IL NUOVO CIMENTO **49 C** (2026) 92
DOI 10.1393/ncc/i2026-26092-5

Colloquia: FATA 2025

High-performance modular TOF-PET imager

R. Pestotnik(1), D. Gascon(4), A. Gola(7), J. M. Benlloch Baviera(8), D. Sanchez Gonzalo(5), R. Ballabriga Sune(10), W. Weber(9), G. Elfakhri(4), J. Barberá Ballester(5), G. Bergamin(10), J. V. Catret Mascarell(4), R. Dolenec(2)(1), J. M. Fernández-Tenllado(4), L. Garrido(4), S. Gómez(6)(4), E. Grauges(4), K. Grogg(4), S. Korpar(3)(1), P. Križan(2)(1), M. Leban(1), S. Majewski(1), T. Marin(4), J. Mauricio(4), D. Mazzanti(4), S. Merzi(7), L. Moliner Martinez(8), Y. Najmaoui(4), S. Nekolla(9), M. Orehar(2), G. Paternoster(7), N. Pavon Hernandez(8), M. Penna(7), S. Portero(4), I. Rauscher(9), G. Razdevšek(1), M. Ribalda(4), A. Sanuy(4), A. Seljak(1), A. Studen(2)(1), O. de la Torre Perez(4), C. A. Trippl(7), M. Viqueira Rodriguez(10), D. Žontar(1), P. Kachru(7), I. Hany(7), G. Palù(7) and L. Parellada Monreal(7)

(1) *Jožef Stefan Institute - Ljubljana, Slovenia*
(2) *Faculty of Mathematics and Physics, University of Ljubljana - Ljubljana, Slovenia*
(3) *Faculty of Chemistry and Chemical Engineering, University of Maribor - Maribor, Slovenia*
(4) *Dept. Física Quàntica i Astrofísica, Institut de Ciències del Cosmos (ICCUB), University of Barcelona (IEEC-UB) - Barcelona, Spain*
(5) *Oncovision - Valencia, Spain*
(6) *Serra Húnter Programme, Polytechnic University of Catalonia (UPC) - Barcelona, Spain*
(7) *Fondazione Bruno Kessler - Povo, Trento, Italy*
(8) *Instituto de Instrumentación para Imagen Molecular (CSIC-UPV) - Valencia, Spain*
(9) *Technical University of Munich (TUM-MED) - Munich, Germany*
(10) *CERN - Geneva, Switzerland*

received 12 March 2026

Summary. — Positron Emission Tomography (PET) is a leading functional imaging modality, yet its global accessibility remains limited due to high cost and infrastructure demands. Recent advances in time-of-flight (TOF) technology enable coincidence time resolutions (CTR) below 100 ps, opening the possibility for radically simplified scanner geometries. We present the concept and development status of a modular, limited-angle TOF-PET imager based on ultrafast flat-panel detectors. The approach leverages sub-100 ps timing performance, improved silicon photomultipliers (SiPMs), and next-generation ASIC readout to compensate for sparseangular coverage while maintaining image quality. Monte Carlo simulations and reconstruction studies demonstrate that a two-panel system can achieve image quality comparable to state-of-the-art clinical scanners with significantly reduced material and cost.

1. – Introduction and clinical motivation

The global burden of cancer continues to rise at an alarming rate. GLOBOCAN 2022 reports nearly 20 million new cancer cases annually, with projections exceeding 35 million by 2050 [1-3]. In parallel, neurodegenerative diseases represent one of the fastest growing causes of disability worldwide. Dementia currently affects over 55 million people globally and is projected to exceed 150 million cases by 2050 due to demographic aging [4-6].

The urgency of strengthening diagnostic infrastructure has recently been emphasized at the global policy level [7, 8]. Functional imaging modalities such as positron emission tomography (PET) are essential not only for oncology, but also for early detection of neurodegenerative diseases, theranostics, and precision medicine.

Despite its clinical impact, PET systems remain expensive and infrastructure-heavy, limiting accessibility worldwide. This motivates the development of scalable, cost-efficient, high-performance imaging architectures.

2. – Timing as the fundamental performance driver

Time-of-flight (TOF) PET improves the image signal-to-noise ratio (SNR) by constraining the annihilation position along the line-of-response (LOR) (fig. 1). The spatial uncertainty introduced by finite coincidence time resolution (CTR) is given by

$$\Delta x = \frac{c_0 \Delta t}{2}, \tag{1}$$

where Δt denotes the CTR and c_0 the speed of light. This relation directly links timing precision to spatial localization.

For the current clinical benchmark of $\Delta t \approx 200\,\text{ps}$, as achieved by Siemens Biograph Vision [9, 10], the localization uncertainty is approximately $\Delta x \approx 3.2\,\text{cm}$. Reducing CTR to 70 ps improves localization to $\Delta x \approx 1\,\text{cm}$, while 50 ps approaches sub-centimeter precision. Such performance represents a qualitative shift in PET reconstruction.

The TOF information improves the image quality by increasing the effective signal-to-noise ratio. In a simplified approximation, the TOF gain in sensitivity scales as

$$\text{SNR}_{\text{TOF}} \propto \sqrt{\frac{D}{\Delta x}}, \tag{2}$$

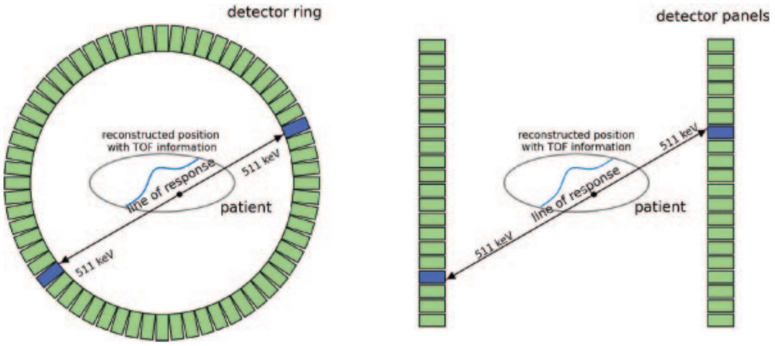

Fig. 1. – Time of flight and limited angle coverage with panel detectors.

where D is the diameter of the object [11]. Thus, improved timing is particularly beneficial for large objects, such as torso or total-body imaging.

Laboratory-scale measurements have demonstrated the physical feasibility of extreme timing performance. Gundacker *et al.* reported CTR values of 58–59 ps FWHM using small LSO/LYSO crystals and high-bandwidth readout electronics [12]. Although such setups are not scalable to full clinical systems, they demonstrate that fundamental detector physics supports sub-100 ps performance.

The ultimate objective of the community, outlined in the "10 ps challenge" roadmap [13], is to approach timing limits where the LOR localization becomes comparable to intrinsic detector dimensions. In this regime, the performance of the PET system becomes dominated by timing rather than angular sampling.

Conventional PET design relies on full 360° detector coverage to suppress reconstruction artefacts. However, limited-angle systems can recover distortion-free images if sufficient TOF information is available. Surti and Karp showed that angular sampling requirements decrease as timing resolution improves [11]. More recently, Razdevšek *et al.* demonstrated via Monte Carlo simulation that CTR values in the 50–70 ps range significantly suppress limited-angle artefacts in panel-based geometries [14].

This observation marks a fundamental shift in PET system design. At 300–500 ps, PET remains geometry-dominated and requires full angular coverage for stable reconstruction. Around 200 ps, TOF improves contrast and noise performance but does not relax geometric constraints. When timing approaches 100 ps, however, localization along the line of response becomes sufficiently precise to compensate for sparse angular sampling. In this regime, PET transitions from a geometry-limited to a timing-dominated modality, with temporal resolution becoming the primary determinant of system performance.

Therefore, timing is not merely an incremental performance parameter, but the primary enabler of simplified modular architectures. Sub-100 ps CTR allows the transition from heavy, cost-intensive ring geometries toward open, limited-angle panel configurations without compromising image quality.

3. – Flat-panel and dedicated PET systems

Limited-angle PET configurations have been investigated primarily for organ-dedicated imaging, where detector proximity compensates for incomplete angular coverage.

Double-panel geometries were demonstrated by Gonzalez-Montoro *et al.* in a proof-of-concept TOF-PET system [15]. The design employed opposing detector heads optimized for dedicated applications, showing that restricted angular sampling can achieve clinically useful image quality when the geometry is tailored to the anatomical target. Similarly, Stiles *et al.* reported a high-sensitivity organ-targeted PET camera [16], demonstrating that proximity-driven sensitivity gain can partially offset reduced solid-angle coverage. Dedicated prostate PET systems have been reviewed in [17], highlighting the advantages of compact detector configurations for improved spatial resolution and lesion detectability.

These systems share several defining characteristics: detector coverage is reduced compared to full-ring PET configurations, the detectors are positioned in close proximity to the region of interest to compensate for limited angular sampling, and the overall design is optimized for specific anatomical targets rather than whole-body imaging.

However, most of these architectures operate in moderate timing regimes (typically 200 ps CTR), where TOF improves contrast but does not fundamentally compensate for missing angular information. Consequently, reconstruction stability in limited-angle configurations remains dependent on application-specific constraints and careful system geometry optimization.

In parallel, the development of long axial field-of-view (LAFOV) and total-body PET systems have demonstrated that increased sensitivity dramatically enhances image quality and quantitative accuracy. While these systems rely on extensive detector coverage rather than angular reduction, they underscore the central importance of sensitivity as a system-level performance metric.

More recently, Razdevšek *et al.* investigated panel-based geometries in the context of extreme timing performance [14]. Their simulations indicate that, as CTR approaches 100 ps, limited-angle artefacts are significantly suppressed. In this regime, TOF information constrains the annihilation position along the LOR sufficiently to reduce the need for full 360° angular sampling.

This marks a conceptual departure from earlier panel-based systems. Whereas previous designs relied primarily on geometric proximity to compensate for limited coverage, ultrafast TOF enables compensation through temporal localization.

4. – Concept of the Modular Ultrafast Flat-Panel PET

The key distinction of the present work is therefore not merely the use of flat panels, but the combination of sub-100 ps coincidence timing resolution, modular scalability, and reconstruction strategies optimized for sparse sampling. By operating in a timing-dominated regime, panel-based PET transitions from a dedicated or niche solution toward a scalable architectural alternative to ring-based systems. Thus, flat-panel PET should not be viewed solely as a specialized organ-dedicated modality, but as a generalizable platform enabled by extreme timing performance.

Building on advances in extreme time-of-flight (TOF) performance, we propose a modular PET architecture based on opposing flat-panel detector modules with adjustable separation (figs. 2 and 3). The central hypothesis is that sub-100- ps coincidence timing resolution compensates for sparse angular coverage.

Conventional PET systems are based on full-ring geometries, providing 360° angular sampling. Such configurations ensure reconstruction stability, but require large quantities of scintillator material, complex mechanical integration, and high system cost.

In contrast, the proposed architecture replaces the ring with two or more planar detector modules arranged in opposition. The system operates in a limited-angle configuration, where reconstruction stability is recovered through temporal localization rather than geometric completeness. When CTR approaches 100 ps, the spatial uncertainty along the LOR becomes sufficiently small to significantly constrain the annihilation position. In this regime, angular sampling requirements are relaxed, enabling open geometries without severe artefacts.

We propose that each panel consists of pixelated scintillator arrays optically coupled to high-performance SiPM matrices and fast multi-channel ASIC readout. The detector modules are conceived as scalable building blocks: independent readout and synchronization, mechanically stackable in the axial direction, configurable in separation distance, and expandable toward multi-panel or total-body systems. The modular approach enables adaptation to different clinical scenarios: head imaging, organ-targeted imaging, intraoperative guidance, and eventually total-body PET.

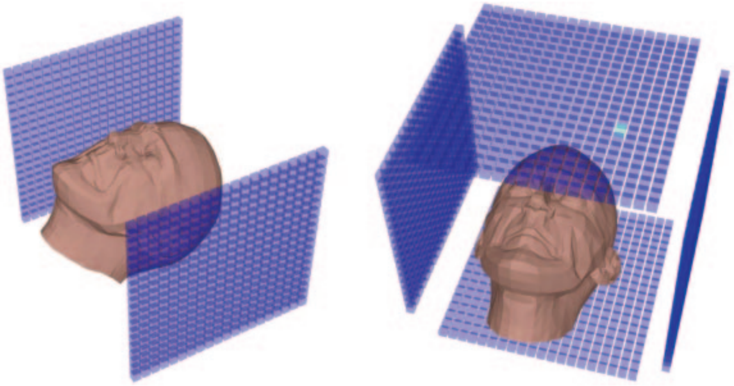

Fig. 2. – PET Imaging with 2 and 4 panels.

Limited-angle configurations are sensitive to parallax errors due to oblique photon incidence. To mitigate this, we foresee incorporating depth-of-interaction (DOI) capability through dual-sided readout in the continuation of the project. DOI information reduces the timing spread due to photon propagation and improves spatial resolution at larger panel separations. Combined with extreme TOF performance, this further stabilizes reconstruction in sparse geometries.

The modular flat-panel concept is inherently scalable. By increasing panel height and stacking modules axially, the system can transition from organ-scale to total-body configurations. Unlike conventional total-body PET, which scales by extending ring length, the proposed architecture scales by extending panel dimensions and optionally adding additional opposing modules.

Such a modular approach provides reduced scintillator usage, simplified mechanical design, flexible field-of-view, and potential cost reduction of 4–5 times compared to full-ring systems. Thus, sub-100 ps timing enables a new class of PET architectures that prioritize modularity, scalability, and accessibility without sacrificing clinical-grade image quality.

5. – Simulation framework

System performance was evaluated using detailed Monte Carlo simulations implemented in the Geant4/GATE framework, followed by image reconstruction with the

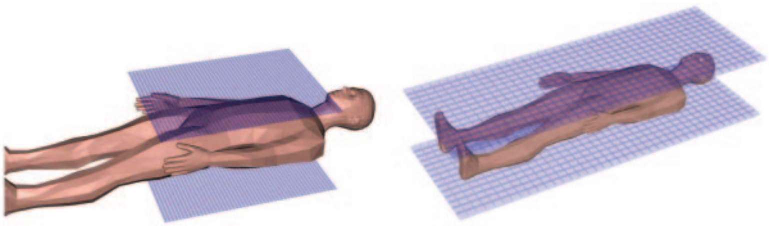

Fig. 3. – Concept of two opposing total body flat-panel detectors with adjustable separation: large field of view and total body configuration.

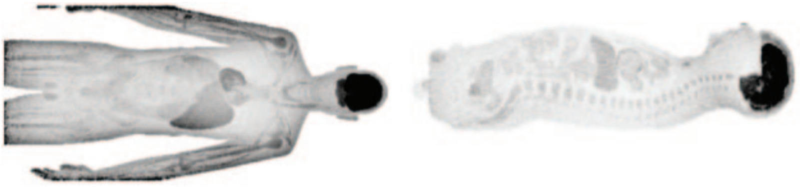

Fig. 4. – Reconstructed image of an XCAT phantom.

CASToR platform using an MLEM algorithm. Digital phantoms, including anatomically realistic XCAT models, were reconstructed for coincidence timing resolutions of 214 ps, 100 ps, 70 ps, and 50 ps to quantify the impact of timing performance on limited-angle imaging (fig. 4).

The simulations indicate that CTR values around 70 ps already significantly suppress limited-angle artifacts and stabilize image reconstruction. Incorporating DOI information further mitigates parallax effects associated with oblique photon incidence in panel geometries. Under these conditions, a two-panel system approaches the image quality of contemporary clinical scanners while using approximately four times less scintillator material.

6. – Detector technology enablers

Effective sensitivity can be approximated as

$$S \propto \eta_{\mathrm{det}} \cdot \eta_{\mathrm{geom}} \cdot \frac{1}{\Delta t}. \tag{3}$$

Improved timing directly increases effective sensitivity, allowing reduced angular coverage without compromising image quality. In addition, scaling the panel dimensions to $120 \times 60\,\mathrm{cm}^2$ enables total-body configurations with 4-minute scans that approach clinical reference performance.

Realizing scalable sub-100 ps coincidence time resolution (CTR) requires coordinated optimization of the photodetector, front-end electronics, and module-level integration (fig. 5). Although scintillator performance establishes fundamental photon statistics, the dominant leverage for practical system improvement lies in SiPM development, electronic bandwidth, and interconnection architecture.

Silicon photomultipliers (SiPMs) are the primary determinant of single-photon time resolution (SPTR). Recent advances in deep trench isolation and metal-filled trench technologies have significantly reduced optical crosstalk, allowing operation at higher overvoltage and thereby improving avalanche build-up timing [18,19]. Next-generation NUV-sensitive devices achieve SPTR values approaching 70 ps per detected photon, which enables multi-photon coincidence timing in the 50–70 ps regime when coupled to optimized LYSO crystals. Reducing microcell capacitance, improving fill factor, and stabilizing temperature-dependent gain variations are critical for maintaining uniform timing performance across large detector arrays.

Front-end electronics must preserve this intrinsic detector performance. The FastIC and FastIC+ ASIC architectures integrate high-bandwidth analog front ends with 25 ps bin time-to-digital converters (TDCs), ensuring that electronic jitter remains well below

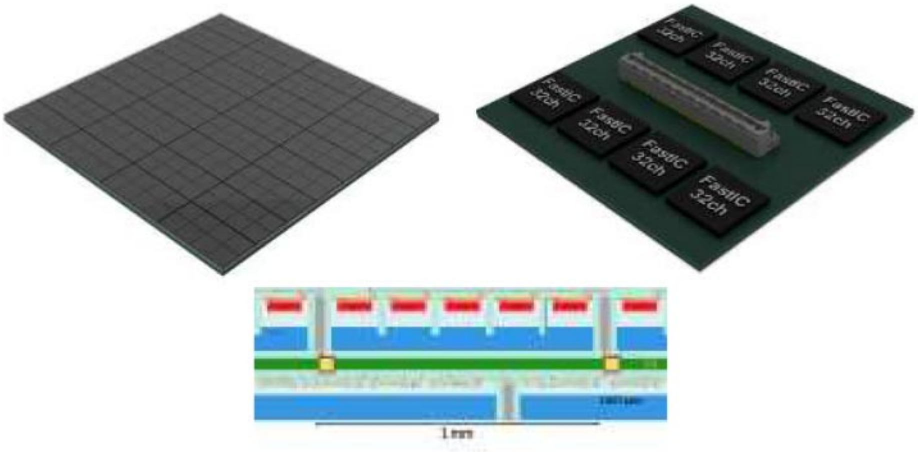

Fig. 5. – Photon detector module comprises an interposer with the silicon photomultipliers on one side and readout ASIC chips with a communication and power supply connector on the other.

the target CTR. Low input capacitance, fast discrimination, and stable threshold control are essential to avoid degrading the earliest photon timing. Next-generation ASIC developments aim to reduce the electronic contribution to below 20 ps, ensuring that system performance remains limited by photon statistics rather than readout noise.

Equally important is the integration strategy at the detector module level. High-density interconnects between SiPM arrays and ASIC electronics must minimize parasitic capacitance and signal distortion. Hybrid 2.5D integration approaches, including micro-bump bonding and through-silicon vias, reduce interconnect length and improve bandwidth, while enabling fine pixel segmentation. Such architectures are essential for translating laboratory-scale timing performance into scalable, mechanically robust flat-panel modules.

System-level synchronization and calibration are integral to preserving timing precision across spatially separated panels. Low-jitter clock distribution, deterministic latency links, and temperature compensation strategies ensure picosecond-level stability in multi-module configurations.

By combining optimized SiPM devices, fast low-jitter ASIC electronics, and high-density integrated detector modules, the proposed architecture targets scalable system-level CTR in the 50–70 ps range. Crucially, the objective is not record laboratory timing, but stable sub-100 ps performance within a modular platform suitable for clinical translation.

7. – Impact and outlook

The proposed modular ultrafast flat-panel PET architecture has implications that extend beyond incremental detector improvement. By transitioning from geometry-dominated to timing-dominated system design, the work enables a fundamentally new class of scalable PET platforms. Conventional PET systems rely on full 360° detector coverage to ensure reconstruction stability, which inherently drives material usage, mechanical complexity, and overall system cost. In contrast, when coincidence time resolution approaches the 50–70 ps regime, sparse angular coverage can be compensated through temporal localization along the line of response. This decouples image quality from rigid ring geometries and opens the possibility of open, reconfigurable, and modular detector layouts.

From an architectural perspective, such timing performance enables a substantial reduction in scintillator material while maintaining clinical-grade image quality. The proposed flat-panel approach replaces heavy ring structures with mechanically simpler, stackable modules. These modules can be arranged with adjustable separation and extended axially toward larger fields of view. As a consequence, the system becomes adaptable to multiple clinical scenarios, ranging from organ-targeted imaging and intraoperative applications to extended axial and total-body configurations. The architecture therefore provides a pathway toward flexible PET platforms rather than fixed-purpose systems.

The societal and clinical implications are significant. Cancer incidence is projected to exceed 35 million new cases annually by 2050, while dementia prevalence may surpass 150 million cases worldwide [1,4]. Expanding access to molecular imaging is therefore not only a technological challenge but a global healthcare priority. Reducing detector material requirements and simplifying system integration may lower capital costs and facilitate installation in smaller hospitals and resource-constrained regions. In parallel, sub-100 ps timing enhances lesion detectability, contrast recovery, and quantitative precision, all of which are critical for theranostics and precision oncology workflows.

From a detector-physics standpoint, the work advances the practical realization of the 10 ps roadmap [13] by translating laboratory-scale timing achievements into scalable detector modules. The integration of optimized scintillators, low-crosstalk SiPMs, 2.5D interconnection strategies, and fast ASIC electronics constitutes a coherent approach toward system-level sub-100 ps performance. Equally important, the architecture provides an experimental framework for studying the interplay between timing resolution, angular sampling density, and reconstruction conditioning, which remains a fundamental question in limited-angle tomography.

The technological maturation of the concept follows a staged trajectory. The initial phase focuses on component-level optimization, including scintillator screening, SiPM evaluation, and validation of FastIC-class ASIC electronics, with the objective of demonstrating stable sub-70 ps CTR under laboratory conditions. Subsequently, detector modules incorporating 2.5D integration and deterministic synchronization will be assembled into scalable panel units. This stage will validate inter-channel timing stability, depth-of-interaction strategies, and multi-module calibration. The next milestone is the realization of a two-panel system demonstrator, enabling phantom imaging and quantitative benchmarking against clinical reference systems. Finally, axial stacking and field-of-view extension will explore the transition toward extended and total-body geometries, together with industrialization pathways.

In the long term, if scalable 50–70 ps performance is achieved, PET system design may undergo a structural shift comparable to the introduction of TOF technology two decades ago. Instead of relying primarily on geometric completeness, future systems may prioritize timing precision and modularity. Such a transition would support compact, reconfigurable PET platforms capable of deployment across a broader range of clinical environments.

The ultrafast modular flat-panel concept thus represents not merely a detector upgrade, but a step toward democratizing high-performance molecular imaging while maintaining scientific rigor and clinical relevance.

8. – Conclusion

Ultrafast TOF performance redefines PET system architecture. When CTR approaches 50–70 ps, timing precision compensates sparse angular sampling, enabling simplified modular geometries without sacrificing clinical image quality.

This work integrates advances in detector physics, SiPM technology, and system-level simulation to propose a scalable, accessible, next-generation PET platform.

* * *

This project has received funding from the European Union's Horizon Europe research and innovation programme under grant agreement No. 10109989 (European Innovation Council Pathfinder Open project PetVision), the Slovenian Research and Innovation Agency research projects No. J7-50229, BI-US/22-24-165, and research programme No. P1-0135.

REFERENCES

[1] Bray F. *et al.*, *CA Cancer J. Clin.*, **74** (2024) 229.

[2] Ferlay J. *et al.*, *Global Cancer Observatory: Cancer Tomorrow (version 1.1)* (International Agency for Research on Cancer, Lyon, France) 2024, available from `https://gco.iarc.who.int/tomorrow` (accessed 17 February 2026).

[3] Bray F. and Møller B., *Nat. Rev. Cancer*, **6** (2006) 63.

[4] Nichols E., Steinmetz J., Vollset S. *et al.*, *Lancet Public Health*, **7** (2022) e105.

[5] World Health Organization, *Dementia: Number of people affected to triple in next 30 years*, WHO News Release (2017).

[6] Xu L. *et al.*, *J. Glob. Health*, **15** (2025) 04156.

[7] World Health Organization, *Strengthening medical imaging capacity*, World Health Assembly Resolution, Geneva (2025).

[8] Hricak H. *et al.*, *Lancet Oncol.*, **26** (2025) 7.

[9] Prenosil G. A. *et al.*, *J. Nucl. Med.*, **63** (2022) 476.

[10] Lan X. *et al.*, *Eur. J. Nucl. Med. Mol. Imaging*, **48** (2021) 3755.

[11] Surti S. and Karp J. S., *Phys. Med.*, **32** (2016) 12.

[12] Gundacker S. *et al.*, *Phys. Med. Biol.*, **65** (2020) 025001.

[13] Lecoq P., *Phys. Med. Biol.*, **65** (2020) 21RM01.

[14] Razdevšek G. *et al.*, *Med. Phys.*, **52** (2025) 2845.

[15] Gonzalez-Montoro A. *et al.*, *EJNMMI Phys.*, **11** (2024) 73.

[16] Stiles J. *et al.*, *Sensors (Basel)*, **22** (2022) 4678.

[17] Gonzalez A.J. *et al.*, *PET Clin.*, **19** (2024) 49.

[18] Najmaoui N. *et al.*, *IEEE Trans. Radiat. Plasma Med. Sci.*, `https://doi.org/10.1109/TRPMS.2025.3619872` (2025).

[19] Mariscal-Castilla A. *et al.*, *IEEE Trans. Radiat. Plasma Med. Sci.*, **8** (2024) 718.