Software Solutions for Nuclear Imaging Systems in Cardiology, Small Animal Research and Education

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Abstract

The sensitivity for observing physiological processes makes nuclear imaging an important tool in medical diagnostics. Different types of nuclear imaging modalities, with emphasis on the software components and image reconstructions, are presented in this thesis: the Cardiotom for myocardial heart studies at the Karolinska University Hospital, the small animal Positron Emission Tomograph (PET) scanners for research and the SPECT, PET, spiral CT and Cardiotom demonstrators for the Royal Institute of Technology medical imaging laboratory.

A modular and unified software platform has been developed for data representation, acquisition, visualization, reconstruction and presentation of the programs of the imaging devices mentioned above. The high performance 3D ML-EM and OS-EM iterative image reconstruction methods are implemented both on Cardiotom and miniPET scanners.

As a result, the in-slice resolution of the first two prototypes of the Cardiotom today is the same as the formerly used filtered back-projection, however the in-depth resolution is considerably increased. Another improvement due to the new software is the shorter time that is required for data acquisition and image reconstruction. The new electronics with the newly developed software ensure images for medical diagnosis within 10 minutes from the start of the examination. The first system from the standardized production of the Cardiotom cameras is in the test phase.

The performance parameters (sensitivity, spatial and energy resolution, coincidence time resolution) of the full ring miniPET camera are comparable to other small animal PET systems.
List of Publications

This thesis is based on the first nine papers in the list. The author’s name is indicated with underlining in each of the publications.


IX. LSO based dual slice helical CT and PET demonstrators, I. Valastyán, A. Kerek, J. Imrek, G. Hegyesi, G. Kalinka, J. Molnár, D. Novák, Accepted for publication in Nucl. Instr. and Meth. A
Other articles and conference contributions to which the author has contributed, but which are not included in this thesis:


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<td>ADC</td>
<td>Analog to Digital Converter</td>
</tr>
<tr>
<td>APD</td>
<td>Avalanche Photo Diode</td>
</tr>
<tr>
<td>CT</td>
<td>Computed Tomograph</td>
</tr>
<tr>
<td>DICOM</td>
<td>Digital Imaging and COmmunications in Medicine</td>
</tr>
<tr>
<td>DSP</td>
<td>Digital Signal Processing</td>
</tr>
<tr>
<td>FBP</td>
<td>Filtered Back-projection</td>
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<tr>
<td>FDG</td>
<td>FluoroDeoxyGlucose</td>
</tr>
<tr>
<td>FFT</td>
<td>Fast Fourier Transformation</td>
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<tr>
<td>FOV</td>
<td>Field Of View</td>
</tr>
<tr>
<td>FPGA</td>
<td>Field Programmable Gate Array</td>
</tr>
<tr>
<td>GUI</td>
<td>Graphical User Interface</td>
</tr>
<tr>
<td>HLA</td>
<td>Horizontal Long Axis</td>
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<tr>
<td>IFFT</td>
<td>Inverse Fast Fourier Transformation</td>
</tr>
<tr>
<td>IP</td>
<td>Internet Protocol</td>
</tr>
<tr>
<td>ISA</td>
<td>Industry Standard Architecture</td>
</tr>
<tr>
<td>KN</td>
<td>Klein-Nishina formula for photon electron scattering</td>
</tr>
<tr>
<td>LEHR</td>
<td>Low Energy High Resolution</td>
</tr>
<tr>
<td>LOR</td>
<td>Line Of Response</td>
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<tr>
<td>LSO</td>
<td>Lutetium orthosilicate</td>
</tr>
<tr>
<td>LVDS</td>
<td>Low-Voltage Differential Signaling</td>
</tr>
<tr>
<td>LYSO</td>
<td>Lutetium Yttrium orthosilicate</td>
</tr>
<tr>
<td>MCA</td>
<td>Multi Channel Analyzer</td>
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<tr>
<td>ML-EM</td>
<td>Maximum Likelihood Expectation Maximization</td>
</tr>
<tr>
<td>OS-EM</td>
<td>Ordered Subset Expectation Maximization</td>
</tr>
<tr>
<td>PACS</td>
<td>Picture Archiving and Communication System</td>
</tr>
<tr>
<td>PCI</td>
<td>Peripheral Component Interconnect</td>
</tr>
<tr>
<td>PEM</td>
<td>Positron Emission Mammography</td>
</tr>
<tr>
<td>PET</td>
<td>Positron Emission Tomograph</td>
</tr>
<tr>
<td>PMT</td>
<td>Photomultiplier Tube</td>
</tr>
<tr>
<td>PSF</td>
<td>Point Spread Function</td>
</tr>
<tr>
<td>PSPMT</td>
<td>Position Sensitive Photomultiplier Tube</td>
</tr>
<tr>
<td>ROI</td>
<td>Region Of Interest</td>
</tr>
<tr>
<td>SiPMT</td>
<td>Silicon Photo Multiplier Tube</td>
</tr>
<tr>
<td>SPECT</td>
<td>Single Photon Emission Computed Tomograph</td>
</tr>
<tr>
<td>SSRB</td>
<td>Single Slice ReBinning</td>
</tr>
<tr>
<td>TCP</td>
<td>Transmission Control Protocol</td>
</tr>
<tr>
<td>TOF-PET</td>
<td>Time Of Flight PET</td>
</tr>
<tr>
<td>UDP</td>
<td>User Datagram Protocol</td>
</tr>
<tr>
<td>VHDL</td>
<td>Very high speed integrated circuit Hardware Description Language</td>
</tr>
<tr>
<td>VLA</td>
<td>Vertical Long Axis</td>
</tr>
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</table>
1. Introduction

Medical imaging is one of the fundamental clinical methods for early diagnosis and assessment of disease. The wide range of modalities enable examinations of different organs in a non-invasive way. Early detection ensures a better prognosis with more therapeutic options and thus results in greater rates of cure or survival.

Nuclear imaging is a method of producing images by detecting radiation from radionuclide-labeled agents that are administrated to the body, mostly intravenously. These radiopharmaceuticals are designed to make their way to a specific organ in order to reveal its physiological function. The imaging techniques are based on detection of photons and the establishment of the coordinates of the point of gamma ray emission. It is therefore necessary to also determine the energy of each photon detected and to be able to distinguish between primary and Compton scattered events.

Different modalities such as gamma camera, SPECT and PET are available in nuclear medicine. These devices are designed for planar 2D (gamma camera) and tomographical 3D (SPECT, PET) imaging of organs.

Nowadays, studying the physiological processes in small animals plays an important role in research. The effect of newly developed drugs can be followed in the bodies of the small animals with the help of nuclear imaging. This type of non-invasive technique for small animals requires not only smaller cameras, but accordingly higher spatial resolution than that currently available in human scanners.

The design of new imaging equipment, or improvements to existing components, is a challenge and is the aim of this thesis. Hardware solutions, based on digital signal processing technique and high performance software are used in modern cameras. By using these solutions, improved performance of sensitivity, spatial and energy resolution as well as a shorter scan time for examination should be achievable.

Multi-core processor computers and computer clusters allow parallel image processing algorithms and software. Parallel processing also increases the calculation speed.

In Chapter 2, the general description of nuclear medical systems, gamma camera, SPECT and PET are presented. Ectomography, a form of limited view angle tomography that has been implemented a mobile gamma camera system, is also described, since the method is not widely known in nuclear medicine.

In Chapter 3 and 4, the data acquisition hardware of the systems used in this thesis is presented. Sections 4.2 and 4.3 describe the technique for identifying detector crystals and the event handling.

Chapter 5 summarizes the image reconstruction algorithms implemented for systems in this thesis and, in Chapter 6, the software platform for developed nuclear medical devices is described.

Chapters 7, 8 and 9 contain the description of the imaging devices designed for clinical, research and educational purposes, respectively.

2. Nuclear medical imaging systems

2.1. Gamma camera

The first functional two dimensional imaging system for single photon emission, the gamma camera, was introduced by Hal. O. Anger in 1958 [1][2]. The development of the
gamma camera made it possible to show, in real-time, blood-flow in a patient, to follow kidney function, and to examine the liver as it generates bile, to give a few examples. Together with the possibility to produce $^{99m}$Tc isotope, the use of the gamma camera started a new era in nuclear medicine.

Most gamma cameras are based on a NaI scintillation crystal coupled to photomultiplier tubes (PMT) and an analog resistor grid connected to the readout electronics. The schematic view of a traditional gamma camera is shown in *Figure 1*. Modern cameras use digital readout of the signals from the PMTs. Recent developments in new gamma ray detection techniques, such as pixelated scintillators and solid-state light detection, or fully solid state detector systems with superior energy resolution, are also on the market. However, regardless of how photons are detected, or at which point of the circuitry the digitalization of the signals is made, the gamma camera technique is still dominated by the basic principles introduced by Anger.

*Figure 1: Schematic view of a gamma camera. The patient with the organ to be depicted emits $\gamma$ photons. The collimator filters the $\gamma$ rays having in different trajectory. The transmitted photons are absorbed by the scintillator and light is emitted. This light is converted to electrical pulses by the PMTs.*

The function of the collimator is to filter the direction from which photons can reach the detector. Most collimators are made from a dense material, usually lead or tungsten, with parallel holes of a few millimeters in diameter. Photons travelling parallel to the collimator holes pass through and will emit flashes of visible light when absorbed in the scintillator. Other photons are absorbed in the collimator septa. Crystal thickness is customized for the energy of the photons (typically 9 to 12 mm NaI scintillator for 140 keV $\gamma$-rays from $^{99m}$Tc).

The readout electronics of a conventional analog camera consist of a resistor grid (the Anger logic) that converts the pulses to $X$ and $Y$ coordinates that describe the position of the incident photon in the detector area. The image acquired is always a 2D projection of the 3D distribution of the radioactive nuclei in the patient. The $X$, $Y$ coordinates are calculated from:

$$X = \frac{X^+}{X^+ + X^-}, \quad Y = \frac{Y^+}{Y^+ + Y^-} \quad \text{(Eq. 1)}$$

where $X^+$, $X^-$, $Y^+$, $Y^-$ are the four Anger (corner) signals from the resistor grid. The detected energy ($E$) is:

$$E = X^+ + X^- + Y^+ + Y^- \quad \text{(Eq. 2)}$$
The readout system, and the X and Y coordinate calculation, are different in full digital gamma cameras. An ADC converts the analog outputs of each PMT to a digital signal. Position calculation may then be performed digitally. The fully digital design results in greater processing flexibility, energy and spatial resolution. The X and Y coordinates are calculated from the pulse with the largest amplitude.

The energy spectrum of the photons detected is continuous from the primary energy down to zero. Non-primary events have usually been scattered and do not represent the true origin of the original photon. The correct localization of detected gamma rays is crucial for imaging. Many of these events can be excluded by applying an energy window positioned about the energy peak of the primary photons. However, some unwanted events still occur within the energy window and lead to reduced image quality. These are due to scattering in the patient, photons not fully absorbed in the collimator or even pile up events. Corrections for these false events are required to produce high quality pictures.

2.2. SPECT

The SPECT (Single Photon Emission Computed Tomography) technique [3] is based on the idea of acquiring 2D projection images of the distribution of radiopharmaca from different directions using a rotating gamma camera. The original 3D radioactivity distribution is then reconstructed from those 2D projections. For this reason the camera head is mounted on a gantry that allows it to rotate around the patient.

To increase the sensitivity and thus shorten the time of examination the SPECT systems can have more detector heads. Figure 2 shows a dual headed SPECT system.

Gamma cameras and SPECT performances are mainly optimized for 140 keV photons, which is the energy of the photons emitted from $^{99m}$Tc isomer decay. The energies of the other medically useful single photon emitting isotopes (e.g., $^{123}$I, $^{133}$Xe and $^{201}$Tl) are close to this value.

Figure 2: Schematic view of a dual head SPECT camera. The two heads are arranged in 90° and rotate around the patient simultaneously.
2.3. **Ectomography**

Ectomography [4] is a limited view angle tomographic method which means the angle between the direction of projection and the axis of rotation is greater than 0° [5][6]. Ectomography is an alternative to conventional SPECT and the ectomographic imaging method was first implemented in the Cardiotom camera [7] and so far no other scanners have been built using this method.

In the Cardiotom, the camera head is stationary and does not rotate around the patient, as in SPECT. Instead a slant hole collimator rotates stepwise through 360° around an axis perpendicular to the detector. The acquisition geometry of ectomography is shown in *Figure 3*. The shape of the FOV is a cone in 3D and it is shown in the drawing as a green triangle.

![Figure 3: Acquisition geometry of ectomography with slant hole collimator. The reconstruction volume is the green triangle. The blue arrow shows the rotation direction of the collimator.](image)

The advantage of ectomography as used in the Cardiotom is its simplicity and thus the possibility for mobile mechanical construction. The system enables bedside application which better facilitates hospital logistics compared to a stationary SPECT.

2.4. **Positron emission tomography**

Positron Emission Tomography (PET) [8] is based on the simultaneous detection of two annihilation photons from positron emitter isotopes. PET is a widely used and rapidly growing nuclear imaging modality. High sensitivity PET studies are used in oncology [9], for brain function measurements or when developing and testing new drugs. In addition, the pharmaceutical researchers frequently test their new radiopharmaceuticals on small animals (like rats, mice, or monkeys).

Since most of the chemical elements found in living tissue also have positron emitting isotopes, a majority of these biomolecules can be labeled with radioisotopes, and signals from the coincident 511 keV annihilation photons can be detected. The positron emitters $^{11}$C, $^{15}$O, $^{18}$F, $^{13}$N are frequently used in PET scans.

PET cameras consist of a ring of scintillation crystals surrounding the object and thus no rotation of the detectors is necessary. The size of the crystal elements should be as small as possible to achieve a good spatial resolution, but on the other hand they should have sufficient thickness for absorption of the 511 keV photons. The principles of a PET system are shown in *Figure 4*. If two coincident photons hit two scintillators, the emitting nucleus is located along the line connecting the crystals (called LOR, line of response). In reality, more events will be
registered. Apart from the true coincident signals, random and scattered events will also be detected. Corrections for these unwanted events are essential for producing reconstructed images of high quality.

Since in PET the two photons are detected in coincidence there is no need to use a collimator and thus the sensitivity of the system will increase significantly. This so called electronic collimation does not depend on the source detector distance which improves the spatial resolution.

However PET technique also has spatial resolution limitations. One is due to the fact that the mean free path of the positron (and thus the point of annihilation) depends on the energy of the positron. This range is 1-2 mm in water for the most commonly used positron emitters. In addition, since annihilation generally occurs prior to the positron reaching a state of rest the direction of the two emitted photons will slightly deviate from $180^\circ$. The range of positrons could be a problem for small animal PET systems where high spatial resolution is required. The deviation from $180^\circ$ of the two photons is on the other hand of no concern for small animal PET systems. Another problem is the depth of interaction in the scintillator material that decreases the resolution, especially in small animal PET systems.

![Figure 4: The PET acquisition geometry with two annihilation points (red and yellow). The black line shows a true coincidence event, blue arrows indicate a scattered coincidence event, caused by the two photons from the red source. A random coincidence event is represented by the two red arrows. Dotted lines show the mis-positioning of the LORs calculated from the scattered and random coincidence events.](image)

Nowadays combined systems (SPECT-CT, PET-CT) are installed in hospitals to provide not only structural information in the same coordinated system of the functional images but also to facilitate accurate CT based attenuation corrections.

### 3. Medical imaging systems presented in this thesis

The thesis is devoted to the description of some medical imaging systems using ionizing radiation. These are gamma cameras based on ectomography such as Cardiotom Mark 2, Mark 3 and the new Mark 4. The small animal PET cameras that are described are
miniPET-1 and miniPET-2. Finally transparent demonstrators have been designed and introduced as a laboratory exercise at KTH. These are the Cardiotom Mark 1 the KTH-SPECT the KTH-miniPET and a dual slice helical CT, the KTH-spiralCT. The devices are summarized in Table 1.

<table>
<thead>
<tr>
<th>Device</th>
<th>Type</th>
<th>Used for</th>
<th>Described in</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cardiotom Mark 2 and Mark 3</td>
<td>Gamma camera</td>
<td>human examinations</td>
<td>Paper II, Paper III, Paper IV</td>
</tr>
<tr>
<td>Cardiotom Mark 4</td>
<td>Gamma camera</td>
<td>human examinations</td>
<td>this thesis</td>
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<tr>
<td>miniPET-1</td>
<td>Small animal PET</td>
<td>test setup</td>
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<td>miniPET-2</td>
<td>Small animal PET</td>
<td>small animal studies</td>
<td>Paper VII</td>
</tr>
<tr>
<td>KTH-SPECT</td>
<td>Gamma camera</td>
<td>laboratory exercises</td>
<td>Paper VIII</td>
</tr>
<tr>
<td>Cardiotom Mark 1</td>
<td>Gamma camera</td>
<td>laboratory exercises</td>
<td>this thesis</td>
</tr>
<tr>
<td>KTH-miniPET</td>
<td>PET</td>
<td>laboratory exercises</td>
<td>Paper IX</td>
</tr>
<tr>
<td>KTH-spiralCT</td>
<td>Dual slice helical CT</td>
<td>laboratory exercises</td>
<td>Paper IX</td>
</tr>
</tbody>
</table>

4. Data acquisition systems in nuclear imaging

4.1. Data acquisition hardware for nuclear imaging

The goal of nuclear medical imaging devices is to locate the position of radioactivity and reveal the functionality of an organ to be studied. Every nuclear medical device is based on the same principle, as mentioned earlier; detecting the position and energy of γ rays emitted by the radiopharmaceutical in the patient. The detectors which convert these photons to electronic pulses are usually scintillator crystals with PMTs or SiPMs as light readout. However, solid state detectors are also used. The hardware of the system consists of a detector, analog front-end electronics, circuits for digitalization of the analog signals and acquisition computer. The general description of the detector is presented in Section 2.1.

The analog front-end electronics perform amplification of the detector pulses, form the Anger signals (see Section 2.1, Equation 1) and might also provide high voltage for the PMT. The outputs of the front-end, which are still analog signals, are connected to the signal processing unit to be digitized and streamed in list mode (all event information, such as position, energy and time stamp) to the computer. A modern signal processing unit is based on FPGA (Field Programmable Gate Array) and DSP (Digital Signal Processing) techniques. The FPGA has the advantage of being able to contain different program codes and is configurable dynamically providing a flexible hardware solution.

The data acquisition and evaluation software in the computer performs the required post-processing of the list mode data, like energy and uniformity correction, filtering, etc., and calculates the projection as well as the reconstruction images. The block schema of a general detecting system is shown in Figure 5.
The imaging devices, included in this thesis, use different types of acquisition systems. *Cardiotom Mark 2, Mark 3* and *KTH-SPECT* that are modernized, and demonstrators, use the conventional analog front-end coupled to a modern DSP unit which may contain FPGA. The *Cardiotom Mark 4* on the other hand was built as a full digital system.

The DSP hardware solutions, based on SAROC, Nallatech and Memec, were all developed at the Institute of Nuclear Research of The Hungarian Academy of Sciences and are presented in the sections below.

### 4.1.1. SAROC

SAROC (Serial ADC ReadOut Controller) is a PC based data acquisition card. It is designed for ISA interface. The card can handle the 4 corner signals. It also requires a start signal which enables signal processing. A Windows XP device driver communicates with the card, it reads its digitized Anger signals and transmits the data to the acquisition software. The SAROC based acquisition system is installed in the demonstrators *Cardiotom Mark 1* and *KTH-SPECT*.

### 4.1.2. Nallatech

This data acquisition hardware is manufactured by Nallatech Ltd. [10]. It is a PCI card and can be plugged directly into one of the slots of a PC. The card consists of two modules named BallyRiff and Strathnuey. The Strathnuey module is the main board with PCI and USB interfaces to transfer data directly to the PC. The BallyRiff daughter module is mounted directly on the Strathnuey card. It contains a XCV1000E-6 Xilinx Virtex II FPGA processor [11] and four 12 bit ADC circuits. The FPGA is programmed in VHDL language.

The two modules, Strathnuey and BallyRiff, contain all the electronic circuits required to digitize four analog signals and send the digitized events to the PC. The block diagram of the Nallatech card is shown in Figure 6. This system also contains an external communication module with Ethernet interface.

The DSP technique, the fast free running 50 MHz ADCs and the high data transfer speed of the PCI bus makes it possible to achieve a high count rate in data collection.

In *miniPET-I* the Nallatech card is used [12], where data are transmitted via the Ethernet interface. *Cardiotom Mark 2* and *Mark 3* are also equipped with Nallatech, but the card is plugged directly into the PCI slot of the acquisition PC.
4.1.3. Memec minimodule

This system is similar to the Nallatech solution, a Xilinx Virtex 4 FPGA [13] is used as a digital signal processor unit. It contains a microprocessor (a PowerPC) with an embedded Linux operating system. The Memec minimodule [14] is equipped with both DDR and FLASH memories for system memory and non volatile storage of software components. The block schema of the Memec system is shown in Figure 7.

The 4 analog input (Anger) signals are digitized by a free running 12 bit 4 channel ADC. The digitized events are processed by the FPGA using the DSP technique and sent to the acquisition PC via an Ethernet network. X, Y position, energy calculations and time stamp generation are implemented in the FPGA. More than one module may be connected to a PC using a standard gigabit Ethernet switch. An external central clock module is necessary for the synchronization of the time stamps between the modules. Each module works independently which facilitates the design of new dedicated systems. The Memec minimodule is used in the miniPET-2 small animal PET scanner, in the KTH-miniPET and in the KTH-spiralCT demonstrators.
4.2. Crystal identification method for position sensitive detectors

PET detectors are built from pixelated crystals optically coupled to a position sensitive light readout. The identification of each crystal in a detector module is a crucial task after data acquisition. The identification means determining the separation in space and numbering each crystal. The numbering makes the reconstruction easier, since the position of a coincidence event is convertible to a crystal number that determines the LOR. The identification and numbering algorithm is then based on the flood field image collected and consists of four steps.

The first step is the calculation of the binary mask which represents the positions of the crystals. This gives the positions of each crystal center.

The second step is the separation of the overlapping crystals. This might be a problem if Position Sensitive Photomultiplier Tubes, PSPMTs are used to collect and identify the position of scintillation light from a pixelated detector block. Terminal crystals may be overlapped in the flood field image because the total size of the crystal array is close to the effective area of the PSPMT. This means that the emitted light cone from the crystal is detected by only one anode at the border of the PSPMT, while more than one anode detects the light in the central part of the tube. Terminal crystals in the flood field picture will either be overlapped or much closer to their neighbors, unlike in the center.

The third step is the numbering of the binary mask which starts from the center of the image and progresses to the borders clockwise.

The fourth step increases the size of the calculated crystal centers (represented by dots in the numbered mask) pixel by pixel. It is an iterative process and stops when the neighboring dots abut.

The crystal identification and numbering algorithm were developed for the miniPET-2 scanner and are described in Paper VII. The algorithm is also used in KTH-miniPET and KTH-spiralCT for crystal identification and numbering.

A part of the properly calculated miniPET-2 crystal discrimination map is shown in Figure 8 (8x9 crystals from the 35x35, where the first two rows are located on the border of the crystal array). Red boxes in the flood field image show the calculated crystal positions.

Figure 8: The top-left part (~10x10 mm$^2$ from the total 48x48 mm$^2$) of a typical crystal identification map of a miniPET-2 detector module. Red boxes show the calculated crystal positions. The side effect can be seen in the first two rows on the top of the image. The distance between the dots (crystal needles) is 1.35 mm, the pixel size is 0.2x0.2 mm$^2$. 
4.3. Event handling

In nuclear imaging the photons emitted by the studied organs, and registered by the detector modules, are the information carriers of the system. The front end hardware transmits this information as digitized events from the detector modules to the data acquisition computer in a byte stream format. Generally, an event contains position, energy and timing information as well as status information.

Different event formats were implemented in the hardware systems, as described in Section 4.1. Different devices based on the same hardware solution, use a similar event format. Table 2 shows the description of the event formats in the devices discussed in this thesis.

Most of the formats contain unused bits which are necessary for keeping the event size in bits proportional to a power of 2. This, besides the compatibility reasons, ensures a faster data processing on the front-end hardware and on the PC. Post-processing, such as energy correction, energy windowing or homogeneity correction, is performed in the PC after receiving an event.

<table>
<thead>
<tr>
<th>Device</th>
<th>Event format</th>
</tr>
</thead>
<tbody>
<tr>
<td>miniPET-1 and Cardiotom Mark 2/Mark 3 (based on Nallatech):</td>
<td>Event size: 3x32 bits (12 bytes)</td>
</tr>
<tr>
<td>2 30 2 2 12 4 12 2 2 12 4 12 bits</td>
<td>id: shows the order of each double words (4byte)</td>
</tr>
<tr>
<td>id timestamp id - X* - X id - Y* - Y</td>
<td>X*, X*, Y*, Y*: corner signals</td>
</tr>
<tr>
<td></td>
<td></td>
</tr>
<tr>
<td>miniPET-2 and KTH-miniPET (based on Memec):</td>
<td>Event size: 128 bits (16 bytes)</td>
</tr>
<tr>
<td>8 8 16 16 16 32 32 bits</td>
<td>status: shows pile-up, overflow events</td>
</tr>
<tr>
<td>status no energy X Y timestamp timeslice</td>
<td>no: event number</td>
</tr>
<tr>
<td></td>
<td>energy, X, Y: the energy, and the coordinates of the event</td>
</tr>
<tr>
<td></td>
<td>timestamp: the lower 32 bits of the 64 bit timestamp in 0.625 ns steps</td>
</tr>
<tr>
<td></td>
<td>timeslice: the upper 32 bits of the timestamp 64 bits timestamp</td>
</tr>
<tr>
<td>KTH-spiralCT (based on Memec):</td>
<td>Event size: 128 bits (16 bytes), similar to the KTH-miniPET</td>
</tr>
<tr>
<td>8 8 16 16 16 16 16 32 bits</td>
<td>status: shows pile-up, overflow events</td>
</tr>
<tr>
<td>status no - X* Y* X Y</td>
<td>no: event number</td>
</tr>
<tr>
<td></td>
<td>X*, X*, Y*, Y*: corner signals</td>
</tr>
<tr>
<td>KTH-SPECT and Cardiotom Mark 1 (based on SAROC):</td>
<td>Event size: 4x32 bits (16 bytes)</td>
</tr>
<tr>
<td>16 16 16 16 16 16 16 16 bits</td>
<td>X*, X*, Y*, Y*: corner signals</td>
</tr>
<tr>
<td>- X* - X - Y* - Y</td>
<td></td>
</tr>
</tbody>
</table>
5. Image reconstruction

5.1. Filtered back-projection

The mathematical model of the tomographic data acquisition is described by the Radon transform, based on the work of the Austrian mathematician Johann Radon. SPECT, PET and CT acquisition is based on this principle [15]. The first and simplest reconstruction method from parallel projections was the back-projection which is the inverse Radon transform. The algorithm gives a perfect solution for an infinite number of projections. In practice, when a limited number of projections is used, the back-projection method causes artifacts, i.e. a star-shaped pattern around the reconstructed object. Part of the star-shaped pattern can be reduced by filtering [16] the projections in the frequency domain, called filtered back-projection (FBP) [15]. Figure 9a shows projections of a point source and the corresponding back-projections are displayed in Figure 9b. The disadvantages of the FBP method are the high noise content and the star shaped pattern, while the advantage is its high speed. An improved technique that reduces artifacts is the iterative reconstruction method.

![Figure 9: (a) Projections generated from a single central point source. (b) Back-projections of a point source using different numbers of projections (3, 6, 180). With finite numbers of projection angles, "star-shaped" artifacts will occur.](image)

5.2. Iterative methods

Iterative reconstruction methods [17] are commonly used in nuclear imaging to obtain images with high resolution and a good signal-to-noise ratio. Iterative algorithms model the data acquisition process as an over-determined equation system. The reconstruction task is to find the relationship between the projection data and the radioactivity distribution by solving an equation system in an iterative manner. The general iterative reconstruction procedure is shown in Figure 10.
5.2.1. **ML-EM**

L.A. Shepp and Y. Vardi in 1982 introduced the Maximum Likelihood Expectation Maximization (ML-EM) iterative reconstruction method [18]. This method reduces the artifacts, but requires considerably more computation time than FBP. The ML-EM algorithm supports both 2D (slice-by-slice) and 3D input data sets and is able to produce 2D slices as well as 3D volumes. The ML-EM algorithm is described by the formula:

$$x_v^{n+1} = \sum_p A_{pv} \cdot \sum_p \left[ \frac{y_p}{\sum_{v'} A_{pv'} \cdot x_{v'}^n} \right]$$

(Eq. 3)

where $x_v^n$: the estimated $v$'th voxel of the reconstructed volume in the $n$'th iteration step

$A_{pv}$: The $p,v$ element of the system matrix. It describes the contribution of the $v$'th voxel in the image space to the $p$'th projection pixel,

$y_p$: $p$'th pixel of the measured projection (total counts in the $p$'th projection line)

The system matrix represents the voxel contribution of the reconstructed volume to the projection data set. Since the system matrix depends only on the camera parameters it is possible to generate the elements independently from the reconstruction process.

5.2.2. **OS-EM**

The Ordered Subset Expectation Maximization algorithm [19] is the accelerated version of the ML-EM. The projection data are grouped into subsets. An iteration passes through all the specified subsets. Further iterations may be performed by passing through the same subsets using the result of the previous iteration as an initial estimation. The OS-EM algorithm is described by the following term:
\[ x_{v}^{n+1} = \sum_{p \in S_k}^{S} A_{pv} \cdot \sum_{p \in S_k}^{S} A_{pv} \cdot \sum_{p \in S_k}^{S} \frac{Y_{p}}{x_{v}^{n}} \]  

(Eq. 4)

where \( S_k \) is the \( k \)th subset. The subsets may be overlapped or non-overlapped.

The iterative reconstruction software, presented in this thesis, is based on the ML-EM and OS-EM methods. The same reconstruction process has been applied in ectomography and also used in small animal PET imaging; both with predefined system matrices and with the help of standardization of the input data format. The input is represented by a vector structure, wherein one element represents the total number of events along one projection line. This kind of projection data handling simplifies the implementation of the image reconstruction software modules.

6. Design of a common software platform for the nuclear medical devices presented in this thesis

All nuclear medical imaging equipment uses computers for acquiring, processing, presenting and storing data. Dedicated, manufacturer specific software is installed on the camera computer. The data acquisition software is based on the manufacturer’s hardware solution. However image presentation and evaluation must follow the standard medical evaluation protocol of the depicted organs.

During the development of imaging devices listed in Table 1, it was understood that there is a need for software tools which can assist in detector tests, communicate with the hardware and provide data for performance tests. Equipment also needs user level applications for performing examinations in a user friendly way.

A unified software platform (called UniM-SDK) was elaborated for the newly designed and updated nuclear medical cameras at both KTH and the Institute of Nuclear Research of the Hungarian Academy of Sciences. Each camera program has its own GUI (Graphical User Interface) and contains camera dependent dedicated parts, but is based on the unified toolkit.

The requirements of the platform were:

- Modular
- Supports old and new hardware solutions
- Runs both on Windows and Linux systems
- OpenGL based visualization
- DICOM input/output for the clinical applications
- High computational power for clinical applications especially for the image reconstruction
- Easy and transparent design for laboratory exercises
- Simple output file format for demonstrators
- Unified data acquisition methods
- Unified implementation of image reconstruction algorithms for the different cameras

The advantage of the modular system is that a given camera uses only those modules which are necessary, on the other hand common modules are developed only once. The relationship between the medical imaging devices (see Table 1) and its hardware and software
components is shown in Figure 11. Software tools (functions, classes, software libraries) for the standard medical evaluation protocols of the different modalities are included in the platform.

The main features of the software platform fulfil the criteria above:

- Unified pixel/voxel representation
- Curve and histogram representation
- Compressed and non-compressed internal file format
- 2D and 3D image transformations
- Displaying 2D slices
- Displaying curves and histograms
- Displaying markers, rulers, axes, text beside the slices and curves
- Displaying and handling ROIs
- Support for creating different window layouts

- Image filtering with low- and high pass filters
- Support sequential processes and applications
- Implementation of geometrical primitives, such as point, line, plane, triangle, rectangle, polygon
- Contour following
- Surface rendering with marching cubes algorithm
- DICOM input/output and conversion support

**Figure 11:** The interconnection between the different modules in the developed software platform. The equipment and the hardware layer are located at the top of the figure. The unified software components are placed in the lower part of the figure.
6.1. Description of the software platform

The software platform is described in Paper I. The platform consists of 6 core modules. These consecutive modules provide common functions for a specific user level application. The platform is called M-Project and the name of each module starts with the capital letter ‘M’.

The software programs for devices described in this thesis are organized into several groups. These groups have a similar conformation to the core modules mentioned above. The hierarchy of the core modules and the grouped applications are shown in Figure 12.

The MCardiotom module contains the common methods and user level applications for the Cardiotom Mark 2, Mark 3 and Mark 4. The M-miniPET-2 module contains programs related to the development of the miniPET-2 scanner. Most of the components are compatible with the miniPET-1, however the software of the miniPET-1 is not based on the M-Project, since the software platform was introduced after the miniPET-1 project was finished.

The M-KTHscanner software module contains the programs designed for laboratory exercises at KTH (see the summary of the devices in Table 1).

The platform and its modules are written in C++. This implementation ensures fast codes, which is crucial in nuclear medicine where large data sets exist, and the speed of image processing, especially the speed of the most time consuming task, i.e. the reconstruction, is critical.

![Figure 12: The hierarchy of the core and application modules. Core modules are color-coded and the lines show the links between the modules.](image-url)

The user interfaces in the M-Project applications were created with the help of the open-source version of the Qt framework [20]. The platform independent Qt from Nokia provides the general and OpenGL [21] windows, buttons, controls for the user interface under Linux, Windows and Mac OS X operating systems.
DICOM compatibility and the reading and writing of the DICOM files are based on the open-source version of the DCMTK [22] software library which provides low level interfaces for accessing the DICOM tags and pixel/voxel values.

Fourier transformation and its inverse transformation are required both for the filtered back-projection reconstruction and also for filtering an image in the Fourier domain. The FFTW library [23] is used in the platform for the calculation of the FFT and IFFT transformations. The FFTW library is also used in the well known mathematical program package, Matlab [24]. The connection between the applied open-source libraries and the M-Project modules is shown in Figure 13.

![Figure 13: The relationship between the open-source libraries (green boxes) and the M-Project modules (blue).](image)

6.1.1. Core modules

The core modules contain the low level library functions for applications. The modules are presented on the left of Figure 12 and listed in Table 3 below.

<table>
<thead>
<tr>
<th>Module name</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>UniM-SDK</td>
<td>This is the main core, contains pixel/voxel handling methods, geometrical primitives, OpenGL visualization. The features are published in Paper I.</td>
</tr>
<tr>
<td>MDetector</td>
<td>Methods for representing a general PET detector module or a SPECT camera head in Cartesian coordinate system and those visualization classes.</td>
</tr>
<tr>
<td>MReconstruction</td>
<td>Methods for 2D and 3D system matrix generation and the conventional sinogram based 2D FBP, ML-EM, OS-EM reconstructions and filter implementations related to the FBP algorithm.</td>
</tr>
<tr>
<td>MAipAcq</td>
<td>Low level functions for a network based data acquisition system. The components were designed to the Memec minimodule based acquisition system.</td>
</tr>
<tr>
<td>MDaq</td>
<td>Universal data acquisition routines. It hides the hardware differences, described in Chapter 4, from the user level applications.</td>
</tr>
<tr>
<td>MStepperMotor</td>
<td>C++ interface for the stepper motor controller [25], applied in the KTH-miniPET, the KTH-spiralICT, the KTH-SPECT and for the moveable heart phantom.</td>
</tr>
</tbody>
</table>

1) **UniM-SDK**: This module is the basis of the M-Project platform. The main features are the unified flexible template-based data representation technique and the easily configurable OpenGL window layouts. Visualization classes are able to display the implemented geometrical primitives as well as curves, histograms, 2D slices, ROIs and surface models. The functions of the libraries are described in Paper I.
2) **MDetector**: This module contains the geometrical representation of crystals and detector modules for conventional PET and SPECT scanners. The crystals can be either box or cylinder shaped. The box shaped crystal is described by its 8 corner points. Cylinder shaped scintillators use the center of the crystal, a normal vector which describes the direction and in addition the radius and the height. The detector module contains the array of crystals (in case of SPECT, only one crystal). The position, the size of each crystal and the module are used during the system matrix generation and also in the post processing of the *miniPET-2* list mode events. The MDetector module contains visualization methods of crystals and detectors in order to check the arrangement and also to make the interface more attractive for the user. As an example, wireframe model of 14 virtual detectors with 8x8 crystal array are shown in *Figure 14*.

3) **MReconstruction**: It contains sinogram based 2D reconstruction algorithms, such as filtered back-projection, ML-EM, OS-EM as well as the Radon transform. Built-in low- and high pass filters improve the quality of the reconstructed images. The available filters are listed in *Table 4*.

*Table 4: Filters in the MReconstruction module.*

<table>
<thead>
<tr>
<th>Filter Type</th>
<th>Filters</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low pass filters:</td>
<td>Butterworth, Cosine, Shepp-Logan, Hann, Hamming</td>
</tr>
<tr>
<td>High pass filter:</td>
<td>Ramp</td>
</tr>
<tr>
<td>Adaptive filter:</td>
<td>Wiener</td>
</tr>
</tbody>
</table>

Unified 2D and 3D system matrix (SM) generation classes, required for the iterative reconstruction methods, are implemented in the MReconstruction module. The track of the photon is represented by a tube. The probability values for the contribution of a given LOR to an image voxel are stored in the SM and are calculated in the same way both in 2D and 3D: using the intersection of the tubes and the points of the FOV. The geometrical representation of the tubes in a given camera is described with the help of the MDetector module. Papers II and VI contain detailed descriptions of the system matrix calculation.

The system matrices are stored in the memory as sparse matrices (without the zeros). The 8 byte real elements are stored only once using an index table pointing to the matrix item. This kind of representation saves memory because the number of different SM elements is much less than the size of the matrix. The storing schema is shown in *Figure 15*.

*Figure 15: Storing method of the system matrix elements. The 2 byte index table pointers mark the non-zero 8 byte real values in the system matrix.*
4) **MAipAcq**: This module implements a general network based data acquisition protocol, extended with the applications necessary for the Memec minimodule data acquisition card. The protocol uses a client-server model for describing the system.

Two types of clients are implemented: the first are the detector clients, which send data, and the second are the data acquisition clients (DAQCL) which receive data from the detector modules and process those. The clients accept control commands, such as start, stop and set parameters. An arbitrary number of detector and DAQCL might be present in the network.

The server controls the data acquisition process and the clients. It sends commands via the network and does not handle the data. The server establishes communication between the clients.

The commands are sent on a command channel, while data are forwarded on a data channel. The client-server architecture and the command and data channels are shown in **Figure 16**. Each client has only one command channel and multiple data channels, which support the connection of a cluster of PC as acquisition computers.

![Figure 16: The client-server architecture of the network based data acquisition system implemented in the MAipAcq module. The command channel between the clients and server is marked with a blue line and the data channels between the detectors and data acquisition clients are red.](image)

The server detects the clients connected to the network dynamically using DYNDISC (DYNamic DISCcovery) protocol. This is realized by a 5-step handshake as described below. In each step a packet is sent to the peer as confirmation. If confirmation is not received within a predefined time interval, the peer sends the information again. After the connection to the command channels is configured, the server sends requests to all clients for building up the data channels. Finally, the system is ready for data acquisition. The command channel uses TCP/IP; the data channel is able to transmit bytes using either UDP/IP or TCP/IP. The five-step handshake is:

a) The clients send broadcast messages with their IP and port number.
b) The server accepts the broadcast messages and sends back the server’s IP and port number.
c) The clients send a request packet for establishing the command channel.
d) The server sends an acknowledge package with the command channel port number.
e) The clients open the command channel and the connection is established.
5) **MDaq**: This is a wrapper for the low level acquisition libraries, such as MAipACQ, functions for SAROC and Nallatech I/O functions. The MDAQ module hides the hardware dependent I/O routines from the user level applications. The MDAQ supports the application of different hardware models within the same data collection program.

6) **MStepperMotor**: The same type stepper motors and programmable 2 axis controllers were mounted in the demonstrators. These controllers are programmable with built-in strings via the serial port of the PC. The MStepperMotor module ensures the serial port I/O and wrapper C++ classes for manipulating the stepper motors. Start, stop, and parameter setting methods (such as velocity and acceleration) are implemented here. The manufacturer’s built-in motor commands do not wait until motion is complete. However this is required in the step-and-shoot acquisition mode (see Chapter 9). The MStepperMotor module overwrites this characteristic and waits until the movement is completed.

### 6.1.2. Application modules

Applications developed for the imaging devices described in Table 1 are also grouped into modules. These modules contain application specific libraries, test routines and several executable programs required for a proper usage of the instruments.

1) **MCardiotom**: It contains programs for the Cardiotom Mark 2 and Mark 3. The full data acquisition, reconstruction and presentation software, camera homogeneity and energy uniformity program and gated acquisition tests applications are implemented. A dedicated user level reconstruction application for the Cardiotom Mark 4 was also placed here. Besides this GUI based software, console applications, such as DICOM-to-raw and raw-to-DICOM converters, as well as the system matrix generator for the camera are available. Screen shots of the programs of the MCardiotom module are shown in Sections 7.1 and 7.2.

2) **M-MiniPET-2**: This module contains libraries and tools for the miniPET-2 scanner tests. One of these tools is the detector tester application, described in Section 8.2.2. Other PET specific tools required both for miniPET-2 and KTH-miniPET are also implemented here. They are the following:

   - **lutCreator**: GUI application for the crystal position map and crystal based energy correction determination.
   - **energyCorrection**: performs the crystal based energy correction on the single events using the crystal position map (see Section 4.2) and applies the predefined energy window.
   - **coincidenceSorter**: processes the output of the energyCorrection program and calculates the coincidences from the singles using a predefined time window.
   - **coincidenceSpectrum**: calculates the coincidence time spectrum.
   - **ssrb**: transforms the coincidence events to sinograms using the single slice rebinning (SSRB) [26] method.

The results of the process chain are the 2D sinograms. These can be reconstructed using the algorithms in the MReconstruction core module.
3) **M-KTHScanner**: This module contains the data acquisition software solutions for the *KTH-spiralCT, KTH-miniPET, KTH-SPECT* and *Cardiotom Mark 1*. This module is designed for educational purposes, which implies transparency. Each step in the data acquisition is manual to provide a better understanding of the event processing from the acquisition to the reconstruction and image presentation. The M-KTHScanner module uses routines not only from the core modules, but also from two other application modules *MCardiotom* and *M-MiniPET-2*. 
7. Developments in clinical nuclear imaging

7.1. Cardiotom Mark 2 and Mark 3

The Cardiotom system is a mobile gamma camera for 3D imaging. It is based on ectomography, which is an alternative method to SPECT. Ectomography [5][27], has been developed and implemented in the Cardiotom systems at the Division of Medical Engineering, Karolinska Institute. The system has been designed to provide a tomographic unit that can be used bedside, when it is not possible to move a patient to the department equipped with a standard SPECT system. Indeed, the Cardiotom suits the emergency environment thanks to its relatively small size and simplicity compared to SPECT systems. The device is designed primarily for early diagnosis of myocardial and cerebral infarction studies in the emergency room [28].

The Cardiotom uses the limited view-angle technique of ectomography and a rotating slant hole collimator [29] to perform tomographic imaging (see Figure 3). In SPECT, the direction of the collimator holes is perpendicular to the axis of rotation, whereas in ectomography, this angle is less than 90 degree. Since only the collimator rotates, the camera can be positioned very close to the patient (approximately 10 mm).

In limited view angle tomography the volume imaged is not covered by all projections. This result in missing data and an incomplete reconstruction of the volume imaged. The volume that can be reconstructed from ectomorphic projections is conical as shown in Figure 3.

The Cardiotom systems, Mark 2 and Mark 3, have been introduced earlier [28][30], but performance was lacking in the areas of data acquisition speed and image reconstruction capabilities. The aim of the work was to replace the Cardiotom hardware with high performance fast electronics and develop new dedicated software for data acquisition and image reconstruction. The original mechanics with the rotating collimator, the camera head with the NaI crystal, the PMTs and the analog resistor grid are the only parts of the old systems that remained unchanged.

7.1.1. Segmented collimators

By segmenting the slant hole collimators, (i.e. dividing the collimator into equal sections in which the slant angle is the same, but orientated in different directions) the sensitivity is increased. This can be used either to reduce the acquisition time or to maintain the same time and reduce the total dose to the patient. However segments make the field of view smaller and more difficult to position the camera head. The projection area shrinks with an increasing number of segments. The common volume for a segmented collimator for all projections constitutes a double cone. The volume outside the double cone is not covered by all the projections resulting in missing data that may cause artifacts in the reconstructed volume.

The relevant geometrical parameters of the multi-segment collimator (cf. Figure 17) are the diameter ($D$), the thickness ($t$), the slant angle ($\alpha$) and the number of sectors ($N$). The geometry of the reconstructed volume is strongly dependent on these parameters.
The thickness of septum between the collimator sectors is described by the term:

\[
s = 2 \cdot t \cdot \tan(\alpha) \cdot \sin(\pi/N)
\]  
(Eq. 5)

This denotes that a non-sensitive region exists between the sectors. The distance \((z_{\text{max}})\) from the collimator, where the radius of the reconstruction volume is largest, is:

\[
z_{\text{max}} = \frac{(D - s)}{2 \cdot \tan(\alpha) \cdot (1 + \sin(\pi/N))}
\]  
(Eq. 6)

Table 5 shows the maximum reconstruction volume radii for the most commonly used collimators. The radius of the reconstruction volume as a function of distance \((z)\) can be calculated from the following equation:

\[
r(z) = \begin{cases} 
  z \cdot \tan(\alpha) \cdot \sin\left(\frac{\pi}{N}\right), & \text{for } z \leq z_{\text{max}} \\
  \frac{D - s}{2} - z \cdot \tan(\alpha), & \text{for } z > z_{\text{max}} 
\end{cases}
\]  
(Eq. 7)

Table 5: Maximum radius of the reconstruction volumes and their distances from the collimator face calculated for a different number of segments. The collimator diameter is 400 mm and the thickness is 30 mm.

<table>
<thead>
<tr>
<th>Collimator</th>
<th>(r(z_{\text{max}})) [mm]</th>
<th>(r(z_{\text{max}})) [% of FOV]</th>
<th>(z_{\text{max}}) [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>2 segment ((\alpha=40^\circ))</td>
<td>87</td>
<td>21.8 %</td>
<td>104</td>
</tr>
<tr>
<td>3 segment ((\alpha=40^\circ))</td>
<td>83</td>
<td>20.8 %</td>
<td>114</td>
</tr>
<tr>
<td>4 segment ((\alpha=40^\circ))</td>
<td>75</td>
<td>18.8 %</td>
<td>127</td>
</tr>
</tbody>
</table>

Cardiotom systems are equipped with a 400 mm diameter and 30 mm thick collimators. Each segment must, however, be large enough to accommodate the 2D projection of the organ of interest. Collimators currently available are divided into 2, 3 or 4 segments with slant angles of 30°, 37.5° and 40° [6]. A typical projection image of a heart acquired with the 3 segments collimator is shown in Figure 18.
7.1.2. The new hardware

The hardware of the Mark 2 and Mark 3 of today is the FPGA based Nallatech system (see Section 4.1.2). The maximum count rate of the system is approximately 60 kcps for an energy window centered ±10% of the 140 keV from $^{99m}$Tc. Acquisition matrix sizes are currently 64x64, 128x128 or 256x256 pixels. A block schema of the updated Cardiotom camera is shown in Figure 19.

7.1.3. Data acquisition and reconstruction software

In addition to fast data acquisition it is equally necessary to have fast and reliable image reconstruction. The fully 3D iterative ML-EM method (see Paper II) is able to reconstruct the
3D volume from the 12-20 projections (usually acquired in 128x128 pixels) in less than 5 minutes. Data acquisition time depends on the number of collimator positions (number of projection images). The total process from the start of the data collection to the display of the reconstructed 3D volume takes about 10 minutes with 20 angles and 18 seconds per frame, when a three segment collimator is used. The new data acquisition and reconstruction software runs on a 3.4 GHz PC. The software is based on the M-Project toolkit which ensures fast image reconstruction and OpenGL visualization.

Collection of projection images and collimator rotation to predefined angles are performed after an investigation has started. After that, reconstruction starts immediately. The only interaction that is needed from the operator is entering the data of the patient and deciding upon the acquisition protocol. The protocol includes preselected parameters for the number of angles, acquisition time, collimator type, isotope and type of study (stress or rest). A screen shot of the program, where a 3D reconstruction of a human heart is presented, is shown in Figure 20.

![Figure 20: The main window of the data acquisition and reconstruction program of the Cardiotom Mark 2 and Mark 3. Reconstructed slices along the short axis from the base to the apex are presented. The right ventricle is clearly visible in the first three images at the top.](image)

The validation and performance parameters of the new iterative 3D ML-EM method are described in Paper II. The measurements were performed using point and line source phantoms and the results are presented in the Master’s thesis of J. Lagerlöf [31]. The results show that the spatial resolution of the new method is significantly better for the in-depth resolution and is the same for the in-slice resolution compared to the formerly used filtered back-projection method.
7.1.4. Gated imaging – The dynamic heart phantom

A dynamic heart phantom has been developed to test the ECG time gating of the Cardiotom system. The design of the phantom enables the applicability in standard SPECT or PET systems. Tests using the Cardiotom Mark 2 together with a small hand held commercial ECG and its signal generator module [32] were performed. Results were compared to data taken with a standard SPECT system. The phantom design, the performance and the result of the comparison with a SPECT study are described in Paper III.

7.2. Cardiotom Mark 4

Mark 4 is a full digital Cardiotom camera built by the gamma camera manufacturer, Inter Medical GmbH [33] in Lübbecke, Germany. A photo of the new camera is shown in Figure 21. Besides the first three prototypes (Mark 1, 2 and 3) this is the first instrument from standardized production. The camera is equipped with the same type of collimator as used in Cardiotom Mark 2 and Mark 3. The FPGA based data acquisition electronics and the data collection software were installed by Inter Medical GmbH. The new system has standard ECG input thus it is able to collect gated images.

However, the image reconstruction software is based on the M-Project toolkit and includes the existing fully 3D iterative methods (ML-EM and OS-EM described in Section 5.2). The data acquisition, reconstruction and presentation programs use DICOM [34] files for sending the results to the next process. The data flow is shown in Figure 22. When the patient logging and all of the projections are collected, the acquisition program stores the projection data in a DICOM file. The reconstruction software reads this file and performs the reconstruction according to the current setup (algorithm type, number of iterations and subsets, voxel size). When the reconstruction is finished, the reconstructed volume is saved into a predefined file. This file is loaded by the presentation program which displays the results, the operator may reorient the images and then store them on a local hard disk or send them to the hospital’s PACS.

*Figure 21: Photo of the Cardiotom Mark 4 under calibration at the manufacturer.*
Figure 22: Communication data flow between the Cardiotom Mark 4 software modules. The communication is realized by exchanging DICOM files. The reconstruction (right side of the figure) is implemented with the help of the M-Project toolkit, while the acquisition and presentation modules (on the left) are based on the manufacturer’s solution.

7.2.1. Image reconstruction software

The implemented image reconstruction software module works together with the manufacturer’s acquisition and presentation programs as described in Figure 22. Beside the reconstruction, pre- and post-processing modules were implemented in the reconstruction software in order to provide high quality pictures.

A Butterworth low pass filter [35] is applied in the frequency domain, before reconstruction, to reduce background noise. The filter function is described by the following term:

\[
F(\nu) = \frac{1}{\sqrt{1 + \left(\frac{\nu}{\nu_c}\right)^p}}
\]  
(Eq. 8)

where \(F(\nu)\) is the filter function, \(\nu\) is the spatial frequency of the projection image \(\nu_c\) is the cutoff frequency, where the filter function is equal to square root of 2, and \(p\) is the power of the filter. The typically used cutoff value in the cardiac studies is 0.25, while the power equals 4. The filter parameters can be changeable in the program. The shape of the filter with the default parameters is shown in Figure 23. Figure 24 shows projection images of a static heart phantom before and after Butterworth filtering. Filtered images contain less noise and a smoother background.
Figure 23: The Butterworth filter in the frequency domain. The cutoff frequency is marked with a black triangle.

![Figure 23](image)

Figure 24: Unfiltered (left) and filtered (right) projection images from a Jaszczak cardiac SPECT phantom insert. (128x128 pixels, pixel size 1.54x1.54 mm$^2$)

![Figure 24](image)

The M-Project’s reconstruction methods were modified according to the special geometry of the ectomographic system. The reconstruction methods (3D ML-EM and OS-EM) were implemented in a dual core processor environment in order to increase the computational speed. The ML-EM algorithm is able to reconstruct gated studies within 5 minutes, while the time necessary for the OS-EM reconstruction is just 2 minutes on a dual core P4 PC. A screenshot of the reconstruction program with reconstruction parameters is shown in Figure 25.
Figure 25: Screenshot of the reconstruction program of the Cardiotom Mark 4. The reconstruction contains 4 steps: pre-filtering (Butterworth), reconstruction, attenuation correction, post-filtering (Wiener).

After the reconstruction the operator may perform distance based attenuation correction. This is an experimental method, where a constant attenuation coefficient is assumed. The mean attenuation coefficient of a heart filled with blood is approximately 0.12 cm\(^{-1}\). An estimation of the attenuation can be calculated and applied in the reconstruction volume slice-by-slice by using the coefficient and the distance to the collimator. Slices parallel to the camera head have the same source collimator distance. Figure 26 shows the result of the attenuation correction of the heart phantom.

A Wiener filter [36], which is a standard filter for SPECT image processing [37][38][39] is used for post-filtering. It reduces the amount of noise present in the signal by comparison with an estimation of the desired noiseless signal. The 1D Wiener filter is described in the frequency domain by the following equation:

\[
W(f) = \frac{1}{MTF(f)} \cdot \frac{MTF(f)^2}{MTF(f)^2 + M/P(f)}
\]  
(Eq. 9)

where M is the total image count, which for Poisson noise is equal to the average value of the noise power spectrum, P is the object power spectrum and MTF is the modulation transfer function. The MTF can be derived from the point spread function (PSF) by Fourier transformation. The measured PSF function of the Cardiotom Mark 4 is shown in Figure 27. The Wiener filter dialog and the result of the filtering are shown in Figure 28.
Figure 26: The depth model based attenuation correction dialog. Three orthogonal slices (short axis, VLA and HLA is presented from left to right). The uncorrected volume is in the top and the corrected is in the bottom.

Figure 27: The PSF of the Cardiotom Mark 4. The blue line shows the fitted linear trend line and its corresponding equation is also presented.
Figure 28: The Wiener filter dialog. The unfiltered volume is at the top and the filtered is along the bottom.

7.3. Monte Carlo simulations of Ectomography with Gate

Simulations are essential tools in the development of nuclear imaging equipments. They are used in the estimation of performance parameters, such as spatial-, energy- and timing resolution and sensitivity, and also in the design of new geometrical concepts.

The most commonly used technique in particle transport simulations is the Monte-Carlo method. Using this, interactions of the simulated particles with the surrounding material can be recorded in space and time. Material properties, such as density and type of containing atoms influence the interaction and thus the track and energy of the particle. Physical processes, e.g. radioactive decay, Compton and Rayleigh scattering, attenuation and collisions are considered during the simulation.

Dedicated Monte-Carlo simulator software for nuclear medical devices is available. Simind [40], SimSET [41], Detect2000 [42], MCNP [43] and Gate [44] are well known programs for simulations of medical systems.

In this thesis the Gate, Geant4 Application for Tomographic Emission, was used for the Cardiotom simulations. Gate is based on the Geant4 [45][46][47] Monte-Carlo simulator. The main application areas of the Geant4 are high energy-, nuclear- and accelerator physics and space sciences. The Geant4 is developed at CERN and used worldwide.

Gate encapsulates the Geant4 libraries in order to achieve a modular, versatile, scripted simulation toolkit adapted to the field of nuclear medicine. Gate provides the capability for modeling time dependent phenomena such as detector movements or source decay kinetics, thus allowing the simulation of time curves under realistic acquisition conditions. Gate is validated for several PET and SPECT scanners.
Since the Monte-Carlo simulations are very time consuming tasks, a cluster of 24 dual-core PCs were applied for speeding up the calculation. The simulation tasks were distributed between the nodes using Sun Grid Engine software [48].

Spatial resolution of the Cardiotom was simulated studying the performance parameters of the camera. A small sphere shaped phantom with a 1 mm radius and 5 MBq of $^{99m}$Tc activity was used. Data were collected for a 3 segment 30 mm thick slant hole collimator, the same as that mounted on Cardiotom Mark 4. Data for 20 collimator positions were collected with different source to collimator positions from 4 to 17 cm, in 1 cm steps. The acquisition time for the simulation code was 20 seconds/position. Approximately 1,000 events were simulated for each collimator position. After the 20 steps of ML-EM iterative reconstruction, the PSF were determined for the different source positions. The simulated in-slice resolution is shown in Figure 29. The slope and the values of the simulated PSF function is close to the measured function (see Figure 27). Differences are due to the attempt to simplify the simulation; for example background activity, scintillation light distribution in the NaI crystal, electronic noise, variance of the PMT parameters were not included in the simulation.

\[ y = 0.0495x + 5.0525 \]

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{figure29}
\caption{Simulated PSF of the Cardiotom Mark 4. The blue line shows the fitted linear trend line.}
\end{figure}

7.4. Scatter correction of Cardiotom images

Compton scattering of photons in the patient as well as in the camera itself is an adverse process in gamma camera imaging. Several correction algorithms were published [49][50][51] for SPECT cameras but none of them have been applied yet to the Cardiotom system.

A spatially varying scatter correction method based on the Klein-Nishina (KN) cross section for a single photon scattering formula [52] was adopted in the Cardiotom system. The scatter model was studied using simulated projection data. Two simulations were performed with a cylindrical phantom surrounded by either air or water. The cylindrically shaped phantom with a 6 cm diameter and 6 cm height was filled homogenously with 370 MBq activity of $^{99m}$Tc. The center of the phantom was placed 10 cm away from the collimator
surface. In order to simulate the Compton scattering in the body, the same phantom was surrounded by an ellipsoid (short axis: 18 cm, long axis: 40 cm, height: 40 cm) filled with water. The simulation geometry of the phantom is shown in Figure 30. The $X$, $Y$ and $Z$ coordinates of the scintillation point in the NaI crystal, the time stamp, energy and number of scattering in the phantom were recorded for each event during the simulation.

![Figure 30: The simulation geometry in Gate. The cylindrical phantom (blue) surrounded by air. The red parts represent the collimator holes in the different sectors. The yellow part is the NaI scintillator and the gray is the lead shield.](image)

Ten equally spaced projections from $0^\circ$ to $120^\circ$ were simulated using the 3 segment collimator designed for Mark 4. The simulated events are described in Table 6. The $Nse$ is the number of simulated events which represents the photons emitted by the $^{99m}$Tc source in phantom, the $De$ is the number of photons detected by the NaI crystal and $Dse$ is the number of scattered photons in the phantom.

<table>
<thead>
<tr>
<th></th>
<th>Cylindrical phantom surrounded by air</th>
<th>Cylindrical phantom surrounded by water</th>
</tr>
</thead>
<tbody>
<tr>
<td>$Nse$</td>
<td>$6.956 \times 10^9$</td>
<td>$6.8 \times 10^7$</td>
</tr>
<tr>
<td>$De$</td>
<td>$2 \times 10^6$</td>
<td>$1.4 \times 10^5$</td>
</tr>
<tr>
<td>$Dse$</td>
<td>$2.73 \times 10^5$</td>
<td>$10^6$</td>
</tr>
</tbody>
</table>

Simulated list mode events were arranged into 128 energy channels using energy discrimination from 15 keV to 200 keV pixel-by-pixel and 128x128 pixel images with 3.08x3.08 mm$^2$ pixel size. In order to achieve sufficient energy spectrum statistics the pixels were grouped into 2x2 super-pixels. A twenty percent energy window around the photopeak, and a range of 115 to 120 keV as the scaling window were chosen in the evaluation.

The first test of the correction algorithm adopted was the estimation of the scatter fraction for the whole energy spectrum in both phantoms. The second was the pixel-by-pixel calculation of the scatter fraction on the projection data and the reconstruction of the projections.
7.4.1. Simulations of the phantom surrounded by air

With the help of the phantom surrounded by air, described above, the number of scattered photons in the phantom itself might be estimated. Figure 31 shows the total energy spectrum, the estimated Compton scattered photon spectrum in the energy window and the corrected energy spectrum. The total number of scattered photons in the 20% energy window equals 86625 which was calculated from the simulated list mode events. The KN scatter correction method is estimated a 88312 events in the same window. The method overestimates the amount of scattered photons; however the difference is not significant, being only 2%.

![Energy spectra](image)

*Figure 31: The uncorrected (blue), the scatter (green) and the corrected (red) energy spectra of the cylindrical phantom surrounded by air. The amount of scattered events and the corrected spectrum were calculated only in the 20% energy window around the photopeak.*

The spatially varied scatter fraction in the photopeak were calculated in each super-pixel and subtracted from the original picture. Projection images at 0° are shown in Figure 32. The picture a) is the uncorrected image, b) is the scatter fraction and c) is the scatter corrected image. OS-EM iterative reconstruction algorithm with 4 iterations and 4 non-overlapped subsets was used in the reconstruction of the phantom. Figure 33 shows both uncorrected and corrected reconstructed central, cross sectional slices, parallel to the collimator surface.

![Projection images](image)

*Figure 32: The uncorrected (a), the scatter fraction (b), and the scatter corrected (c) projection of the phantom surrounded by air. The pixel size is 3.08x3.08 mm².*
7.4.2. Simulations of the phantom surrounded by water

Simulations were performed in order to test the scatter correction method when the photons scatter not only in the phantom itself but also in the body. The simulation conditions were the same as described in the previous section, except that the cylindrical phantom was surrounded by an ellipsoid filled with water. The size of the body compartment is described above. Figure 34 shows the total energy spectrum, the estimated Compton scattered photon spectrum in the energy window and the corrected energy spectrum. Figure 35 shows the uncorrected, scatter fraction and corrected projections, respectively. The reconstructed central transaxial slices without and with correction are shown in Figure 36.
Correction of Compton scattered photons has an important role, since a large number of photons scatter in the body and in the heart itself. The adopted method is easy to implement however the support of the pixel-by-pixel energy spectrum is required. The data acquisition software of the Cardiotom Mark 4 does not yet support this type of acquisition mode.
8. Developments of PET for small animal imaging

The PET technique is widely used in human studies and recent technical developments have made it suitable also for small animal research. Small animal PET imaging is one of the most suitable methods for testing newly developed PET radiopharmaceuticals or for studying the characteristics of several tumor diseases. Small animal PET scanners require a good spatial resolution and a high sensitivity over a field of view that suits the size of the object being imaged. As mentioned in Section 2.4 a limiting factor is the positron range prior to annihilation which might be a few millimeters depending on the isotope used.

8.1. The miniPET-1 project

A small animal PET device, the miniPET-1, is presented in Papers V and VI in the thesis. It has been designed and built at the Institute of Nuclear Research at the Hungarian Academy of Sciences (ATOMKI), Debrecen, Hungary. Other collaborators of the miniPET project were the PET Center of the University of Debrecen, Hungary, Mediso Ltd, Budapest, Hungary and KTH, Stockholm.

The project is completed and the miniPET-1 serves today as a test bench for technologies in design of a full ring camera for small animal PET imaging. In Paper V the hardware of the miniPET-1 and its Ethernet based data acquisition system are presented.

8.1.1. The miniPET-1 hardware

The miniPET-1 consists of 4 detector blocks mounted on a rotatable gantry. The distance between the detector blocks is 93 mm. A photo and drawing of the miniPET-1 are shown in Figure 37. Each detector block is built up of an 8x8 array of LSO scintillation crystal needles, each with a size of 2x2x10 mm$^3$. The crystal array is optically coupled to a Hamamatsu R8520-00-C12 PSPMT [53]. The signal outputs of the PSPMTs are connected to a Nallatech based data acquisition system (see Section 4.1.2).

![Figure 37: A photo of the miniPET-1 with the 4 detector modules (labeled 1-4) on the left and a drawing on the right indicating the applied coordinate system.](image-url)
8.1.2. Event processing

MiniPET-1 events are stored in list mode and the data processing consists of several steps. The event format is described in Table 2. First the crystal location index and the signal amplitude for energy discrimination have to be calculated. The positions of the incoming photons are calculated from the Anger signals discussed earlier (see Section 2.1). The evaluation of events in the LOR is based on the topological information and the offline coincidence sorting. The coincidence time window can be varied between 2-20 ns. The final tasks are the image reconstruction using the ML-EM algorithm and the attenuation correction. The reconstruction uses a cluster of PCs to increase the speed of the system matrix calculation. The image reconstruction of the miniPET-1 is described in Paper VI. The miniPET-1 software is not based on the M-Project toolkit; however M-Project can handle the miniPET-1 event format and is also able to reconstruct the coincidence data.

8.1.3. Test and performance measurements

The miniPET-1 was tested with line- and a cylinder shaped phantoms. The $^{68}$Ge calibration line source measurement and the corresponding images and results are shown in Figure 38 and Figure 39, respectively. The data acquisition was performed by rotating the detectors over 180° with one minute data collection time per gantry position. Coincidence events above 400 keV energy were selected. The FWHM of this solid phantom, having a diameter of less than 0.6 mm, in the transaxial slice was found to be 1.8 mm.

![Figure 38: Photo of the $^{68}$Ge calibration source and the visible detector blocks 1-4 of the miniPET-1 camera. The solid 7.4 MBq source less than 0.6 mm in diameter is encapsulated in a 105 mm long steel cylinder.](image)
Spatial resolution measurement was performed with the most commonly used isotope, $^{18}$FDG. The phantom, which was a hypodermic needle (inner diameter < 0.5 mm) attached to a syringe, was filled with 11.1 MBq of activity. To minimize scattering the needle was surrounded by air. Images were acquired in the center of the field of view and at distances of 1, 2, 3, 4 and 6 mm from the center in the X direction. The acquisition time was 5 minutes for each angle, with a total of 40 minutes. The spatial resolution of the *miniPET-1* is shown in *Figure 40*.

Another phantom, a syringe with an 8 mm inner diameter, was filled with 1 ml $^{18}$F activity. The length of the 1 ml liquid in the syringe was 17 mm. This is the size of the field of view in the Z direction. The reconstruction of the axial and coronal slices and the 3D view of the cylinder are displayed in *Figure 41*.

*Figure 39: The reconstructed axial and coronal slices and the 3D view of the $^{68}$Ge line source. The image size is 21 mm x 21 mm with pixel size 0.1 mm x 0.1 mm. The profile curve (dots) and the Gaussian fit on the right shows the intensity distribution of the corresponding axial slice.*

*Figure 40: Spatial resolution of the *miniPET-1* in the radial direction.*
8.1.4. Mice studies

Mice are animals small enough to fit into the limited FOV of the miniPET-1 scanner. Mice injected with human pancreas tumor were investigated in collaboration with University of Szeged, Hungary. The purpose of the studies was the testing of the effectiveness of a new experimental drug, developed at University of Szeged, and at the same time the presentation of the applicability of the miniPET-1 scanner.

The mice were injected with $^{18}$FDG and after 30 minutes relaxation time, data were acquired in 8 gantry positions from $0^\circ$ to $180^\circ$ with $22.5^\circ$ angular steps and an 8 min/position acquisition time. A photo from the study and the results are shown in Figure 42. The tumor is located on the back of the mouse. A small tear of FDG was placed close to the tumor to show the orientation in the pictures. High FDG uptake in the tumor was detected.

Figure 42: The photo on the left shows a mouse in the miniPET-1 scanner and the corresponding reconstructed axial slices are displayed on the right. Each slice shows the tracer distribution in the 16 mm diameter FOV. The tumor is shown in the top images and the FDG marker is shown in the lower part. The red circle on the photo shows the location of the tumor.
8.2. The miniPET-2

The aim of the miniPET-2 project was to develop a full ring small animal PET scanner with an FOV large enough for mice and rat imaging using the acquired knowledge from miniPET-1. The second generation of our small animal PET scanner, called miniPET-2, consists of 12 individual detector modules.

The port diameter of the scanner is 206 mm and the FOV is 75 mm in diameter and 48 mm deep. A photo of the miniPET-2 scanner is shown in Figure 43.

![Figure 43: The miniPET-2 small animal PET scanner. The 12 detector modules are mounted on the gantry and connected to a gigabit Ethernet switch.](image)

8.2.1. The miniPET-2 hardware

Each detector module is composed of one Hamamatsu H9500 PSPMT [53] and 1225 pieces of LYSO [54] scintillation crystals (in a 35x35 array configuration, crystal size: 1.27 x 1.27 x 12 mm$^3$). The 256 anode outputs of the H9500 PSPMT are connected to the analog front-end with the resistor grid, which determines the four Angler signals (Section 2.1). These corner signals are connected to the Memec minimodule (Section 4.1.3) based data acquisition system. The PC processes the incoming list mode data, and performs coincidence sorting energy, windowing and image reconstruction.

An external clock generator provides the modules with a 50 MHz central clock signal.
8.2.2. Tests of the detector modules

The test of the individual modules was based on acquiring flood field images (with 256x256 matrix size), while list mode events are used in the energy uniformity corrections and coincidence tests.

The individual tests, like the determination of the uniformity of the PSPMTs or the quality check of the crystal coupling to the PSPMT, are relatively simple tasks.

The detector module tester software was developed for the assessment of the quality of the modules. It is able to acquire flood field images, present the energy spectrum in a given region, perform energy windowing, store the list mode data, as well as perform crystal identification and crystal based energy correction. The detector tester program is based on the M-Project toolkit. A screen shot of the program is shown in Figure 44.

![Screenshot of the detector tester program.](image)

To test the central clock generator and the timestamp algorithm, implemented in the FPGA of the Memec module, list mode data were recorded and the coincidence timing spectrum was determined between facing detectors, using a 40 ns wide coincidence window. The measured coincidence timing spectra and the corresponding values are presented in Paper VII. The mean FWHM of the coincidence timing spectrum of the opposite detector pairs was found to be 2.76 ns.

8.2.3. Crystal identification

Crystal identification and numbering from 1 to 1225 (the total number of crystals) are necessary for PET reconstruction. The method described in Section 4.2 and in Paper VII was used.

The proper crystal identification is affected by three effects: border effect, self activity of the LYSO crystals and the peak-valley ratio. Eight per cent of the crystals per detector
modules are affected by the border effect (102 crystals out of 1225) thus classification is needed for proper image reconstruction. The flood field images used in the identification were established from coincidence events.

8.2.4. Small animal studies using miniPET-2

To illustrate the performance of miniPET-2 for its intended purpose, results of a cardiac study of a rat is shown in Figure 45. The spatial resolution of the system proved to be the expected 1.4 mm. The images were collected at the PET Center of University of Debrecen.

![Figure 45: Short axis view of a rat heart.](image)

8.2.5. Industrial application of the PET technique

Besides the phantom and rat measurements the applicability of miniPET-2 for industrial studies was also investigated. The PET technique is able to follow chemical processes, wherein positron emitter isotopes can be used as a marker. One of these processes is the absorption of gases in catalysts. The goal of this project was to study the applicability of the PET technique and miniPET-2 scanner for 3D imaging of the structure of a heterogeneous catalyst.

Heterogeneous catalysis is fundamentally important in the petrochemical industry using small metal particles supported on a solid surface. Synthetic materials, made from the catalysts, are used every day in products ranging from fuels to fertilizers or as a catalytic converter in cars.

The imaging of the catalytic reaction helps in understanding the kinetics and surface dynamic of catalysis i.e. the choice of the catalyst type. Different experimental conditions influence the speed of a chemical reaction and yield, as well as adding to the understanding of the reaction's mechanism on surface sites.

Heterogeneous absorber, granulated ascarite [55], was packed into small glass tubes. The length of the samples was 30 mm. Ascarite is a sodium hydroxide coated non-fibrous silicate which absorbs CO₂ gas. The two samples, a thin (3 mm inner diameter) and a thick (10 mm inner diameter) glass tube, were tested in the miniPET-2.
$^{11}$C ($T_{1/2}$=20.4 min) labeled CO$_2$ gas was flowed in the tube. The total absorbed activity was 500 µCi in each sample. After 15 minutes of data acquisition, coincidence events were decay corrected and the images reconstructed using the ML-EM iterative reconstruction algorithm. Photos of the thick and the thin samples and the results of the PET study are shown in Figure 46 and Figure 47, respectively. The activity distribution profiles along the longitudinal axis of the samples were calculated and are presented in Figure 48. The analysis of recorded images shows the binding of the radioactive gas molecules in the catalyst. The activity is higher in the front part of the absorbers, which means most of the gas molecules are absorbed in this region, while the back part remains clean.

**Figure 46:** Absorber in the thick glass tube (top picture). The radioactive $^{11}$C-methanol gas was flowed across the tube and the black arrow shows its direction. PET reconstruction of the sample is presented in the bottom pictures. The cross-sectional and the longitudinal slices in the position of the cursor are shown on the left and right, respectively. The pixel size is 0.65x0.65 mm$^2$.

**Figure 47:** The thin absorber (at the top) and the corresponding cross-sectional and longitudinal slices. The pixel size is 0.65x0.65 mm$^2$. 
The study of heterogeneous catalytic processes requires quantitative information on the concentration of $^{11}\text{C}$-labeled reactants, intermediates and products as a function of reaction time, catalyst temperatures and position along the catalyst bed. The high resolution miniPET-2, as shown above, is a feasible tool for the visualization of these factors.
9. Development of demonstrators for medical imaging courses

To get an insight into the different imaging modalities, laboratory exercises are an important part of a university course in medical imaging and are necessary for understanding the principles of complex imaging systems.

Computed Tomographs (CT), Positron Emission Tomographs (PET) and Gamma cameras are the most commonly used structural and functional imaging devices in medicine. Therefore, there is a need for transparent demonstrators where the principles of the different modalities and their functions are presented. The aim of the developments discussed in this chapter was to demonstrate the major building blocks of the different scanners for undergraduate students.

Demonstrators (TOF-PET [56], X-Ray wire chamber [57], CT [58] and gamma camera [59]) have been developed earlier at the Physics Department of KTH. To illustrate the most commonly used imaging modalities for anatomical and functionality studies in clinical situations, SPECT, Cardiotom, PET and helical CT demonstrators have been developed and added to these systems at KTH with the primary goal of providing transparency for understanding the physical processes and at the same time allowing data collection for image reconstruction.

Some of the technologies applied in human investigations and small animal imaging were also used in the construction of the new demonstrators. The designed or upgraded demonstrators such as KTH-SPECT, Cardiotom Mark1, KTH-miniPET and KTH-spiralCT are based on different hardware solutions but all of them were built using the components of the developed unified software platform, M-Project (described in Section 6).

Conventionally, the detector rotates around the patient in most of the medical imaging devices (Figure 49a), except in the PET where the detectors surround the patient. In the KTH-SPECT, KTH-miniPET and KTH-spiralCT the phantom rotates around its own axis and data acquisition is performed with the stationary detector system (Figure 49b). This is equivalent to a reverse rotating detector around a patient. This solution for the demonstrators simplifies the mechanical and electrical arrangement of the systems. The rotation is controlled by stepper motors.

Step and shoot acquisition technique [60] was implemented in the demonstrators, except the KTH-spiralCT which is able to acquire data during continuous motion. After the data collection in a given position (angular or bed position or both of them), the motor rotates the phantom to its next position.

![Figure 49: (a) Conventional mode: the detector rotates around the patient. (b) The implemented rotation mode in the demonstrators: the phantom rotates around its own axis and the detector is in a fixed position.](image-url)
9.1. KTH-SPECT

An old gamma camera of type Starcam Mobile 300 A/M manufactured by General Electric Medical Systems was donated to the medical imaging courses at KTH by Danderyd Hospital, Nuclear Medicine Department.

A rotating phantom has been built. It is placed in front of the camera head and rotated to collect 2D projections, substituting the standard SPECT geometry. The phantom is a hollow cylinder (a drum with 108 mm diameter) with three $^{57}$Co radioactive point sources attached to its surface. A massive plastic cylinder can be inserted into the drum to simulate gamma absorption (the absorption coefficient is similar to that of human tissue). Figure 50 shows a photo of the camera and the SPECT phantom and Figure 51 shows the schematic representation of the system. The camera together with the phantom is able to demonstrate 3D SPECT imaging.

The phantom rotates around its own axis, as described in Chapter 9, and data acquisition is performed with the static camera head and a LEHR parallel-hole collimator. The phantom and its performance are described in Paper VIII.

Figure 50: A photo of the updated KTH-SPECT (GE Starcam 300) camera with SPECT box in the bottom left corner. The point sources are glued on the transparent drum; the solid black cylinder simulates the absorption of the body.
Figure 51: Schematic view of the SPECT phantom. The stepper motor rotates the drum with the point sources attached through 360°. The absorber can be placed into the drum to simulate tissue equivalent absorption. Lead glass windows on the top and the 1 cm thick steel holder provide radiation protection.

The system can illustrate the advantages and limitations of SPECT imaging in the laboratory exercise or project work. Students can measure the effects of different distortions, such as spatial blurring or attenuation. They can compare image quality for different reconstruction settings. The 2D filtered back-projection was used from the M-Project’s MReconstruction module (see Section 6.1.1) to reconstruct the 3D radioactivity distribution from projections. It also demonstrates the presence of well known star shape artifacts in the reconstructed images.

As an example, a study of three point sources is shown in Figure 52. Figure 53a shows the reconstructed axial slices where two of the point sources are located on the same slice (sources at the top of Figure 52). Figure 53b shows the 3D view of the three reconstructed point sources. The projection images have 256x256 pixels and the reconstructed volume consists of 256 parallel slices with 128x128 pixels.

Figure 52: The screen shot of the data acquisition program. Projection image of 3 point sources and the corresponding X and Y projections are shown in the display. The axis of the rotating drum carrying the radioactive sources is oriented vertically.
Figure 53: (a) The reconstructed axial slices (from slice number 121 to 129). The star shaped patterns (a well known artifact of filtered back-projection) are present around the images of the point sources. The angular step between the projections was 9°. (b) The 3D view of the reconstructed point sources. The two points on the right hand side are shown in the same axial plane. The yellow cylinder shows the orientation of the phantom. The wire-box represents the field of view of the camera.

In Paper VIII the effects on the reconstructed image, in cases when the coordinate system of the 3D projection and the rotating phantom are not completely aligned, is also discussed. It is shown that with a misalignment of as little as 1° the distortion of the reconstructed image is clearly visible.
9.2. Cardiotom Mark 1

*Cardiotom Mark 1* was the first prototype of the Cardiotom cameras. The camera head is a General Electric MaxiCamera 400 equipped with the collimator rotation system. The original data acquisition computer was a DEC PDP-11. The camera is equipped with a 1 segment collimator with 30 cm diameter and 5 cm thickness (c.f. *Figure 17*).

The aim of the present work was to convert this camera to a demonstrator at the KTH. The camera head, the resistor grid for the Anger logic, the high voltage power supply and the analog electronics were kept untouched. The camera is moveable using its original gantry.

The data acquisition computer and the interface between the 4 analog corner (Anger) signals have been replaced with a PC and the same type of SAROC (see Section 4.1.1) data acquisition card as used in *KTH-SPECT*. The collimator rotation is automatic in *Cardiotom Mark 2/3/4*, but in *Mark 1* it is more transparent. The motion is controlled by a motor with a tristate switch (slow-, fast- and no motion). A digital scaler registers the steps and the students can stop the movement manually when the collimator reaches its proper position.

The data acquisition software is based on the M-Project unified software toolkit and the same image reconstruction module is used as in *Cardiotom Mark 4*. The modularity of the M-Project platform enables the using of the data acquisition libraries from the *KTH-SPECT*.

The software follows a step-by-step graphical user interface, where the GUI guides the students through the acquisition process. The photo of the updated camera is shown in *Figure 54*.

![Figure 54: Photo of the Cardiotom Mark 1. The analog electronics with the high voltage power supply are located in the black boxes in the rack. The PC with the data acquisition card is placed above the analog electronics.](image)
The converted system is able to demonstrate a working model of a gamma camera as well as ectomography. The collimator rotation and its mechanics are shown in *Figure 55*. The students can understand the principles of the camera head, control the step-and-shoot acquisition technique, as well as the procedures required for 3D imaging.

![Figure 55: The camera head with the collimator and its open rotating mechanics. The bicycle chain moves the collimator which is driven by a stepper motor.](image)

Before using the *Cardiotom Mark 1* for 3D imaging the students perform quantitative studies to determine the parameters of the collimator, such as slant angle and thickness, as well as perform the pixel-to-mm calibration, and measure the point spread function of the camera using point source. For instance, from the projection image of two sources, the distance between the points can be expressed in pixels. A screenshot of the acquisition of program of the *Cardiotom Mark 1* is shown in *Figure 56*, with the acquired projection image of the two sources. The calculated pixel size is 1.25 mm using 256x256 matrix.

![Figure 56: Screenshot of the data acquisition program. The distance between the point sources is 100 mm and the distance in pixels is equal to 80. The matrix size is 256x256 pixels. The energy spectrum of the $^{57}$Co source is displayed on the right.](image)
The steps of the data acquisition, which are similar to the other Cardiotoms used in the hospitals, are shown in Figure 57.

The students start the program, place a $^{57}$Co source under the camera and start monitoring. They verify that the source is in the field of view and study the energy spectrum acquired.

The next step is the setup of the examination parameters, such as number of angles (collimator positions) and the acquisition time per position.

The camera collects a projection image and then the students move the collimator to the next position to acquire the new projection.

After the acquisition is finished, the program reconstructs the 3D radioactive distribution from the projection images. Students can choose different reconstruction methods (ML-EM or OS-EM) and other parameters. One of the tasks during the exercise is the comparison of the reconstruction algorithms, its running time and the optimal number of iterations.

The last step is the image presentation and storing. The post processing (FWHM calculation, profile generation) of the reconstructed images is done using Matlab [24].

*Figure 57: Flow sheet of the data acquisition.*

Students can also test reconstruction algorithms implemented for the Cardiotoms used in hospitals. Three orthogonal slices of a reconstructed $^{57}$Co point source are shown in Figure 58.

*Figure 58: Three orthogonal slices (axial, coronal and sagittal) of a $^{57}$Co source using ML-EM reconstruction algorithm and 15 iterations.*
9.3. KTH-miniPET

A small PET (called KTH-miniPET) and a dual energy, dual-slice helical CT (called KTH-spiralCT) have also been designed to serve as demonstrators for educational purposes. The two demonstrators are presented in Paper IX.

The photon detectors of both demonstrators are based on pixelated LSO crystals, similar to the miniPET systems described earlier. The scintillator light is detected by a PSPMT connected to FPGA based digital data acquisition systems. Memec minimodule based data acquisition electronics (the same as implemented in the miniPET-2) and data acquisition software, based on the M-Project platform, were installed to serve the modularity and transparency of the systems.

The KTH-miniPET (Figure 59) contains two opposite mounted detector blocks that do not surround the whole field of view. However if we rotate the phantom (i.e. Ken, the doll) between the two stationary detector modules according to Figure 49b, the whole FOV can be covered. A block schema of the KTH-miniPET is shown in Figure 60.

Two $^{22}$Na positron emitter point sources are placed into Ken’s head, which simulates tumors in the FOV. The camera acquires the annihilation photons, originating from $^{22}$Na sources and rotates the doll to the next position. The main parameters of the KTH-miniPET are described in Table 7.

Figure 59: Photo of the KTH-miniPET. The two stationary PSPMTs are mounted close to the head of the doll.
Figure 60: Block-schema of the KTH-miniPET. The two stationary detector modules are placed in front of each other. The doll rotates around its own axis and it is moveable in the horizontal direction (along the Z axis). The motion is controlled by stepper motors. The two 8x8 LSO crystal matrices are optically coupled to the center of each PSPMT.

Table 7: Parameters of the KTH-miniPET demonstrator.

<table>
<thead>
<tr>
<th>Crystal type:</th>
<th>LSO</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of crystals:</td>
<td>8x8 per detector module</td>
</tr>
<tr>
<td>PSPMT type:</td>
<td>R2486 and R8520</td>
</tr>
<tr>
<td>Field of View:</td>
<td>Diameter: 16 mm, Depth:16 mm</td>
</tr>
<tr>
<td>Spatial resolution:</td>
<td>~ 2mm</td>
</tr>
</tbody>
</table>

9.3.1. Detector modules

Each detector module consists of LSO scintillation crystals and a Hamamatsu PSPMT. The effective area of the scintillator matrix is $16 \times 16 \text{mm}^2$ and it is shown in Figure 61. The PSPMT determines the spatial position of the incoming photons and the amplitude of the output pulse represents the energy of the photons. Two different types of PSPMT were used, a square shaped R8520 which has 6x6 cross plate anodes and an effective area of 22 mm x 22 mm, and a round shaped Hamamatsu R2486 with 50 mm active area equipped with 16 x 16 anodes wires.

Figure 61: Photo of the 8x8 LSO crystal array.

The different types of PSPMT do not affect the PET principle, since the individual crystals are identifiable in both detector blocks.

The photocathode sensitivity of a PSPMT is not homogeneous. This means that if two photons with the same energy hit the photocathode at different positions, the detected energies will be different. As a consequence the position of the photopeak in the energy spectrum will be different and its FWHM will be wider, depending on the position of the hit. Energy correction during data acquisition is therefore necessary for both PSPMTs to compensate for this effect. Non-uniformity of the R2486 tube, as measured by the supplier, Hamamatsu, is
shown in Figure 62. The density of the black area indicates homogeneity; the white areas show the reduced sensitivity. The correction value of the \( i \)th crystal is calculated using the formula:

\[
c_i = \frac{P_{total}}{P_i}
\]  

(Eq. 10)

where \( P_{total} \) is the position of the photopeak in the total energy spectrum and \( P_i \) is the position of the photopeak in the energy spectrum of the \( i \)th crystal.

The energy correction values in each crystal positions were calculated and used during the data acquisition. The correction values of the R2486 and R8520 PSPMTs are shown in Figure 63. The uncorrected and corrected energy spectra of the individual crystal needles are shown in Figure 64. The resolution of the corrected total energy spectrum (sum of the spectra of each crystal) is equal to 16%, while the resolution in the uncorrected case is 45% using the R8520 PSPMT. The energy resolution of the R2486 PSPMT is 25% without correction and 10% after correction.

![Figure 62: Non-uniformity map of the round shaped R2486 PSPMT as measured by Hamamatsu.](image)

![Figure 63: Energy correction maps of the R2486 and R8520 PSPMTs. The pictures show that 80% difference is possible between the amplifications under the crystals. In order to calculate the correction tables with high accuracy, 50 million single events were acquired.](image)
Figure 64: The first two graphs show the uncorrected and corrected energy spectra of the 64 LSO crystals, coupled to the R2486 PSPMT. The third and fourth graphs show the uncorrected and corrected spectra of the R8520 PSPMT.
The mechanical arrangement ensures the centers of the two PSPMTs are located along the same axis and are perpendicular to the rotation axis of Ken. The two crystal blocks are coupled to the center of the two PSPMT respectively. Position histogram images and energy spectrum form the square and round shaped PSPMTs as shown in Figure 65.

![Figure 65: The crystal position map and energy spectrum of the R8520 and R2486 PSPMTs. The 64 crystals are represented by the dots. The second peak in the energy spectra is the photopeak.](image)

9.3.2. Data acquisition electronics

Each PSPMT has 4 output ("corner") signals, named as $X^+$, $X^-$, $Y^-$, $Y^+$. These signals are as mentioned above connected to the Memec based data acquisition modules. Each PSPMT has individual data acquisition electronics, however a common clock synchronization is necessary to give a correct time stamp for the events. Therefore the acquisition module of the R8520 PSPMT shares its own clock signal with the module of the other PSPMT.

9.3.3. Coincidence sorting, event processing and image reconstruction

The DAQ program developed for the KTH-miniPET is not only responsible for the data acquisition but also controls the doll movement and rotation. The program is based on the M-Project unified toolkit (see Figure 12).

The single events acquired by the detectors are transmitted to the PC. In order to select the two 511 keV photons, produced in the positron annihilation, the acquisition program selects the coincidence events from the single events using the time stamp (the event format is described in Table 2). This is made by selecting the time correlated events out of the two singles dataset. Because of this offline coincidence sorting dedicated electronics for coincidence detection during data acquisition is not necessary. A coincidence time spectrum is shown in Figure 66. All event processing is done by the PC such as energy windowing (by selecting the 511 keV photopeak), position discrimination, homogeneity correction and ordering the coincidence events into pairs along the LORs and sinogram generation.
Figure 66: The coincidence time spectrum of the KTH-miniPET. The red curve shows the Gaussian fit of the coincidence time spectrum. The FWHM is 3.06 ns.

The reconstruction algorithms, implemented in the MReconstruction module of the M-Project platform, are available in the software for the students.

Sinograms of two point sources are shown in Figure 67. The corresponding FBP reconstruction (3 orthogonal slices) of the $^{22}$Na point sources is shown in Figure 68.

Figure 67: Sinograms of two $^{22}$Na point sources. The missing data in the sinogram indicates that the whole field of view is not covered by the detector.

Figure 68: Reconstructed orthogonal slices of two $^{22}$Na point sources (mounted in doll’s head). Axial, coronal and sagittal slices are displayed from left to right. The star shape pattern of the FBP is visible in the left picture. The pixel size is 2x2 mm.
9.4. **KTH-spiralCT**

The aim of this project was to demonstrate the working principles of computed tomography in general and especially the design of a multislice and helical CT. The *KTH-spiralCT* (see *Figure 69*) is based on similar solutions as the *KTH-miniPET*.

To achieve a simple mechanical design some compromises had to be made in the construction of the *KTH-spiralCT* demonstrator as described in Paper IX. The compromise included simplification in the choice of radiation source and also in selection of the detector array. By using a radioactive source that emits photons in the energy region of an X-ray tube, but at much lower intensity, we were able to adopt the position sensitive single photon detection technology that is used in the *KTH-miniPET*. Protecting students against radioactivity is easier using a low intensity source than a real X-ray tube. This also allows the visual inspection of the system during the laboratory exercise.

By applying the photon detection technology developed for PET we obviously lost the possibility of collecting sufficient data in a short time. On the other hand, single photon detection permits analysis of each photon acquired. Thus, since the emitted photon energies are known, one is able to discriminate scattered photons. In addition CT data can be collected simultaneously for different energy regions (17.8 keV for low energy and 59.5 keV for high energy imaging).

*Figure 69: Photo of the KTH-spiralCT.*

**9.4.1. Mechanical setup**

The detector and the X-ray tube rotate around the patient in 3rd generation CT scanners. To achieve a simple mechanical design, the phantom rotation method (*Figure 49b*) is implemented in *KTH-spiralCT*. The movement of the doll is controlled by stepper motors. The first stepper motor rotates the doll around its own axis which simulates the rotation of the detector and X-ray source assembly. The detector acquires the γ photons from the X-ray source and the motor turns the doll to the next position. The second stepper motor moves the doll along the Z direction, which simulates the patient’s bed motion. The schematic view of the *KTH-spiralCT* is shown in *Figure 70*. 
9.4.2. The X-ray source

As mentioned above, instead of an X-ray tube, a 10 mCi $^{241}$Am low energy gamma-ray source is used to simplify the operation. The source is placed in a tungsten alloy cylinder with a wall thickness of 1 cm, which absorbs the radiation for the highest energy gamma-rays by a factor of $10^5$. A 1.5 mm diameter conical hole collimates the beam. Figure 71 shows the energy spectrum of the $^{241}$Am source, acquired by the KTH-spiralCT.

![Figure 71: Energy spectrum of the $^{241}$Am source. The left peak is the superposition of the 13.8 keV, 17.8 keV and 20.8 keV characteristic X-ray peaks. The right peak corresponds to the 59.5 keV gamma photons.](image)

9.4.3. Detector module

The detector module consists of one Hamamatsu R2486 PSPMT and an LSO scintillation crystal array. The crystals are arranged in a 2x24 array configuration (two rows each contain 24 crystals). The same type and size crystals were used both in KTH-miniPET.
and *KTH-spiralCT*. The photo of the crystal array is shown in Figure 72. The crystal array is attached to the center of the PSPMT.

![Figure 72: Photo of the LSO scintillation crystals used in KTH-spiralCT.](image)

The mechanical arrangement ensures that the center of the PSPMT is located along the same axis as the $^{241}$Am source and perpendicular to the rotation axis of the doll. Nevertheless a small error can be observed between the alignment of the detector arrays and the $X$-$Y$ axis of the PSPMT, in Figure 73. The cause of the misalignment comes from the non-linearity of the PSPMT. This non-linearity will result as curve shaped patterns in the image, but it is obvious that its origin is from the corresponding rigid crystal array. Besides the homogeneity correction, energy correction is also necessary for proper data acquisition. The same type of correction method is implemented in the acquisition software as used in the *KTH-miniPET*.

![Figure 73: Crystal position map of the R2486 PSPMT. The crystals are represented by dots. Some dots have higher activity compared to others because the picture is displayed without homogeneity correction.](image)

### 9.4.4. Data acquisition electronics

The same data acquisition system was used as designed for the *KTH-miniPET* (Section 9.3.2), except that no timestamp generation is necessary for the CT events.

### 9.4.5. The DAQ (Data acquisition) software

The DAQ program of the *KTH-spiralCT* is also based on the M-Project toolkit. It controls the doll movement and rotation and data acquisition and it performs energy and homogeneity correction and also processes the events using the low energy and high energy windows. Finally it also generates the sinograms and performs the image reconstruction.

Steps in the data acquisition are similar to a real CT acquisition but contain additional steps which are necessary for a better understanding of the principles of the CT. The steps are described in Figure 74.
The students open the acquisition program and start monitoring. In this mode the program shows the position map of the scintillation crystals and the energy spectrum. After the selection of the proper energy ranges, the program acquires a topogram from the doll. The students can select the region they intend to scan. Two topograms are collected at the same time, one using low energy and the other high energy photons. The resulting topograms are shown in Figure 75.

The next step is the selection of the acquisition parameters. Here the students can select between conventional CT mode and helical CT mode. They can set the number of acquisition angles, the speed of the bed and the acquisition time.

The camera acquires the data and presents it as two sinograms simultaneously, one for the low and one for the high energy range. An example is shown in Figure 76.

After acquisition is complete, the program reconstructs the 3D radioactive distribution from the sinograms. Students can choose different reconstruction methods (FBP or ML-EM) and parameters. One of the tasks during the exercise is the comparison of the reconstruction algorithms, their running time, and the optimal number of iterations.

Figure 75: Low and high energy topograms of the Barbie doll. The body contour is clearly visible in the low (17.8 keV) energy image (left), while the steel bullet on the doll's back is distinguishable in the high energy (59.5 keV) picture (right).
Image reconstruction algorithms implemented in the MReconstruction module are available in the software. Examples for reconstructed low and high energy slices of the doll are shown in Figure 77.

Figure 76: Low and high energy sinograms of a cross sectional slice of the Barbie doll.

Figure 77: Low (left) and high energy (right) cross sectional slices of the doll. The corresponding sinograms are displayed in Figure 76. The size of displayed area is 48 mm.
10. Conclusion and outlook

Software platform

A unified software platform has been developed for the equipment presented and listed in this thesis, and for further medical as well as non-medical applications. As shown, the components of the platform can be used in a wide range of applications, from clinical devices to a small animal PET scanner (miniPET-2) and in demonstrators for teaching at KTH.

Further plans are to extend the fields of application also to be suitable for a new type of detector system; such as the ongoing project, which has the aim of developing a position sensitive detector system using microchannel plate. Another plan is to replace the existing software of the Atomki’s Palmtop MCA [61] (Multi Channel Analyzer) with a high performance modular program based on the M-Project platform.

Since the platform contains a large number of software tools for data handling and image reconstruction, many other detector systems might be candidates for its application. The candidates might be systems based on the SiPM light sensor, APD, or fully solid state detector modules.

Cardiotom

Cardiotom Mark 2 and Mark 3 were updated and new data acquisition, reconstruction and presentation programs have been developed. The two modernized cameras are used now in Karolinska University Hospital in order to test and validate the ectomographical acquisition method for cardiac studies.

Cardiotom Mark 4 was built by a gamma camera manufacturer and reconstruction software, based on the M-Project software platform, with the required pre- and post-filtering algorithms has been developed and installed.

Monte-Carlo simulations have been used in order to validate the parameters of the Mark 4 and to determine the scatter fraction and its correction on the projection images. Further developments might be the elaboration of a resolution recovery method for the iterative reconstruction.

The overwhelming majority of SPECT systems manufactured and installed in hospitals today are based on the traditional scintillator-PMT technique. However it is obvious that in the near future, these traditional and heavy gamma cameras will be replaced with much smaller and lighter detectors (scintillators equipped with SiPM or solid state detectors). One of the benefits of the ectomographic 3D imaging technique is that its mechanical construction based on a stationary camera head which makes a mobile design possible. This mobility would further be improved by using the new types of small size detectors which require less weight for radiation shielding as well as less weight and size for the gantry.

Candidates for dedicated systems, based on ectomographic imaging technique, are not only the heart but also other superficial small organs.

miniPET

The goal of the miniPET-1 collaboration was to design, construct and test a PET scanner built with four detector blocks. One important objective was also the development of
the data acquisition, image reconstruction and image processing software necessary for the project. Mice studies were performed to show the performance of the camera.

The second generation, full ring small animal PET scanner, the miniPET-2, has been built and used for small animal examinations. We can conclude these small animal PET detector modules can separate the neighboring crystal needles and thus give a superior system resolution with the FWHM=1.3 mm in the center of the FOV. In addition the time resolution is less than the typically used coincidence windows (5ns or 10ns) which permits high count rates resulting in low random coincidence events. The first results of the miniPET-2 setup, based on individual detector modules, show that the detectors are eligible for construction of a scalable full ring small animal PET scanner and the whole system is applicable both for preclinical as well as non-medical studies.

Detector quality control software has been designed for the miniPET-2. Automatic crystal position calculation and detector homogeneity, as well as crystal based energy correction calculation methods have been implemented and tested on miniPET-2, KTH-miniPET and KTH-spiralCT.

Applicability of the PET technique, especially the small animal Positron Emission Tomography, in non-medical fields has been studied with catalysts. The preliminary results are promising.

One of the major advantages of the miniPET system presented in this thesis is its construction based on individual and modular detector blocks. Therefore the geometrical arrangement of the detector modules for dedicated PET systems is only limited by the FOV and can even change during scanning to optimize the data collection rate and/or quality. For instance, this gives possibilities for designing dedicated PET for the breast, Positron Emission Mammography (PEM), or scanners for pediatric purposes where different geometrical arrangements are required.

Besides the possibility of a changeable geometrical arrangement of the detector modules, new types of detectors can be tested with the existing hardware and software setup. One of the aims of our new ongoing EU project (known as the Central Nervous System Imaging project) is to build a small animal PET camera based on SiPMs. The scintillation crystal matrix, the geometrical arrangement, the readout system and the software will be the same type as designed for the miniPET-2.

**Demonstrators**

Demonstrators (KTH-SPECT, Cardiotom Mark 1, KTH-miniPET and KTH-spiralCT) for the medical imaging course at KTH have been designed and/or updated and the required acquisition, reconstruction and presentation software has been developed. The software for the equipment is based on the M-Project software toolkit. The devices demonstrate the principles of planar, SPECT, ectomography, PET and CT imaging modalities. The installed software shows transparently the steps of data acquisition, data processing (e.g. energy window selection, coincidence sorting in PET, sinogram generation) and image reconstruction with different methods and parameters.

A near future development of the KTH-miniPET would be to increase of the FOV by replacing the detector blocks with the ones used in miniPET-2, which have 35x35 LYSO scintillator crystals. This would make PET imaging more “visible” and enable imaging of more realistic 3D objects.
Another radical improvement of the *KTH-spiralCT* would be to build a more “realistic” spiral CT based on the existing mechanics but with the replacement of the detector module with a modern pixelated 2D solid state system. Since the pixel size of such flat panel detectors is considerably smaller (50-200 µm) a greatly improved spatial resolution would be achieved. However, due to the limited number of photons emitted from the $^{241}$Am source used, a true X-ray source would also be needed.
11. Description of my own contribution to the papers included in this thesis

Paper I: Unified software platform

My tasks were the elaboration, design and development of the software libraries as well as the whole software platform and implementation of the device specific user level applications.

Papers II, III and IV: Cardiotom Mark2 and Mark3

The Cardiotom with the present development of hardware and software is a sufficiently fast imaging device for acute heart studies. The high performance data acquisition with the DSP technique, and the fast fully 3D iterative image reconstruction method have improved the spatial resolution and sensitivity of the existing prototypes, Mark 2 and Mark 3. Image artifacts and noise are reduced, too.

My contribution to the revival of the Cardiotom project was the development of data acquisition, reconstruction and presentation software. I have created a new reconstruction software based on the fully 3D ML-EM and OS-EM algorithms and installed this software in the Cardiotom systems. Furthermore, this reconstruction is installed on the Cardiotom Mark 4 camera.

I also contributed to the performance parameters measurements as well as in the test of the gated acquisition method.

The system as it works today can provide images for interpretation within 15 minutes from the start of the examination.

Papers V and VI: miniPET-1

The miniPET-1 project is now completed. My work in the project, presented in Paper VI, was the development and implementation of the fully 3D iterative ML-EM algorithm for image reconstruction. I have participated in the test measurements and in the validation and interpretation of the reconstructed images. I assisted during the whole software development of the miniPET, for instance with GUI design and implementation of the presentation program and algorithm implementation for image processing (e.g. image transformation, interpolation).

Paper V describes the hardware of the miniPET-1 camera, wherein I helped with the phantom measurements, reconstruction and interpretation of the reconstructed images.

The miniPET-1 camera is in use at the PET Center of Debrecen and serves as a test bench for phantom measurements.

Paper VII: miniPET-2

The miniPET-2 project is also completed. I assisted in the development of the detector modules, I have developed software for the determination of detector and crystal performance parameters as well as algorithms for the position and energy discrimination and homogeneity correction. I took part in the initial system tests and non-animal studies.
Paper VIII : KTH-SPECT

The SPECT and gamma camera is an imaging system that has been added to the KTH laboratory for courses in medical imaging. The camera is also used to demonstrate SPECT imaging by collecting 2D projections from a rotatable cylindrical phantom. I have developed the data acquisition and image reconstruction software based on filtered back-projection written in C++, using the developed unified software platform.

Paper IX : KTH-miniPET and KTH-spiralCT

Two demonstrators, a spiralCT and a small PET have been added to the KTH laboratory courses. Both devices demonstrate the principles of the CT and PET modalities for undergraduate students. My tasks were the development of a transparent acquisition, reconstruction and presentation software as well as the communication software running in the FPGA of the Memec minimodule based data acquisition card.
References


[5] S. Dale: ECTOMOGRAPHY - Theory and Implementation in Gamma Camera Imaging, Department of Medical Engineering, Karolinska Institute and Department of Clinical Physiology, Thoracic Clinics, Karolinska Hospital; 1989


[21] www.opengl.org

[22] http://dicom.offis.de/dcmtk


[30] M. Persson: Development of a Mobile Tomographic Gamma Camera Based on Ectomography – Cardiotom. Department of Medical Laboratory Sciences and Technology, Division of Medical Engineering, Karolinska Institute; 2001


[32] Labtech Ltd.: www.labtech.hu

[33] Inter Medical GmbH: www.intmed.de

[34] Nema standard: http://medical.nema.org/


[40] Simind : [http://www.radrys.lu.se/simind](http://www.radrys.lu.se/simind)


[53] Hamamatsu Ltd.: http://www.hamamatsu.com


[57] X-ray Imaging with Wire Chamber: KTH Particle Physics laboratory exercise, 2006


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