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PET performance and MRI compatibility evaluation of a digital, ToF-capable PET/MRI insert equipped with clinical scintillators

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Abstract
We evaluate the MR compatibility of the Hyperion-II\textsuperscript{13} positron emission tomography (PET) insert, which allows simultaneous operation in a clinical magnetic resonance imaging (MRI) scanner. In contrast to previous investigations, this work aims at the evaluation of a clinical crystal configuration. An imaging-capable demonstrator with an axial field-of-view of 32 mm and a crystal-to-crystal spacing of 217.6 mm was equipped with LYSO scintillators with a pitch of 4 mm which were read out in a one-to-one coupling scheme by sensor tiles composed of digital silicon photomultipliers from Philips Digital Photon Counting (DPC 3200-22). The PET performance...
degradation (energy resolution and coincidence resolution time (CRT)) was evaluated during simultaneous operation of the MRI scanner. We used clinically motivated imaging sequences as well as synthetic gradient stress test sequences. Without activity of the MRI scanner, we measured for trigger scheme 1 (first photon trigger) an energy resolution of 11.4% and a CRT of 213 ps for a narrow energy (NE) window using five $^{22}$Na point-like sources. When applying the synthetic gradient sequences, we found worst-case relative degradations of the energy resolution by 5.1% and of the CRT by 33.9%. After identifying the origin of the degradations and implementing a fix to the read-out hardware, the same evaluation revealed no degradation of the PET performance anymore even when the most demanding gradient stress tests were applied. The PET performance of the insert was initially evaluated using the point sources, a high-activity phantom and hot-rod phantoms in order to assess the spatial resolution. Trigger schemes 2–4 delivered an energy resolution of 11.4% as well and CRTs of 279 ps, 333 ps and 557 ps for the NE window, respectively. An isocenter sensitivity of 0.41% using the NE window and 0.71% with a wide energy window was measured. Using a hot-rod phantom, a spatial resolution in the order of 2 mm was demonstrated and the benefit of time-of-flight PET was shown with a larger rabbit-sized phantom. In conclusion, the Hyperion architecture is an interesting platform for clinically driven hybrid PET/MRI systems.

Keywords: ToF-PET, MRI, PET/MRI, PET/MRI interference, PET performance, MR compatibility, digital SiPM

(Some figures may appear in colour only in the online journal)

1. Introduction

Positron emission tomography (PET) is a powerful tool in nuclear medicine, enabling e.g. the visualization of metabolic processes. As a tracer-based imaging modality, PET images contain only limited anatomic information. This complementary information is commonly added by combining PET with computed tomography (CT). After having introduced the first combined PET/CT systems in the late 1990s, fully integrated solutions are nowadays prevailing in clinical practice (Beyer and Townsend 2006). In spite of this development, a combination of PET with magnetic resonance imaging (MRI) was first discussed in the 1990s and gained interest over the years. In contrast to CT, MRI provides a decent soft-tissue contrast and avoids the administration of an additional radiation dose as MRI makes use of static (the so-called magnetic $B_0$ field) and time-varying magnetic fields to acquire image information. Furthermore, MRI offers a wide range of different contrast mechanisms emphasizing a potential benefit of PET/MRI over PET/CT. However, in contrast to PET/CT, the integration of a PET and MRI system is a complex task as the extensively used magnetic fields require a higher level of compatibility. The associated induction effects or occurrences of Lorentz forces might either lead to degradation effects on the PET side or on the MRI side. Whereas for the latter, typical problems range from $B_0$ distortions over signal-to-noise ratio (SNR) losses to spatial encoding disturbances, the PET system and its sub-components might show a sensitivity for magnetic fields resulting in either a malfunction or a performance degradation. Several groups reported such kind of performance degradation when e.g. radio frequency (RF) or gradient pulses are applied during PET operation (Schlyer et al 2006, Weirich et al 2012): for instance, Weirich
et al (2012) reported count-rate losses associated with the application of gradient-intense MRI sequences.

Despite these potential risks, a couple of different research systems were developed in recent years whereby the integration topology changed over time: starting from a ‘separate system approach’, in which the scintillation crystals are separated by exploiting optical fibers from the photomultiplier tubes operated outside the magnetic field because of their strong magnetic field sensitivity, a higher level of integration was achieved with the utilization of solid-state photon detectors. Devices of such kind allow an operation inside the strong magnetic field of the MRI scanner and thus enable a higher level of integration. Avalanche photodiodes (APD) as utilized in Pichler et al (2006), Schlyer et al (2007) and Delso et al (2011) and more recently Silicon Photomultipliers (SiPM) (see e.g. Schulz et al (2009), Hong et al (2013), Weissler et al (2014), Weissler et al (2015)) are mainly chosen in this regard. Apart of the choice of photon detection technology, the downstream read-out platforms differ from its topology and integration level. In contrast to more conservative approaches in which most of the electronics are installed outside the MRI system, our group presented a more aggressive approach (Weissler et al 2014, 2015): here, the digitization takes place directly inside the MRI bore and thus reduces the need for analogue signal transmission over long routing and hence minimizes the risk of signal quality degradation related to RF pulses or switching gradients. Furthermore, in a second generation, the utilized analogue SiPMs of the Hyperion-I scanner were replaced by a digital version, presented in 2009 by Philips Digital Photon Counting (PDPC). These so-called digital SiPM (dSiPM) devices digitize the signal directly on sensor level by counting the number of micro-cell (also called single photon avalanche photodiode (SPAD)) breakdowns, meaning that no analogue signal has to be transferred to separate digitization-performing electronics such as an application specific integrated circuit (ASIC). A potential downside of the highly integrated approach is the presence of most of the electronics inside the MRI system’s field-of-view (FoV), thus intensifying MR compatibility issues. In a previous study, we assessed the MR compatibility of our PET architecture in detail on the MRI as well as on the PET side: we found a decent performance on both imaging modalities. However, a sensitivity of our PET electronics to gradient switching resulting in a degradation of the energy and timing performance (coincidence resolution time, CRT) was revealed. As this study was performed with a detector configuration suitable for preclinical imaging (1 mm crystal pitch, light guide, preclinical configuration, figure 1(a)) which does not allow for the best possible timing performance achievable with our platform, we tested the influence of the gradient switching on a test setup equipped with a detector employing a clinical scintillator configuration (4 mm crystal pitch, one-to-one coupled, clinical configuration, figure 1(b)) using two PET modules in coincident mode (Schug et al 2015a). Due to the highly improved CRT of about 240 ps, we found a degradation of about 30%. Moreover, a test with the PET modules at a position which is clinically more relevant, namely closer to the cover of the MRI bore and thus closer to the gradient coils, showed a degradation of about 52%. As the PET platform with its good timing performance is interesting for clinically driven PET systems, we extended this gradient compatibility study in three ways: first, in the building phase of a copy of the Hyperion-II scanner with the preclinical crystal configuration, we built as an intermediate step a prototype PET ring with the clinical crystal configuration. Second, we tested the impact of the gradient switching on this prototype using the synthetic gradient test protocols as presented in Wehner et al (2015). As the degradation of the energy and timing resolution can be associated with an instability of supply voltages, we equipped the prototype scanner with an improved version of the interface (IF) board (building the interface between a sensor tile and the PET module’s main PCB) yielding a more stable voltage supply in a third step. This modified hardware was evaluated in the same way and compared with the original version.
In addition, we extended the scope of this work and also evaluated the PET performance of the prototype scanner in more detail: here, we acquired a sensitivity profile, tested the configuration in a high-activity scenario, and measured the energy and timing performance for different sensor configurations. We performed imaging experiments using a hot-rod phantom to evaluate the spatial resolution and conducted a demonstration experiment using a rabbit-sized phantom to show the time of flight (ToF) capabilities of our PET platform. Since the clinical demonstrator existed only for a short period of time, only an initial PET performance evaluation was conducted. A full characterization according to NEMA NU 2-2012 (clinical) or NEMA NU 4-2008 (preclinical) was not possible.

2. Materials

2.1. Hyperion-II PET Insert

The Hyperion-II\textsuperscript{D} platform equipped with detectors in the preclinical configuration (figure 1(a)) was described in detail in Weisser\textit{et al} (2015). In Schug\textit{et al} (2015b), we gave a detailed overview of the PET scanner components which are relevant for gamma detection. The clinical crystal configuration used in this work (figure 1(b)) was described before in Schug\textit{et al} (2015a). Therefore, we will only briefly summarize the scanner configuration used in this work.

\textbf{Figure 1.} Sketches of the detector stack configurations. (a) The preclinical detector configuration employs a lightguide between the LYSO scintillator and the sensor tile. (b) The clinical detector configuration used in this work uses a one-to-one coupling of crystals to readout channels.
The detector stack consists of a pixelated scintillation crystal array of $8 \times 8$ optically isolated cerium-doped lutetium yttrium orthosilicate (LYSO) crystals with a height of 10 mm and a pitch of 4 mm. Although thicker crystals with a height of about 20–30 mm are commonly used for clinical PET systems, 10 mm thick crystals were chosen for three reasons: firstly, we aimed for the maximum possible CRT performance and secondly, the system’s sensitivity was not the main focus of the investigation. Thirdly, thicker crystals intensify the parallax error in the event positioning. With a rather small system bore (see below), crystals with a thickness commonly chosen for clinical whole-body PET systems with a much larger bore would therefore not be an appropriate choice.

To read out the scintillation light, the crystal array is glued to a sensor tile which is the same as presented in our previous works. The sensor tile is $32.6 \times 32.6 \text{ mm}^2$ in size and is made up of 16 Digital Photon Counters (DPC 3200-22) sensors from PDPC with $2 \times 2$ read-out channels (also referred to as pixels) (Degenhardt et al. 2009, Frach et al. 2009, Degenhardt et al. 2010, Frach et al. 2010) resulting in $8 \times 8$ readout channels per sensor tile each one-to-one coupled to a single crystal.

The 3200 single-photon avalanche diodes (SPADs) per DPC pixel can be deactivated individually in order to reduce the overall dark count rate of the sensor. A trigger scheme can be set between 1–4 and requires $1, 2.33 \pm 0.67, 3.0 \pm 1.4$ and $8.33 \pm 38.0$ mean number of SPAD breakdowns, respectively, per pixel in order to generate a trigger. Low trigger schemes result in a better timing performance but are more prone to thermally induced SPAD breakdowns which may result in a higher dead time and lower sensitivity. The row-trigger-line refresh feature can help to reduce the dark noise induced triggers and is used throughout this work (for details and explanation see Schug et al. (2015b) and Marcinkowski et al. (2013)). When the trigger condition is met, a time stamp is generated and causes the DPC to enter a validation phase of programmable length. The configurable validation scheme determines the threshold of SPAD breakdowns which has to be reached during the validation phase of programmable length for a trigger to be validated, the integration phase to be started and a DPC hit to be read out. Details of the different trigger and validation schemes and a comprehensive description can be found in Philips Digital Photon Counting (2014) and Tabacchini et al. (2014).

Each sensor tile is connected to an FPGA-based (field programmable gate array) control and read-out board (IF board) (Dueppenbecker et al. 2012) (figure 2(a)). The FPGA configures and reads out each individual DPC of the sensor tile. Besides that, the supply voltages, in particular the bias voltage defining the operating voltage of the SPADs, are controlled by this FPGA. As mentioned in the introduction, we also used an improved version of the IF board: an additional voltage regulator necessary to stabilize the voltage supply of the FPGA,
as well as an additional feed forward capacitor used to reduce a ripple on the bias voltage, were incorporated on the improved IF board design.

In contrast to previous works, we equipped only one of the three possible detector rings by mounting \(2 \times 1\) detector stacks (figure 2(b)) on a singles-detection module (SDM) (Weissler et al. 2012a) and used ten of these PET modules to build up the PET ring. The resulting PET FoV has an axial extension of 32 mm and a transaxial crystal-to-crystal spacing of 217.6 mm. All PET modules are connected to a synchronization unit, which distributes a reference clocking and trigger signals to each single PET module. The PET ring, the synchronization unit and the supporting infrastructure (e.g. mechanics, power supply) are mounted on a trolley which can be rolled over the MRI patient table (figure 3).

As already stated in the introduction, the clinical configuration as described here was an intermediate step created during the build-up of a fully populated preclinical PET insert.

2.2. MRI system

To test the influence of the switching gradients on this clinical prototype, we used a clinical Philips Achieva 3-T MRI system equipped with a Dual Quasar gradient system. This gradient system employs two gradient amplifiers which allows to increase the available gradient slew rate up to \(200 \text{ T m}^{-1} \text{s}^{-1}\). The MRI system was running software release 3.2.1.0 and we installed a system patch implementing the synthetic gradient test protocols as described in Wehner et al. (2015).

For MRI acquisition, we equipped the PET insert with dedicated PET-transparent radio-frequency transmit and receive (RF Tx/Rx) coils. We used mainly a small mouse coil (12-leg birdcage resonator, high-pass version) with an inner diameter of 46 mm. To measure a large rabbit-sized phantom, we installed a larger RF Tx/Rx coil with an inner diameter of 160 mm (Weissler et al. 2015).

2.3. Phantoms and test sources

We used two kind of radioactive PET phantoms: point sources as well as fillable phantoms.
The point sources were used to calibrate the PET scanner and to evaluate the basic PET performance parameters such as energy and timing resolution for the different sensor configurations applied. Up to five $^{22}$Na NEMA cubes with an edge length of 10 mm, an active area of 0.25 mm in diameter with an activity of 1.0–1.3 MBq were used.

We used three different phantoms which were filled with a $^{18}$F Fluorodeoxyglucose (FDG) solution. A tube with a diameter of 10 mm and a length of 25 mm was used to test the behavior of the system at high activities. Two custom-made structured phantoms served to investigate the spatial resolution and to demonstrate the ToF benefits of the system. To determine and investigate the spatial resolution of the system, data from a hot-rod phantom with several structured regions with rod diameters and spacings of 1.0 mm, 1.5 mm, 2.0 mm, 2.5 mm, 3.0 mm and 4.0 mm surrounded by an ring of activity with an outer diameter of 43 mm was acquired. To assess the benefits of ToF image reconstruction, we used a large rabbit-sized, high-contrast phantom with an outer diameter of 114 mm. The structured insert provided a regular rod structure. The rods were aligned on a Cartesian grid, had a diameter of 3 mm and a height of 20 mm (a sketch of the phantom can be found in Schug et al (2015c)). The center-to-center spacing of two adjacent rods were two times the diameter.

3. Methods

3.1. PET system configuration and processing parameters

The cooling temperature of the PET system was set to $-5 \, ^\circ\text{C}$ leading to a temperature of about 0–5 $^\circ\text{C}$ measured under operation on the sensor tiles.

For this operating temperature, we measured the breakdown voltage $V_{\text{break}}$ of each sensor tile whereas $V_{\text{break}}$ is defined as the bias voltage at which the SPADs start to breakdown. The bias voltage applied during operation is the sum of $V_{\text{break}}$ and an additional overvoltage $V_{\text{ov}}$. In our experiments, we used a conservative value of $V_{\text{ov}} = 2.5 \, \text{V}$. To reduce the dark noise of the sensor tile, we chose to disable 20% of the cells showing the highest dark count rates.

As a validation configuration of the DPCs, we applied the validation scheme 0x00:OR yielding a threshold of 52 photons. The validation length was set to 10 ns and the integration time was chosen to be 165 ns. The neighbor-triggering capabilities of the DPCs were not used (Schug et al 2012).

Pixel photon values were corrected for saturation and DPC hits were temporally clustered using a default cluster window of 5 ns. Exploiting the one-to-one coupling, the scintillating crystal was identified from the pixel with the highest photon count of the cluster. Afterwards, the energy was calculated using the photon sum of the four pixels of the DPC comprising the main pixel. To improve the time stamp performance and reject detector scatter and pile up events, a minimal light fraction of the main pixel of more than 60% of the total photon sum of all DPC pixels of a cluster is required. This decreases the system’s sensitivity.

A narrow energy (NE) window of 461–561 keV was used to obtain the best timing performance and a wide energy (WE) window of 250–625 keV was applied to yield a higher sensitivity. Those filters were applied before performing a coincidence search with a sliding coincidence window (CW). We used a CW of 5 ns for measurements acquired with low activities. For measurements using higher activities, we reduced the CW down to 500–1000 ps. Coincident events with more than two singles were discarded.
3.2. Computation of the energy resolution and CRT

The energy resolution was determined using the energy spectrum of accepted coincident singles. A Gaussian was fitted using a fit range of $-0.5$ to $1$ FWHM around the photopeak, whereby no background removal or modeling was performed. The FWHM used was determined by an initial, preliminary Gaussian fit.

The CRT was calculated by evaluating the FWHM of the histogram of the measured time differences corrected for the known source positions. As for the energy resolution, a Gaussian fit was performed in the fit range of $-0.5$ to $0.5$ FWHM around the peak. Again, no background modeling or removal was applied and the FWHM used for the fit range was determined using an initial, preliminary fit.

3.3. MRI compatibility investigations

The MRI compatibility of the Hyperion-II\textsuperscript{D} platform equipped with preclinical detector stacks was investigated in Wehner \textit{et al} (2014), Wehner \textit{et al} (2015), Weissler \textit{et al} (2015). Based on these results, the clinical scintillator configuration as described in section 2.1 was investigated using two PET modules each equipped with six detector stacks in coincidence in Schug \textit{et al} (2015a). In these works, we identified the interaction of the PET hardware with the gradient system, especially the $z$-gradient, as the main and almost solely interference phenomena on the PET side resulting in a PET performance degradation. Starting from those results, we evaluated in the present work the influence of the gradient system on the CRT and energy resolution using realistic imaging protocols as well as synthetic gradient stress tests. These protocols are described in more detail in Wehner \textit{et al} (2015).

To have the highest sensitivity for the CRT degradation under gradient stress tests and to obtain the best CRT possible, we used a point source configuration which consisted of five $^{22}\text{Na}$ point-like sources with a combined activity of $5.59$ MBq distributed along the $z$-axis (axial). This allowed a very precise determination of the CRT as each line of response (LOR) can be unambiguously assigned to one of the point sources.

Using this source configuration, we tested the clinical configuration with the original IF board: first, we applied five MRI imaging sequences (sequence parameters are summarized in table 1), whereby the sequence parameters were motivated by clinical imaging protocols preinstalled at the MRI system. Here, we applied T1- and T2-weighted turbo spin echo (TSE) sequences (T1WATSE and T2WTSE), T1- and T2-weighted fast field echo (FFE, gradient echo) sequences (T1WFFE and T2WFFE) as well as an FFE based echo planar imaging (EPI) sequence (EPI7, acceleration factor $= 7$). Secondly, we used synthetic gradient test sequences as sketched in figure 4. The sequence gives full control over the gradient system, meaning that gradient strength, slew rate, switching duty cycle as well as switching direction are accurately adjustable. Using this sequence, we tested all combinations of three different gradient strengths (GS) ($10$ mT m$^{-1}$, $20$ mT m$^{-1}$ and $30$ mT m$^{-1}$), five different switching duty cycles (SDC) ($20\%$, $40\%$, $60\%$, $80\%$ and $100\%$) defined as the percentage in which the gradients are in a switching state (see figure 4) and five slew rates (SR) ($25$ T m$^{-1}$ s$^{-1}$, $50$ T m$^{-1}$ s$^{-1}$, $100$ T m$^{-1}$ s$^{-1}$, $150$ T m$^{-1}$ s$^{-1}$ and $200$ T m$^{-1}$ s$^{-1}$). As switching direction, we chose the $z$-axis in all experiments as previous experiments showed the highest PET performance degradation in relation to this direction.

Each parameter set defined one MRI test sequence which was applied during PET acquisition in a time window of $20$ s length. The resulting PET data was gated taking into account a small safety margin of $2$ s. One measurement cycle comprised five different gradient parameter sets (constant GS, constant SDC and varying SR) yielding 15 individual PET data
acquisitions for all 75 gradient test parameter sets. The energy resolution and CRT were determined for each gradient test parameter set and compared to reference measurements which were acquired without gradient activity (before and after the MRI sequence). PET performance changes are calculated as relative changes and the statistical uncertainty is obtained from the aforementioned fit. For the PET data acquisition containing the worst case gradient sequences (SDC = 80%, GS = 30 mT m⁻¹), we evaluated the coincidence rate over time to study the impact on the count rate performance. Besides this worst-case scenario, we also show a measurement series with a SDC of 20% and GS of 30 mT m⁻¹ which was used to make a comparison to the results presented by Weirich et al (2012) and approximately matches the gradient parameters of that work.

Table 1. Overview of the imaging sequences applied (T1WATSE = T1-weighted turbo spin echo sequence (TSE), T2TSE = T2-weighted TSE sequence, T1WFFE = T1-weighted fast field echo (FFE, gradient echo) sequence, T2WFFE = T2-weighted FFE sequence, EPI7 = FFE sequence with echo planar imaging (EPI) readout (acceleration factor = 7)).

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Imaging sequences</th>
<th>T1WATSE</th>
<th>T2TSE</th>
<th>T1WFFE</th>
<th>T2WFFE</th>
<th>EPI7</th>
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Note: The following abbreviations were used: TR = Repetition Time (given in ms), TE = Echo Time (given in ms), TF = Turbo Factor, FA = Flip Angle (given in °), VS = Voxel Size (given in mm³), ST = Slice Thickness (given in mm), MS = Matrix Size, MSR = Maximum Slew Rate (given in T m⁻¹ s⁻¹), MSDC = Maximum Switching Duty Cycle.

Figure 4. Sketch of the newly implemented gradient test sequence: it gives complete control over the gradient system meaning that strength, slew rate, switching direction as well as asymmetries or duty cycle can be adjusted accurately (reprinted with permission from Wehner et al (2015)).
After upgrading the platform with the improved IF board, these experiments were repeated to evaluate the updated platform.

3.4. PET performance evaluation

To get a better understanding of the PET performance, we carried out some basic investigations with the upgraded platform. These investigations range from measurements using point sources to determine typical performance parameters up to measurements with structured phantoms to demonstrate the imaging capabilities of the clinical configuration.

3.4.1. Point source measurement. As for the MRI compatibility investigation, we used the five $^{22}$Na point-like sources with a combined activity of 5.53 MBq distributed centrally along the axial axis (positions: $(x, y, z) = \{(-11.9, -3.1, -9.2), (-0.7, -10.4, -8.1), (-2.0, -4.2, 0.0), (9.0, -6.7, 5.1), (-3.2, 1.6, 9.5)\}$ mm). We configured all different trigger schemes (1–4) consecutively and recorded PET data for approximately 5 min for each trigger scheme. The energy resolution, CRT as well as the prompt rate is determined for both energy windows and each measurement. With respect to the prompt rate, we selected a signal region of 10 mm around the nominal source positions and regarded the resulting rate as the trues rate (motivated by the NEMA NU4-2008 standard).

3.4.2. Sensitivity profile. To acquire a sensitivity profile, we moved one $^{22}$Na point-like source with an activity of 1.1 MBq axially through the center of the FoV in approximately 1 mm steps. At each position, we acquired PET data for at least 1 min and the trues rate was determined as the prompt rate in a 3D signal region with a radius of 10 mm around the nominal source position (motivated by the NEMA NU4-2008 standard). Then, the trues sensitivity is calculated by dividing the trues rate by the source’s activity whereby the latter was corrected for the branching ratio of the $^{22}$Na $\beta^+$ decay which is 0.906. The entire measurement series was measured in two parts on separate days. The PET data was processed with a cluster window of 2 ns and a coincidence window of 1 ns.

3.4.3. High activity measurement. To test the behavior of the system at high activities, we used the FDG-filled cylindrical volume. It was placed in the axial center of the scanner with an offset in the transaxial plane at the position $x = 2.69$ mm, $y = 13.28$ mm. At the beginning of the experiment, the activity of the phantom was approximately 200 MBq and we performed measurements down to about 45 MBq.

During the FDG decay, we measured with trigger schemes 1–3 for approximately 30 s, whereby trigger scheme 1 was only characterized at lower activities due to power consumption and current limitations on the power supply chain. The acquired data sets were processed with a cluster window of 2 ns, a CW of 500 ps and we applied both the NE and the WE window. To estimate the trues and randoms rate, we defined a signal region with a radius of 20 mm around the central axis of the cylinder of the activity volume. Since the phantom is not compliant with the NEMA NU4-2008 standard, we did not calculate a scatter rate. The trues rate is determined by the prompts rate obtained in the signal volume corrected for the randoms rate. We estimated the randoms rate from the measured singles and prompts rate as proposed in Oliver and Rafecas (2012) in the stated signal region. The random signal region is comparable to the random region defined for a mouse-sized scatter phantom in the NEMA NU-2008 standard and the evaluation should shield comparable results with regards to the randoms rate. As an equivalent to the noise equivalent count rate (NECR) without accounting for scattered events, we calculated the square of the trues rate divided by the difference of the trues rate and randoms rate.
3.5. Imaging experiments

To demonstrate the spatial resolution of the clinical configuration, we used the hot-rod phantom providing various rod size sections and measured for 2264.8 s with an activity of 9.81 MBq.

The rabbit-sized phantom was measured in five bed positions to cover the whole phantom with the FoV of the PET insert (see table 2). For this phantom, an image reconstruction once with ToF information and once without was performed.

All imaging experiments have been conducted using the trigger scheme 1. The acquired raw PET data were processed with a cluster window of 2 ns and a CW of 1 ns.

3.5.1. Image reconstruction. For the image reconstruction, we used a 3D reconstruction (Salomon et al. 2011, 2012) implementing an ordered subset expectation maximization (OSEM) algorithm (Hudson and Larkin 1994).

For the image reconstruction of the smaller hot-rod phantom, we employed a self normalization, but did not correct for scatter or attenuation. In contrast, scatter and attenuation were modeled for the larger rabbit-sized phantom using an artificial cylindrical attenuation map. We used an isotropic reconstruction voxel size of 0.5 mm, 32 iterations with 8 subsets without filtering the activity distribution at any point. For the ToF capabilities of the image reconstruction, a ToF kernel of 215 ps for trigger scheme 1 is assumed.

2D image slices are shown for the different imaging experiments. The slice thickness used is 1 mm for the hot-rod phantom and 15 mm for the rabbit-sized phantom.

4. Results

4.1. MRI compatibility investigations

4.1.1. Original IF board. The results of the PET performance degradation induced by imaging sequences are listed in table 3. With respect to the energy resolution, only the T2FFE sequence showed a measurable degradation of about $0.58 \pm 0.25\%$. For the CRT measurements, only the EPI7 sequence had a measurable impact. A degradation of $2.95 \pm 0.70\%$ was observed. In contrast to these two measurements, all other imaging protocols did not provoke a significant performance degradation ($p \geq 5\%$).

The degradation maps of the synthetic gradient stress tests are shown in figure 5. The worst degradation of the energy resolution was measured as $5.1 \pm 0.3\%$ for a GS of 30 mT m$^{-1}$, a SR of 200 T m$^{-1}$ s$^{-1}$ and a SDC of 80%. For the CRT, the worst degradation was observed for a GS of 30 mT m$^{-1}$, a SR of 200 T m$^{-1}$ s$^{-1}$ and a SDC of 60% as $33.9 \pm 1.2\%$.

For the PET acquisition containing the gradient test parameter set which provoked the highest energy resolution degradation (GS of 30 mT m$^{-1}$ and a SDC of 80% and SRs ranging from 25–200 T m$^{-1}$ s$^{-1}$), the relative coincidence rate as function of time is shown in figure 6.
Here, a drop of the coincidence rate of about 2% was observed for the highest SR applied, namely 200 T m$^{-1}$ s$^{-1}$, the value for which the highest energy resolution degradation was found, too. For the next smaller SR of 150 T m$^{-1}$ s$^{-1}$, the coincidence count rate drop is halved to about 1% and all further test sequences with lower SR did not show a measurable decline of the coincidence rate.

In addition to that, the relative coincidence rate as function of time for the gradient stress tests with a GS of 20 mT m$^{-1}$ and a SDC of 20% and SRs ranging from 25–200 T m$^{-1}$ s$^{-1}$ is shown in figure 7 for comparison reasons. In that case, no measurable count rate losses were observed during this measurement.

### 4.1.2. Improved IF board

The PET performance degradation measurements obtained with the prototype scanner equipped with the improved IF boards during the application of the imaging sequences are listed in table 4. No significant degradation of the energy resolution or the CRT was measured ($p \geq 10\%$) in all cases.
The degradation maps acquired using the gradient stress test are depicted in figure 8. With the improved IF board, the gradient stress tests did not cause a significant degradation of the energy resolution or the CRT.

For comparison reasons, the relative coincidence rate as function of time for the gradient stress tests with a GS of 30 mT m\(^{-1}\) and a SDC of 80% and SRs ranging from 25–200 mT m\(^{-1}\) ms\(^{-1}\) is shown in figure 9. In contrast to the measurement shown above, no drop of the coincidence rate was measurable.

**Figure 6.** Relative coincidence rate over time for 5 s time bins acquired with the platform equipped with the original interface boards. The time periods with gradient stress test sequences with a GS of 30 mT m\(^{-1}\) and a SDC of 80% for different SRs ranging from 25–200 mT m\(^{-1}\) ms\(^{-1}\).

**Figure 7.** Relative coincidence rate over time for 5 s time bins acquired with the platform equipped with the original interface boards. The time periods with gradient stress test sequences with a GS of 20 mT m\(^{-1}\) and a SDC of 20% for different SRs ranging from 25–200 mT m\(^{-1}\) ms\(^{-1}\) are marked. This evaluation is used as a comparison to the results presented in Weirich et al (2012).
4.2. PET performance evaluation

4.2.1. Point source measurement. The results of the point source measurements are summarized in Table 5: while the energy resolution is almost the same for all trigger settings with approximately 11.4%, the CRT depends on the trigger scheme and ranges from 215 ps for trigger scheme 1 and the NE window to 621 ps for trigger scheme 4 and the WE window. The sensitivity loss associated with trigger scheme 1 is approximately 5–6% compared to the sensitivity obtained with trigger schemes 2–4. For all trigger schemes, the CRT using the WE window is about 10% worse compared to the CRT with the NE window. The sensitivity of the WE window is about 77% higher compared to the NE window.

4.2.2. Sensitivity profile. Figure 10 shows the axial sensitivity profile of the scanner measured with trigger scheme 1 for the NE and WE window. The NE window shows an isocenter sensitivity of 0.41 ± 0.01% and the WE window 0.71 ± 0.01%.

Table 4. Energy and CRT degradations for the imaging sequences measured with the improved IF board.

<table>
<thead>
<tr>
<th>MRI sequence</th>
<th>ΔE/E (FWHM) (%)</th>
<th>rel. deg. (%)</th>
<th>CRT (ps)</th>
<th>rel. deg. (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>No sequence</td>
<td>11.388 ± 0.013</td>
<td>0.013</td>
<td>213.0 ± 0.7</td>
<td>0.013</td>
</tr>
<tr>
<td>T1WATSE</td>
<td>11.427 ± 0.023</td>
<td>0.34 ± 0.23</td>
<td>213.7 ± 1.2</td>
<td>0.30 ± 0.66</td>
</tr>
<tr>
<td>T2WTSE</td>
<td>11.385 ± 0.022</td>
<td>−0.02 ± 0.23</td>
<td>214.4 ± 1.2</td>
<td>0.63 ± 0.64</td>
</tr>
<tr>
<td>T1WFSE</td>
<td>11.415 ± 0.025</td>
<td>0.24 ± 0.25</td>
<td>213.2 ± 1.3</td>
<td>0.09 ± 0.69</td>
</tr>
<tr>
<td>T2WFSE</td>
<td>11.349 ± 0.024</td>
<td>−0.34 ± 0.24</td>
<td>214.1 ± 1.3</td>
<td>0.51 ± 0.70</td>
</tr>
<tr>
<td>EPI7</td>
<td>11.398 ± 0.024</td>
<td>0.09 ± 0.24</td>
<td>214.7 ± 1.3</td>
<td>0.80 ± 0.68</td>
</tr>
</tbody>
</table>

Note: The no sequence row is used as a reference to calculate the degradation as relative change for the imaging sequences.

Figure 8. Degradation maps of the gradient stress test measured with the improved IF board. The top row shows the energy resolution degradation and the lower row the CRT degradation. Each column shows the degradation maps for a fixed gradient strength. Individual degradation maps show the degradation for different slew rates and switching duty cycles.
4.2.3. High activity. Measurements were recorded between ∼46–193 MBq for trigger scheme 3 and between ∼44.5–163 MBq for trigger scheme 2. The trigger scheme 1 showed stability problems at high activities and only allowed a stable operation at an activity of ∼42.1 MBq. For the other two measurements with trigger scheme 1 at ∼49.5 MBq and ∼58.3 MBq one detector stack was not operational due to the high power consumption and did not deliver any hit data resulting in a lower sensitivity of the scanner. Nonetheless, these measurements allowed a determination of the energy resolution and CRT of the scanner. Two measurements with trigger scheme 2 and 3 were conducted between those two trigger scheme 1 measurements. Both measurements deliver a CRT which is unexplainably worse than the expectation based on the other trigger scheme 2 and 3 measurements. As a result, the activity range including those four measurements showing either a sensitivity loss due to a missing detector stack or an unexplainable degradation of the CRT is marked with a grey shaded bar in the result plots (figure 11).
The energy resolution as function of activity is depicted in figure 11(a). It degrades linearly by about 2.9%/100 MBq (measured relatively).

The CRT as function of activity is shown in figure 11(b) for all trigger schemes. Depending on the trigger setting and the energy window used, the CRT degrades relatively towards higher activities by 2.7–4.1%/100 MBq.

The sensitivity for the cylindrical activity distribution and the defined signal region as function of activity is shown in figure 11(c), where a decline in sensitivity can be observed with increasing activity.

The random fraction evaluated in the described signal region as function of activity is presented in figure 11(d) and shows a linear increase for the WE window of 1.38%/100 MBq and for the NE window of 1.14%/100 MBq for trigger schemes 2 and 3.

The ratio of the trues rate squared and the difference of the trues rate and the randoms rate is measured as function of activity and shown in figure 11(e), whereby the rates are evaluated in the signal region. This measure does not peak in the evaluated activity range.

4.3. Imaging experiments

4.3.1. Phantoms. The reconstructed images from the hot-rod phantom with rod sizes between 1.0–4.0 mm and a ring of activity surrounding the structured region shows clear rod separation down to 2.0 mm for a slice thickness of 1 mm (figure 12). The 1.5 mm region can not be resolved properly anymore in this phantom.

The large rabbit-sized phantom was reconstructed without (figure 13(a)) and with the application of a ToF kernel (figure 13(a)). In both cases, 2D slices with a slice thickness of 15 mm are presented. The reconstruction with ToF information shows a clear improvement. Image artifacts, which are caused by the PET ring geometry, are visible in the case of non-ToF reconstruction. They are resolved when the ToF information is included in the reconstruction.
5. Discussion

5.1. MRI compatibility investigations

The MRI compatibility of the Hyperion-II D platform equipped with preclinical LYSO crystal arrays was investigated in Wehner et al (2014), Wehner et al (2015), Weissler et al (2015): the high resolution scintillator configuration in conjunction with a light guide and the employed
trigger scheme 3 delivered an energy resolution of 12.87% and a CRT of 555 ps. Using the synthetic gradient stress tests as described above and in Wehner et al (2015), a maximum degradation of the energy resolution of about 5.8% and the CRT of about 14.3% was provoked. Especially the CRT degradation played a subordinate role in the preclinical scenario (in combination with a mouse Tx/Rx RF coil), as the extent of the objects under investigation are too small in order to profit from ToF-PET.

Figure 12. (a) MRI image of the hot-rod phantom (acquired with the T2WTSE sequence). (b) 2D slice of the PET reconstruction (OSEM, attenuation correction, 32 iterations, 8 subsets). A 0.5 mm voxel pitch and ToF information with a ToF kernel of 215 ps were used. The shown slice thickness is 1 mm.

Figure 13. 2D slices of the multi-bed reconstruction (OSEM, attenuation correction, 32 iterations, 8 subsets) of the rabbit-sized phantom. A 0.5 mm voxel pitch was used and the shown slice thickness is 15 mm. For the reconstruction shown in the left image, no ToF information was used during reconstruction (a). In a second reconstruction, the ToF information with a ToF kernel of 215 ps was used (b).
In a configuration with rather clinical parameters, in which e.g. trigger scheme 1 is applied to provide a much better CRT, ToF-PET gains importance. With a superior CRT performance, the influence of the gradient switching is expected to become more substantial. Moreover, an investigation using trigger scheme 1 is expected to be more sensitive to gradient switching. As a result, a first evaluation of two modules equipped with clinical detector stacks was conducted with respect to gradient switching (Schug et al 2015a). In the referenced work, we used a demanding sequence with a high gradient strength and duty cycle based on a normal EPI sequence (EPI factor: 49, gradient strength: 30 mT m\(^{-1}\), slew rate: 192.3 T m\(^{-1}\) s\(^{-1}\), TE/TR: 12/25 ms and switching duty cycle: 67% with the gradient in z-direction) as presented in Wehner et al (2014). The better timing resolution proved to offer a higher sensitivity to the degradation induced by the gradient sequence compared to the preclinical LYSO crystal configuration. Whereas, the energy calculation in the one-to-one coupled read-out scheme proved to be more stable. The degradations measured were 3.3% for the energy resolution and 30% for the CRT for two modules mounted on the same gantry as used in this work.

In the present work, the investigations were extended to a full-ring geometry allowing for an initial PET performance evaluation of the clinical crystal configuration as discussed earlier. In addition, the full synthetic gradient test protocol was applied in order to investigate the degradation phenomena in detail. The maximum degradations found in this work are 5.1 ± 0.3 % for the energy resolution and 33.9 ± 1.2% for the CRT. For the imaging scenarios tested, the degradation levels are more than one order of magnitude smaller compared to the maximum degradation level observed. This is expected, as the SDCs of the imaging sequences are below 20% (see table 1 and Wehner et al (2015)). For the test sequences causing the highest energy resolution degradation, a loss of coincident events of about 2% was observed. This loss is caused by the event filtering with the NE window in combination with the degradation of the energy resolution. A loss of raw detector signals was not observed (Wehner et al 2014, 2015).

In contrast to that, the CRT degradation did not cause a loss of events in this measurement as a very large CW was employed.

Based on these studies, the reason of the degradations was investigated, identified and addressed by improving the interface board. The improved interface boards were installed and tested on the prototype scanner equipped with clinical scintillators presented in this manuscript. The upgraded platform did not show any measurable PET performance degradation in terms of energy resolution, CRT and count rate performance even under the most demanding synthetic gradient test sequences. Hence, the Hyperion-IID platform equipped with the improved interface boards is expected to deliver the same PET Performance outside and inside an MRI system even under the most demanding gradient sequences without measurable degradations. The interference tests presented in this work were conducted on a demonstrator with a bore size which is small compared to a whole-body PET insert. In case of a whole-body system, the bore would be larger and the PET modules would be placed closer to the gradient coils and would therefore experience a higher magnetic flux of the gradient fields. Experiments with modules at this maximum position were already conducted in Schug et al (2015a). There, the increased magnetic flux led to e.g. a CRT degradation of up to ~50% in contrast to the worst degradation level of 30% measured at the nominal position. As also discussed in Wehner et al (2015), the degradation scales with the magnetic flux density and therefore with the proximity to the gradient coils. Although we did not test the improve IF-board at the maximum position, a serious performance degradation is not expected as the degradation worsening (without the hardware fix) between the nominal and maximum position was less than a factor of 2 and because no discontinuous scaling behavior of the degradation levels with the distance to the gradient coils was observed.
Only few other groups investigated the influence of MR operation on the PET performance at the level of detail we presented here and in our previous works. In Weirich et al. (2012), the influence of gradient sequences on the count rate for the Siemens 3T MR-BrainPET scanner has been investigated. The highest reduction of the count rate was observed for a $z$-gradient with a GS of 20 mT m$^{-1}$, SR of 154 mT m$^{-1}$ ms$^{-1}$ and a SDC of 10.6%. An energy window of 420–580 keV and a CW of 12 ns was used. The observed coincident count rate loss was reported as 1.2%. A comparable investigation with the Hyperion-II$^D$ platform equipped with the original interface board and comparable gradient sequence parameters are shown in figure 7 and does not show a loss of coincident events for the given sets of gradient parameters.

Therefore, we propose to scan the full parameter space of the gradient system in order to acquire the full understanding of the influence of the gradient system on the PET performance, as MR imaging sequences may not be sensitive enough to show degradations and only investigate a single set of parameters of the gradient system. This might be in particular critical, as this level of investigations does not allow a reliable compatibility investigation. In case of more demanding imaging experiments, degradations might occur which were not observed in the supposed compatibility investigations. In contrast, synthetic test protocols as discussed in Wehner et al. (2015) and applied in this work, allow to distinguish between gradient switching direction and the influence of each parameter of the gradient system and do not include any RF induced interferences, which might lead to degradation of the PET performance, as well. Hence, a more detailed understanding of interference phenomena becomes possible.

5.2. PET performance evaluation

5.2.1. Point source measurement. The timing resolutions obtained with the scanner are the best values obtained with an imaging-capable MR-compatible PET system. Because of the one-to-one readout scheme, the photon density on a single DPC pixel is increased in comparison to a light-sharing readout scheme as used for the preclinical Hyperion-II$^D$ scanner. This results in a lower CRT degradation going from trigger scheme 1 to higher trigger schemes (Schug et al. 2015c). Moreover, since the one-to-one coupling does not need to combine hit data of multiple DPCs and because consequently higher validation thresholds are chosen, the sensitivity does only degrade by about 5% when trigger scheme 1 is configured (in comparison to higher trigger schemes). In contrast to that result, we measured a sensitivity loss of about 40% with similar cooling temperature and operating voltages on the preclinical scanner exploiting a light-sharing readout scheme (Schug et al. 2015c). However, even for the clinical scintillator configuration, the sensitivity loss is expected to be higher if the insert is operated at higher temperatures.

5.2.2. Sensitivity profile. The sensitivity profile shows the expected triangular shape and the measurements taken on two separate days match nicely showing the reproducibility and stability of the investigation.

5.2.3. High activity. Trigger scheme 1 showed problems when operated at activities above 43 MBq. This has been observed before for the platform equipped with preclinical LYSO crystal arrays (Schug et al. 2015c). Origin of this restriction is the increased power consumption associated with trigger scheme 1: the required current strength reaches the current limitations implemented on stack level. This results in a supply voltage drop and prohibits the operation of trigger scheme 1 above the mentioned activity. Trigger schemes 2 and 3 are not affected by this limitation and can be operated without degradations up to the highest activities investigated. The only exception are two measurements taken between the problematic trigger scheme 1 measurements.
We couldn’t identify the exact cause of the slight CRT degradations for these two measurements. As there are trigger scheme 2 and 3 measurements before and after the problematic measurements, we conclude that both measurements are affected by a misconfiguration of the system’s operational parameters which we could not identify in the later data analysis. A further analysis is not possible because the clinical scintillator configuration was only installed for a short period of time. Therefore, a repetition of the high activity measurement is not possible.

The energy resolution and CRT only show slight degradations towards the highest measured activities. The degradation coefficient of the energy resolution with the activity is only a third of the coefficient measured with the platform equipped with preclinical LYSO crystal arrays (Schug et al 2015c). The corresponding coefficient for the CRT is on the same level for both read-out schemes.

With only two stacks per SDM, the employed one-to-one coupling and higher validation threshold, the raw data rate per SDM is much lower compared to the fully equipped preclinical SDMs. Up to activities of almost 200 MBq, we did not observe saturation effects. Such saturation effects were observed for the preclinical configuration with six stacks at activities of about 25–55 MBq depending on the trigger and validation scheme (Schug et al 2015c).

The sensitivity shows a decline, most likely due to pile up effects which could be reduced with a more sophisticated clustering and coincidence processing. The randoms fraction is very low due to the narrow coincidence window and stays below 2.7% even up to the highest activities and largest energy windows investigated. The NECR (without accounting for scatter) does not show the peak in the measured activity range.

5.3. Spatial resolution

As expected, the spatial resolution of the PET scanner is in the range of half the crystal array’s pitch: the hot-rod phantom shows rod separation down to at least 2 mm. However, the 1.5 mm rods cannot be resolved anymore. Thus, we conclude that the presented scanner configuration is expected to provide a spatial resolution in the order of 2 mm.

5.4. ToF benefit for reconstruction

As demonstrated by the imaging experiments presented in figure 13, adding the ToF information to the reconstruction clearly improves the visual impression of the image. Artifacts associated with the PET scanner’s geometry can be substantially reduced. As expected from ToF-PET, the SNR improves with increasing CRT performance depending on the diameter of the activity distribution. This effect was also shown for the preclinical configuration (Schug et al 2015c).

6. Conclusion

The PET performance of the Hyperion-II$^D$ platform in the presented configuration with the improved IF board design, the clinical crystal configuration and using the preclinical gantry geometry seems not to be harmed in any measurable degree by the operation of the employed 3 T MRI scanner even under the most extreme conditions the system is able to generate. This is a promising result if the platform should be used to built up gantries with a different geometry or if the PET modules should be integrated directly next to the gradient coils where they might experience stronger gradient field strengths.
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