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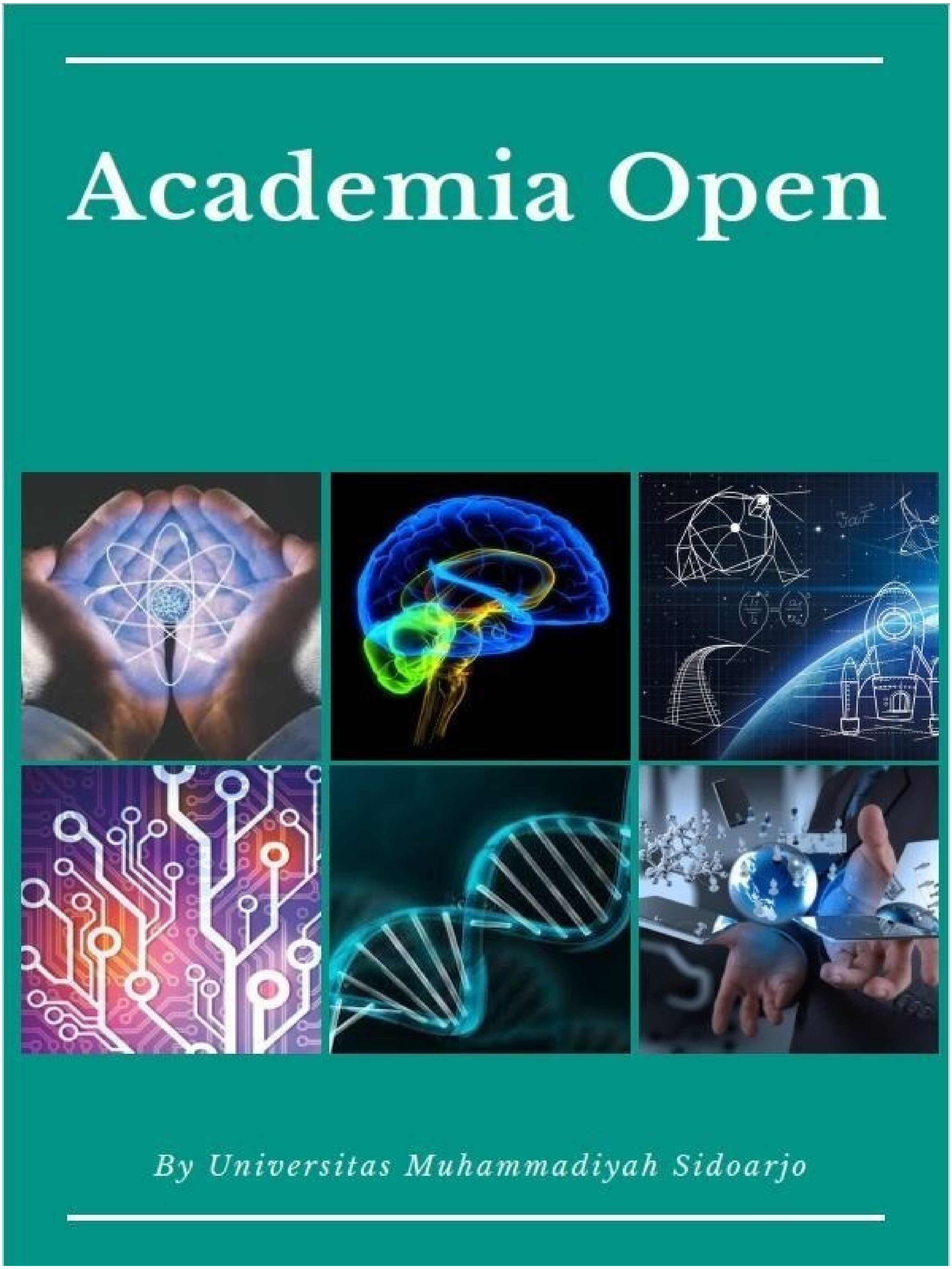

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Applications of nuclear physics in medical diagnostics: A comparative study between SPECT and PET

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Abstract

Nuclear physics has fundamentally transformed medical diagnostics through Single Photon Emission Computed Tomography (SPECT) and Positron Emission Tomography (PET), which provide functional and physiological information beyond conventional imaging modalities. This study conducts a comprehensive comparative analysis of SPECT and PET by examining their nuclear physics principles, technical instrumentation, image quality parameters, and clinical applications. Employing systematic literature review methodology, the research synthesizes peer-reviewed articles from major scientific databases published between 2015 and 2025. The analysis reveals that PET demonstrates superior spatial resolution (4-7 mm versus 8-12 mm), enhanced sensitivity (1-2% compared to 0.01-0.03%), and greater quantitative accuracy due to coincidence detection and higher photon energy (511 keV). Conversely, SPECT maintains advantages in cost-effectiveness, radiotracer accessibility through on-site synthesis, and longer half-lives suitable for extended protocols. This research integrates physical, instrumental, and patient-dependent factors influencing image quality while exploring emerging developments including hybrid imaging and artificial intelligence applications. The findings establish evidence-based criteria for modality selection, emphasizing the complementary nature of these techniques in advancing precision diagnostics across oncology, cardiology, and neurology.

Keywords : Nuclear Medicine Imaging, SPECT Modality, PET Technology, Radiotracer Applications, Gamma Photon Detection, Medical Diagnostics

Highlight :

- PET demonstrates superior spatial resolution (4-7 mm) and sensitivity (1-2%) compared to SPECT's performance metrics.
- Hybrid imaging systems combining PET-CT achieve widespread clinical acceptance, particularly enhancing oncology diagnostic capabilities.
- Collimator geometry primarily limits SPECT resolution (8-12 mm), while coincidence detection enables PET's improved quantitative accuracy.

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Introduction

The main aim of modern medicine is to give the detailed, comprehensive image of the human organs or tissues. Standard forms of imaging are unable to generate images that have complete details. In X-ray imaging, e.g. CT, the images are created through absorption of photons and are planar. The most developed methods like MRI will detect certain tissue properties because of the nuclear magnetic dipole of hydrogen atoms but the generated images will not give an exact picture of organs and tissues. Thus, nuclear medicine was put forward to identify the different diseases at an early stage before their symptoms manifest by analyzing the functions and physiology of cells instead of just anatomy.

Nuclear medicine imaging processes have become crucial in clinical in practice, especially in oncology, neurology and cardiology, through the use of artificial radiosubstances (radiotracers) [1]. Radiotracer is made up of an atom of a radioactive isotope coupled with a certain carrier molecule, which has a specific target cell. Radiotracers should be of a certain type, they should have an energy range between 70 and 511 Kev, a clean decay that is gamma-based and medium life (minutes to hours). Radiotracer can be delivered to patients through intravenous (IV) injection, oral route (taking a tablet), or even by inhaling the radiotracer, based on the disease and targeted organ under investigation. After entering the body, the material decays until it reaches the part of the body it is targeted and then releases its energy in the form of a radiation, which can be a gamma ray or a positron among others, depending on the kind of injected source. During the purchase of a successful image, the most significant detail is the correct choice of a radiotracer. Table 1 provides a list of typical radiotracers in nuclear medicine imaging, that includes data on physical properties of the radiotracer, as well as examples of clinical and research use [2], [3]. Unlike the traditional X-ray imaging, where the photons are transmitted through the body and a detection system records the transmission images, in nuclear medicine, photons are emitted in the body and the detection system captures them.

Table 1. Common Radiotracers in Nuclear Medicine.

Isotope	Half-life	Radaition Type	Energ y (kv)	Modality	Clinical Application
Technetium-99m (Tc-99m)	6 hours	Gamma	140	SPECT	Bone scans, brain, heart, liver, lungs, kidneys, thyroid
Iodine-123 (I-123)	13 hours	Gamma	159	SPECT	Thyroid imaging, brain, kidenys
Fluorine-18 (F-18)	110 minutes	Positron	511	PET	Brain studies, cancer imaging,
Oxygen-15 (O-15)	2 minutes	Positron	370	PET	Blood flow, brain, heart research
Carbon-11 (C-11)	20 minutes	Positron	390	PET	Brain research

The scanner, also known as a detection system, gamma camera, or positron camera is a system that consists of a collimator, a detector, a photomultiplier tube and a data processing system. The collimator is used to form the directions of the gamma rays released by the body of the patient. The detector is made of a scintillation crystal; a NaI(Tl) crystal and a photomultiplier tube (PMT). The gamma radiation loses its energy to the scintillation crystal to produce the light signal. The PMT amplifies the single photon and transforms it into an electric signal and passes on to a data processing system.

Single Photon Emission Computed Tomography (SPECT) and Positron Emission Tomography (PET) are the two commonest imaging modalities used in nuclear medicine. Choi et al explain that radiotracers are the type of radiotracer that determines the diagnostic SPECT or PET imaging a patient gets. Recent studies indicate that, in some circumstances, scanning patients with SPECT and scanning with PET has a higher accuracy level particularly in patients with a high body mass index (BMI) where factors of patient thickness influence the attenuation factor [4]. It is through the merit of PET over SPECT of showing a higher sensitivity because of the mode of detection.

To further explain, in this paper, we have given a detailed comparison of both modalities regarding such physical factors that affects the quality of images as imaging resolution and accuracy. To deal with these factors, one should speak about the differences in the principles of working.

Fundamental Nuclear Physics Principles

A. Photon emission of radioactive decay

In SPET, gamma-emitting isotopes are used. These isotopes undergo decay, resulting in the emission of a single photon. For example, Technetium-99m and Iodine-123 undergo radioactive decay, emitting a single gamma photon with sufficient energy (140 keV and 132 keV, respectively) to penetrate the patient's body, as shown in Figure 1 (A) [5]. In contrast, the PET uses positron-emitting isotopes. The isotopes, such as Fluorine-18, decay via positron emission. The positron travels a short distance through the patient's body (0.1-1.0 mm) and interacts with tissues, annihilating with an electron [6]. The energy is released in the form of two gamma photons, each with an energy of 511 keV, in opposite directions (angle 180°). Figure 1 (B).

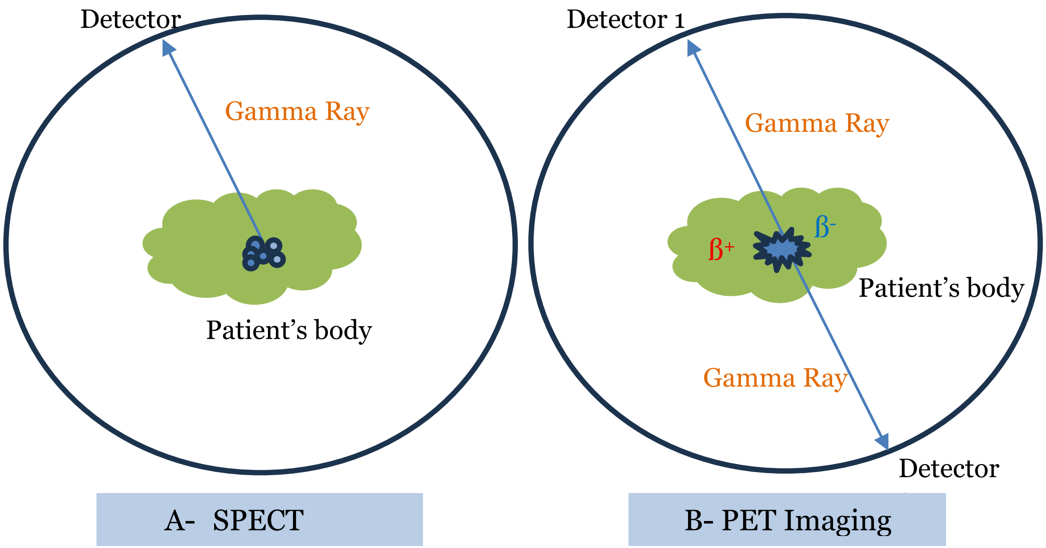

Figure 1. Schematic representation of (A) SPECT and (B) PET detection.

B. Image Acquisition

Figure 2 presents key elements of the gamma camera that would be applicable in the SPECT imaging system. The system spins around the patient to obtain information at different angles thus generating 2D projection images. Then a 3D image will be formed by a mathematical reconstruction algorithm. The first one is collimators, consisting of a piece of lead material containing thousands of parallel holes, which designate the direction of the gamma photons. The scattered photons (red) or photons, which are transmitted at an angle (blue) are blocked by collimator and the photons that travel directly without altering the path are recorded by the detector (green). That is, these are the photons that only help in the formation of the image. The collimator is also thought to be the key limiting factor to the sensitivity of SPECT since it absorbs about 99% of emitted photons with only a fraction of less than 1% being absorbed. Figure 3 gives the several types of collimators such as parallel-hole, pin-hole, covering and diverging collimators, which are used in fan-beam and cone-beams of applications. The use of these collimators depends on the clinical usage. An example of this is a small organ that is to be studied using a pinhole collimator as it is necessary to magnify the images at a small field of view (FOV). Nevertheless, in the case of the bone scanning where the fineness of a detail matters, long-septum collimators may be employed since their sensitivity is low. The produced image is determined by the distance between the source and collimator, the farther the source and collimator the worse the image will be. The second element of a gamma camera is the detector, which is a scintillation crystal, which registers the photons that came through the collimator. The size of the detector is about 40 cm x 50 cm and its thickness is between 6 and 12 mm. There are some specifications that the scintillation crystal must have, which include the material being used being of high Z to eliminate all the incident of the gamma rays. The radiation that is received by the scintillation crystal is then converted to light photons that are translated into an electric signal by the third element, the photomultiplier tube (PMT) array, by amplifying the signal with a preamplifier. To minimize the noise the scintillation crystal must have a high light output. The interaction position of gamma photons in the crystal is computed using array output [7].

For the PET system, the imaging setup is slightly different due to the coincidence detection of annihilation photons by a positron camera. The position camera is based on multiple scintillation detectors. The signal can be read out in the form of a 3D image. As the concept of imaging relates to the spontaneous emission of two opposite annihilated photons, there is no need to use collimators because the detectors can be arranged along the path of the detection photons. The primary component of the PET camera is the scintillation detectors that surround the imaging volume. When one detector records a photon, all other detectors are checked for the recording of the opposite photon. A coincidental event can be recorded only if the time window between the two opposite recorded photons is in the range of 6-15 nanoseconds. The detectors that recorded two coincident photons are connected by a created line called the line of response (LOR). Each pair of detectors is capable of creating LOR. Many LORs can create a single or several projections. Due to this, the detectors can record an actual coincidence event (true coincidence), where two detected photons originate from the same annihilation process (Figure 4-1). In the scattered coincidence, one or both photons are scattered in the patient and successfully detected, leading to inaccuracy (Figure 4-2). Moreover, a random coincidence event occurs when two photons are detected simultaneously but originated from different annihilation processes, resulting in image noise (Figure 4-3). Finally, a reconstruction algorithm utilized all the LORs to create a 3D image (indicated by the blue dashed line in Figure 4).

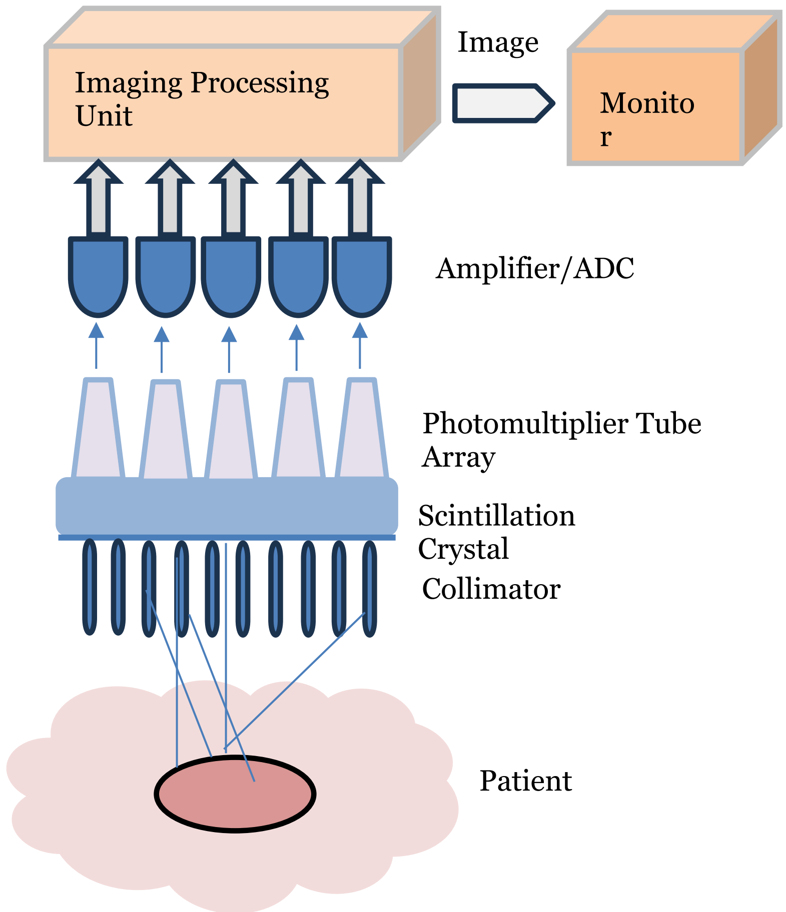

Figure 2. Schematic representing a SPECT Gamma camera**.**

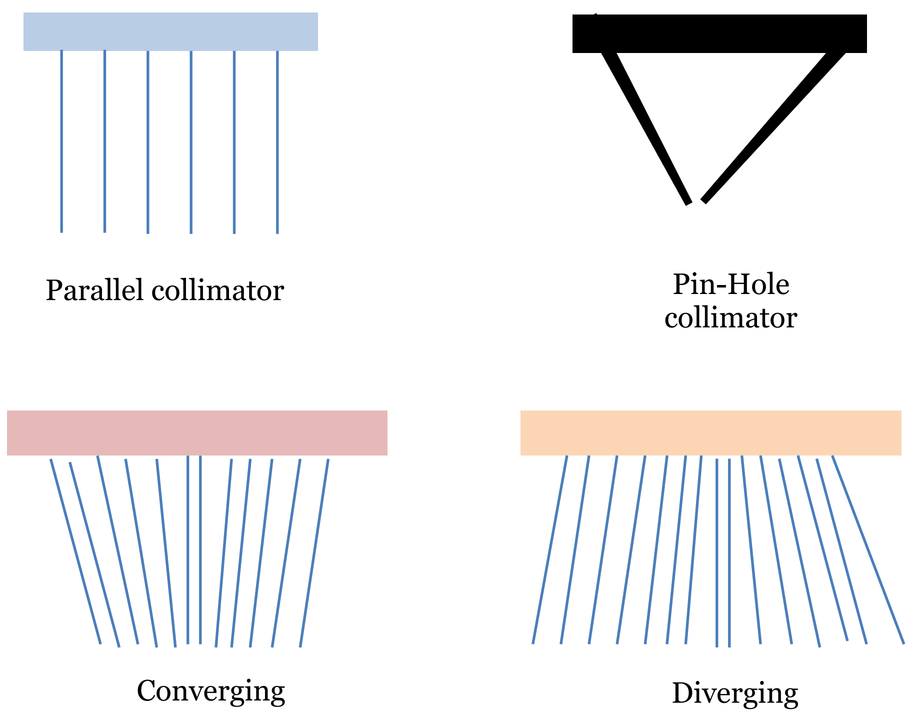

Figure 3. The schematic represents different types of collimators.

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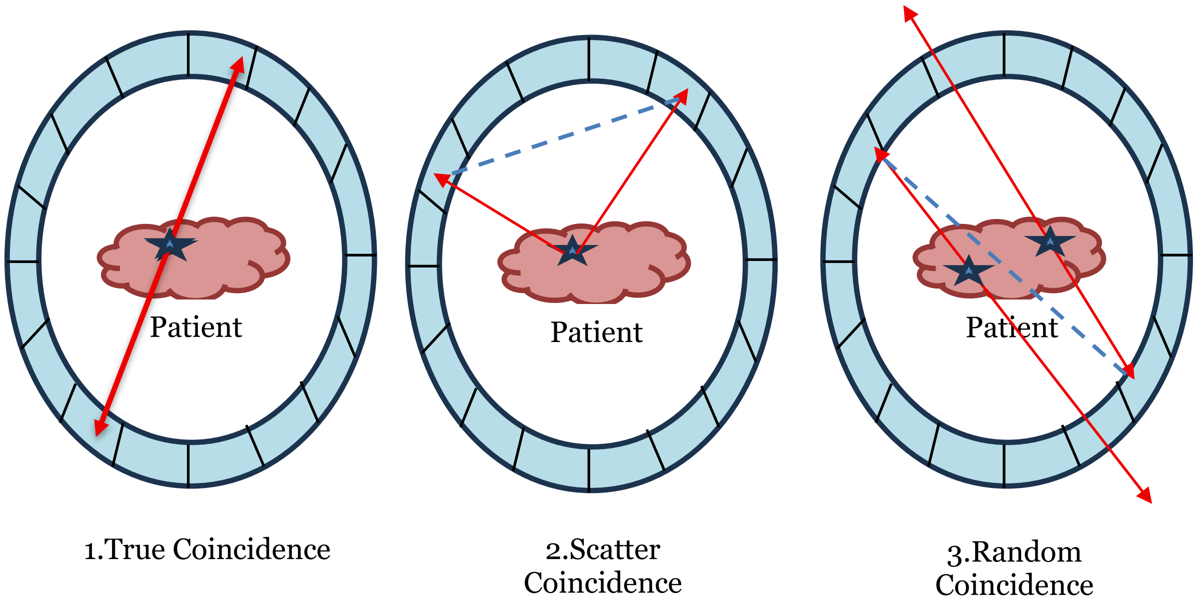

Figure 4. Photon detection in PET. The red arrows represent the photon annihilation, and the blue dashed lines represent LOR.

C. Image Quality Parameters

Several factors that influence the quality of the image have to be considered. These factors can be categorized into three main groups: physical characteristics factors, instrumental factors,and patient-dependent factors. The first one refers to the properties of radiation and how it interacts with the surrounding tissues. The second one is related to the characteristics of the imaging system. The last one considers the patient's preparation. In exceptional cases, a correction should be administered, such as the attenuation correction that applies in PET to compensate for the attenuation that occurs due to the long traveled distance by the photon through the patient [5].

D. Image Quality Parameters

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4.1 Spatial Resolution

The ability of the system to distinguish two closely spaced objects is known as spatial resolution. It is an essential factor in medical imaging. The image resolution in the SPECT modules is limited (8-12 mm) by the collimator geometry (length in the case of the parallel collimator or the collimator hole size in the case of a pinhole collimator). The type and size of the detector crystal limit the image resolution in the PET module. This is because gamma rays penetrate the crystal before interacting and do not interact at the surface of a crystal. Another factor that also influences the resolution is the deviation of the emitted photons after annihilation from the optimal degree (180^{o}). The distance travelled by a positron since emission untill annihilation is an additional factor that impacts the spatial resolution [8].

The energy of photons also influences the spatial resolution. The amplitude of the measured signal is determined by the energy deposited in the crystal. Therefore, the spatial resolution measured in the crystal with 140 keV is inferior compared with that measured in the crystal with 511 keV. Generally, PET has a superior resolution (ranging from 4 to 7) compared to SPECT.

4.2 Sensitivity

Sensitivity, the ability of the system to record the emitted photons, can be increased by increasing the spatial resolution. Again, excitation of the collimator reduces the sensitivity of SPECT. PET sensitivity is superior (1-2%) compared to SPECT (0.01-0.03%). In other words, PET can detect the same amount of photons in a shorter scan time.

4.3 Quantitive Accuracy

This term determines the system's ability to measure the absolute concentration of the radiotracer. In both SPECT and PET, quantitative accuracy requires a linear reconstruction algorithm, correction for the absorbed and scattered photons, and calibration using the kBq/cm^3 unit. SPECT is reconstructed using photon counts (counts per voxel), while PET uses radioactive concentration. Due to the lower count statistics and photon energy-dependent attenuation, SPECT has less robust accuracy than PET [9].

Methodology

Strategy of literature review

This comparative study was conducted across major scientific databases (PubMed, ScienceDirect, Springer, Google Scholar, and random Google searches), focusing on articles published between 2015 and 2025, and examining physical principles, instrumentation, applications, and limitations. All searches were based on the following keywords:

1. Nuclear medicine

2. SPECT vs PET
3. Nuclear imaging
4. Gamma camera
5. Positron emission
6. radionuclide
7. Medical physics

Results and Discussion

Both SPECT and PET are available in preclinical and clinical practice, enabling whole-body imaging with unlimited signal detection. Both techniques offer higher sensitivity and accuracy compared to CT and MRI [10].

A. Comparison factors

The main differences between the two techniques lie in the type of radiotracer and the method of signal detection, as well as the conversion to 3D imaging. These differences are the sources of the different image quality. In some aspects of the clinical application, PET is superior to SPECT; however, the latter has the majority of the advantages. Table 2 summarizes the main comparison elements between the two techniques. For instance, the ideal radiotracers for SPECT are those radionuclides with a half-life in the range of hours.

Table 2 Comparison between SPECT and PET

Feature	SPECT	PET
Decay Type	Gamma Emission	Positron emissiom follwing anihillation
Photon Energy	70 – 364 keV	511 keV
Detection mechanism	Collimator and Gamma camera	Positron camera
Radiotracer half-life	hrs	minutes
Comon Radiotracer	Tc-99m, I-123	F-18
Scanning Time	Long (20minutes-1hour)	Short (several minuties) 20 min as maxium
Image quality	Meduim	High
Cost	Less expensive	Expensive

Both physical effects and deficiencies of the equipment influence the accuracy and resolution of imaging. The spatial resolution of SPECT is limited by several factors, including detector thickness, collimator geometry, patient-to-collimator distance, the readiotracer energy, and the size and density of the patient's organ or body being imaged. However, the sensitivity is limited mainly by collimators.

B. Future prospective

Various developments are being implemented to enhance the performance of modalities. The release of a new generation of PET and SPECT is discussed in multiple studies, both in terms of hardware (for instance, material and size of the crystal or collimators) and software (for example, imaging reconstruction). Complication factors are improved by addressing new ideas to overcome any restrictions.

Moreover, the recent prospects target new ideas. Hybrid imaging is one such idea. Where combining SPECT or PET with the conventional CT or MRI. The combined system can provide both functional and anatomical imaging. PET-CT has received wide clinical acceptance, particularly in oncology practice [5]. Due to its favorable outcomes, scanner manufacturers have been encouraged to replace PET scanners with PET-CT units. SPECT-CT doesn't follow the same trend despite the increasing number of installed units [11], [12].

Artificial Intelligence (AI) is a promising tool for imaging. The next generation of experts must deal with one of the most powerful tools [13]. Recently, machine learning algorithms have been employed in simple imaging techniques, such as ultrasound [14]. However, applying the same concept to nuclear medicine imaging requires additional time because it necessitates a considerable amount of input data, which is not yet organized across different institutions and organizations worldwide [5].

Conclusion

This review is the comparative analysis of Single Photon Emission Computed Tomography (SPECT) and Positron Emission Tomography (PET), with the emphasis on the complementary nature of nuclear medicine in the modern medical diagnostics. In the analysis, the physics underlying each of the two imaging modalities are shown: the detection of individual gamma emission, versus the detection of two coincidence gamma emission as a result of the annihilation process, the differences between the two instrument setups that cause the formation of images of different quality of imaging.

PET is the better modality, in respect of spatial resolution, sensitivity, and accuracy. These benefits make it the leading request in clinical applications that must have high accuracy. Nonetheless, the modality that is used more is SPECT due to multiple reasons. The first advantage other than the accessibility and cost-effective nature is that radiotracers can be synthesized in the hospital without the need of a cyclotron to manufacture the synthetic.

It should be noted that the two modulities are complementary and not competitive. In order to identify the modality that is to be used clinically, there are a number of factors that should be put into consideration. The remaining essential issues are the level of quantification which is needed, availability of radiotracers and financial considerations.

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