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Silicon and Silicon Carbide Technologies for Nuclear Medicine: Performance, AI-Augmented Imaging, and Future Prospects

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Abstract

Silicon-based technologies have become central to modern nuclear medicine, owing to their superior spatial resolution, mature microfabrication infrastructure, compact form factor, and seamless compatibility with advanced readout electronics. In parallel, silicon carbide (SiC) has emerged as a promising wide-bandgap semiconductor for radiation detection and dosimetry in demanding environments, particularly where radiation hardness, thermal stability, and high-temperature operability are required. This study examines the physical basis of silicon detectors, their operation principles, and their applications in nuclear medicine imaging, dosimetry, and medical electronics, with emphasis on positron emission tomography (PET), single-photon emission computed tomography (SPECT), and hybrid imaging systems. The article also compares silicon and SiC in terms of bandgap, charge transport, radiation tolerance, breakdown strength, energy resolution, and fabrication maturity. In addition, the role of silicon technologies in the development of digital medicine and artificial intelligence-assisted imaging is discussed, along with the current progress of nuclear medicine infrastructure and initiatives in Saudi Arabia. The study concludes that silicon remains a mature and highly effective detector material for high-resolution nuclear medicine applications, while SiC offers a strong future pathway for next-generation dosimetry and radiation-hard detector systems.

Keywords: Silicon detectors; Nuclear medicine; Radiation dosimetry; Silicon carbide detectors.

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1. Introduction

Silicon (Si, atomic number 14) is the second most abundant element in the Earth's crust and the seventh most plentiful in the universe, occurring naturally as silicon dioxide (SiO_2) in sand, silicate rocks, and clay minerals rather than in elemental form [1,2]. It was first isolated in 1824 by the Swedish chemist Jöns Jacob Berzelius, who produced it by reducing potassium fluorosilicate with potassium metal and carefully washing the resulting product [3]. Industrial production today involves heating sand (SiO_2) with carbon at temperatures approaching 2200°C, followed by further purification to yield electronic-grade material [1,5]. At room temperature, silicon exists in two allotropic forms: crystalline silicon, which exhibits a grayish hue with a metallic sheen and a diamond cubic crystal structure, and amorphous silicon, which appears as a brown powder. High-purity single crystals of the crystalline form are grown by the Czochralski technique, producing the wafers that underpin virtually all modern solid-state devices. When doped with trivalent elements such as boron or gallium, or with pentavalent elements such as phosphorus or arsenic, silicon acquires the controlled electrical conductivity required for transistors, solar cells, rectifiers, and microchips [1,5]. Beyond microelectronics, silicone-based materials - organosilicon polymers derived from silicon- have been used in biomedical applications for over sixty years, having entered commercial markets in 1946 [3,4]. Their combination of hydrophobicity, chemical and thermal stability, low surface tension, inherent biocompatibility, and bio-durability makes them suitable across a wide range of medical device, pharmaceutical, and personal-care applications, in forms ranging from gels, fluids, and adhesives to elastomers for lubricating, bonding, and damping [4]. In the context of nuclear medicine and radiation physics, silicon's most consequential property is its capacity to generate electron–hole pairs when traversed by ionising radiation, requiring only approximately 3.6 eV per pair - far less than the ~15 eV needed in gas-filled detectors and orders of magnitude less than scintillator systems -which translates directly into superior energy resolution and signal yield [1,6]. Combined with fast carrier mobility, compatibility with CMOS microfabrication, and the availability of ultra-pure crystals, silicon has become the material of choice for radiation detectors, medical imaging electronics, and dosimetry devices in modern nuclear medicine [7,8]. This study examines these roles systematically, from fundamental material properties through clinical applications to the evolving landscape of silicon-based nuclear medicine in the Kingdom of Saudi Arabia. The present work offers three original contributions: a systematic side-by-side comparison of Si and SiC detector characteristics grounded in published experimental data; a hypothetical quantitative simulation of detector signal yield and energy resolution under controlled conditions; and an integrated assessment of silicon-based nuclear medicine technologies within the context of Saudi Arabia's Vision 2030 national health strategy.

2. Physical and Radiological Properties of Silicon

Silicon (atomic number Z = 14, atomic mass ≈ 28.085 u) is a group-IV metalloid with a face-centered cubic (diamond cubic) crystal structure. The four valence electrons form covalent bonds with four neighboring atoms, giving silicon its characteristic semiconductor behavior [5]. As illustrated in Figure 1, the three categories of solids- metals, semiconductors, and insulators are distinguished by their electronic band structures: metals exhibit overlapping valence and conduction bands allowing free electron flow, whereas insulators possess a wide bandgap (E_g) that prevents thermal excitation of electrons under normal conditions. Silicon occupies an intermediate position as a semiconductor, with a moderate bandgap of 1.12 eV at 300 K. At absolute zero, all electrons are tightly bound; as temperature rises, thermal excitation promotes electrons across this bandgap, creating mobile electron–hole pairs (shown in the lower panels of Figure 1 as the symmetric n and p carrier distributions about

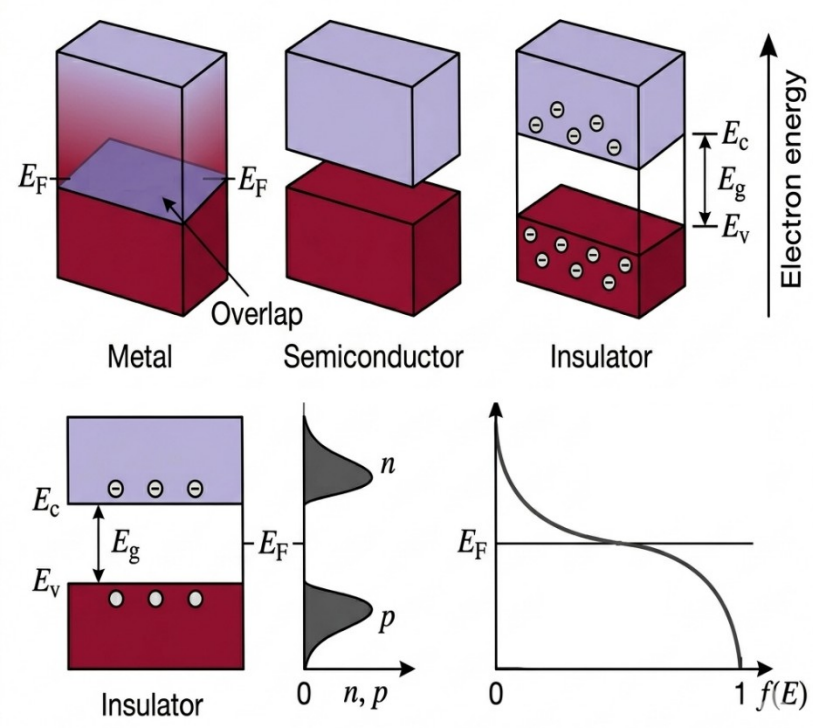


Figure 1: Band structure classification of solids (metal, semiconductor, and insulator), illustrating silicon's intermediate position as a semiconductor with a bandgap of 1.12 eV at 300 K

E_F) that underlie The Fermi–Dirac distribution f(E) confirms that at finite temperatures, a non-zero probability exists for electrons to occupy states above E_c, enabling conduction. These intrinsic material properties acquire particular clinical significance in nuclear medicine, where the sensitivity of PET and SPECT systems depends critically on the charge yield and timing characteristics of their constituent detectors. Silicon detectors have become a cornerstone of nuclear medicine -particularly in high-resolution Positron Emission Tomography (PET) and Single Photon Emission Computed

Tomography (SPECT), due to their exceptional signal characteristics and advanced fabrication potential [9]. These intrinsic material properties translate directly into measurable detector advantages; Table 1 summarises the three core physical properties of silicon, their technical specifications, and their clinical relevance in PET imaging.

Table 1: Core Physical Advantages of Silicon Detectors and Their Clinical Relevance in PET Imaging

Property	Key details	Clinical impact	Ref.
Detection efficiency Low ionisation energy & high resistivity	~3.6 eV per e-h pair (vs ~34 eV for air ionisation chambers; 400–1000 eV for scintillators); wide depletion zone minimises dark current	Superior energy resolution; high signal-to-noise ratio essential for low-activity medical imaging	[6,28]
Speed & material quality High carrier mobility & crystal purity	Electron mobility ~1450 $cm^2/V \cdot s$; Czochralski and float-zone growth yield ultra-low impurity concentrations and improved radiation hardness	Fast coincidence timing for PET; stable long-term performance in clinical radiation environments	[6,10]
Spatial resolution & integration CMOS compatibility and strip/pixel segmentation	Standard wafer processing enables on-chip ASIC readout; strip pitch 50–200 μm; position resolution below 10 μm achievable in laboratory conditions	Sub-millimetre resolution in clinical PET; far exceeds the 1–2 mm limit of conventional scintillator-based systems	[7,11]

Achieving the material purity required to exploit these properties demands extensive processing of naturally occurring silicon sources. Silicon does not occur in elemental form in nature; it is found principally as silicon dioxide (SiO_2) and in complex silicate minerals. Representative concentrations in geological materials are: volcanic rocks (281,500 ppm), sandstone (368,000 ppm), shale (73,000 ppm), and limestone (24,000 ppm). In aquatic environments, silicon concentrations are 6.5 ppm in fresh water and 3 ppm in seawater. Biological systems also incorporate silicon: land plants accumulate 120–6,000 ppm, and certain marine organisms up to 70,000 ppm [2]. Industrial production of metallurgical-grade silicon involves carbothermic reduction of quartz (SiO_2) with carbon at approximately 2200 °C. Electronic-grade silicon requires further purification via the Siemens process or fluidized-bed reactor deposition of trichlorosilane, achieving purity levels exceeding 99.9999999% (nine nines). Single-crystal boules are then grown by the Czochralski technique and sliced into wafers that form the substrates for detector and integrated circuit fabrication [1,5].

3. Applications of Silicon in Nuclear Medicine

3.1 Silicon Radiation Detectors: Operating Principles

Owing to its high purity, fast radiation response, and well-established fabrication technology, silicon is one of the most widely used semiconductor materials in nuclear radiation detectors [29,31]. Silicon radiation detectors operate on the principle of reverse-biased p-n junctions. Under an applied reverse bias voltage, a depletion region devoid of mobile charge carriers forms across the junction, creating a sensitive detection volume. When ionizing radiation (gamma rays, X-rays, or charged particles) traverses this depletion region, it generates electron–hole pairs proportional to the deposited energy. The built-in electric field separates these carriers, driving electrons toward the n-type contact and holes toward the p-type contact, producing a measurable current pulse at the output [6,7]. As illustrated in Figure 2, a silicon strip detector comprises p^+-type implant strips separated by SiO_2 passivation on a bulk n-type substrate (~300 µm thick), with a backplane n^+-type contact connected to a positive bias voltage; the generated electrons and holes drift in opposite directions under the electric field E and are collected at their respective electrodes, with the resulting signal amplified by an external preamplifier

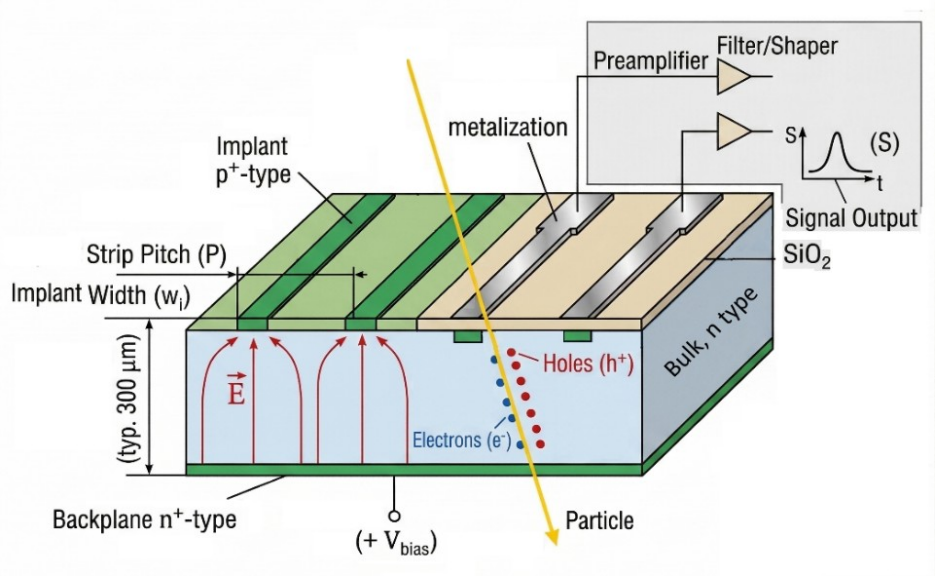


Figure 2: Principles of operation of a silicon strip detector

and filter/shaper chain. The device architecture, shown in Figure 3, consists of a p^+ top layer, a lightly doped n^- substrate of thickness D, and an n^+ bottom contact, all connected to an external amplifier under reverse bias, confirming the same charge-collection mechanism across detector geometries [31,32]. Common detector configurations include p-n, n-p, and n-in-p geometries. The n-in-p configuration is particularly favored for high-radiation applications, as electrons dominate charge collection and are less susceptible to trapping caused by radiation-induced bulk defects [14,15]. Silicon strip detectors segment the junction into parallel strips with a typical pitch of 50–200 µm, providing one-dimensional position information, while double-sided silicon strip detectors (DSSDs) add

orthogonal strips on both faces for 2D position resolution below 10 μm. Silicon photomultipliers (SiPMs) represent a further advancement: arrays of avalanche photodiode microcells operating in Geiger mode, offering single-photon sensitivity, compactness, and immunity to magnetic fields — properties that conventional gas-based photomultiplier tubes cannot match [16,17].

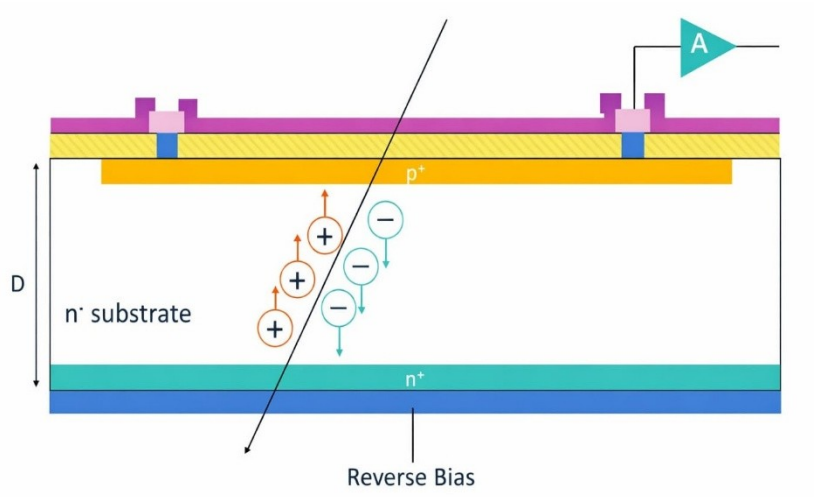


Figure 3: Silicon p^+-n^--n^+ radiation detector under reverse bias showing electron-hole pair generation and charge drift to the external amplifier.

3.2 Medical Imaging: PET and SPECT

3.2.1 Positron Emission Tomography (PET)

PET is a non-invasive functional imaging modality based on the coincidence detection of 511-keV annihilation photons produced when positrons emitted by a radiotracer annihilate with electrons in tissue [18]. The most widely used radiotracer is ^{18}F-fluorodeoxyglucose (FDG), which accumulates preferentially in tissues with elevated glucose metabolism, enabling detection of malignant tumors, neurological disorders, and cardiac dysfunction [18,19]. As illustrated in Figure 4, a PET ring is lined with silicon-based SiPM detectors that record the two 511-keV annihilation photons emitted simultaneously in opposite directions. In modern PET scanners, silicon-based SiPMs have largely replaced traditional photomultiplier tubes as the photodetection elements coupled to scintillating crystals such as LYSO or BGO, offering superior timing resolution critical for time-of-flight PET (ToF-PET), a compact form factor enabling MRI-compatible hybrid systems, and stability across clinical temperature ranges [25]. Silicon microelectronics additionally perform coincidence logic, time stamping, list-mode data acquisition, and iterative image reconstruction central to PET operation [9,26,27].

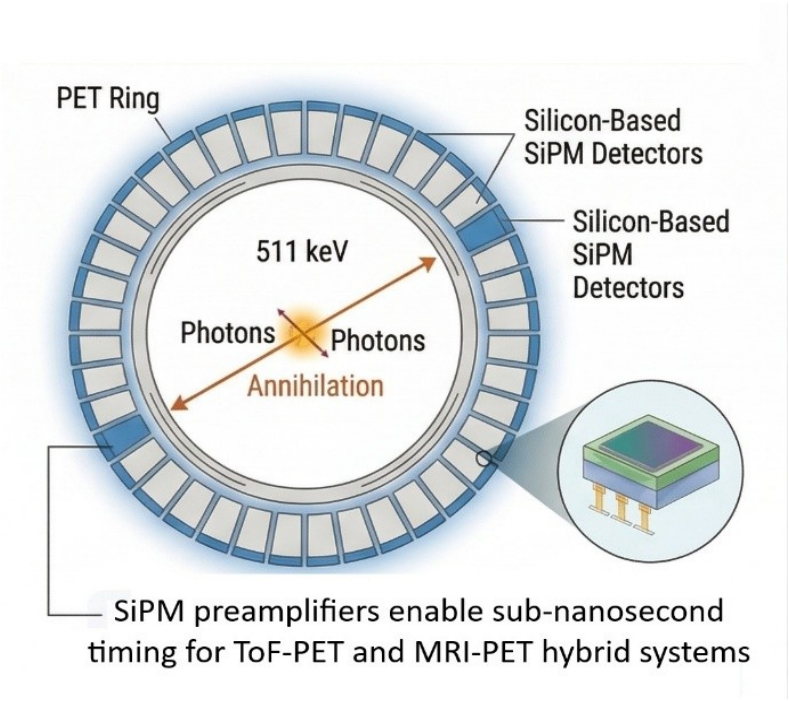


Figure 4: Silicon-based SiPM PET ring enabling 511-keV coincidence detection for ToF-PET and MRI-PET hybrid imaging.

3.2.2 Single-Photon Emission Computed Tomography (SPECT)

As illustrated in Figure 5, a clinical SPECT system consists of a sliding patient table, a pair of rotating dual-head gamma cameras that encircle the patient, and a dedicated workstation for image

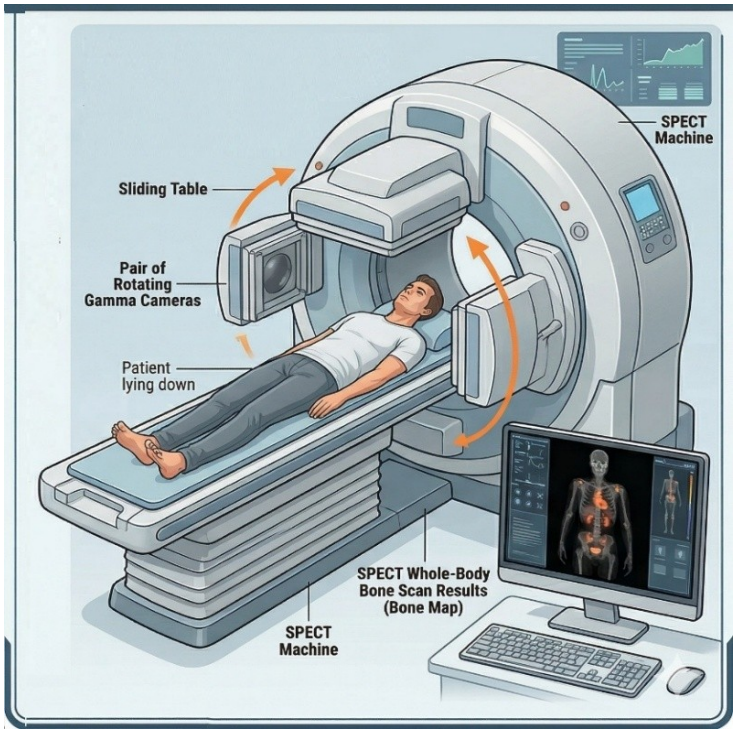


Figure 5: Clinical SPECT scanner with rotating dual-head gamma cameras and whole-body bone scan reconstruction.

reconstruction. The system measures the spatial distribution of gamma-emitting radiotracers, generating 3D tomographic maps of tracer concentration that are routinely used to assess myocardial perfusion, hepatobiliary function, renal perfusion, bone metabolism, thyroid pathology, and tumor staging [18,28]. Silicon detectors, particularly Compton cameras based on position-sensitive silicon strip detectors, offer an alternative imaging approach that eliminates the mechanical collimator, thereby recovering detection efficiency. As shown in Figure 6, the Compton camera design employs two silicon strip detector planes: a position-sensitive scattering detector that generates a recoiling electron, followed by an absorption detector that captures the scattered photon, enabling source localization without mechanical collimators. In combined SPECT/CT configurations, silicon detectors are integral to the CT subsystem for anatomical co-registration, as well as to data acquisition and image correction electronics [9].

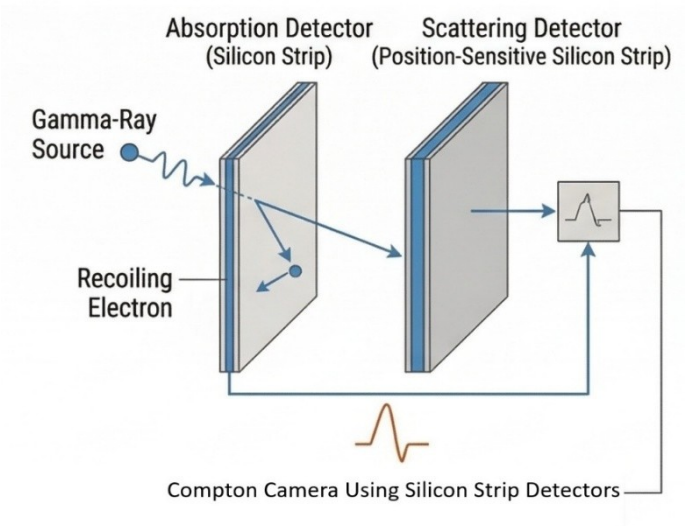


Figure 6: Compton camera using silicon strip detectors for collimator-free gamma-ray source localization

3.3 Silicon in Radiation Dosimetry

Accurate radiation dosimetry is paramount in both nuclear medicine imaging and radiotherapy to ensure therapeutic efficacy while protecting surrounding healthy tissue. Silicon diode dosimeters are widely deployed for this purpose, exploiting two key material advantages: near-tissue equivalence (Z_{eff} ≈ 14 vs. Z_{eff} ≈ 7.4 for soft tissue), which minimises energy-dependent correction factors, and a significantly greater charge yield per unit volume compared with air ionisation chambers, owing to silicon's substantially higher density (2.33 g/cm³) and lower electron–hole pair creation energy (3.6 eV vs. ~34 eV in air) [6,32]. For advanced radiotherapy modalities involving small radiation fields, steep dose gradients, and ultra-high dose rates such as FLASH radiotherapy, silicon dosimeters offer

spatial resolution below 1 mm², sub-millisecond response time, and the ability to be fabricated in body-implantable form factors for real-time in vivo dosimetry [6,33]. Beyond conventional silicon, silicon carbide (SiC) has emerged as a promising wide-bandgap material for next-generation radiation dosimetry, particularly in carbon ion therapy environments; a detailed comparative analysis is presented in Section 4 [33].

4. Comparative Analysis: Silicon vs. Silicon Carbide Detectors

Silicon carbide (SiC) has emerged as a wide-bandgap alternative to silicon for radiation detector applications requiring operation in high-temperature, high-radiation, or high-electric-field environments [21,33]. Among its polytypes, the 4H-SiC configuration has attracted the greatest research interest owing to its combination of a wide bandgap, high carrier saturation velocity, and superior thermal conductivity. Table 2 presents a systematic comparison of the principal physical properties of 4H-SiC and silicon at room temperature, and Figure 7 shows the cross-sectional architecture of a representative SiC p-n diode detector fabricated for dosimetry research.

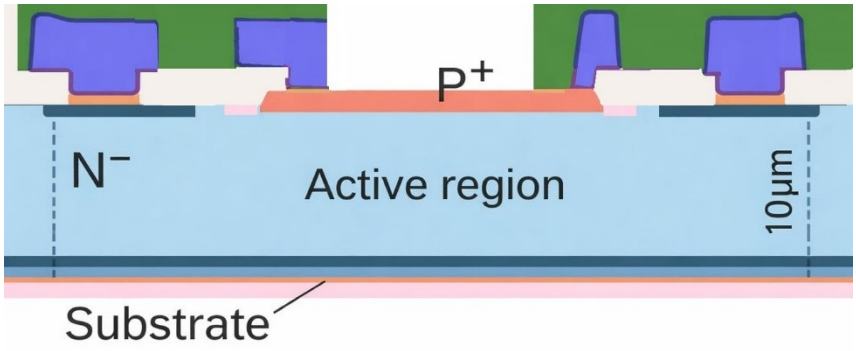


Figure 7: Cross-section of a SiC p-n diode with n⁻ epitaxial layer on substrate

As Table 2 demonstrates, SiC exhibits a bandgap approximately three times larger than silicon (3.23 eV vs. 1.12 eV), thermal conductivity roughly three times greater (4.9 vs. 1.5 W/cm·K), and a breakdown electric field ten times higher (3–4 × 10^6 vs. 3 × 10^5 V/cm). These properties make SiC inherently resistant to thermal noise and radiation-induced bulk damage, conferring an operational advantage near the Bragg peak in carbon ion therapy, where both high local ionisation density and elevated linear energy transfer (LET) present significant challenges for conventional silicon detectors. Matsumoto et al. investigated LET distribution measurements using an energy-dispersive SiC dosimeter exposed to pristine and spread-out Bragg peak (SOBP) carbon beams at 290 MeV/u, conducted across the plateau, SOBP, and post-Bragg peak regions. Relative biological effectiveness (RBE) was estimated using a linear-quadratic model based on dose-averaged LET values,

demonstrating the potential of SiC detectors for therapeutic dose calculation in carbon ion therapy, though further validation remains necessary [33].

Table 2. Comparative physical properties of 4H-SiC and Si at room temperature [21,33].

Property	4H-SiC	Si
Bandgap E_{gap} (eV)	3.23	1.12
Breakdown field E_breakdown (V/cm)	$3–4 \times 10^6$	3×10^5
Electron mobility μ_e ($cm^2/V \cdot s$)	800	1450
Hole mobility μ_h ($cm^2/V \cdot s$)	115	450
Saturation velocity V_sat (cm/s)	2×10^7	0.8×10^6
Atomic number Z	14/6 (average)	14
Relative permittivity ε_r	9.7	11.9
e-h pair creation energy (eV)	7.6–8.4	3.6
Density (g/cm^3)	3.22	2.33
Displacement energy E (eV)	30–40	13–15
Thermal conductivity (W/cm·K)	4.9	1.5

Silicon, however, retains decisive advantages in standard clinical environments: its lower e-h pair creation energy (3.6 eV vs. 7.6–8.4 eV in SiC) produces proportionally larger signal charges per unit deposited energy, its higher carrier mobilities enable faster timing, and its vastly more mature fabrication ecosystem delivers lower cost, better wafer uniformity, and seamless integration with CMOS readout electronics [6,7]. Table 3 quantifies these differences in detector energy resolution under alpha-particle irradiation, confirming that silicon achieves superior resolution under standard operating conditions.

Table 3. Measured ($FWHM^{det}$) and statistical-limit ($FWHM^{stat}$) energy resolutions for SiC and Si alpha-particle detectors [35].

Detector	$FWHM^{det}$ (keV)	$FWHM^{stat}$ (keV)
SiC	10.3	5.0
Si	7.3	3.5

The choice between Si and SiC is therefore application-driven: silicon is preferred for conventional PET/SPECT imaging and standard dosimetry owing to its superior signal yield and proven integration pathways, while SiC becomes the material of choice in high-radiation-dose or elevated-temperature environments, such as neutron dosimetry in reactor fields or heavy-ion therapy monitoring [33,35].

4.1 Hypothetical Simulation: Comparative Signal Analysis

To further illustrate the practical significance of the material differences summarised in Tables 2 and 3, a hypothetical simulation was conducted comparing Si and SiC detector performance under identical irradiation conditions. Assuming an absorbed energy of 5 MeV, pair creation energies of W_{Si} = 3.6 eV and W_{SiC} = 8.0 eV, and a Fano factor F = 0.1, consistent with established semiconductor detector theory [6], the number of generated electron–hole pairs N and the energy resolution R were computed according to:

$$N = \frac{E}{W} \qquad (1)$$

$$R = 2.35 \times \sqrt{\frac{F \times N}{N}} \qquad (2)$$

The results, presented in Table 4, confirm that silicon generates 2.22 times more electron–hole pairs than SiC under identical conditions, yielding proportionally superior energy resolution. This quantitative advantage directly underpins silicon's dominance in standard clinical PET/SPECT applications, while SiC retains its operational edge in high-radiation environments as discussed above.

Table 4. Hypothetical simulation results for Si and SiC detectors at E = 5 MeV, F = 0.1.

Parameter	Si	SiC
Pair creation energy W (eV)	3.6	8.0
e-h pairs N	1,388,889	625,000
Energy resolution R	6.31×10^{-4}	9.40×10^{-4}
FWHM (keV)	3.16	4.70
Relative signal strength	Strong	Moderate

Note: Values computed from standard detector physics formulae; W values from Table 2; F = 0.1 as typical for semiconductors [6]

As shown in Figure 8, silicon generates 2.22 times more electron–hole pairs than SiC under identical conditions (1,388,889 vs. 625,000 pairs), yielding a proportionally superior energy resolution (FWHM = 3.16 keV vs. 4.70 keV). This quantitative advantage directly underpins silicon's dominance in standard clinical PET/SPECT applications, while SiC retains its operational edge in high-radiation environments.



Figure 8. Hypothetical simulation comparing electron–hole pair yield, energy resolution, and signal strength for Si and SiC detectors under identical irradiation conditions.

5. Silicon detector advantages and the role of artificial intelligence in nuclear medicine

Semiconductor radiation detectors offer a set of advantages that collectively explain their dominance in nuclear medicine instrumentation. Their high density - 2.33 g/cm³ for silicon and 3.22 g/cm³ for SiC - maximises stopping power per unit thickness, permitting compact thin-detector designs with reduced parallax error in tomographic imaging [6,7]. The small diffusion length of charge carriers in solid-state materials, combined with fine strip or pixel segmentation, yields position resolution below 10 μm, far exceeding the capability of gas-filled detectors [7,14]. At 3.6 eV per electron–hole pair, silicon generates approximately 4–10 times more signal carriers than gas ionisation chambers (20–40 eV per ion pair) and 100–270 times more than scintillator systems (400–1000 eV per photoelectron), producing intrinsically superior energy resolution [6]. Silicon's full compatibility with CMOS fabrication further allows the detector and its readout electronics to be co-integrated on the same substrate, reducing parasitic capacitance, electronic noise, and system complexity, while carrier transit times below 30 ns support the tight coincidence timing windows required in PET systems and permit high-count-rate operation without significant pile-up distortion [7,8]. Engineered structures such as p-spray insulation and oxygen-enriched bulk silicon extend these advantages to radiation-tolerant designs adequate for cumulative doses below 1 kGy per year, which encompasses the majority of clinical nuclear medicine environments [7]. These hardware strengths are substantially amplified by artificial intelligence algorithms running on silicon-based processors — CPUs, GPUs, and custom neural processing units — which perform real-time image reconstruction, attenuation correction, tumour segmentation, lesion classification, and standardised uptake value (SUV) quantification in PET and SPECT workflows, as illustrated in Figure 9. Deep convolutional networks applied to low-count PET acquisitions have demonstrated reductions in administered radiotracer activity of up to 80% while preserving diagnostic image quality, automated tumour staging performance comparable to expert nuclear medicine physicians, and simultaneous multi-pathology classification on chest scintigraphy. The integration of AI inference engines directly into silicon ASIC readout chains, enabling event-level classification and adaptive thresholding at the point of acquisition, represents the next frontier in this convergence [19].

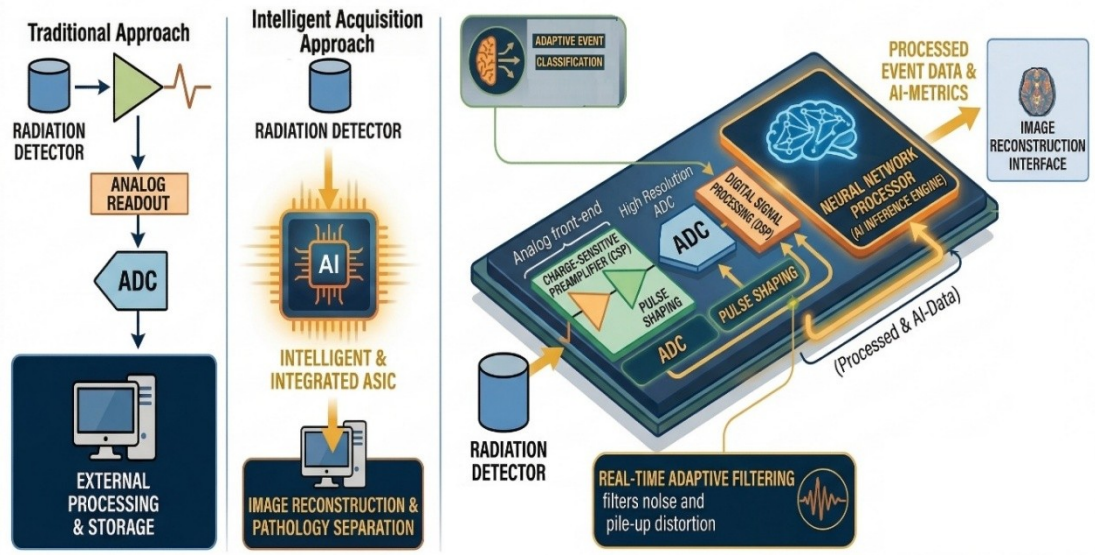


Figure 9: AI-integrated ASIC readout architecture for radiation detectors, enabling real-time event classification, adaptive filtering, and on-chip image reconstruction at the point of acquisition

6. Nuclear Medicine in Saudi Arabia: Current Status and Vision 2030 Initiatives

Saudi Arabia has made substantial investments in nuclear medicine infrastructure over the past two decades. The Kingdom operates 51 nuclear medicine facilities under government, private, and Ministry of Health management, collectively performing approximately 37,655 general nuclear medicine investigations and 12,387 cardiac studies annually, with a 2018 national survey documenting 21 PET/CT systems, 55 SPECT/CT systems, and 35 standalone SPECT and gamma camera units distributed across the Kingdom [37]. Major nuclear medicine centres are co-located with leading tertiary hospitals including King Abdulaziz Medical City, King Faisal Specialist Hospital and Research Centre, King Saud Medical City, and Almana Hospitals, offering comprehensive services encompassing oncology, cardiology, neurology, endocrinology, and nephrology [37,40]. The Nuclear and Radiological Regulatory Commission (NRRC) has established radiation safety regulations aligned with International Atomic Energy Agency (IAEA) standards and WHO guidelines. National Diagnostic Reference Levels (NDRLs) for radiographic, CT, and hybrid SPECT/CT examinations have been developed and validated through multicentre studies, providing an evidence-based framework for optimising patient doses while maintaining diagnostic quality [38,39,41]. Building on this foundation, the Kingdom's Vision 2030 health transformation agenda targets the systematic deployment of PET/CT for oncological, neurological, and cardiac imaging; the integration of nuclear medicine and medical physics curricula at Princess Nourah bint Abdulrahman University and King Saud University with American Board certification pathways for Saudi nationals; the development of

comprehensive regulatory frameworks for radiation source management and emergency preparedness under the NRRC; active participation in IAEA technical cooperation programmes for technology transfer and quality-assurance capacity building; and support for precision medicine and biomedical imaging research at King Abdullah University of Science and Technology (KAUST) [37]. These coordinated initiatives collectively position Saudi Arabia to transition from a technology-importing nation in nuclear medicine to one capable of regional leadership in advanced molecular imaging and silicon-based detector research.

7. Challenges and Future Research Directions

Despite the considerable maturity of silicon detector technology, several technical and translational challenges must be addressed before its full potential in nuclear medicine can be realised. The first and most established challenge is radiation damage and long-term stability. Prolonged exposure to high-energy particles induces bulk crystal defects (primarily vacancy clusters and interstitials), as well as surface oxide damage that progressively degrades detector performance by increasing leakage current and introducing charge trapping centres. Oxygen-enriched silicon and cryogenic operation at −20°C or below mitigate these effects but add system complexity and cost; future research should therefore prioritise defect-engineering strategies and the development of novel silicon allotropes with intrinsically enhanced radiation hardness [7]. A second frontier concerns ultra-high dose-rate dosimetry for FLASH radiotherapy. This emerging modality delivers therapeutic doses in microsecond pulses at rates exceeding 40 Gy/s, a regime in which conventional silicon diodes exhibit pronounced ion recombination and polarisation effects that compromise dosimetric accuracy. The development of diamond-like carbon thin-film sensors and SiC-based diodes specifically optimised for FLASH conditions represents a critically needed and actively pursued research direction [22]. Miniaturisation for in vivo dosimetry presents a third challenge. Implantable silicon dosimeters with wireless readout capability would enable real-time dose monitoring during brachytherapy and external-beam radiotherapy fractions; silicon's biocompatibility for implantable applications is well established, with silicone materials in clinical use since 1946 [3,4, 40], but achieving reliable power delivery and bidirectional data transmission within the body without introducing artefacts in concurrent MRI or CT imaging remains a significant engineering obstacle. The commercial scalability of SiC detectors constitutes a fourth constraint. Unlike silicon, SiC cannot be grown from a melt phase and requires vapour-phase epitaxy processes that are substantially more expensive and lower in throughput than the Czochralski method used for silicon. Achieving wafer-scale SiC fabrication with the uniformity and yield required for clinical nuclear medicine applications remains an open materials science challenge; hybrid Si/SiC detector arrays combining silicon's signal yield with SiC's radiation hardness in a single

module represent a pragmatic intermediate solution meriting systematic investigation [22,35]. Finally, the co-design of AI-capable silicon ASICs defines a fifth emerging frontier. Future nuclear medicine systems will increasingly require on-detector AI inference, with neural network engines embedded directly in the ASIC readout chain, enabling event-level classification, pile-up rejection, and adaptive thresholding in real time. Engineering silicon ASICs that incorporate neural processing capability while maintaining the noise floor, power consumption, and timing resolution demanded by medical detector operation represents an open and consequential problem at the intersection of semiconductor physics and machine learning [19].

8. Conclusions

This study has demonstrated that silicon occupies a central and indispensable position in the technological ecosystem of nuclear medicine. Its favourable physical properties (particularly low electron–hole pair creation energy, fast carrier mobility, compatibility with CMOS fabrication, and broad commercial availability) make it uniquely suited for radiation detection, PET and SPECT imaging systems, nuclear medical electronics, and precision dosimetry. The systematic comparison with silicon carbide (SiC) reveals complementary strengths: silicon excels in energy resolution and signal magnitude under standard conditions, while SiC's wider bandgap and higher radiation hardness offer advantages in extreme environments such as heavy-ion radiotherapy and high-neutron-flux facilities. This conclusion is further supported quantitatively by the hypothetical simulation, which demonstrates that silicon generates 2.22 times more electron–hole pairs than SiC under identical irradiation conditions, directly confirming its superior signal yield in standard clinical environments. Judicious selection of detector material based on application-specific requirements is therefore essential. The convergence of silicon hardware capabilities with artificial intelligence algorithms is driving a qualitative transformation in nuclear medicine diagnostics, enabling low-dose imaging, automated image interpretation, and adaptive scanning protocols with quantitative accuracy exceeding prior approaches. At the national level, Saudi Arabia's expanding nuclear medicine infrastructure and Vision 2030-aligned initiatives in workforce development, regulatory standardisation, and international scientific collaboration position the Kingdom to play an increasingly significant role in the adoption and development of silicon-based nuclear medicine technologies. Future research priorities should address radiation damage mitigation in silicon detectors, FLASH radiotherapy dosimetry, implantable in vivo dosimetry systems, scalable SiC fabrication, and the co-design of AI-capable silicon ASICs for next-generation nuclear medicine instrumentation.

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Data Availability Statement

Not applicable.

Conflicts of Interest

There is no conflict of interest, the writers assert.

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