Positron Emission Tomography: status of the art and future perspectives

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1 P.E.T.: present features

Positron Emission Tomography (PET) is among the most sensitive methods to trace amounts of molecules in vivo. PET provides a kind of metabolic information that other imaging methods are unable to provide; therefore this technique is used to measure in man or in living animals biochemical and physiological processes in any organ with 3-D resolution. The last 25 years have seen a rapid and still ongoing development in the production of positron emitters, radiochemical labelling techniques, tomograph technology and image reconstruction algorithms.

2 The method

The use of positron emitters for radioactive labelling offers important advantages compared to single photon emitters[1]. Some of the physiologically most interesting chemical elements like Carbon, Nitrogen and Oxygen have only positron emitting short-lived isotopes. This makes a detector sensitive to positron decay highly desirable.

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Positrons travel only a short distance in tissue[1], then annihilate with an electron into two 511 keV γ emitted back-to-back. When these two photons are detected by two opposite detectors the decay event can be located along the line connecting the two detectors. The two 511 keV gamma rays are detected by a coincidence circuit with a narrow time window, usually about 15 ns. These detectors capture the photons emitted by the isotope from within the system under study at a total rate of up to hundreds of millions per second, while the transducers (such as PMTs) convert them to electrical signals sending them to proper electronics. A proper mathematical algorithm corrects the raw data for scatter, attenuation, accidental coincidence; it normalizes for differences in detector efficiencies and reconstructs the spatial distribution of the radioactivity density inside the organ or the system under study.

Since both photons have the same energy the detection probability is almost independent from the point between the two detectors where the annihilation occured. Therefore the corrections for attenuation in tissue, which are generally huge in these measurements, can be done very accurately.

Commercial tomographs consist of thousands of detectors, which are arranged in several parallel rings surrounding the object to be investigated. Pairs of opposite detectors are connected in coincidence. Originally this was restricted to opposing detectors in the same ring or in neighbouring rings with septa shields in between them to get rid of unwanted scattered radiation. More recently, to increase sensitivity and acceptance the septa were removed and coincidences between many rings were registered. Now the most advanced systems have no shielding at all between the detectors. The accepted solid angle is increased by extending the axial Field Of View (FOV) with more rings and tightly packed detectors with a high stopping power arranged in a compact geometry. This 3D or volume data aquisition, where all possible coincidence lines inside the detector volume are registered, induces, besides higher counting rates of true coincidences, a large increase of scatter and random coincidences. A scatter coincidence is a detected event where one or both photons were deflected by Compton scattering and are therefore assigned to wrong coincidence

[1]The β^+ emitted by radioactive fluorine annihilates with a nearby electron after a path of about 1.4 mm to over 13 mm depending on the radioisotope emploied.

lines. Scattered coincidences may reach the same amount as the true events and very elaborate correction algorithms have become necessary to subtract scatter contributions for accurate quantitative results. A random coincidence is an event where two photons from different decays are detected during the short coincidence time window. Random coincidences can be subtracted online using a second delayed coincidence window.

3 New developments

The improvements of the PET tomographs spatial resolution was a primary goal since the beginning [1]. Early tomographs used Tallium doped NaI (NaI(Tl)) crystals as scintillation detectors[2]. NaI(Tl) crystals were soon substituted by BGO (Bismuth Germanate) crystals which have higher detection efficiency and are not hygroscopic.
Commercial tomographs using BGO reached a spatial resolution of approximately 4 mm. PET detectors require that the detector material be of high enough Z to maximize the photoelectric cross sections for 511 keV, be fast enough to handle the high counting rates involved, and be able to separate genuine coincident events. Furthermore, detectors for PET should show a reasonable pulse height resolution for 511 keV photons so that scattered events in the material under study can be rejected.
Presently[3] the first tomographs with a new scintillator material LSO (Lutetium Oxyorthosilicate) are being built. LSO has similar high detection efficiency as BGO, but with a five fold higher light output and an eight times faster light decay time. This allows to push spatial resolution close to its physical limit of 1-2 mm, which is determined by the positron range in tissue and the small noncollinearity of the annihilation photons. Increasing spatial resolution requires smaller scintillation crystals. When the size of the PMT became the resolution limiting element, the one-to-one coupling between crystal and PMT

[2]Brownell and Sweet at Massachusetts General Hospital made the first positron medical image in 1951. The imaging device emploied two simple NaI(Tl) crystals, moved manually to scan brain tumors. In the 1960's and 1970's positron imaging devices used detector's arrays.

[3]New improvements in PET tomographs deal with crystals with better energy resolution (LSO) and shorter decay time (Gadolinium Oxyorthosilicate).

was given up and an Anger camera principle was applied in the block detector concept. For example, 8x8 crystals were cut into a single scintillator block forming a light guide to 4 PMTs. By comparing the signals of the 4 PMTs the individual crystals are uniquely identified. An even more cost effective scheme, a quadrant sharing arrangement of PMTs allows to further increase the number of crystals per PMT. This was recently achieved in the first PET tomograph for neurological studies in humans with LSO scintillators, the High Resolution Research Tomograph (HRRT), manufactured by SIEMENS/CTI, Knoxville, TN, USA and installed at the Max-Planck-Institute for Neurological Research in Koln. It consists of 120,000 single crystal elements, 2.1 mm x 2.1 mm each, arranged in 104 rings giving a reconstructed spatial resolution of less than 2.5 mm in a 20 cm diameter volume. This made it also necessary to measure the depth of interaction (DOI) of the incident photons in the 15 mm deep detectors to avoid ambiguities when photons penetrate several crystals by oblique incidence. From several ideas how to obtain DOI information, a scintillation phoswich with two crystal layers with different light decay times was chosen. Pulse shape discrimination allows then to separate scintillation events in the two layers.

After correction for scatter and random events, attenuation and dead time an image of the activity distribution can be reconstructed. Originally this was done with standard 2D filtered backprojection algorithms giving a stack of transverse image slices. With 3D data aquisition this was extended to volume reconstruction, which required hours of computing time. Therefore the algorithms were parallelized and run on multiple processors. Only recently, with the rapid developments in computer hardware, iterative 3D reconstruction is becoming routinely feasible.

Most recent trend in industrialized countries is the use of hybrid PET/CT systems[9], which combines a dedicated PET with an X-ray computerized tomography (CT) scanner in the same instrument. The CT images provide a map for PET attenuation correction and an anatomic framework for the PET metabolic information.

4 Future trends for PET tomographs

PET detector development progressed significantly since the first PET tomographs. The number of elements increased from a few ten elements in the first tomograph at St. Louis Hospital to more than 100,000 in the new HRRT that has been delivered to Max-Planck-Institute in Koln on February 1999[2]. The HRRT is the first commercial LSO tomograph and it represents a major change in the detector technology for PET. The ECAT HR+ at that time was the highest resolution commercial PET tomograph, using BGO detector crystals. Commercial tomograph with the same detector as the ECAT HR+ but having six rows of detectors rather than four is the ECAT HR++, delivered to the MRC Hammersmith Hospital in London in 1997. The ECAT HR++ has 27,648 detector elements. These BGO tomographs have the highest feasible density detector elements and use large numbers of expensive photomultipliers (1728 for the HR++). Since BGO has limited light output and relatively slow decay the pratical limit in image resolution is reached for these tomographs.
LSO allows the use of fewer PMTs for the PET detectors and at the same time increases the perfomance in counting rate, sensitivity and image resolution. As is evident from data in Table 1, LSO has allowed

	PET/ SPECT	ECAT HR+	ECAT HRRT
Sensitivity ($\cdot 10^3$)	850	1200	4000
Image resolution	4.5 mm	4.5 mm	2.5 mm
Peak true count rate	100,000	200,000	900,000
number of PMTs	198	1152	1120

Table 1: Performance comparisons of three tomographs

an improvement of the HRRT over the HR+ of more than a factor three in sensitivity, almost a factor of two in space resolution, a factor of 4.5 in count rate and still the HRRT has fewer PMTs. This is a breakthrough in detector technology not only for research tomographs, but also for clinical tomographs.

4.1 Scintillator choice

NaI detectors operated in Anger logic mode have served the gamma camera market well in the past, although the hygroscopic nature of the detecor material does not favour the development of such desirable features as crystal segmentation. This feature is indeed valuable when the detector is operated without a collimator and thus requires high count rate and good linearity performances. The nuclear market trend toward Fluorine Deoxy Glucose (FDG) imaging creates the need to acquire images at 511 keV in a reasonable time, which is very inefficent with current NaI detectors. On the other hand, the current BGO detector for PET has insufficient light output to provide adequate energy resolution for the 140 keV currently used in SPECT. Both BGO and NaI have relatively long light decay time (300 ns and 230 ns, respectively), which may limit their counting rate capability. The technology breakthrough in positron imaging is the discovery of LSO crystal which has a number of advantages over the other scintillators used for nuclear medicine instrumentation[4]. Table 2 compares the main parameters of NaI, BGO, BaF_2, LSO crystals. For 511 keV

Parameter	NaI	BGO	BaF_2	LSO
Density (g/cm^3)	3.7	7.1	4.9	7.4
Mean free path (cm at 511 keV)	2.9	1.1	2.2	1.2
Index of refraction	1.85	2.15	1.50-1.54	1.82
Hygroscopic?	yes	no	no	no
Decay time (ns)	230	300	0.8-630	40
Light output [NaI(Tl) = 100]	100	15	5-21	75

Table 2: Properties of NaI(Tl), BGO, BaF_2 and LSO for 511 keV photons

photons, LSO has a mean free path almost equal to that of BGO and is thus an as efficient scintillator with the same potential for high resolution imaging. The light output of LSO is three-fourths of NaI, thus much larger than that of BGO or BaF_2, resulting in a much better energy resolution. The LSO crystal is very rugged and is non-hygroscopic. The scintillation light decay time for LSO is 40 ns compared to 230 ns for NaI, 300 ns for BGO and 0.8 (630) ns for the fast (slow) scintillation process of BaF_2. It provides a

rapid time response and may give a time resolution of 750 ps FWHM. Thus a coincidence time window of about 4 to 6 ns is feasible, which should significantly reduce random coincidences relative to current PET cameras, and improves the signal-to-noise ratio. Moreover the event processing time for LSO can be almost six times shorter than that achievable with BGO or NaI, allowing better counting rate capabilities.

5 Tracer production

The commonly used positron emitting nuclides in PET are ^{11}C,^{13}N,^{15}O and ^{18}F. The first three isotopes are the most abundant elements in organic compounds[1]. This allows the labelling of naturally occuring biomolecules or drugs without altering the compounds defined courses in living systems. The ^{18}F isotope can be used to substitute hydrogen or hydroxyl groups. Because of their short half-life, ranging from 2 min for ^{15}O to 109 min for ^{18}F, they have to be produced close to their application. Various small, compact cyclotrons specially built for the production of short lived isotopes in a hospital environment are commercially available. There are cyclotrons with external proton and deuteron beams, self-shielding machines with internal targets down to 3 MeV beam energy for the exclusive generation of ^{15}O and negative ion machines allowing simultaneous irradiations on separate target ports. Computer controlled automated robot systems synthetize in heavely shielded hot cells radiolabelled compounds in optimized short times of few minutes, including purity controls.

The most commonly used PET radiotracer is FDG, a glucose analog, metabolic imaging agent giving precise and regional information of energy metabolism in brain, heart, other organs and tumors[9]: its uptake in tissue can be easily quantified as glucose metabolic rate. Biochemical processes of the body's tissue, such as metabolism of glucose, are altered in virtually all diseases and PET detects these changes by identifying areas of abnormal metabolism, which are indicated by high photon emission. Cancer cells, for instance, typically have a much higher metabolic rate, because they are growing faster than normal cells, thus they absorb 60 to 70 times more sugar than

normal cells and consequently emit more photons.
The production of ^{18}F used for PET is based on the nuclear reaction $^{18}O(p,n)^{18}F$, where water with 98% of $H_2^{18}O$ as target material is employed; the reaction produces 800 mCi, corresponding to 32 GBq[3]. The relatively high half-life (109 minutes)[9] of ^{18}F allows the radiotracer distribution to local hospitals after its production in a regional center. The half-life of 109 minutes also provides an appropriate decay time for searching tumors and metastases by patient whole body scans.
The high uptake of FDG by the tumor produces very high-localized intensity PET images in which is very difficult to see other organs and structures. To evaluate the exact location of a tumor for staging a patient, for treatment or follow-up planning, it is necessary to setup an anatomic framework for the metabolic information provided by PET. In addition, for better and more exact quantification of FDG uptake, an anatomical map of the structures imaged by PET is needed. These two reasons lead to the development of hybrid PET/CT systems in which both are combined in one instrument. By combining PET and CT in the same instrument, images of both approaches are automatically registered. PET images have an anatomical reference and photon attenuation can be corrected with higher accuracy[9].

6 Complete Body Screening (3D-CBS)

Dario Crosetto is the inventor of the 3D-CBS system and founder of 3D-Computing[6]. Crosetto studied how to improve, by over 400 times, the efficiency of the current PET machine for whole-body examination. These efficiency improvements are made possible because of the novel electronics and detector assembly of the 3D-CBS machine, which is integrated with a CT scan. The 3D-CBS allows the use of a larger area of economical crystal detectors thereby exposing the patients to approximately 4% of the radiation they currently receive.
Before the advent of this invention, screening of the entire body was not advisable because current PET machines expose the patients to over 10 times the radiation recommended by the International Commission of Radiation Protection. This high dose is required,

because most of current PET tomographs detect with less accurate measurements only one out of 10,000 photons emitted from the patients body. In addition, current PET examinations are slow and cost prohibitive.

Crosetto's improvements stem from an innovative way to detect more accurately a larger number of photons emitted by the tracer medium than current designs can do. The patent-pending device will allow for detection of more photons, more accurately, thus reducing radiation doses to the patient, and improving the image quality. Furthermore, it allows the examination of more patients per hour, therefore reducing costs.

The key features of his invention are supported by simulations and hardware implementations. Crosetto's radical improvements in PET efficiency are supported by the new architecture of his electronics using a set of DSP (Digital Signal Processor) on each electronic channel providing the capability to exchange information received from neighboring detector elements and to execute complex algorithms that can measure more accurately the total energy and the spatial resolution of the incident photon. It eliminates also the parallax error of the oblique photons, allowing reduction of false positives, false negatives and an increase in image sharpness. Another supporting feature is the way the signals from the detectors are connected to the set of DSPs on each electronic channel and the new way the detector is assembled as a single (or few) camera(s) made of hundreds of sensors, each capable of finding a photon candidate, versus current PET, which is an assembly of hundreds of small cameras, each with lower energy and spatial resolution at the edges and corners with respect to the center.

The innovations described above allow increasing the length of the PET detector, using economical crystals or different kinds of cheaper detectors, from the current 16 cm to over one meter (when the actual length of the detector is doubled, the number of photons captured is increased by a factor of four).

Untill a few years ago there was the belief that the only way to improve PET efficiency was to improve the efficiency of the crystal detector and not to improve the electronics. During the past 25 years, improvements in PET design have achieved higher efficiency by a factor of only 2 to 3 times every 5 years.

Crosetto's novel technique shows that PET efficiency can be improved by using a special massively parallel-processing system with digital signal processing on each electronic channel. None of the new PET components (crystals, PMT, Avalanche Photo Diode or the new Flat Panel Sensors that replace the PMT) can improve PET efficiency by more than a factor two or so.
The 3D-CBS higher sensitivity will more effectively show abnormal biological processes at the molecular level, before the cancer exhibits sympthoms and before an anatomical change occurs in the body tissue, which is normally detected by CT. Past experience suggests that earlier detection achieved with regular screening using very sensitive devices can dramatically improve survival rates.

7 Current PET limitations

The electronics of the current PET limits its performance; it is not fully capable of extracting the complete characteristics of the interaction between the photon and the detector from signals arriving at high data input rates from thousands of sensors[5]. The electronics has been the main impediment to extending the axial FOV; the increases in efficiency that would justify extending the axial FOV are not possible with the electronics of the current PET. The current PET electronics inefficiencies in detecting photons occur because there is no independent digital signal processing at each electronic channel and there is no communication between adjacent electronic channels. This limitation affects sensitivity and spatial resolution. Sensitivity is lowered when photons striking a crystal coupled to the border of two sensors, causing them to release half (or less) of their energy in two (or more) adjacent electronic channels, are not recognized as photons because each channel receives less than the nominal energy to be considered as a 511 keV photon of a PET emission event.
Spatial resolution suffers at the edge of each 2x2 PMT block because the centroid algorithm cannot weight the PMT signals from both sides of the PMT closest to the point at which the photon struck the crystal. This causes a reduction of the overall sensivity, which translates into greater patient exposure to radiation, poorer image quality and longer scanning time.

The image quality of the current PET is poor because it has:

1. a short FOV, limited by an inefficient electronics that does not offset the cost of the detector if the FOV were increased;

2. no accurate time-stamp assigned to each photon limiting the detection of neighboring photons emitted within a short time interval, causing long dead-time of the electronics and increasing randoms (i. e. photons in time coincidence belonging to two different events), most PETs do not have any photon time-stamp assignment;

3. analog signal processing on the front-end electronics limiting photon identification because of poor extraction of the characteristics of the incident photon and no capability to improve the signal-to-noise (S/N) ratio;

4. detector boundary limitations to 2x2 PMT blocks, no correlation between signals from neighboring detector blocks, no full energy reconstruction of the photons that hit the detector; most of the current PET do not attempt to make any energy reconstruction of the event, but make decisions in accepting or rejecting a photon first and later an event based on the threshold of a single signal.

5. dead-time of the electronics. Dead-time of the electronics is due to any bottleneck (e.g. data multiplexing from many lines to a single line, input saturation, processing, output saturation) present at any stage of the electronics;

6. saturation of the electronics at the input stage due to its inability to detect and process two nearby photons that hit the detector within a short time interval.

7. saturation of the electronics at the output stage due to the limiting architecture of the coincidence detection circuit.

8. a high number of randoms due to the non accurate measurement of the photon arrival time and to the long (about 12 ns) time window used when determining if two photons belong to the same event.

9. poor measurement of the attenuation by different tissues at different locations in a patient's body. These measurements are necessary to calculate the attenuation correction coefficients for PET scans.

Furthermore, high radiation dose delivered to the patient is required by the current PET because each examination needs to capture the amount of photons which provide a sufficient statistics to yield a good image. The short FOV and the inefficient electronics allow accumulation of fewer than 2 photons in coincidence for every 10,000 emitted. This inefficiency requires one to administer a necessarily high radiation dose to the patient in order to keep the examination time within an hour.

8 How to improve current PET efficiency

Before attempting to improve any system, it is necessary to determine where the inefficiencies are, how large they are, how they can be reduced and by how much.

Crosetto considered a total body PET scan where the radiotracer used is water with ^{15}O[8].

The initial number of photon pairs emitted per second by the tracer in the patient body (1424 million) and the number of photon pairs per second captured by current PET (0.2 million) are not in question, because those quantities have been measured by hospitals and universities and are in agreement with measurements done by PET manufacturers.

Of the initial quantity of photon pairs emitted from within the body, some 1210 million pairs per second are scattered or are absorbed by the body. This quantity of capturable photons leads to 15% efficiency for the first stage.

Due to the short FOV (length of the detector), another 196 million photon pairs per second, which emanate from the part of the body not covered by the detector, are lost in the second stage. This leaves only 18 million photon pairs per second remaining to be captured. This quantity is estimated by dividing the length of the FOV of the detector (16 cm) by the average length of the human body (180 cm), yielding an efficiency figure equivalent to 8.5% for this stage applying

this percentage to the 214 million capturable pairs per second leaving the body.
Some photons from within the detector area are also lost. Some others emanating from the part of the body covered by the detector leave the body at angles that escape the detector through the openings between the detector segments. This number can be calculated as a percentage of the perimeter of a circle drawn around the lengthwise cross section of the entire detector not covered by the two 16 cm segments. Thus, (178-32)/178 = 0.82, i. e. 82% of the 18 million photons pair per second remain to be captured after the third stage, which is equivalent to a total 18% efficiency for this stage.
Detector crystals do not have perfect stopping power and do not capture every photon in range. Crystals have efficiency ranging between 80% and 95%. Therefore, applying such rate to the remaining 3.2 million pairs per second still capturable after the third stage in some low cost crystal detectors, 20% are lost, or 0.65 million, and 80% remain potentially capturable, i. e. 2.5 million pairs per second. Current PET capture only 0.2 million pairs per second of the original 1424 million photon pairs per second emitted by tracer within the patient's body. Of the 2.5 million photon pair per second remaining after the fourth stage, the loss of 2.4 million pairs per second is accounted for by deficiencies in the electronics and the detector design. The efficiency of stage 5 and 6 can be calculated as equivalent to 8%, as derived by subtraction from the total inefficiency and the sum of all other inefficiencies.
It is obvious from this analysis that the section needing serious study and improvements is the last one, which provides only 8% efficiency. the first stage has the efficiency related to a natural phenomenon that cannot be changed. The second and third stage can be increased in length and solid angle only if the electronics of stage 5 and 6 are not overwhelmed by the increasing of the data acquisition rate. Stage 4, although the one in which much effort and money has been invested during the past decades, can only be improved from about 80% to something over 95% (among the so called ideal crystals, that is LSO).

9 The 3D-Flow

Crosetto's key innovation on the electronics (on which the 3D-Flow^{TM} is based) consists in a parallel-processing architecture enabling execution of a complex real-time algorithm, calculating different types of depth of interaction with zero dead time and with data exchange with neighboring processors for a time interval longer than the time interval between two consecutive input signals[7].
This provides better energy measurement, helps to reject scatter events more efficiently and provides a way to improve spatial resolution by measuring more accurately the location where the incident photon hits the detector. It also increases sensitivity in accepting oblique photons by eliminating the parallax error accurately measuring the interaction depth.
Another aspect of the invention is the way these concepts are implemented in hardware.
The 3D-Flow^{TM} parallel-processing architecture allows the execution of complex algorithms with neighboring signals correlation in real time and provides the capability to extract more accurate information from the signal generated by the interaction between the incident photon and any type of crystal detector[8]. This allows a more efficient use of economical crystals. The coupling of the detector to the electronics is made in such a way that there are no boundaries or fixed detector segmentations; rather, each sensor of the detector (PMT, Avalanche Photo Diodes, etc.) is an element of a large array with the capability to act as the center of a cluster of elements, all providing information. Finally, the one-to-one mapping of the detector array with a single array of electronic processing channels remedies the inefficiency of current PET in capturing fewer photons, less accurately at the edge and corner of each of the hundreds (or thousands) of small cameras or at the edge of detectors with fixed segmentation.
A data aquisition and processing board has been developed for high efficiency photon detection in PET/CT. The board includes 68 3D-Flow^{TM} processors, each capable of executing up to 26 operations in a single cycle. The 3D-Flow^{TM} DAQ IBM PC photon-detection board has the capability to execute different real time algorithms for photon detection and can be interfaced to different types of daughter analog-to-digital boards. The daughter boards provide signals carrying the

amplitude information from an Analog to Digital Converter and the time information from an analog or digital Constant Fraction Discriminator (CFD) for 16 detector channels, coupled to different types of crystals (e.g. slow: NaI(Tl), BGO, or fast: LSO, Gadolinium Oxyorthosilicate (GSO), etc.).

9.1 3D-Flow^{TM} parallel-processing architecture

This architecture takes the parallelization process one step further than DSP and its software tools allow creation, in only a few hours, of a new application with different algorithms executed on thousands of processors. Each of the 3D-Flow^{TM} processors of one layer of the 3D-Flow^{TM} stack[4] executes in parallel the real time algorithm, from beginning to end, on data received from the PET detector, while processors at different layers of the 3D-Flow^{TM} stack operate from beginning to end on different sets of data received from the PET detector.

The extension by the 3D-Flow^{TM} architecture of the execution time in a pipeline stage beyond the time interval between two consecutive input data is illustrated by the following example: an identical circuit (a 3D-Flow^{TM} processor) is copied four times: A, B, C, D; the number of times the circuit is copied corresponds to the ratio between the algorithm execution time and the time interval between two consecutive input data. A bypass switch coupled to each processor in each 3D-Flow^{TM} in layer A sends one data raw to its processor and passes along to the next layer three input data packets and one output result from its processor. The bypass switches on the 3D-Flow^{TM} processors at layer B send two input data packets along to the next layer, one output result received from layer A and one result from its processor, and so on. Only the processors at layer A are connected to the PET detector and these receive only input data. The processors at layer D send out only results. This architecture simplifies the connection in a parallel processing system and does not require a high fan-out from the detector electronics to send data to different processors of a parallel-processing system. All connections

[4]A stack is a set of several layers, assembled one adjacent to another to make a system.

are point-to-point with several advantages in low power consumption and signal integrity.

10 Conclusions

In conclusion, major developments in PET technology have already played a major part in defining and establishing the role of this imaging modality in oncology. The introduction of PET/CT is the process of refining this role. New software developments combined with the introduction of new scintillators and PET detector designs hold the potential to improve the throughput of this technique and open the way to new clinical applications.
In the present project we plan to tackle the issue of the cost of the electron detectors substituting the expensive small crystals with Multigap Resistive Plate Chambers (MRPCs) which are cheaper gas detectors and can be built and assembled in large dimensions.

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