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Design study of a cardiac-dedicated PET system

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ABSTRACT

The work studies the feasibility of developing a cardiac-dedicated PET system using Monte Carlo simulation tools. The proposed system comprises a dual-panel geometry and both panels utilize fine crystal elements (~2–3 mm) to help improve high spatial resolution. The system performances were studied with respect to photon detection sensitivity, count-rate performances, spatial resolution, depth-of-interaction (DOI) capability, and potential contrast recovery improvement due to time-of-flight (TOF). For a 2:3 dual panel configuration (front panel: 24.0×18.0 cm^2 and back panel: 36.0×27.0 cm^2), the system is able to achieve a peak photon sensitivity of ~13.0% at the location of the heart. For count-rate performances, the system is able to yield a continuous increase of noise equivalent count rate (NECR) as a function of the total activity of radiotracers between 10 mCi and 50 mCi. When crystal elements of $2 \times 2 \times 20$ cm^3 are deployed in the front panel and crystal elements of $3 \times 3 \times 20$ cm^3 are deployed in the back panel, the proposed system is not only able to revolve 2 mm diameter spheres in the image slices parallel to the panels but also demonstrates good resolution uniformity of less than 5%. On the other hand, significant resolution degradation occurs in the direction perpendicular to the panels due to the following two factors: limited angular coverage and finite DOI resolution. Such degradation was quantitatively analyzed for five different DOI resolutions (0 mm, 2.5 mm, 5 mm, 10 mm and 20 mm), as well as three different crystal configurations. Finally, the contrast study with a heart-like phantom (comprising lumen, aortic wall, myocardium and small lesions) indicates that TOF capability (time resolution: 250–500 ps) does not significantly improve image quality and lesion detectability, in terms of contrast-to-noise ratio (CNR).

1. Introduction

Positron Emission Tomography (PET) is a widely used imaging modality in diagnosis and prognosis of heart diseases [1–5]. Back to the 1980s, its initial applications were mainly in myocardial perfusion imaging and viability studies [1,2]. Its uses have recently been extended to the detection of vulnerable plaques in atherosclerosis [6,7]. Whole body PET scanners are routinely used in these applications and are able to achieve a photon detection sensitivity of ~2–3% at the center of field-of-view (FOV) and a spatial resolution of ~4–8 mm [8], largely limited by the large-ring geometry (~70–80 cm bore size) and coarse crystal elements deployed. These two limitations pose several challenges in cardiac PET imaging.

For instance, in myocardium perfusion imaging and viability studies, the evaluation of the inferior myocardial wall is commonly degraded by the interferences from nearby extra-cardiac structures such as the liver or stomach wall, due to both partial volume and spillover effects [9,10]. Intense uptake in these organs can overlap with the adjacent heart wall and consequently reduce the ability to accurately assess the true myocardial uptake. A PET system with improved spatial resolution is expected to be less affected by this challenge and to provide more accurate quantitative measurements. In addition, for the detection of vulnerable plaques in atherosclerosis, several studies using whole body PET scanners have suggested that the relatively poor spatial resolution is a major limitation for detecting small plaques in the coronary arteries [7,11,12]. This is not only because coronary arteries themselves are of very small size (~2 mm or less), but also because the myocardium can have significant background uptake which results in interferences with the aortic walls and thus degrades image contrast. A PET system with improved spatial resolution, photon sensitivity and contrast recovery has a great potential to address such a challenge.

Over the past several years, there has been an increasing push in the medical imaging community to develop organ-dedicated PET systems. These systems tend to be easier to use than whole-body imaging systems, require less physical space and are typically less expensive, yet they provide improved imaging performances to larger, much more expensive imaging systems. For instance,

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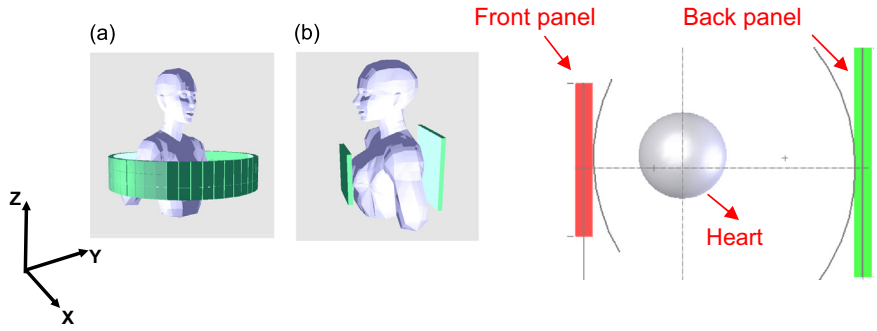

Fig. 1. Illustration of a large ring PET system (a) and a dual-panel cardiac-dedicated system (b). The separation between the two panels is adjustable for patients of different sizes. Two panels (or partially curved) can have different areas and/or crystal dimensions.

systems such as the GE Molecular Breast Imaging and the Naviscan PEM systems have been developed in an effort to provide optimal imaging for breast cancer. Cardiac-specific SPECT scanners have also been developed to optimize the way the cardiac perfusion imaging is performed [13]. Commercial cardiac-dedicated PET systems are currently available through Positron Corporation. While focusing on achieving a high photon detection sensitivity, the system performances suffer from two limitations: poor time resolution (due to the use of BGO crystal materials) and poor spatial resolution (crystal width: 8.5 mm). One important question leads to the study in this paper is whether a cardiac-dedicated PET system of improved photon detection sensitivity and improved spatial resolution, can be used to enhance the role of PET in cardiac imaging applications described above.

2. Methods

2.1. System description and Monte-Carlo simulation

The proposed dual-panel system is shown in Fig. 1. It comprises two detector panels that can be placed in close proximity to the heart. The front panel is placed adjacent to the chest wall, and the back panel is placed against the patient's back. For simplification, three assumptions made in the simulation are as follows: (1) the separation between the front panel and the back panel is 32 cm; (2) the distance from the center of the heart to the front panel is half of the distance to the back panel; and (3) the heart locates within a rectangular cube (size: $12.0 \times 9.0 \times 6.0$ cm^3). Two different ratios of panel areas (2:3 and 2:4) between the front panel and the back panel were studied and indicated as A:B. For example, a panel configuration of "2:3" indicates a front panel with $2\times$ cross-sectional area of the heart (24.0×18.0 cm^2) and a back panel with $3\times$ that area (36.0×27.0 cm^2).

GATE, a photon transport simulation tool, was used for the study [14]. To reduce simulation time, only back-to-back 511-keV photon pairs were simulated. The dead-time configuration of the proposed system was set to be 200 ns (non-paralyzable). Scintillation material was selected to be LYSO (cerium-doped lutetium yttrium orthosilicate). For the dual-panel system configurations, the height of crystal elements was fixed at 20 mm and a gap of 50 μm width between adjacent crystal elements (i.e., optical reflectors) was included. For comparison, a standard cylindrical PET system (referred to as "large ring") was also simulated (Fig. 1), based on a Siemens Biography Accel system (84.2 cm diameter, axial FOV: 21.6 cm). Each detector module comprises a 13×13 LYSO crystal array and the dimension of each crystal element is $4 \times 4 \times 25$ mm^3. The detector energy resolution was selected to be 12% full-width-at-half-maximum (FWHM) at 511 keV and the coincidence time resolution was selected to be 1 ns FWHM, based upon the experimental measurements [15].

2.2. Coincidence photon sensitivity

A point source of 100 μCi was translated from the center to the edge of the FOV (see Fig. 1 for definition of the axes). The energy window and time window settings were set to be twice the energy resolution and time resolution values in Section 2.1, respectively. The sensitivity is defined as the ratio between the total number of coincidence events detected and the total number of 511 keV photon pairs generated in the GATE.

2.3. Count rate performances

Noise equivalent count rate (NECR) [16] as defined in formula (1), was studied in this section.

$$NECR = \frac{T^2}{T+S+R} \quad (1)$$

where T, S and R are the count rates of true, scatter and random events, respectively. The details regarding how to identify the types of events were provided in the previous work [17]. A simplified digital phantom was used, comprising an upper-body torso enclosing the heart compartment (Fig. 1). The activity concentration in the torso was set to be 0.1 μCi/cm^3 (uniform distribution) and the activity concentration ratio was set to be 10:1 between the heart and the torso. A layer of lead shielding (4 cm thick) was placed on the four outer faces of each panel (except the two outer faces parallel to the XZ plane). As with a clinical whole body PET scanner, such lead shielding is required to help reject those 511 keV photons originating from other organs outside of the FOV, such as brain and bladder. Otherwise, the system would suffer from an increased number of random events and thus a lower NECR.

The dependency of NECR on various energy window (6–48%) and time window (1–12 ns) settings was studied. In clinical cardiac imaging, the total injected activity varies for different radiotracers (i.e., ~10 mCi for FDG and N-13 perfusion, ~50 mCi for the ^{82}Rb). For example, a higher activity is required for ^{82}Rb due to its very short half-life (76 s). As a result, the peak NECR was studied as a function of the total activity ranging between 2.5 mCi and 100 mCi. Only the 2:3 panel configuration was tested in this step.

2.4. Reconstructed sphere resolution

For the resolution study, spheres of different diameters (1–4 mm) were set up in a 4×4 array configuration within each quadrant of a slice parallel to the XZ plane. The distance between the centers of two adjacent spheres was twice the diameter. Within each quadrant, three regions were selected from the center towards the edge of the