sensor using bump bonding and flip-chip technologies. Fabricated in IBM 0.25 μm technology, the circuit contains 9600 pixels with a 130 μm pitch, arranged in 80 columns of 120 elements each. Its advanced features include a high counting rate exceeding 100 GHz/cm², a low noise detection level of 100e⁻ RMS, dual energy thresholds, and rapid image read-out in under 2 ms per frame [54]. While *XPAD 3* is a valuable technology for research environments, it is not suitable for the industrial CT scanner considered in this thesis. This limitation arises because, when bonded to a silicon semiconductor material, its energy range reaches only up to 35 keV, restricting its applicability in industrial settings. The maximum energy range can be extended by using an alternative substrate, such as CdTe. However, this increases material costs and reduces market competitiveness.

Similarly, the *Pilatus 4* detector is designed for synchrotron sources and utilizes a 450 μm thick silicon semiconductor, allowing signal acquisition within an energy range of 6 keV to 40 keV [55]. While it achieves a high count rate of 13.7 GHz/cm², its spatial resolution is lower compared to the ARCADIA MD3 chip, as its pixel pitch is six times larger.

The *PIXIE-III* chip, developed by Pixirad Imaging Counters s.r.l., is fabricated using 160 nm CMOS technology and features a 62 μm pixel pitch in a 512×402 matrix, covering an area of 32 mm×25 mm. Each pixel contains two 15-bit counters, driven by two independent discriminators with programmable thresholds. With a broad energy range of 2 keV to 100 keV and a high count rate of approximately 26 GHz/cm², the sensor is well-suited for industrial applications. Additionally, it supports Pixel Summing Mode (PSM), which fully compensates for charge-sharing effects and enables the acquisition of two-color images in a single exposure with high chromatic purity [56]. However, the system requires cooling to -30° during operation, and the CdTe crystal operates at a bias voltage of 400 V. These factors limit its feasibility for use in an in-line CT scanner apparatus.

The key distinction between these sensors and the ARCADIA MD3 chip lies primarily in its unique architecture and cost competitiveness. Hybrid pixel sensors generally have a higher production cost than MAPS due to their more complex fabrication process, which requires the creation of two separate components that must be subsequently connected. Additionally, the models described above offer specialized functionalities for X-ray imaging detection, further increasing the final product cost. Therefore, when tiling an area of approximately 20 cm×20 cm, the cost difference can reach two to three orders of magnitude.

Consequently, the ARCADIA MD3 chip emerges as the most suitable sensor for the MiCROTEC industrial CT scanner system, despite its limitations in count rate and data acquisition performance.

6 Results

This thesis explores the state-of-the-art ARCADIA MD3 pixel sensor chip with its integrated read-out electronics, assessing the potential for industrial applications. The study begins with an evaluation of the sensor response to X-ray particle interactions, complemented by GEANT4 simulations of a complete CT scanner structure. Subsequently, experimental tests are conducted to validate the sensor imaging capabilities and to provide a comparison with an existing industrial CT detection system.

In this section, the findings of previous analyses are summarized and their implications for industrial environments are discussed in detail.

6.1 X-rays interaction within a CT scanner

In the energy range commonly used for industrial applications, X-rays primarily interact through the photoelectric effect and Compton scattering. Considering the Compton scattering mechanism, an investigation into the Klein-Nishina angular distribution reveals that photons with energies of a few keV tend to scatter almost equally in both forward and backward directions. However, there is a null probability of detecting a scattered photon along the same direction as the incident particle because of interactions with electrons bound to the atomic nucleus. Therefore, as the incident photon energy increases, forward scattering becomes more dominant. Nevertheless, across all incident energy levels, the region with the lowest scattering occurs as a ring at 90° relative to the initial beam direction.

The study of Compton scattering process demonstrates that scattered X-rays are a constant feature in a complete CT scanner setup unless an anti-scatter grid is positioned in front of the detection area. Additionally, to minimize photon escape from the apparatus, the X-ray tube source must emit orthogonally to the gantry entry and exit regions.

Another important consideration regarding X-rays interacting via Compton scattering with a sample is their energy distribution. The photon cross-section plot for water, shown in Figure 2.9, indicates that for photon energies up to approximately 30 keV, interactions are dominated by the photoelectric effect. Beyond this energy, Compton scattering becomes the most likely interaction process until the pair production threshold is reached. Consequently, an X-ray tube operating at a few hundred kV is not ideal for minimizing scattering. To address this, a compromise must be made: the energy must be high enough to ensure complete penetration of the material thickness, allowing for reliable statistical signal acquisition at the detection point within the specified time window, but the maximum energy should not exceed approximately 100 keV, in order to reduce the scattering probability.

GEANT4 simulations of the Mito250 CT scanner are performed to evaluate the amount of X-rays that scatter within a sample and to subsequently detect them at the exit area of the gantry. The sample, a polyethylene (PE) block measuring $500\text{ mm} \times 150\text{ mm} \times 150\text{ mm}$, is positioned on the conveyor belt, with its center aligned beneath the X-ray emission point, in order to emulate a set of food products passing through the scanner. The area where scattered X-rays are detected corresponds to the major segment of a circle with a radius matching that of the gantry exit tube, placed 320 mm from the center of the PE block. The segment is implemented to capture only

events scattered within the region above the belt. The energy distribution of photons reaching the detection area is illustrated in Figure 6.1, while Figure 6.2 displays a position map of X-rays hitting the detector. The mean energy of the scattered X-rays is 55.4 keV, closely matching the initial mean energy. This similarity is attributed to the characteristic X-ray emission peak and the attenuation of the spectrum at both lower and higher energy values. The 2D map reveals that the top and lateral regions are the "hot spots" where most scattering occurs. Depending on the shape of the object being scanned, bulkheads could be introduced to block these particles.

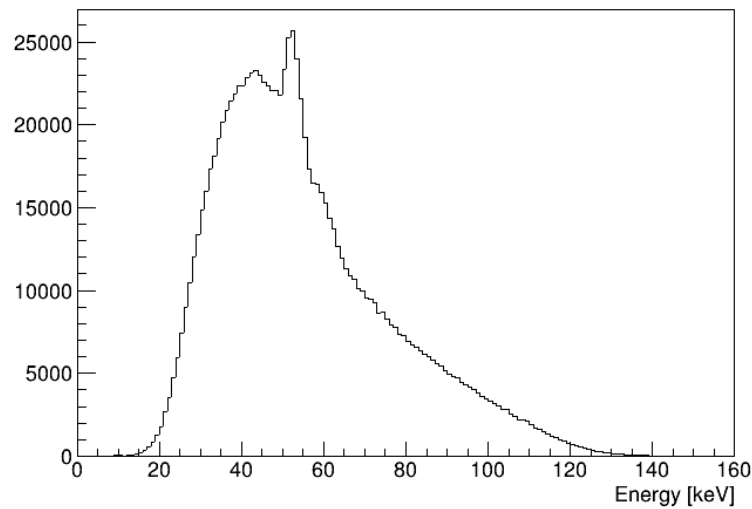


Figure 6.1: Energy spectrum from simulations of X-rays scattered inside the PE block which reach the exit area of the Mito250 gantry.

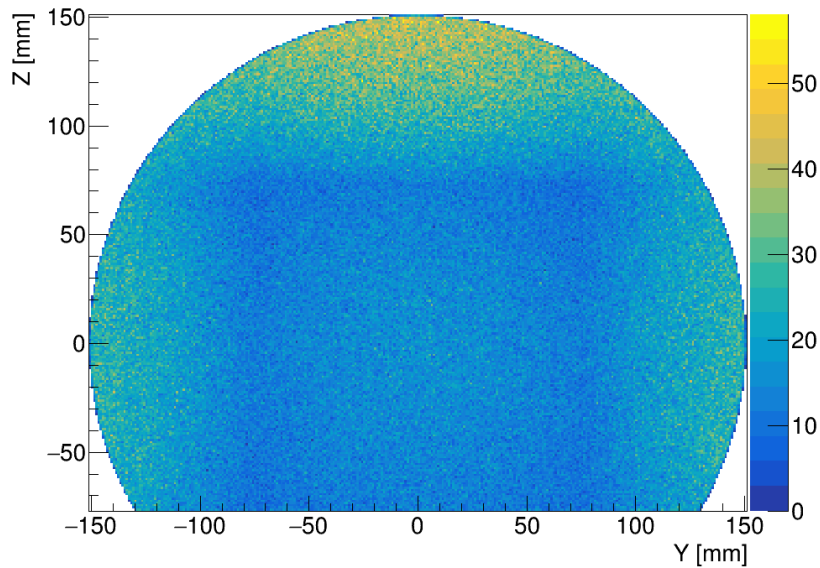


Figure 6.2: Map of scattered X-rays which exit from the Mito250 gantry.

Simulations conducted with the X-ray tube operating at a nominal voltage of 160 kV reveal that approximately 62 % of the X-rays emitted from the source interact with the

polyethylene (PE) block. Of these, 62 % are scattered in various directions, while only 0.8 % exit the gantry. For a high-density PE object with a thickness of 150 mm, less than 25 % of the initial X-rays emitted by the tube are attenuated by the materials within the gantry and reach the Mito250 detection system. Among these, approximately 20 % are scattered by the object. The remaining X-rays are either completely absorbed by the scanner materials or undergo Compton scattering, deviating from their original path.

The total number of X-ray events reaching the GOS detection plane, which has an area of 350 mm $\times$ 400 mm and a thickness of 200 μ m, is about 25 % of the emitted X-rays from the source. Among these, the detected events, comprising both X-rays transmitted through the sample and those interacting via Compton scattering, account for approximately 15 %, corresponding to a detection efficiency of about 60 % for the GOS material. However, around 22 % of the X-rays detected by the GOS plate are scattered from the PE object, contributing to the generation of artifacts in the reconstructed images.

As a comparison, Section 2.3.1 presents the results of GEANT4 simulations involving a sample placed in front of a silicon layer designed to emulate the MD3 ARCADIA chip. These simulations indicate that the number of X-rays scattered and detected within the energy range of 0 keV to 160 keV increases with sensor thickness. However, the proportion of scattered X-rays remains nearly unchanged despite the thickness variations of the sample scanned. For a 800 μ m sensor layer, less than 0.3 % of incident X-rays are detected as scattered events from a 100 mm thick water target.

Nonetheless, the 1.4 % of scattered X-rays are able to leave the gantry of the CT scanner, contributing to the external ambient dose. The energy spectrum plays a crucial role in calculating this dose and understanding the amount we have to expect. The external ambient dose rate, calculated as the sum of all X-rays reaching the major segment area, is 53 mSv/h. This value must be reduced by five orders of magnitude to meet the limit set by the ICRP. As a result, an effective external shielding system is mandatory.

6.2 Shielding system for industrial CT apparatus

Chapter 4 presents a detailed study of the materials and geometries for the optimal shielding structure for the new Mito500 CT scanner. These same considerations can be applied to the existing Mito250 scanner, with a focus on implementing and simplifying the current shielding system.

Lead and stainless steel are selected for shielding due to their advantageous properties. Lead is a high-density material with a large atomic number, making it highly effective at blocking X-rays. However, its toxicity to the human body presents a significant concern, meaning it cannot directly contact the food scanned by MiCROTEC machines. The optimal solution is to use lead as a coating material. Stainless steel, on the other hand, is favored for its density (greater than plastics), high X-ray interaction probability, and low backscattering of photons. Other materials in contact with food are typically plastics, which are nearly transparent to X-rays, as their transmission rate exceeds 95 % in the considered energy range. Interestingly, a coating thickness of just 1 mm is sufficient to yield similar amounts of backscattered particles from Mylar, HDPE, and stainless steel. However, increasing the thickness reduces the backscatter

from stainless steel compared to plastics. Finally, both lead and stainless steel are competitive materials in the market.

The geometry study on potential configurations reveals that changing the conveyor path direction and including diaphragms after each angular deviation reduces the detected ambient dose rate at the scanner exit region. Among the various configurations that meet the requirements, the company selects two options: a single angular deviation of 90° and two angular deviations of 60° each, within the same plane of the conveyor belt.

The combination of the selected geometry and materials provides the solution for the new scanner developed by the company. In fact, the ambient dose rate measured in the external area of the shielding structure is below the limit set by the ICRP. However, the main drawback is the high cost of the entire structure. While both lead and stainless steel are competitive in price, the coating process is a niche operation with a very high cost. As a result, the final CT scanner product becomes less accessible in the market.

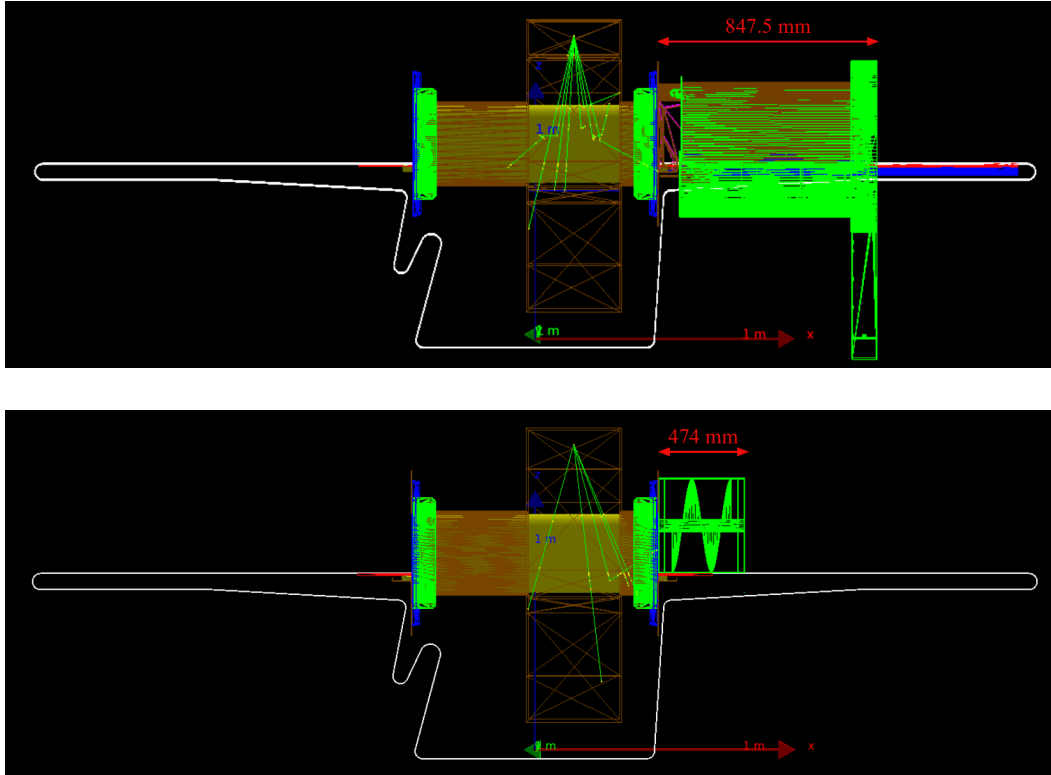


Figure 6.3: Mito250 CT scanner with a shielding system on the right side of the gantry. The two images, taken from GEANT4 simulations, display the length difference considering a new possible apparatus for blocking the exiting X-rays. The advantage is clearly visible, in fact, even in the worst configuration, with an helix of 25 cm pitch, the total scanner length is reduced of about 75 cm.

An alternative configuration could reduce both the material usage and the coating costs of the scanner: the screw conveyor. The ambient dose rate, calculated from GEANT4 simulations in Section 4.3, shows that a helical structure with 4 mm thick stainless steel for one and a half turns, enclosed by two layers of 2 mm lead coated with 1 mm stainless steel, does not achieve the same dose rate values as the previous

configuration. However, increasing the number of helical turns enhances the amount of X-rays blocked. Additionally, using stainless steel for the entire helix reduces manufacturing costs. The helical structure also develops height rather than length, reducing its spatial footprint. This is an important factor in industrial settings, as many factories already have limited space for existing equipment. The addition of large apparatuses would pose a significant challenge, making it a major barrier to purchasing the scanner.

Figure 6.3 shows the length difference between the current shielding structure and the alternative design featuring one and a half turns of a helix enclosed in a box of lead coated with stainless steel for the Mito250 CT scanner. The new shielding structure is preferred for the Mito250 due to its smaller size, while for the Mito500, the larger size requires a doubled helix radius, along with stable support and a more powerful rotational engine.

The helical structure shown in the figure uses a 25 cm pitch screw, suitable for applications such as polystyrene trays. By adding half a turn to the helix to enhance the blocking power, the total length of the scanner is reduced by approximately 35 cm. For industries producing products like yogurt or jam jars, a helical structure with a 15 cm pitch is recommended, resulting in a reduction of about 55 cm. The helical shielding system requires a box of lead coated with stainless steel to contain the structure and block the scattered X-rays. While this still introduces the drawback of higher manufacturing costs, the amount of coated material is reduced compared to the geometry with angular deviations.

6.3 ARCADIA sensor

Considering the X-ray tube selection criteria outlined in the Compton Scattering section, the features must align with the properties of the detection system. For example, the CdTe array pixel sensor *X-Card ME3* [42] is designed to detect X-rays in the energy range from 20 keV to 160 keV, meaning the tube potential must be at least 20 kV higher than the minimum energy required to fully penetrate the sample. Although the CdTe sensor is more efficient than a silicon detector in detecting X-rays, using this system increases the probability of observing scattering interactions. On the other hand, the ARCADIA MD3 chip is sensitive from about 3 keV, selecting the VCSN value that minimizes noise. As a result, the MD3 chip is preferred when using an X-ray tube with a nominal voltage lower than 100 kV.

Additionally, the energy spectrum is crucial when designing a CT scanner as a non-customized product, as it must meet the needs of a diverse range of clients. These clients may need to scan various products, from glass jars of about 10 cm in height to low-density polystyrene trays of about 3 cm in depth. For this reason, the ARCADIA MD3 detector is tested with different voltage values: 40 kV, 80 kV, 120 kV, and 160 kV. The X-ray tubes used in the experiments share the same anode material but are designed for two distinct applications: the 40 kV nominal voltage tube is intended for laboratory tests, operating at high power for short periods, while the 160 kV tube is designed for industrial use, operating at 500 W nearly 24 h a day, seven days a week.

6.3.1 Efficiency

The ARCADIA D-MAPS sensor does not offer high efficiency due to its use of silicon, a material that, while advantageous for its low energy requirement for electron-hole creation, moderate band gap, room temperature operation, wide availability, easy manufacturing, and low production cost, is not optimal for high efficiency X-ray detection. Specifically, the Monolithic sensor provided by the ARCADIA collaboration is designed with minimal alterations to standard fabrication processes, ensuring compatibility with commercial manufacturing techniques. For the X-ray energy range considered, simulations indicate that the efficiency of a single silicon layer of 200 μm thickness, accounting for both photoelectric and Compton scattering interactions, is 5.3%. Of this, 0.03% of the detected X-rays are recorded multiple times within the sensor due to scattering. This efficiency is influenced by the ability of the material to capture X-rays, and silicon relatively low atomic number and density limit its performance. In contrast, the MD3 chip improves efficiency through two main factors: it is a doped *p-n junction*, and it is fully depleted via a reverse bias applied from the back-side, increasing both the interaction probability and charge collection speed at the electrodes.

The interaction probability can be increased through two distinct approaches: increasing the sensor thickness or creating a stack of chips with an absorber layer placed between two consecutive detectors.

The primary advantage of the first solution is the reduction in the number of read-out systems that need to be controlled and synchronized, which also helps reduce material costs. Additionally, it allows for the detection of X-rays that scatter only within the sample. However, the main drawback is the need to increase the bias required to fully deplete the sensor. The second solution involves stacking chips, which introduces additional challenges from both a mechanical and software perspective. The sensors must be physically aligned and synchronized to ensure accurate detection timing. The advantage of this approach is energy selection: by inserting known materials of specific thicknesses between the sensor layers, it becomes possible to discriminate between different energy ranges, providing additional information from the detection system. The main disadvantage, however, is the increased probability of detecting scattered events in the last sensor, which can reduce the accuracy of reconstructing the particle trajectory from the source to the detection point, leading to increased image artifacts.

Therefore, the optimal solution appears to be producing a thicker sensor, which minimizes the artifacts associated with particle scattering. A 600 μm thick ARCADIA D-MAPS chip, equivalent to three layers of the current sensor, is currently in development and will undergo testing in 2025.

The efficiency of the sensor is influenced not only by the particle interaction probability but also by the data acquisition process. Figure 5.13 highlights the primary issue with the ARCADIA demonstrator used in the experiments. When the DAQ system cannot handle the acquisition rate, information from the lower portion of the sensor matrix is lost, and the acquisition rate saturates at a value determined by the FPGA. The maximum acquisition rate specified by the architecture design is 100 MHz/cm². However, the tested 512×512 pixel matrix achieved a rate below 18 MHz/cm². This limitation stems from the acquisition software bottleneck and the lack of the DAQ system optimization for this task. To address these issues, an agreement has been reached with the company to resolve the identified shortcomings and develop a custom sensor

tailored for their scanners.

The new detector should be specifically designed for the company requirements, incorporating adjustable features to support high-rate signal acquisition. The Mito250 scanner currently operates with an X-ray tube rated at 160 kV and 3.12 mA, generating a particle rate of approximately 6.56 GHz/cm² at the detector location. Based on simulations with a 200 μm silicon layer, only 5.3% of the particles interact with the detector, reducing the interaction rate to 323 MHz/cm², which is three times higher than the sensor design limit. By adjusting the tube settings, such as reducing the voltage to 80 kV and increasing the current to 6.25 mA, the X-ray emission rate at the same distance drops to 3.05 GHz/cm², corresponding to an interaction rate of 152 MHz/cm². While this value still exceeds the design limit, it represents a more feasible target with minor modifications to the sensor design.

The current detection system of the company employs pixels sized at 800 μm $\times$ 800 μm , making the 25 μm pixel pitch of the ARCADIA MD3 chip a significant improvement in spatial resolution. However, the increased data volume poses challenges for the acquisition rate. One potential solution is to integrate in-pixel memory while enlarging the pixel pitch. A memory structure, such as a counter, could register the events detected within a predefined time range before transmitting the signal to the periphery. This approach allows events to be accumulated rather than discarded. Associating the summed events with a single *timestamp* is not problematic for the company tasks, provided it stays within half the integration time.

In the MD3 chip, approximately 420 μm^2 of each pixel area is occupied by the diode. Therefore, additional electronics must occupy less than 200 μm^2 to maintain the current pixel size; otherwise, the pixel size will need to increase. A single counter requires around 20 μm^2 , making it feasible to include two or three bits without altering the pixel pitch. For an X-ray tube with a power of 500 W, two counters suffice at a voltage of 80 kV, while three counters are necessary at 160 kV.

Another potential implementation for the needs of the company involves utilizing the Time-over-Threshold (ToT) method. ToT is a time-based signal processing technique used to estimate particle energy. Unlike conventional methods, which convert detector signals from incident radiation into voltage pulses and measure their pulse heights to determine energy, ToT estimates energy by measuring the pulse width of the digital output. In the ToT approach, a comparator compares the voltage pulses to a threshold voltage and generates corresponding digital pulses. A digital circuit, such as an FPGA, can be connected directly to the ToT output and measures the pulse widths. The ToT method eliminates the need for an ADC circuit, making it simpler to implement. However, ToT exhibits a non-linearity between the incident radiation energy and the output pulse width. To address this drawback, two specialized integrated circuits could be considered:

- Dynamic ToT (dToT): in this method, the threshold voltage is dynamically adjusted to track the input signal following a pulse detection, rather than remaining constant. This adjustment improves the linearity between the incident energy and the output pulse width, enhancing the accuracy of the energy estimation.
- Slew Rate Limited ToT: this approach incorporates a slew rate-limited CR-RC shaping into the signal processing chain, ensuring consistent slopes for

the rise and fall of the waveform. As the signal charge increases, the waveform broadens, effectively enhancing linearity in the relationship between the incident energy and the output pulse width.

Both methods offer viable solutions for enhancing the performance of the ToT system while addressing its inherent limitations. An example of a low-power, low-noise synchronous front-end chain with a design that features a shaper-less Charge-Sensitive Amplifier (CSA) with constant current feedback, enabling triangular pulse shaping to facilitate linear Time-over-Threshold (ToT) charge measurement, is implemented in a commercial 65 nm CMOS technology, designed for future pixel upgrades at the CERN LHC [57].

6.3.2 Signal to Noise Ratio

An experimental campaign conducted with the CT scanner from MiCROTEC demonstrates the capability of the ARCADIA MD3 chip to acquire signals from an X-ray tube operating at 160 kV and currents up to approximately 1 mA. A comparison is presented between the detection system currently employed in the Mito250 scanner and the data acquired during tests with the D-MAPS chip.

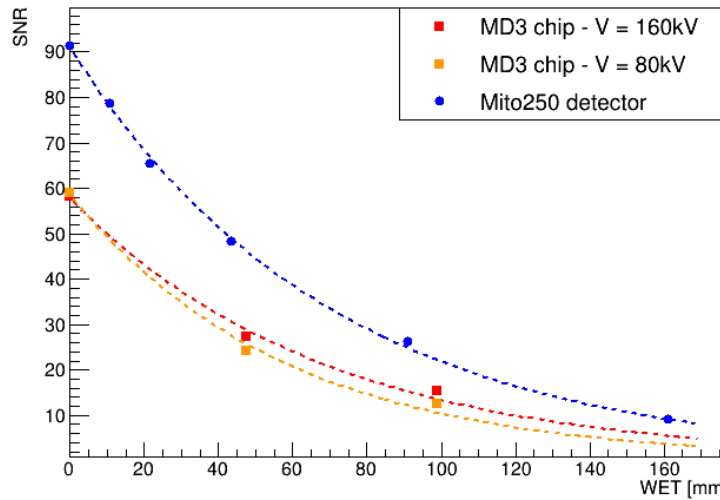


Figure 6.4: Comparison of Signal to Noise Ratio from industrial experiments with Mito250 CT scanner considering the ARCADIA MD3 chip and the detection system of the apparatus. The points are interpolated by an exponential curve, showing no intersection regions.

Figure 6.4 presents data acquired using the same CT scanner with a different detection system, while also varying the voltage of the X-ray tube. The ARCADIA MD3 chip, with a thickness of 200 μm , is placed on the conveyor belt of the Mito250 structure. As a result, a renormalization of the distance is required for a direct comparison with the detection system of the apparatus. Additionally, the values are normalized with respect to the voltage and current used during the tests, resulting in the three curves displayed in the plot. The red and orange markers represent the SNR values measured with the MD3 chip, while the blue markers denote the SNR values obtained from the Mito250 detection system. Error bars, accounting for error propagation, are

calculated but are not visible in the plot. All data points are interpolated with an exponential curve, which provides a good fit.

The low efficiency of the D-MAPS sensor impacts the performance of the new detector compared to the existing one. Notably, there is no thickness for which the SNR of the MD3 chip surpasses that of the current system. Thus, while the sensor demonstrates the ability to acquire signals from an X-ray source at rates comparable to industrial environments, these developments are ultimately ineffective, as the detection system already used in the Mito250 CT scanner proves to be superior. Nevertheless, the detection system inside the Mito250 scanner cannot be shielded from the X-ray Compton scattering generated by the PMMA samples. This scattering effect generally increases the acquired signal at all points except at 0 mm. As a result, the SNR difference decreases, eventually reaching similar values for both detection systems when the water equivalent thickness is approximately 150 mm. However, since this thickness does not correspond to real products scanned by the Mito250 apparatus, the ARCADIA MD3 chip cannot be effectively used in its current form.

To achieve the Signal to Noise Ratio performance of the current detection system, the depleted volume of the D-MAPS sensors must be increased. The results of simulations with thicker silicon layers are presented in Table 6.1. Considering that SNR values tend to approach the theoretical limit provided by NIST tables, a sensor thickness of 800 μm would be necessary to match the SNR values of the current detection system of the Mito250 scanner.

PMMA thickness [mm]	X-ray tube voltage [kV]	Silicon thickness					
		200 μm		400 μm		800 μm	
		GEANT4	NIST	GEANT4	NIST	GEANT4	NIST
0	80	87.2	248.0	98.7	269.1	108.0	284.3
	160	37.7	92.3	47.6	111.5	59.2	134.1
40	80	8.9	24.4	12.5	33.6	17.8	45.6
	160	10.7	27.7	15.1	38.8	20.7	53.7
83	80	3.4	10.8	3.2	15.1	6.5	20.8
	160	5.2	15.3	7.0	21.4	10.0	29.9

Table 6.1: Signal to Noise Ratio evaluated from the GEANT4 simulations and NIST attenuation tables at three possible ARCADIA chip thicknesses. The X-ray tube considered has a nominal voltage of 80 kV and 160 kV. The SNR numbers represent the lower and higher edges of the real SNR, as described in Chapter 5.

As discussed in the previous paragraph, D-MAPS chips with thicknesses exceeding 200 μm are currently under development. Notably, the ARCADIA project has designed and fabricated a 600 μm thick fully depleted sensor, which is scheduled for testing in 2025. Indeed, increasing sensor thickness is a key focus for the collaboration, particularly for medical and space applications. In this context, there is a high probability of achieving a single layer with a thickness of 800 μm , along with improved efficiency resulting from the use of a larger silicon sensor. In fact, continued advancement in technology, design, and manufacturing of DMAPS chips has the potential to enhance sensor performance and production viability, thus expanding the capabilities of detection systems. Although the chip used for the *proof of principle* tests discussed in

this work is relatively unsuitable for the CT scanner apparatus, due to its low SNR, future iterations of this technology will solve this limitation and meet the performance requirements for industrial applications.

6.3.3 Computed Tomography

The ARCADIA MD3 chip, tested inside the MiCROTEC CT scanner, is used to perform a complete CT scan of an iron screw measuring $5.8\text{ mm} \times 8.0\text{ mm}$. The scan utilizes the X-ray tube from the Physics Department. The reconstruction, processed using *Fiji* software, produces a visible 3D volume image, shown in Figure 5.12. Each projection is acquired over approximately 10 s, with the screw that rotates incrementally around its axis. For reconstruction, the analysis considers the counts per pixel corresponding to 1 s of integration time. The resulting 3D volume effectively delineates the threads along the screw body, although the horizontal indentation of the slotted screw drive is less clearly visible.

The feasibility of using the chip in the company CT scanner depends on the quantity of X-rays captured within the selected integration time. A key difference between the Mito250 industrial scanner and the academic laboratory X-ray emission system lies in their power outputs. The Mito250 operates at 500 W, with a voltage of 160 kV and a current of 3.125 mA, whereas the laboratory system operates at 80 W, with 40 kV and 2 mA. The number of X-rays emitted by a tube is proportional to the current, integration time, and the solid angle, while it scales with the square of the voltage. Thus, moving from 160 kV to 40 kV introduces a factor of 16. For example, with the industrial scanner parameters (160 kV, 3.125 mA), the number of collected counts per pixel (for an area of $25\text{ }\mu\text{m} \times 25\text{ }\mu\text{m}$) is 8.69, assuming an acquisition time of 2.5 ms and a VCASN value of 1.

Conversely, under the laboratory system settings (40 kV, 2 mA) with no object between the source and detector, the number of X-rays per pixel in the same acquisition time (2.5 ms) is only 0.15. This implies that, on average, each pixel records one event every 17 ms, compared to 300 μs with the company source. The significant difference is due to the lower current and the 2 mm aluminum filter used in the laboratory setup, which blocks photons with energies below 10 keV.

These calculations are derived from experimental data, ensuring that the chip parameters remained consistent: the VCASN value is set to 1, and a 32×32 sub-matrix of the chip is used for both acquisition. To achieve the same event count per frame as in the industrial scanner, the required integration time for the laboratory system is determined to be 8.81 ms. Using this integration time, another CT scan of the screw is conducted to evaluate the MD3 chip capability to capture the necessary number of particles for image reconstruction. Although the number of X-rays collected per pixel is limited to a few tens of photons, it is sufficient for a complete 3D reconstruction of the iron screw. The resulting image is shown in Figure 6.5.

The threads of the screw are clearly visible in this case as well; however, a general blur is noticeable compared to the image reconstructed with an integration time of 1 s. Specifically, the indentation of the slot screw drive appears only as a slight recess along the edges of the head. Despite this, the result demonstrates the feasibility of using the ARCADIA D-MAPS chip for a complete CT scan with settings comparable to those of industrial food scanners like the Mito250.

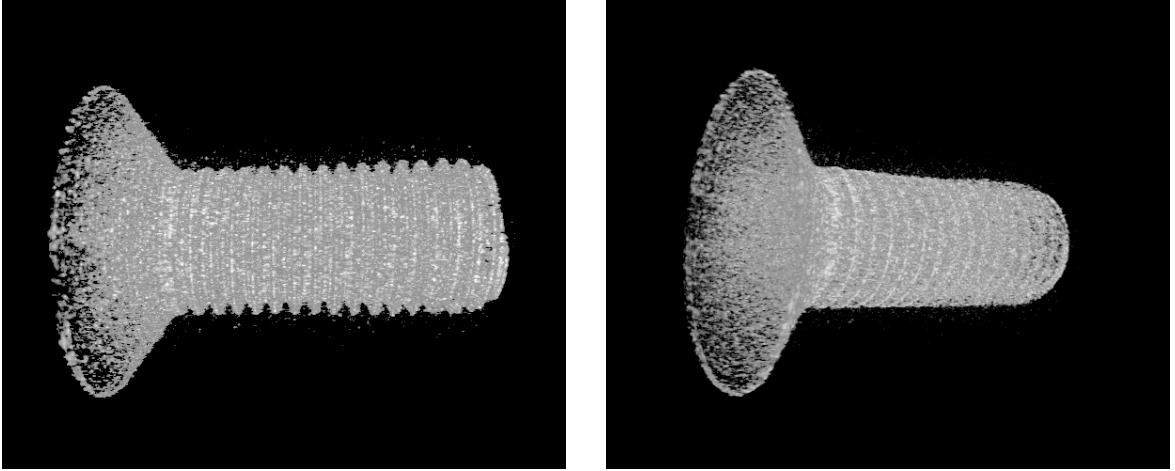


Figure 6.5: 3D image reconstruction of an iron screw considering a single projection acquisition time which provides a comparable number of X-rays hitting the detector as in the case of Mito250 scanner.

Finally, a brief discussion regards the amount of X-rays that deposit their energy within the $200\text{ }\mu\text{m}$ silicon sensor is warranted. An initial analysis of the ambient dose rate in the detection area is performed to compare it with the performance capabilities of the new D-MAPS sensor. GEANT4 simulations, employing a silicon layer of $200\text{ }\mu\text{m}$, estimate that the dose deposited in this region is significantly lower than the threshold required to reduce the detection efficiency of the new sensor. Specifically, the simulations show that less than 1.5 krad/h is released under the maximum current settings used by the company, while experiments with similar detectors have established that a dose of approximately 1 Mrad/h is necessary to impact sensor efficiency.

7 Conclusion and future trends

The FLECT project represents a transformative effort aimed at addressing the limitations of current industrial CT scanning technologies. By leveraging advanced Depleted Monolithic Active Pixel Sensors (D-MAPS) originally developed for High-Energy Physics, this initiative promises to redefine the standards of industrial CT systems, making them faster, more cost-effective, and safer.

To conclude, I will now address the multifaceted outcomes of the project, the challenges, and the broader implications for industries reliant on precise, non-invasive imaging solutions.

7.1 Enhancing CT scanning with advanced detection technology

Industrial CT scanning has traditionally relied on indirect conversion detectors that incorporate scintillators and photodiodes. While effective at their time, these systems have several limitations, including slow signal acquisition, energy inefficiency, and susceptibility to photon pile-up. The FLECT project directly addresses these shortcomings by integrating D-MAPS technology, which offers a monolithic design, direct X-ray conversion, and fully depleted detection volumes. These advancements allow for:

- **Improved imaging resolution and speed:** The direct conversion mechanism of D-MAPS eliminates intermediate steps, such as the need for scintillation and subsequent conversion to electrical signals. This results in significantly reduced signal acquisition times, enabling faster imaging without compromising resolution.
- **Enhanced material differentiation:** Simulation and experimental results confirm the ability of D-MAPS to distinguish between materials with similar absorption coefficients, such as glass and water. This capability is critical in applications like food safety, where detecting contaminants or foreign objects is essential.
- **Cost and energy efficiency:** D-MAPS technology reduces dependency on high-energy X-rays, lowering operational costs and energy consumption. The monolithic integration of readout electronics further simplifies system design, making it more economically viable for widespread industrial adoption.

7.2 Validation through simulation and experiments

A cornerstone of the FLECT project has been the rigorous validation of D-MAPS sensors through simulations and experimental campaigns. Monte Carlo simulations using the Geant4 toolkit are instrumental in modeling X-ray interactions with silicon detectors, assessing their performance under realistic industrial conditions. These simulations provide valuable insights into optimal detector configurations, such as pixel

pitches and silicon thicknesses, to balance resolution and efficiency. Key experimental findings include:

- **Radiographic imaging:** Initial tests using a 40 kV X-ray tube demonstrate the ability of the sensor to capture high-resolution images of small objects, such as metal screws and glass bulbs, with clear material differentiation.
- **CT reconstruction:** Rotating samples under the X-ray beam enabled the creation of accurate 3D reconstructions. For example, the successful reproduction of screw threads with a 0.5 mm pitch validated the sensor potential for high-precision industrial CT applications.
- **High-Energy testing:** Testing with MiCROTEC industrial CT systems operating at 160 kV confirmed the sensor ability to acquire low-energy X-ray signals. While challenges remain in absorption efficiency at higher energies, proposed solutions such as increasing sensor thickness and exploring stacked detector configurations show promise.

Therefore, the FLECT project represents an innovative effort to revolutionize industrial CT scanning. By integrating advanced D-MAPS technology, the initiative addresses long-standing challenges in detection efficiency, image quality, and cost effectiveness. Through rigorous simulation, experimentation, and collaboration with industry leaders, the project has demonstrated the transformative potential of this technology.

7.3 Addressing challenges in absorption efficiency and radiation safety

One of the primary challenges in adapting D-MAPS for industrial CT scanning is the lower absorption efficiency at high X-ray energies. Silicon, the primary material used in D-MAPS, exhibits reduced absorption for X-rays above 20 keV compared to alternative materials like CdTe. The project explores multiple approaches to address this limitation:

- **Increasing sensor thickness:** Simulations suggest that extending the silicon substrate thickness to 600 μm or 800 μm would increase the number of absorbed X-rays, enhancing detection rates. However, this approach requires higher bias voltages and more robust system designs.
- **Stacking sensors:** Layering multiple D-MAP sensors can improve overall absorption efficiency even if the detection probability of a scattering photon increases. Nevertheless, by including absorber material layers of varying thicknesses between consecutive chips, it is possible to recover energy information.

Radiation safety is another critical consideration. Industrial CT systems, especially those used in food processing, operate with continuous X-ray emission, leading to scattered radiation. Simulations using CAD geometries of MiCROTEC systems guide

the design of effective shielding solutions. A combination of FCMs and lead is identified as optimal for reducing ambient radiation, ensuring compliance with safety standards without compromising system performance.

7.4 Collaborative efforts and industrial relevance

The collaboration between the Ph.D. project, MiCROTEC, and the INFN ARCA-DIA initiative underscores the practical focus of this effort. MiCROTEC expertise in non-invasive scanning for food and wood industries provides a real-world testing ground for the D-MAPS technology. Key outcomes of this collaboration include:

- **Adaptation to industrial systems:** The integration of D-MAPS sensors into existing MiCROTEC CT systems demonstrate their compatibility and potential for future industrial applications. Adjustments to system parameters, such as tube voltage and sensor configuration, optimize performance for specific use cases.
- **Economic feasibility:** By reducing energy requirements and simplifying system architectures, D-MAPS technology addresses the cost barriers that have precluded the adoption of advanced detection systems in cost-sensitive industries.
- **Safety enhancements:** The project radiation safety studies not only ensure compliance with regulatory standards, but also pave the way for a new shielding system configuration in industrial settings, protecting both operators and end-users, and reducing the entire CT apparatus size.

7.5 Broader implications and future directions

While the immediate focus of the FLECT project is on food and wood processing, the advancements achieved have far-reaching implications across various sectors:

- **Medical Imaging:** The speed and precision of D-MAPS technology could enhance diagnostic CT systems, enabling faster scans with reduced radiation doses.
- **Airport Security:** High-resolution imaging capabilities can improve baggage scanners, making them more efficient and safer for both passengers and operators.
- **Cultural Heritage preservation:** Non-invasive analysis of artifacts and historical objects could benefit from the material differentiation and high-resolution imaging offered by D-MAPS sensors.
- **Research Applications:** The versatility of D-MAPS technology, as demonstrated in cross-disciplinary experiments like those at Fermilab, highlights its potential for scientific research in fields ranging from particle physics to space exploration.

Looking ahead, the FLECT project will focus on several key areas to further advance the technology:

- **Refinement of sensor prototypes:** Continued development of D-MAPS sensors, including optimization of pixel designs and material properties, will enhance their performance and adaptability.
- **Optimization of data acquisition systems:** Innovative DAQ systems that complement the high-speed capabilities of D-MAPS will be essential for maximizing their potential in industrial applications.
- **Exploration of stacked configurations:** The feasibility of layered detector designs, including the integration of absorber materials, will be explored to improve absorption efficiency and broaden the applicability of D-MAPS.

Therefore, as the initiative progresses, the FLECT project promises to deliver safer, faster, and more reliable imaging solutions for a wide range of industries. The advancements achieved not only enhance current industrial applications but also pave the way for innovative uses in medical imaging, security, cultural heritage, and scientific research. In doing so, the FLECT project sets a new benchmark for the future of non-invasive inspection systems, reaffirming the importance of cross-disciplinary innovation in addressing global challenges.

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