First full in-beam PET measurements of 62 MeV protons onto a PMMA target

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Introduction

Positron emission tomography (PET) is a valuable technique to monitor in-situ and non-invasively the delivered dose in ion beam therapy, exploiting the beta+ activity produced in nuclear interactions along the beam path within the target volume (Figure 1) [Enghardt04, Parodi08, Attanasi11]. Due to the high beam-induced radiation flux and low statistics of annihilation photons, data are usually acquired during beam pauses or after the irradiation [Shakirin11]. The main challenge to be solved in in-beam PET applications is data acquisition also during therapeutic irradiation (full in-beam measurement).

We have implemented a dedicated PET system with faster frontend electronics, that halves the dead time of the system. For the first time, the PET system has been able to sustain the single photon count rates and acquire coincidences during the beam, in conditions of sub-clinical beam currents. A study on the paralyzation conditions and dead time losses under different beam currents is presented and the feasibility of a full in-beam PET scanner is discussed.

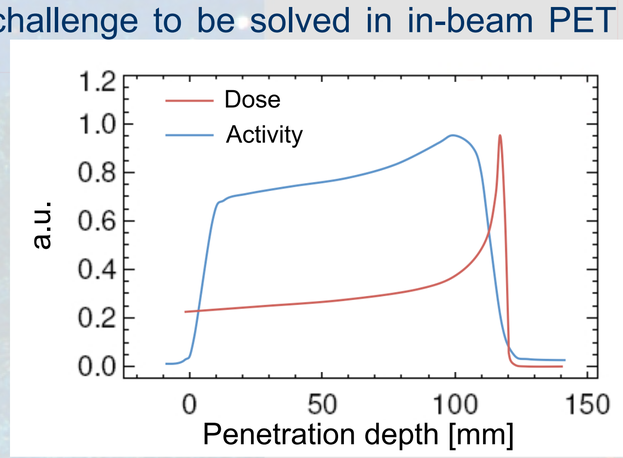

Figure 1

System description

The system is the development of the DoPET monitoring prototype [Vecchio09]. It consists of two opposite planar heads of roughly 10 cm x 10 cm mounted on a variable aperture gantry (Figure 2, a) plus a dedicated acquisition system [Sportelli11a]. Each head contains four detector front-end modules, which are tightly assembled and optically coupled to the LYSO scintillator matrices (Figure 2, b). The matrices are made of 23 x 23 pixels, with a surface area of 1.9 x 1.9 mm^2, 16 mm thick and with a 2 mm pitch. Acquisition electronics is based on short dead-time pulse discriminators, a series of peak detectors for position and energy acquisition and a synchronous coincidence processor operating at 288 MHz with a coincidence window of 5.2 ns [Sportelli11b].

Timing characteristics

Figure 3 shows the pulse amplification and discrimination electronics. Pulse discrimination dead time has been measured as the maximum interval during which CFD arm and trigger signals are activated, using the ^{176}Lu radiation background of the LYSO crystals. The obtained dead time is 129 ns (σ = 13 ns) and the intrinsic timing resolution is 1.4 ns FWHM.

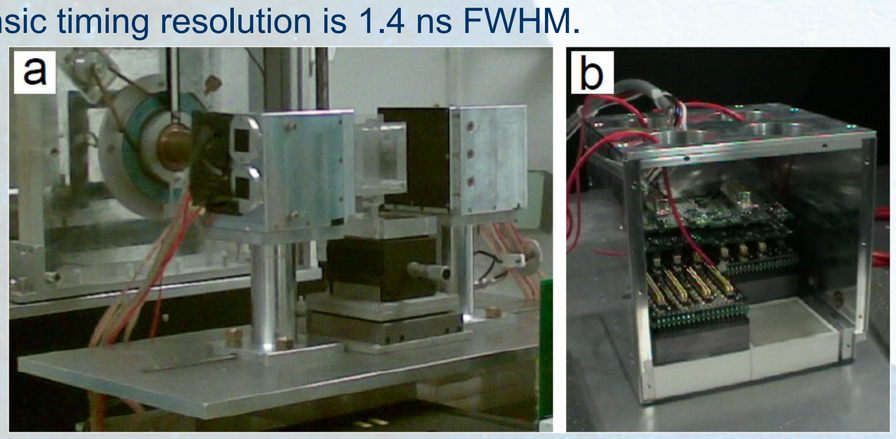

Figure 2

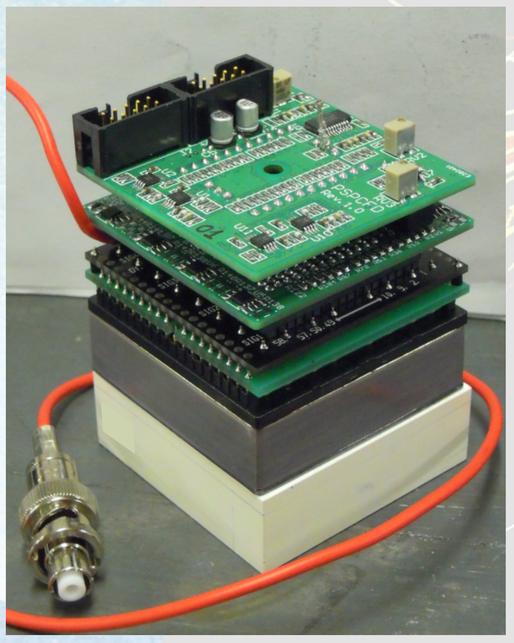

Figure 3

Experimental setup

Passively collimated proton beams of 36 mm diameter and 62 MeV initial energy were delivered at the cyclotron facility of INFN-LNS. The two heads were placed as in Figure 4, giving a geometrical efficiency of ~20% at the center. Only 4 detector modules (i.e., 2 versus 2) out of 8 have been used.

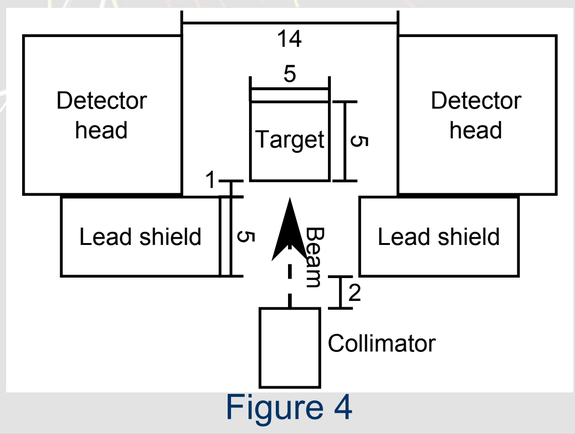

Figure 4

Results

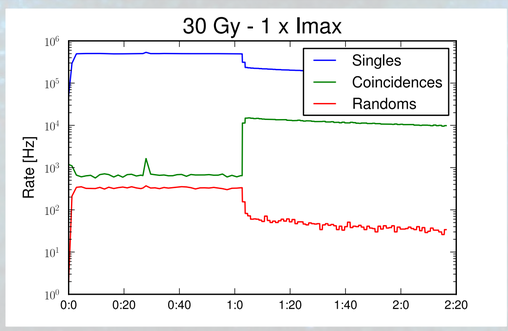

Figure 5

Beam current	Singles rate	Coinc. rate	Rand. rate	Randoms fraction
1.0	501 k	567	319	0.56
0.35	1.09 M	11.7 k	1.4 k	0.12
0.12	588 k	13.6 k	359	0.03

Table 1: Rates are in Hz, beam currents are expressed as fractions of Imax

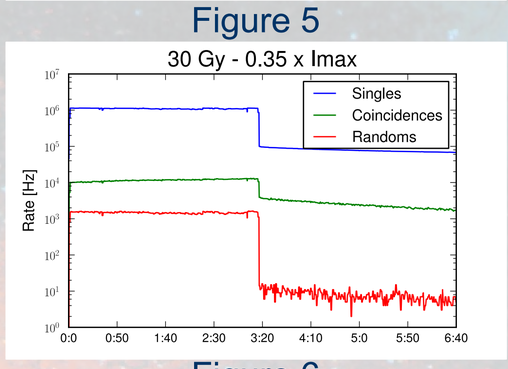

Figure 6

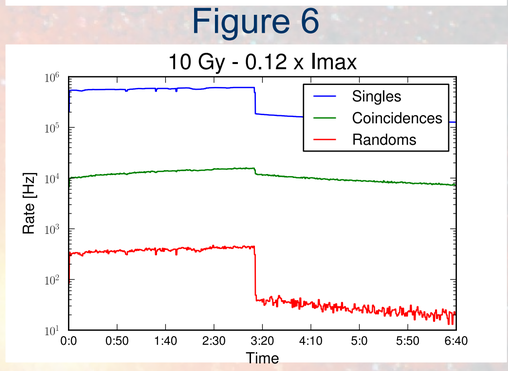

Figure 7

Beam current has been varied at various fractions of the maximum current used in clinics for ocular melanoma (Imax = 30 Gy/minute, Figures 5-7), in order to assess the effect of randoms and dead-time losses due to the high flux of prompt and background photons.

The aim has been to find under which conditions the fraction of randoms to total acquired coincidences were smaller than 0.2 as in standard clinical PET imaging.
As it is shown in Table 1, with the used system this is already achievable using currents correspondent to roughly 10 Gy/minute.

Conclusion

We have measured the response of the DoPET monitoring prototype in terms of counting characteristics with detectors 14 cm apart and beam currents of the order of those used in treating ocular melanoma. It has been shown that reducing the input count rates to roughly one third of the reference clinical current, it is possible to reduce the ratio of random counts over the total acquired coincidences to 0.12. Such a ratio is comparable to standard clinical PET cases and it is sufficiently low for successful image reconstructions.
With a more realistic gantry aperture of 30 cm, by simple geometrical considerations, we expect that the acquisition would be possible also at full currents. This is specially true for other types of tumours where lower doses and lower dose rates (in the order of 2 Gy/minute) are used, as well as for synchrotron pencil beams, i.e., raster scanning, which are less affected by collimation noise.

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