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Intrinsic spatial resolution limitations due to differences between positron emission position and annihilation detection localization

Limitações da resolução espacial intrínseca devido às diferenças entre a posição da emissão do pósitron e a detecção da localização de aniquilação

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Abstract

Since its successful implementation for clinical diagnostic, positron emission tomography (PET) represents the most promising medical imaging technique. The recent major growth of PET imaging is mainly due to its ability to trace the biologic pathways of different compounds in the patient's body, assuming the patient can be labeled with some PET isotope. Regardless of the type of isotope, the PET imaging method is based on the detection of two 511-keV gamma photons being emitted in opposite directions, with almost 180º between them, as a consequence of electron-positron annihilation. Therefore, this imaging method is intrinsically limited by random uncertainties in spatial resolutions, related with differences between the actual position of positron emission and the location of the detected annihilation. This study presents an approach with the Monte Carlo method to analyze the influence of this effect on different isotopes of potential implementation in PET.

Keywords: nuclear medicine imaging, PET, Monte Carlo simulation.

Resumo

Desde sua implementação bem sucedida, a tomografia por emissão de pósitrons (PET) representa uma das técnicas de imagem mais promissoras para diagnóstico clínico. O grande crescimento recente da imagem por PET é principalmente devido à sua capacidade de rastrear o caminho biológico de diferentes compostos no corpo do paciente, assumindo que o paciente possa ser marcado com algum isótopo PET. Desconsiderando o tipo de isótopo, o método de imagem por PET é baseado na detecção de dois fótons gama de 511 keV, sendo emitidos em direções opostas, com quase 180° entre eles, como consequência da aniquilação do par elétron-pósitron. Desta forma, este método de imagem é intrinsicamente limitado pelas incertezas aleatórias na resolução espacial relacionada às diferenças entre a posição real de emissão do pósitron e a localização da aniquilação detectada. Este estudo apresenta uma abordagem pelo método Monte Carlo para estudar a influência deste efeito para diferentes isótopos com potencial implementação em PET.

Palavras-chave: imagem em medicina nuclear, PET, simulação Monte Carlo.

Introduction

Positron emission tomography (PET) is one of the more important nuclear medicine imaging techniques being currently used. It is actually considered to have the capability to change the whole impact role of nuclear medicine; not because it does everything better than conventional single photon emission imaging like SPECT, but because it has the impact and public relations of the fastest growing diagnostic specialty[1]. Nowadays, PET is a powerful imaging technique which utilizes almost exclusively ^{18}F tracer agents, like fluorodeoxyglucose (FDG) to infuse patient in order to produce three-dimensional (3D) images of functional processes in the body. The imaging system is based on the detection of the pairs of gamma rays emitted indirectly by a positron-emitting radionuclide tracer, which

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is introduced into the body on a biologically active molecule[1,2]. Tracer concentration images can be acquired in three-dimensional spaceat different times, therefore constituting a four-dimensional technique. Images are acquired within the body and they are further reconstructed by computer analysis. Modern scanners accomplish dual single-photon emission computed tomography/computed tomography (SPECT-CT) or PET/computed tomography (PET-CT) acquisition in the same procedure.

The most significant fraction of electron-positron decays result in two 511-keV gamma photons being emitted at almost 180º to each other; hence becoming possible to localize their source along a straight line of coincidence (LOR). In practice, the LOR has a finite width, as the emitted photons are not exactly 180º apart. Therefore, employing detectors having high enough time resolution, it becomes possible to localize the event to a segment of a chord, whose length is determined by the detector timing resolution. In this sense, improving time resolution may obtain better signal-to-noise ratio (SNR); and therefore requiring fewer events to achieve the same image quality[1].

Different radionuclides may be appropriate for PET scanning. However, isotopes having short half-life are typically used[1], as reported in Table 1.

One of the most relevant features of PET imaging techniques is its capability to trace the biologic pathway of different compounds within patient, provided it can be radiolabeled with some PET isotope. Therefore allowing to perform almost any kind of specific processes[1,3]. Actually, great efforts are devoted to research and characterization of radiotracers for new target molecules.

The potentiality of new radiotracers is determined by many different factors, including costs and complexity for its production as well as efficiency performance for specific target imaging. Therefore, as a consequence of the imaging mechanism based on the detection of the pair of annihilation gamma rays, it results in intrinsic spatial resolution uncertainties associated with the annihilation localization, which may differ from the actual positron emission position. This effect should be added to others, like detection system, electronic noise and image reconstruction algorithms and eventually patient motion, in order to quantify all the components contributing to the total spatial uncertainty.

The impact of the positron flight on spatial resolution has been recently analyzed by different authors. Studies have been conducted experimentally[4,5], through theoretical calculations[6] or by Monte Carlo methods[7,8]. Actually, Sánchez-Crespo et al.[7] investigated the influence of positron distance of flight in various human tissues on the spatial resolution in PET for positrons from different radioisotopes.

However, it can be demonstrated that almost all cases can be approximately described by positrons travelling in water.

This work presented investigations about the cloud of annihilation points around different positron sources in water performed with the aim of studying and characterizing the intrinsic spatial resolution limitations due to uncertainties arising from differences between positron emission position and actual annihilation localization. Different isotopes of potential use in PET (Table 1) have been investigated, disregarding other properties, like production reliability and practical reasons for utilization convenience.

Table 1. PET radionuclides half-life.

Isotope	Approximate half-life [minutes]
^{11}C	20
^{13}N	10
^{15}O	2
^{18}F	110

Materials and methods

A full stochastic Monte Carlo technique has been developed in order to be the start point for the study of the influence to spatial resolution arising from uncertainties due to differences between positron emission position and annihilation localization. Specific subroutine has been developed, based on the PENELOPE v. 2008 main code in order to simulate a point source isotropically emitting positrons with energy distribution, according to the actual emission properties of each radioisotope. The computer code system PENELOPE v. 2008[9] performs Monte Carlo simulation of coupled electron, positron and photon transport in arbitrary materials, with energy ranging within 10^2 to 10^9 eV. Charged particles (electrons and positrons) are simulated by means of a mixed procedure consisting of dividing detailed simulation for “hard” events, while implementing a condensed approach for “soft” events. The distinction between soft and hard events is determined by user-defined thresholds regarding angular deflection and energy loss in the interaction. The PENELOPE code has been largely applied or different applications on nuclear medicine, including imaging as well as therapy techniques[10,11].

The PENELOPE v. 2008 distribution includes specific packages dedicated to material file creation by means of physical properties included in internal database along with suitable analytical models. In addition, there is the PENGEOM package exclusively devoted to handle user-defined simulation geometry in base on quadric surface approach.

With the aim of performing suitable characterization of positron transport within aqueous media a specific and dedicated simulation code has been developed. This subroutine package allows computing complete full stochastic positron transport, taking into account all radiation interaction mechanisms by means of mixed particle tracking approach. The considered interaction events are soft events, hard elastic collisions, hard inelastic collisions,

Bremsstrahlung emission, inner-shell impact ionization, annihilations and delta ray interactions. The corresponding water cross-sections extracted from PENELOPE databases are reported in Figure 1.

Therefore, once positron cross-sections are already established, it becomes necessary to study the corresponding penetration distances, which are strongly correlated with particle range and, of course, the mean travelled distance between consecutive collisions, defined as mean free path (MFP), usually called l.

Figure 2 presents the corresponded ranges and MFP obtained from PENELOPE cross-sections database as a function of positron kinetic energy.

In this work, different PET radioisotopes (^{11}C, ^{13}N, ^{15}O and ^{18}F) have been considered to investigate the effect of annihilation localization uncertainties. The considered radioisotopes emission spectra have been extracted from validated database[3] and they are reported in Figures 3 and 4.

The simulation geometry used to perform these investigations considered an isotropic homogeneous medium of water equivalent material extended within a 100-mm radius sphere.

In order to assess mean traveled distance before positron annihilation, it is not necessary to consider the whole imaging system. However if a complete description about

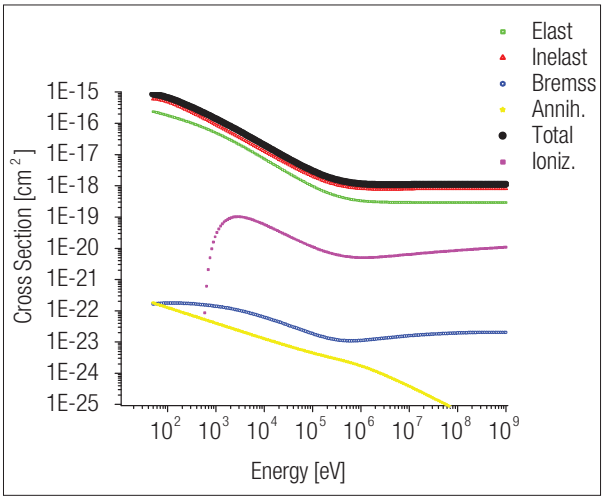

Figure 1. Water cross-sections for positrons: elastic (green), Inelastic (red), Bremsstrahlung (blue), annihilation (yellow), inner-shell ionization (magenta) and total (black) extracted from PENELOPE database, according to the Bethe formalism in the Born approximation.

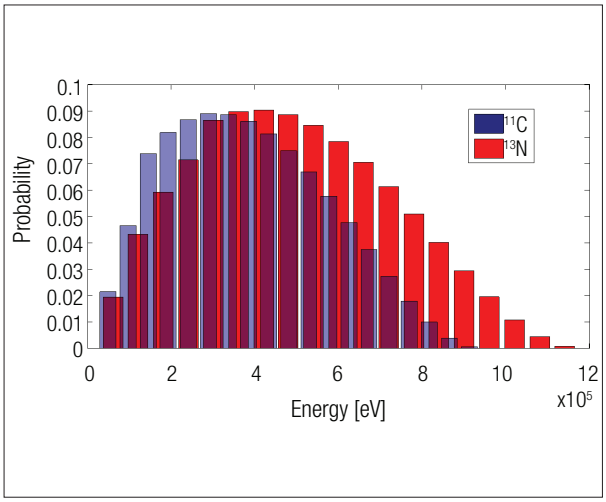

Figure 3. ^{11}C and ^{13}N positron emission spectra used for Monte Carlo simulations. Emission spectra are reported as normalized emission intensity probability per energy channel.

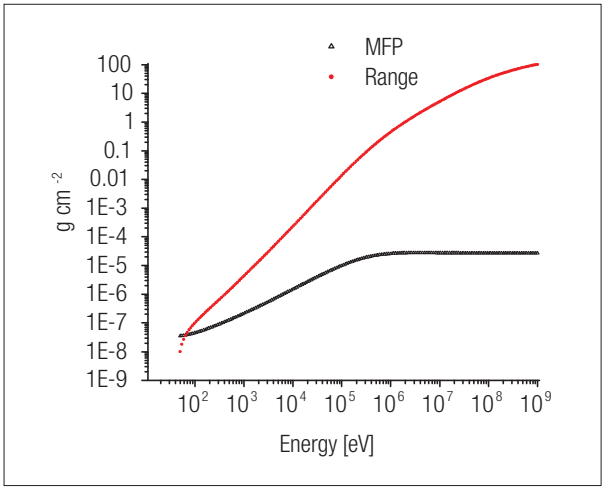

Figure 2. In water ranges (red circle) and mean free path (MFP – black triangle) calculated using PENELOPE cross-section database.

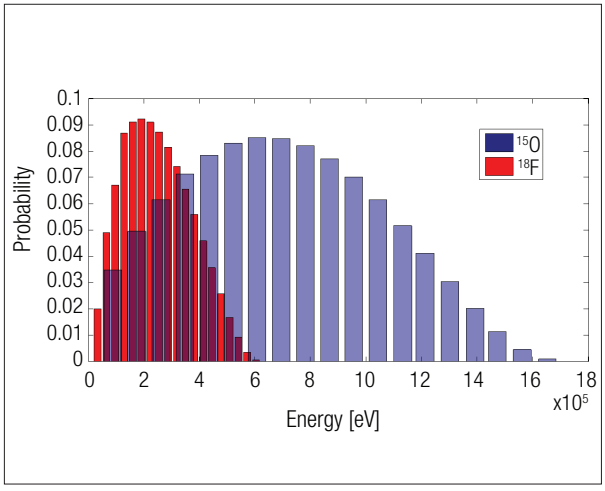

Figure 4. ^{15}O and ^{18}F positron emission spectra used for Monte Carlo simulations. Emission spectra are reported as normalized emission intensity probability per energy channel.

PET imaging spatial resolution would be the goal of the study, it would be mandatory to consider the complete imaging system including specific phantom/patient geometry, mass distribution, isotope activity and distribution and of course collimation and detection devices.

Results and discussion

It is noticeably that even when positron range increases continuously with energy, there is a remarkable plateau for positron MFP at energies greater than 1 MeV (Figure 2), approximately. This threshold is in correspondence with the stabilization plateau for the total cross-section, as expected.

As mentioned, isotropic point source has been placed at the origin of Cartesian coordinates and the developed program allowed to determine the annihilation position for different monoenergetic positron sources or emission spectra.

Figures 5A and 6A show examples of the 3D representation of annihilation positions for 10^6 primary showers per run obtained considering a typical PET radioisotopes (^{15}O and ^{18}F). Once, annihilation localizations have been already determined, it becomes straightforward to calculate the travelled path distribution as the distance from origin to annihilation localization, as shown in Figures 5B and 6B. This study has been performed for different radioisotopes and for a wide range of monoenergetic sources - some of the obtained results are reported in Table 2.

The obtained results show reasonable trends when comparing with the corresponding emission spectra.

As expected, the behavior of the obtained results as a function of the energy seems to be in good agreement with the corresponding mean ranges weighted according to the emission spectra, which may be calculated from analytical

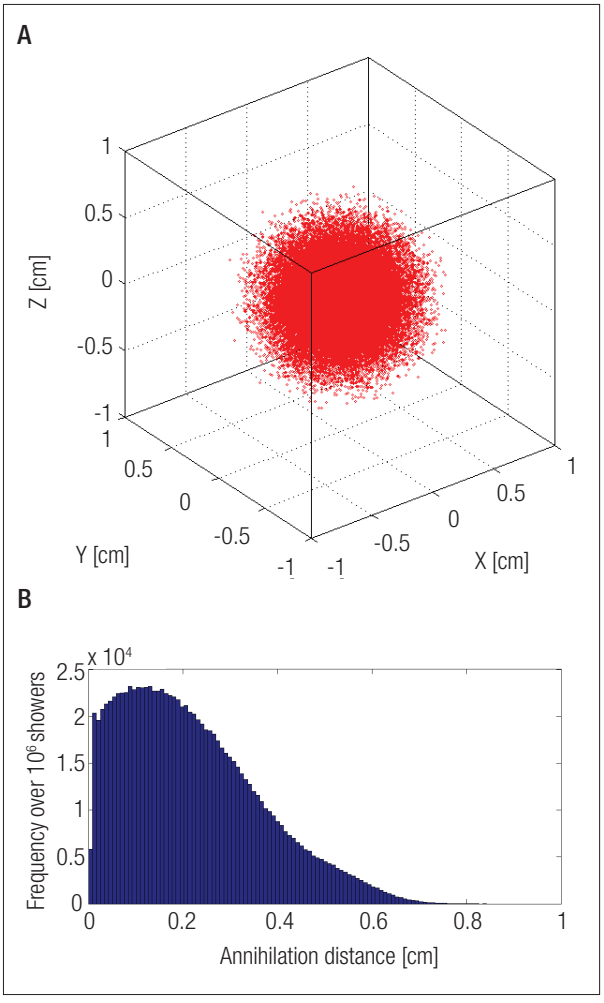

Figure 5. Three-dimensional representation of in-water annihilation localization for 10^6 primary ^{15}O positrons isotropically emitted from a point source at the origin (A) and the histogram of the corresponding traveled path distribution of annihilation localization (B).

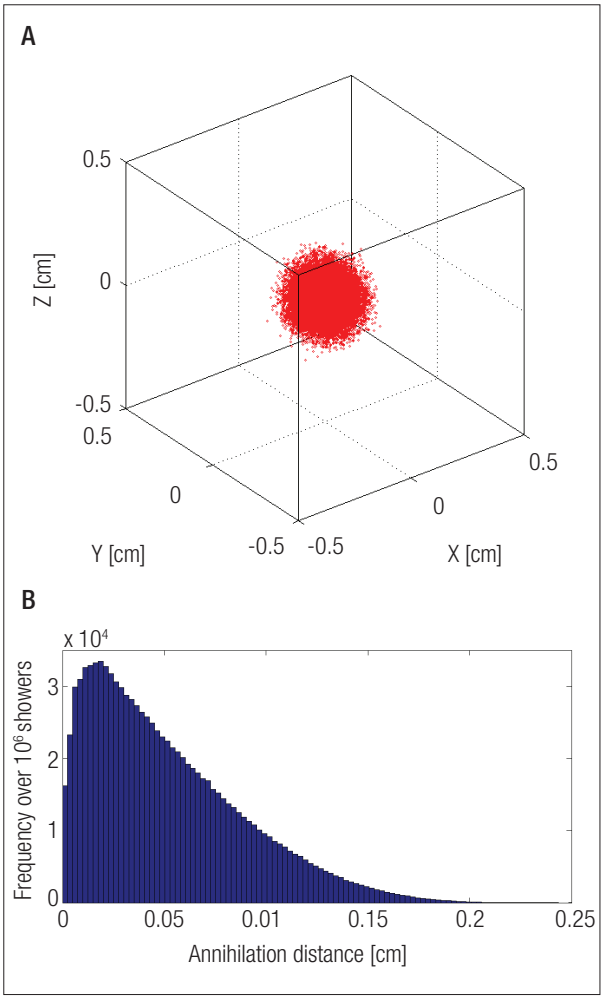

Figure 6. Three-dimensional representation of in-water annihilation localization for 10^6 primary ^{18}F positrons isotropically emitted from a point source at the origin (A) and its corresponding travelled path distribution of annihilation localization (B).

Table 2. Monoenergetic e^+ source and PET radionuclides in-water mean path.

Isotope	Mean value (cm)	SD (cm)	Experimental Data (cm)[4]
50 keV	0.0029	0.0008	
100 keV	0.0097	0.0024	
1 MeV	0.30	0.06	
^{11}C	0. 097	0.06	0.111
^{13}N	0.14	0.09	0.142
^{15}O	0.22	0.14	0.149
^{18}F	0.52	0.038	

SD: standard deviation.

methods or obtained from standard databases[9,10]. Greater differences between emission and annihilation positions correspond to higher energies or harder spectra. Along with practical features, like product costs and reliability, this intrinsic limitation may be pointed out and eventually taken into account when evaluating the potentiality and relative convenience of the different radioisotopes.

As reported in Figures 5 and 6, it is clear that the MFP distribution of emitted positrons does not exhibit Gaussian trend. The obtained Poisson distribution may be main reason for contributing to differences between positron emission position and annihilation localization. In this sense it results convenient to employ stochastic approaches unlike deterministic analytical models.

Conclusions

A suitable method for investigating the intrinsic limitations to PET spatial resolution due to differences between emission and annihilation positions has been proposed. A dedicated Monte Carlo subroutine has been developed for this purpose. As reported in the presented results for static emission sources, intrinsic uncertainties due to differences between emission and annihilation positions may actually arise to non-negligible limitations for the spatial resolution. However, this effect may be even more significant when considering dynamic emission sources, as may be the case of organ motion within patients. Actually, efforts are being dedicated to the development of time-dependent analogue algorithm, for the simulation of moving sources, in order to assess the influence of this effect in a more realistic clinical configuration.

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