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***Abstract*— The purpose of this study was to show how advanced concepts of compact, lossless and "Time Of Flight" capable electronics similar to those foreseen for the LHC and ILC experiments could be fairly and easily transferred to the medical imaging field through Positron Emission Tomography scanners. As a wish of explanation, the two overriding weaknesses of PET camera readout electronics, namely timing resolution and dead-time, were investigated analytically and with the help of a Monte-Carlo simulator presently dedicated to this task. Figures have shown there was rather space available for count rate enhancement, especially through a huge decrease of the timing resolution well below the nanosecond. A solution retained and proposed here for the electronics has been partly drawn from the long experience led in High Energy Physics where this last requirement is compulsory. Also appreciable, thanks to its structure entirely pipelined, this scheme enables problems of dead time to be overcome.**

I. Introduction

Since the build up of the first PET camera, they have been recognized as very powerful and sensitive instruments mainly dedicated to brain studies, cardiac imaging along with early cancer diagnosis and therapy. Their uniqueness resides in having the image that stems from within the patient through back-to-back 511 keV photons emitted in time coincidence as a consequence of positrons annihilations. For reasons of cost and complexity, the Field Of View (FOV) of these instruments is rather limited and ranges from 15 to 25 cm only. This penalizes the geometrical acceptance, especially to true pairs of photons (trues) as they may only arise from within the FOV. Adding the discrimination of the incoming events by the time coincidence requirement leads to a very poor overall detection efficiency.

In their way, photons may scatter before eventually hitting the detector, thus defining meaningless Lines of Response (LOR) yet regarded as trues. This can bring into play photons which would normally not have reached the detector as emitted from outside the FOV. Moreover, uncorrelated photons may be found in time coincidence and deceptively define a wrong LOR. Again, they might have come from anywhere in the body. The proportions of scattered and random events relative to the trues do not evolve linearly with the patient size, the activity and the geometry, at the expense of trues.

The restricted acceptance of PET scanners to "trues" along with the high level of noise assigned to scattered and random events contribute to spoil the sensitivity. In 3D PET, selection of events naturally applies in coincidence and is performed by the electronics only. A coincidence window is hardly implemented and serves as a tolerance for the two photons to hit any part of the detector wherever lies the annihilation in the patient. This allowance also integrates the time dispersions throughout the detection chains. Unfortunately, the setting is unique for any location at the detector surface, leading as a drawback in a non fully optimized selection for each coincidence configuration. Therefore, the discrimination of events yet occupying the electronics is performed inefficiently and the latter saturates when it should not have to. Dead-times are generated along the electronics chain and involve part of the information of interest to be lost.

The motivations behind our study was to evaluate the limitations in term of sensitivity of "recent" scanners and the potential improvement left when going to an architecture with performances pushed to their limits. We have considered PET scanners as a whole and started with the development of a Monte-Carlo simulator to undertake the physics of photon emission. Results that outcome from the latter were re-injected into a simple behavioral model of the entire detection chain which computes intermediate and overall dead-times as a function of the input variables. The other point we went through was the timing resolution with an investigation of respective contributions of the various components that form the acquisition system. We checked the influence of this factor on count-rate performances when lowered to its physical limit using the simulator. At last, we merged the two ideas and figured out the net overall gain that could be drawn from a dead-timeless and high timing resolution design before suggesting a synoptic of what such an electronics could be.

In the remainder of this paper, we will present a study that sticks to a given architecture of camera, namely the HR+ from CTI. A simple reason is that most of the interesting information and literature we used to cover this subject were regarding this scanner in particular. In the model of the readout system, originally devoted to this scanner, the dead-times entries were re-adjusted to account for the evolutions made in the field since this instrument has been offered for sale. We paid special care to make the comparison with an up-to-date instrument, the Accel from CPS Innovations. Both cameras belong to the same family, as Siemens subsidiaries, and share nearly the same ring physical properties. Literature about the Accel [1] and the HR+ [3] indicates that both instruments were developed with common electronics philosophy, at least for the timing discrimination. No doubt they presumably use similar working principles for the energy recovery as well. Hence, replacing initial BGO scintillator of the HR+ by LSO to match the Accel on this point too, we tuned parameters of the dead-time model so that the new HR+, yet hypothetical, finally yields count rate performances close to the Accel [2].

II. Count rate performances

As mentioned in introduction, a simulator was specifically developed to run the physics of emission of PET cameras along with the behavior of the incoming events through the acquisition system.

A. Tomograph description

The HR+ is a whole body PET scanner made up of 4 rings of 72 BGO scintillator blocks each. The latters are "scored" in 8x8 arrays so as to obtain nearly individual crystals. This design allows for light sharing by assuring that the light produced by one element optimally spread out in a well defined manner at the 4 reading Photo-Multiplier Tube (PMT) surface. To reduce the cost and the electronics processing, blocks are bundled in groups of 12 with 4 packed up axially and 3 transaxially as outlined by the sketch in figure 1. The ring diameter is of 82.7 cm which enables a patient port diameter of 56.3 cm and extends by 15.6 cm (axial FOV), whereas crystal thickness is of 3 cm for that instrument. All dimensional parameters, features and settings of the HR+ stated in this document were principally found in [3], [4] and [5].

B. Dead-time modelling

Among the often complicated Monte-Carlo simulators developed to compute dead-time in PET, Moisan et al. [5] pitched on a simple analytical approach to assess losses that occur at each stage of the HR+ electronics. The detection efficiency ϵ_d is scaled down each time to take account for losses at the subsequent stages. In Figure 2, a synoptic of the camera electronics is given with locations where dead-times appear.

When an incoming photon strikes a "pixel" of the detector with sufficient energy to rise above the threshold, an integration starts on the 3 switched integrators. During this period

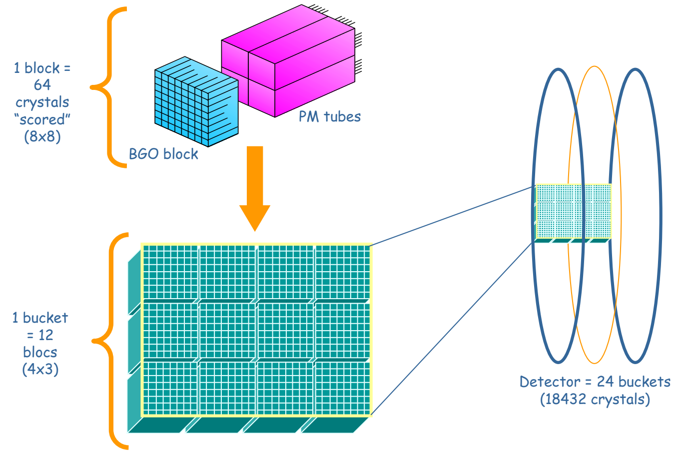

Fig. 1. Ring architecture of the HR+ camera

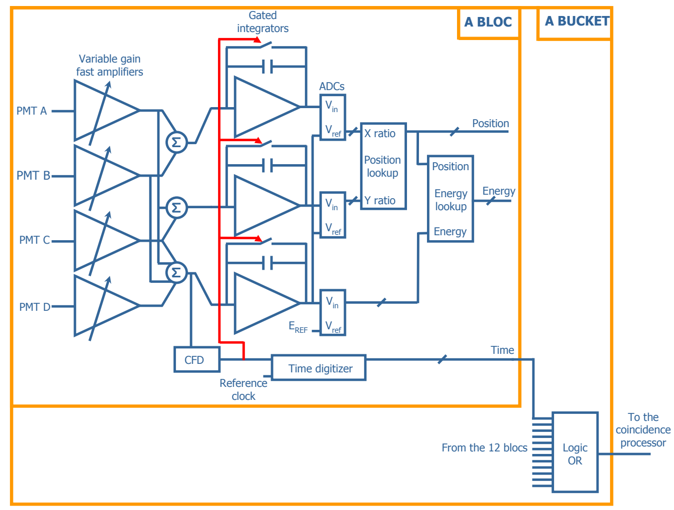

Fig. 2. Synoptic of the HR+ electronics

(τ_{block}), photons that comes at the block surface are simply ignored. It produces a dead-time, considered by lowering the detection efficiency for singles as follows (1).

$$\epsilon_d \rightarrow \epsilon_d' = \epsilon_d \times \frac{exp(-\tau_{block} N_s(\epsilon_d)/n_{block})}{1 + \tau_{block} N_s(\epsilon_d)/n_{block}} \quad (1)$$

where $N_s(\epsilon_d)$ is the remaining photon rate after the detector and n_{block} is the total number of blocks.

At the bucket side, only one event from one block can be sent to the coincidence processor, that comes at the next stage, in a given clock cycle τ_{buck} [3]. This second "bottle neck" leads inevitably in events being lost again as shown by equation (2) below.

$$\epsilon_d' \rightarrow \epsilon_d'' = \epsilon_d' \times \frac{1}{1 + \tau_{buck} N_s(\epsilon_d') n_{blbk}/n_{block}} \quad (2)$$

where n_{blbk} is the number of blocks per bucket.

At last, losses are likely to arise after the coincidence processor, namely at the acquisition system of limited bandwidth

N_{BWL}. This implies the trues, randoms and scattered to be re-scaled according to equation (3).

$$n_i'(\epsilon_d'') = n_i(\epsilon_d'')\frac{N_{BWL}}{\sum_i n_i(\epsilon_d'')} \quad if \quad \sum_i n_i(\epsilon_d'') \geq N_{BWL} \tag{3}$$

This contribution will not be regarded in the remaining of this study because solutions now exist to make up for bandwidth issues.

C. Simulator and settings

The Monte-Carlo software is constrained to the case of an annular camera, the HR+, and presumes the object to be investigated is a cylindrical phantom fully filled with active water and centered in the FOV. It generates and rules the behavior of singles, scattered, non-scattered, randoms and trues events. Encoding of the end-shields and energy discrimination were also performed to match the specifications [2] and routine clinical settings respectively [4].

The counting statistics that comes out is re-injected into the dead-time modelling just presented. Rates of Trues (T), Scattered (S) and Randoms (R) that escaped correspond to the counts effectively recorded and used for images reconstruction. From the latters, the Noise Equivalent Counting Rate (NECR) is deemed according to formula (4).

$$NECR = \frac{T^2}{T + S + 2R} \tag{4}$$

NECR has been introduced many years ago as a standard figure of merit to compare PET performances irrespectively of architectural changes that affect either the trues and randoms. This yields an assessment of the effective rate at which trues would be collected if noise were in trues alone [3]. Such a formalism find its usefulness in predicting how changes in trues, scattered and randoms proportions affect image quality. The factor 2 in equation (4) is needed in the case of the HR+ camera as it counts random events twice, one passively during the normal acquisition and a second in a delayed window so as to subtract them statistically latter on [3] [5].

III. Timing Resolution

In State of the Art tomographs with fast scintillators like LSO, timing resolution of 3ns (fwhm) can be achieved. In [1] W.W. Moses and M. Ullisch presented a study on factors influencing timing resolution in a commercial PET camera, the CPS Accel, by investigating individual contributions along a typical detection chain. The experimental setup was as described in figure 3.

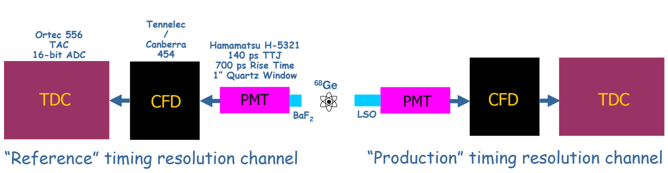

Fig. 3. Experimental setup of the timing investigation

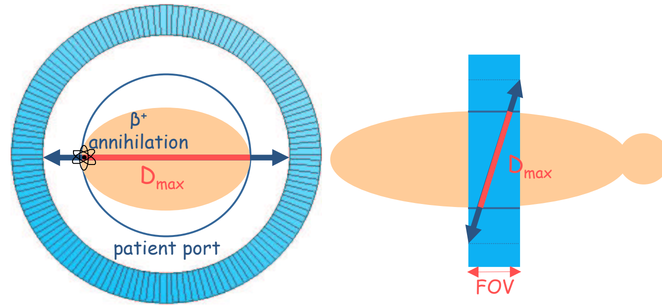

Fig. 4. Sketch of the least favorable location of emission

The channel on the left was left unchanged along the study and consisted in a sequence of very high-end electronics parts in order to serves as reference. The crystal used here is one of the fastest crystal available (BaF_2), well designed for timing. On the right hand side, the same components were used, replacing in turn only one reference element by a production one similar to those of the real camera. In this way, individual contributions were specified as shown in table I.

TABLE I
Individual contributions on timing resolution

Component	Contribution (fwhm)
LSO crystal ($3 \times 3 \times 30mm^2$)	336ps
Light sharing (block)	454ps
PMT	422ps
PMT array	274ps
CFD	1354ps
TDC	2000ps

Value for the LSO contribution was drawn from an earlier publication of W.W. Moses [6] so as to better fit with the HR+ crystals size requirement. It clearly appears that the CFD and TDC bring the main contributions and hence limit somewhat the overall resolution attainable. To be noticed, the value for the CFD is a "raw" value and would have to be increased to account for shaping traditionally in use but not implemented here.

The worst case scenario involving a positron annihilation at the far end of a patient while into contact with the instrument gantry (as described in figure 4) imposes a minimum coincidence windows of (5) :

$$\tau_{coinc_{min}} = \frac{D_{max}}{c} + \sqrt{2} \times \sqrt{\sum (contributions)^2} \tag{5}$$

A numerical application of equation (5) with the informations provided leads to a minimum coincidence window of nearly 6ns. It fits well the factory setting of the Accel.

IV. Reducing Dead-Time and Noise Through Electronics

Enhancing of the amount of interesting information in PET requires first of all to avoid losses that can be avoided. A solution for dead-times τ_{block} and τ_{buck} to be lowered would be to consider crystals individually through a full detector pixelisation. A minimum integration time would thus be allowed while leaving the surrounding crystals active. An even more powerful solution would be to use free-running electronics that constantly integrate the incoming signals and clears out count losses that might occur. Doing so would suppress the light sharing and PMT array contributions to the time resolution.

Successive events passing through the very front-end electronics so freely would require to be discriminated one from the others both in energy and time. As there would be no trigger to valid entries, conversion to digital along with algorithms would be necessary to do the job at the next stage. No entry timing means no CFD and TDC any more, hence cancelling two of the three major contributions to time accuracy. Whatever the electronics performances, the ever remaining and yet impeding geometry contribution will still be present unless TOF algorithm is employed. The latter consists in a computation of the relative times of flight of "coincident photons" from measurements of their respective time of arrival.

If we follow all these ideas where only the scintillator and the photodetector remain, an effective coincidence window can be re-deemed following equation (6).

$$\tau_{coinc_{min}} = \sqrt{2} \times \sqrt{\tau_{crystal}^2 + \tau_{PMT}^2} \qquad (6)$$

Calculation made in agreement with the latter formulae gives 750ps for the minimum effective coincidence window attainable in the present case.

V. Results and discussion

A first simulation was performed for the modified HR+ camera with a phantom of 40cm diameter and 175cm length to sham a typical patient as described in figure 5.

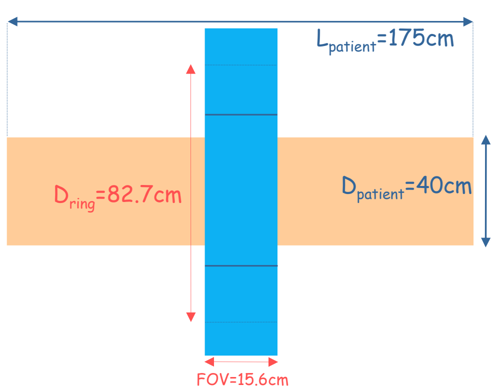

Fig. 5. Specification of the setup considered in simulations

The result of this serves as our reference in term of count rate. Our hypothetic instrument obviously accounts for the replacement of BGO by LSO, the modification of τ_{block} (90ns) and τ_{buck} (64ns) in accordance and the new coincidence window value (6ns). In a second time, the coincidence window was lowered down to the 750ps found when TOF is implemented.

In both cases, the activity "injected" to the phantom was typical of regular clinical exams, that is to say 5.5kBq/cc or 0.15μCi/ml. For the same reason, the NECR has been computed each time at that particular activity only. Indeed it seemed reasonable to us to restrict the study to this value as quite representative of everyday instrument use. The results are summarized in table (II).

TABLE II
RESULTS OF SIMULATIONS

Parameter	Series 1	Series 2
Activity	5.5kBq/cc	5.5kBq/cc
τ_{block}	90ns	-
τ_{buck}	64ns	-
τ_{coinc}	6ns	750ps
NECR	24kHz	90kHz
Gain (electronics)	1.2	-
Overall gain	1.2	3.75

Results obtained for series 1 well agree with expectations as the electronics do saturate but leaves only a 20% gap for possible improvement if compared with a perfect acquisition system, which is not much. It proves that actual tomograph electronics are not absolutely obsolete as many believe. By nature, PET instruments are subjected to enormous amount of noise that force them to saturate when they should not have to. They would not saturate without noise, but rejecting it requires high rate reading ability. Hence, there is a need to changing the philosophy by adopting a configuration capable to sustain high rates with no losses at input while discarding noisy events shortly and very efficiently. In this way, series 2 shows that the NECR increases roughly by a factor 3 just by setting a narrower coincidence window whereas it goes up to 3.75 when dead-times are suppressed too.

VI. Proposal of electronics

All the key-parameters of the future electronics architecture we plan to realize have already been emphasized and discussed previously. A synoptic of this proposal is shown in figure 6.

A free-running sampling ADC with an estimated clock frequency of 50Mhz is intended to be used in front of the trigger logic which will process "raw fast" informations to extract pulses amplitude and timing with high accuracy. Overall, the fully pipelined architecture will discard dead-times and enable in-line events selection so as to register only those of interest for later images reconstruction. For digital time recovery from samples, the incoming signal needs to be reproducible. We investigated LSO raw signal and the conclusion is that they suit well such a requirement, even before shaping.

VII. Conclusion

It has been known for long that enhancing the timing resolution of the detection chain in PET scanners significantly improves their events collection efficiency and hence the image quality that comes with. A Time of flight algorithm appears

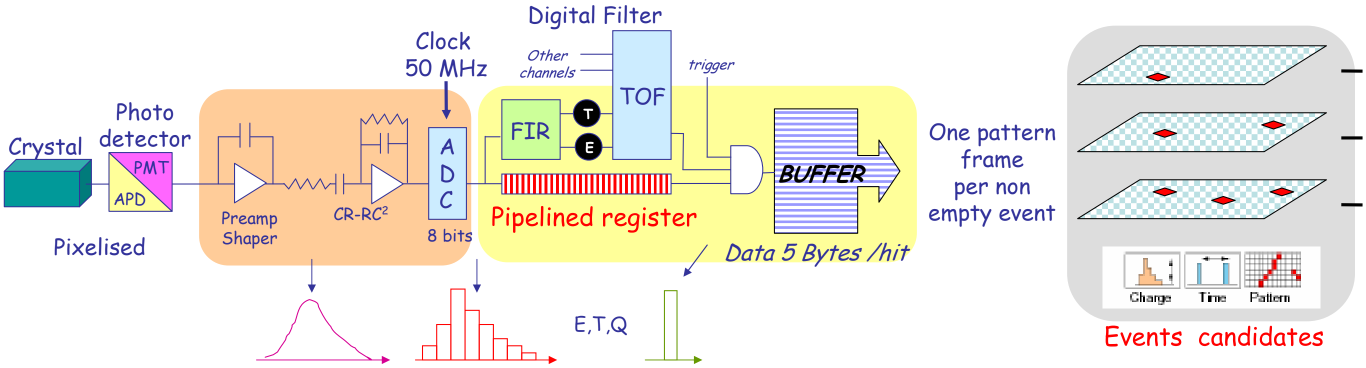

Fig. 6. Synoptic of the lossless and TOF capable electronics proposed

to be a good candidate as it would enable high noise rejection through an effective coincidence window below the nanosecond. Merging this to a fully pipelined readout architectures to prevent from losses at high rates could therefore allows for performances unattained so far. A gain of 3 seems to be reasonably expected on NECR (at 5.5kBq/cc) while keeping the present geometry. In this case, the same would go for the time required to run full-body exams ($\simeq$10 minutes) and let easily imagine PET with FOV extension of up to 70-80 cm to reduce exams time length even more ($\simeq$2 minutes). Moreover, it could open new possibilities such as real-time dosimetry by using, for example, PET in hadrontherapy.

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