

A METHOD FOR TIME CALIBRATION OF PET SYSTEMS USING FIXED β^+ RADIOACTIVE SOURCE*

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A method for the time calibration of the Time-of-Flight Positron Emission Tomograph (TOF-PET) systems using fixed sources is described. Compared to the commercially used calibration methods, the new method gives a chance to run the calibration during the medical scan. Reduction of the time needed for calibration can increase the number of patients examined by PET. The process of calibration of the Jagiellonian Positron Emission Tomograph (J-PET) detector is shown as an example.

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1. Introduction

The Positron Emission Tomography (PET) is one of the most popular imaging techniques of the human body. During the PET scans, a positron from the β^+ emitter given to the patient, directly or after forming a positronium, annihilates with an electron from the patient, with emission of photons. Registration of produced photons allows one to reconstruct the distribution of radioisotopes in the patient's body, further interpreted as the metabolic image. The imaging of metabolism can be improved by measurement of the time difference between registration of the two photons in coincidence (Time-of-Flight (TOF)) [1]. In the case of the TOF-PET scanners, the time resolution of the detection system and its calibration is crucial.

The Jagiellonian Positron Emission Tomograph (J-PET) detector is an example of the TOF-PET system, constructed at the Jagiellonian University in Kraków, which is based on plastic scintillators and very fast electronics

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[2–8]. Analysis of the data collected by the J-PET detector is performed by the dedicated framework software [9], which allows one to incorporate specific algorithms for image reconstruction and time calibration [10, 11]. Additionally, the J-PET detector is the first detector that is able to create a unique positronium lifetime image [12, 13] and to study quantum entanglement in positronium atoms [14–16].

In this article, the new method of the time calibration for the TOF-PET scanners will be presented. Its performance will be shown based on the results from the calibration of the J-PET detector.

2. Calibration methods of TOF-PET systems

There are a couple of established time calibration methods developed for the TOF-PET systems widely used in hospitals [17, 18]. Almost all of them need additional calibration measurement to be done before the medical scan. Calibration usually focuses on creating line-of-responses (LORs) between two detectors that registered photons from the positron annihilation. Assuming that the position of annihilation is known, one can retrieve the relative delay between the two chosen detectors from the distribution of time differences between registration of photons on the reconstructed LOR [17].

One set of calibration methods uses a radioactive source placed in the fixed position in the center of the tomograph, that is situated in a large test object for imaging purposes (phantom), in which positron can annihilate. Large phantom is usually a large cylinder and ensures that one detection module can be connected by the reconstructed LORs with many other modules, like in Fig. 1 (a). In the first phase of the calibration process, one detection module is chosen as a reference and the relative delays are calculated based on the possible LORs that can be created between the chosen module and phantom used in the calibration. In the next phase, the procedure is repeated for detection modules that were calibrated in the first phase. Iterations are performed until relative delays are calculated for every detection module [18]. It is worth mentioning that there is also a method with the rotating source, which while rotating it, is mimicking large phantom [19].

On the other hand, calibration of the TOF-PET system can be performed also by using an additional detection module with point-like source instead of using large cylindrical phantom. It is used as a reference detector to which all of the other modules are calibrated. An example of this method is presented in Fig. 1 (b), where the reference external detector is rotating around the radioactive source situated in the center of the tomograph [20].

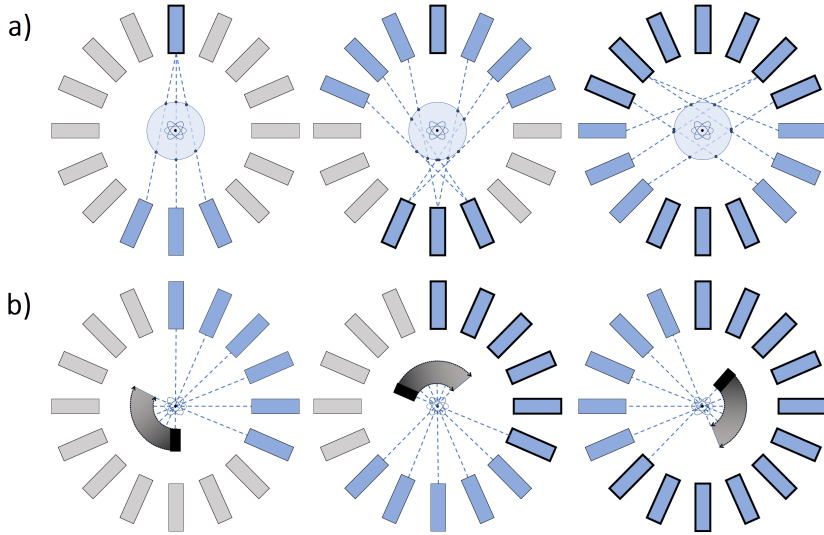


Fig. 1. Schemes of the calibration methods of the TOF-PET scanners using (a) a large phantom and (b) a rotating reference detector. From the knowledge of (a) the dimensions of the phantom and (b) the position of the reference detector, relative delays can be calculated for each detection module. Figure adapted from [17].

Line-of-responses that is used for time calibration in the aforementioned methods originates from the annihilation of a positron and an electron in the used phantom. However, there are other sources of radiation which can be used in the calibration, for example, cosmic rays, commonly treated as background in measurements on the PET systems [21]. LOR can be then created from the track of a single muon that deposited energy in two different modules. Relative delay is then calculated based on the knowledge about the position of modules that registered signals. Using cosmic rays allows to calibrate simultaneously all the modules of the TOF-PET systems, even during the medical examinations, but because of low rate of cosmic rays comparing to the activity of radioisotopes used in standard methods, this calibration method is not developed for commercial applications.

3. Calibration of the J-PET detector by fixed beta-plus radioactive source

Currently established methods of time calibration of the TOF-PET systems demand either to perform additional measurement with the specific setup, which takes time that could be used for scans of patients. It could also be run simultaneously during the PET scans but because of low cosmic

rays rate, it is a very long process, which limits the frequency of calibration. In this article, we present a method of calibration which combine the advantages of both approaches: high rate of calibration signals and potentiality to calibrate during the scan of the patient.

The new method of time calibration is focused on characteristic of β^+ emitters which can introduce additional photons that is not geometrically correlated with photons coming from the annihilation of the positron. This opens a possibility to calibrate all detection modules at once without the need for creating LORs. Additional photons come from the fact that some radioisotopes such as, for example, ^{22}Na or ^{44}Sc [22, 23], after the emission of positron from the β^+ decay transmute into nucleus in an excited state, which further deexcitate with the emission of photon. Photon coming from the annihilation of positronium and deexcitation of the nucleus can be distinguished based on the energy deposited in the detection module [24].

For the calibration measurement, the ^{22}Na source was placed in the center of the J-PET detector. Information about the position of the source is used for the calculations of the time of the photon emission. Time of the emission is defined as the time of the photon registration by a given module subtracted by a factor correlated with the distance of the source and a chosen module. In the calibration procedure, only events where detector registered one photon coming from the deexcitation of the nucleus and only one photon coming from the annihilation of the positron are considered. For selected events, the time difference distribution between emission of photon from annihilation and emission of deexcitation photon can be determined for each detection module pairs separately (Time difference = $\text{Time}_{\text{annihilation}} - \text{Time}_{\text{deexcitation}}$). Time offset between detectors can be defined as a mean of the time difference distribution.

For simplicity, one can focus on a single module for which there can be two time difference spectra produced. The first one is created from all the cases for which photon coming from the annihilation hit chosen module. The cases where the photon coming from the deexcitation of the source hits a given module will generate the second time difference distribution. Next, one can make a pair of time difference spectra for a given module, from the two cases described above. For a given pair of time difference distributions, time offsets can be calculated and compared with itself. If they are the same it means there is no difference whether we add the emission time (of annihilation photon) or whether we subtract the emission time (of deexcitation photon). If there is no significant difference, one can treat the module as calibrated. When the time offsets are not the same, miscalibration measure for a given module can be calculated as a difference of the time offsets. A correction for every module is applied and the calculation procedure is run until miscalibration measure is negligible for all of the modules. Results of

calibration on the J-PET detector are shown in Fig. 2, where one can see the measure of miscalibration as a function of a detection module identifier (ID) before calibration procedure and after it. Resolution of the time difference before calibration was estimated to 770 ps in Full Width of Half Maximum (FWHM), where after calibration resolution improved to 610 ps in FWHM.

The calibration method presented in the article uses information about the position of the radioisotope. If the position is not known, it can be obtained from the registration of the two photons coming from the annihilation of a positron and an electron. Additionally, the properties of the phantom do not influence calibration procedure, therefore, there are no contraindications in running it during the scan of the patient and, additionally, because of the high activity of a radioisotope used in the scan, the calibration process will be fast.

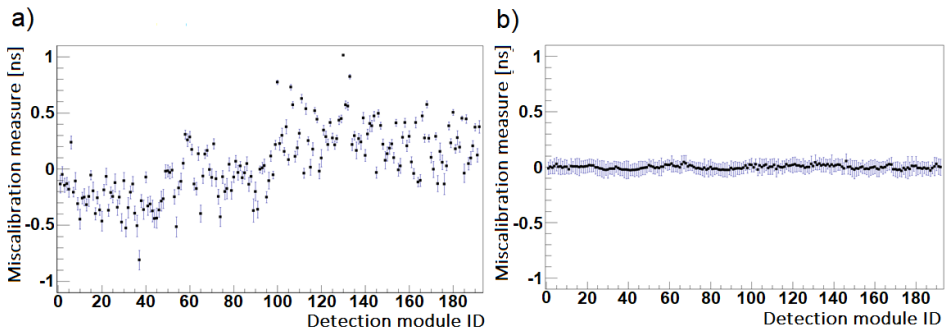


Fig. 2. Measure of miscalibration (a) before calibration procedure, (b) after calibration procedure.

4. Summary

In this article, a new method of the TOF-PET scanners time calibration is introduced. It is based on the use of the time correlation between annihilation and deexcitation photons which are angularly uncorrelated. The presented method requires usage of a specific class of the radioisotopes that introduce deexcitation photon into the measured setup, which is a reference to the time of the positron annihilation. The new approach makes the calibration fast and allows to perform it simultaneously during the scan of the patient which makes the whole process easier. Calibration was tested by calibrating detection modules of the J-PET detector.

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REFERENCES

- [1] S. Vandenberghe *et al.*, *EJNMMI Physics* **3**, 3 (2016).
- [2] P. Moskal *et al.*, *Nucl. Instrum. Methods Phys. Res. A* **764**, 317 (2014).
- [3] P. Moskal *et al.*, *Nucl. Instrum. Methods Phys. Res. A* **775**, 54 (2015).
- [4] P. Moskal *et al.*, *Phys. Med. Biol.* **61**, 2025 (2016).
- [5] S. Niedźwiecki *et al.*, *Acta Phys. Pol. B* **48**, 1567 (2017).
- [6] P. Kowalski *et al.*, *Phys. Med. Biol.* **63**, 165008 (2018).
- [7] M. Pałka *et al.*, *JINST* **12**, P08001 (2017).
- [8] G. Korcyl *et al.*, *IEEE Trans. Med. Imag.* **37**, 11 (2018).
- [9] W. Krzemień *et al.*, *Acta Phys. Pol. A* **127**, 1491 (2015).
- [10] P. Moskal *et al.*, *Acta Phys. Pol. B* **47**, 509 (2016).
- [11] P. Moskal *et al.*, *Eur. Phys. J. C* **78**, 970 (2018).
- [12] P. Moskal *et al.*, *Phys. Med. Biol.* **64**, 055017 (2019).
- [13] P. Moskal, B. Jasińska, E.Ł. Stepień, S.D. Bass, *Nature Rev. Phys.* **1**, 527 (2019).
- [14] B.C. Hiesmayr, P. Moskal, *Sci. Rep.* **7**, 15349 (2019).
- [15] B.C. Hiesmayr, P. Moskal, *Sci. Rep.* **9**, 8166 (2019).
- [16] M. Nowakowski, D. Bedoya Fierro, *Acta Phys. Pol. B* **48**, 1955 (2017).
- [17] W.W. Moses, C.J. Thompson, *IEEE Trans. Nucl. Sci.* **53**, 5 (2006).
- [18] Xiaoli Li *et al.*, *IEEE Trans. Nucl. Sci.* **63**, 3 (2016).
- [19] A.E. Perkins *et al.*, *IEEE Nucl. Sci. Symp. Conf. Rec.* **5**, 2488 (2005).
- [20] M. Skurzok *et al.*, *Acta Phys. Pol. A* **132**, 5 (2017).
- [21] M. Silarski *et al.*, *Bio-Algorith. Med-Syst.* **10**, 19 (2014).
- [22] A. Dash, R. Chakravarty, *Am. J. Nucl. Med. Mol. Imag.* **9**, 30 (2019)
<http://www.ajnmml.us/files/ajnmml0086228.pdf>
- [23] M. Sitarz, J. Cussonneau, T. Matulewicz, F. Haddad, *Appl. Radiat. Isotopes* **155**, 108898 (2020).
- [24] P. Moskal *et al.*, *Phys. Med. Biol.* **64**, 055017 (2019).