PET/CT in radiation oncology

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PET/CT is an effective tool for the diagnosis, staging and restaging of cancer patients. It combines the complementary information of functional PET images and anatomical CT images in one imaging session. Conventional stand-alone PET has been replaced by PET/CT for improved patient comfort, patient throughput, and most importantly the proven clinical outcome of PET/CT over that of PET and that of separate PET and CT. There are over two thousand PET/CT scanners installed worldwide since 2001. Oncology is the main application for PET/CT. Fluorine-18 deoxyglucose is the choice of radiopharmaceutical in PET for imaging the glucose uptake in tissues, correlated with an increased rate of glycolysis in many tumor cells. New molecular targeted agents are being developed to improve the accuracy of targeting different disease states and assessing therapeutic response. Over 50% of cancer patients receive radiation therapy (RT) in the course of their disease treatment. Clinical data have demonstrated that the information provided by PET/CT often changes patient management of the patient and/or modifies the RT plan from conventional CT simulation. The application of PET/CT in RT is growing and will become increasingly important. Continuing improvement of PET/CT instrumentation will also make it easier for radiation oncologists to integrate PET/CT in RT. The purpose of this article is to provide a review of the current PET/CT technology, to project the future development of PET and CT for PET/CT, and to discuss some issues in adopting PET/CT in RT and potential improvements in PET/CT simulation of the thorax in radiation therapy. © 2008 American Association of Physicists in Medicine. [DOI: 10.1118/1.2986145]

Key words: PET/CT, 4D CT, cine CT, MIP CT, average CT

I. INTRODUCTION

PET/CT was developed in 1998. It was not until 2001 when the first commercial PET/CT scanner became available. Prior to the technology of PET/CT, the CT and the PET data were acquired in two different scanners. Fusion of the PET and CT data was performed with software techniques. Registration of the PET and CT data was more successful for the brain studies with rigid transformation but less accurate for the other regions of the body, in particular in the thorax and the abdomen, due to the difficulty in repositioning the patient in two separate sessions and the nonrigid nature of the organs. Average error in the fusion of separate PET and CT brain images was in the order of 2–3 mm. This error increased to 5–11 mm when fusing separate PET and CT body images of the thorax. The advent of PET/CT scanners has facilitated the hardware fusion of PET and CT data sets by transporting the patient between the PET and CT components of the scanner without repeating patient setup. It has been shown that hardware fusion is more accurate than software fusion in diagnosis, staging and restaging of many cancer types, in particular in the area of tumor infiltration of adjacent structures that could not be conclusively assessed using the separate CT and PET data.

The fast scan speed of CT has shortened a normal imaging session of about 1 h with a stand-alone PET to less than 30 min with a PET/CT. This time savings has a major impact on patient comfort in particular for patients who may need to raise their arms over the head during the PET/CT scan. The CT images are adapted for attenuation correction and tumor localization of the PET data. There are over 2000 PET/CT scanners installed worldwide, and stand-alone PET scanners have not been in production since 2006. Fluorine-18 deoxyglucose (\(^{18}\text{F}-\text{FDG}\)) is the major radiopharmaceutical in PET/CT and has been approved for diagnosis and staging of non-small-cell lung cancer (NSCLC), colorectal cancer, esophageal cancer, head and neck cancer, lymphoma, and melanoma since 1998. In addition, PET/CT has been approved for staging and restaging and for therapeutic monitoring of breast cancer.

The average \(^{18}\text{F}-\text{FDG}\) PET sensitivity and specificity across all oncology application are estimated at 84% (based on 18 402 patient studies) and 88% (based on 14 264 patient studies), respectively, according to Gambhir et al.\(^9\) from a collection of 419 articles from 1993 to 2000. Specifically, the sensitivity of PET ranged from 84% to 86%, the specificity ranged from 88% to 93%, and the accuracy ranged from 87% to 90%. The addition of PET significantly improves the diagnosis of lung cancer with the sensitivity of 96% and specificity of 73% (as compared to sensitivity of 67% for CT alone), and the staging with the sensitivity of 83% and specificity of 91%, respectively (as compared to sensitivity of 64% and specificity of 74% for CT alone).

The additional clinical values of PET/CT to PET alone or separate PET and CT have also been documented. The first
As opposed to the accurate setup of patients in the treatment position with external laser lights and a flat table support needed for radiation oncology. Collaboration between the two disciplines is critical to the success of PET/CT for RT. Application of PET/CT in nuclear medicine is a task of detection based on the standardized uptake value (SUV). The physicians can sometimes read through the artifacts such as the ones caused by respiratory motion, implanted metal such as dental filling, pace makers, prosthetics, and surgical clips, and intravenous or oral contrast media. Radiation oncologists may not be familiar with the images in PET/CT imaging and their associated artifacts. They generally depend on the diagnosis and staging of the nuclear medicine physicians for RT planning. Both the location and extent of tumor are critical in RT and may not be fully captured in the nuclear medicine report. In addition, there is still no standard for delineating the gross tumor volume (GTV) in PET images. The workstations used for evaluating PET/CT images in nuclear medicine are usually not equipped with planning software utilities required for RT. Reproduction of SUV may be difficult once the PET/CT images are transferred to planning workstations in RT. It is critical to ensure the PET/CT scan data are acquired, processed, and transferred correctly for RT. Migration from stand-alone PET to PET/CT has improved registration between the CT and PET data from hardware fusion, in particular in the area outside the thorax and the abdomen such as the head and neck area. However, PET/CT has also introduced a new problem, namely the misalignment between the PET and CT data in the thorax and the abdomen due to patient respiration and the different scan speeds of CT and PET. The first clinical investigation addressing the impact of misalignment was conducted by Osman et al. They observed that curvilinear cold artifacts paralleling the dome of the diaphragm at the lung bases were frequently noted in PET/CT images. However, clinically they reported that this artifact resulted in significant inaccurate localization of lesions for only 2% in a study of 300 patients. Gould et al. reported in a cohort of 259 patients for cardiac PET/CT imaging, 40% of the patient data exhibited false positive defects due to misalignment. It is evident from these two studies that the severity of misalignment is dependent on the area of interest. Misalignment between PET and CT may not be a significant issue in a whole body PET/CT scan when the lesions may not be in the lungs, liver, and esophageal and not close to the diaphragm. However, it becomes a significant issue in cardiac PET/CT imaging when the heart is right above the diaphragm. Coaching patients to breath-hold in an effort to mitigate the artifact was not very successful. In a study of 100 patients undergoing PET/CT with breath-hold at midexpiration CT by Pan et al., 50% of the patients exhibited the curvilinear cold artifacts in the PET images caused by respiratory motion. In a recent study of 216 consecutive patients undergoing free breathing CT in PET/CT by Chi et al., 68% of the patient data exhibited either the curvilinear cold artifacts or tumor misalignment. Misalignment due to breathing has been shown to cause variation in SUV, critical to delineation of gross tumor volume in the PET data. We
will highlight and project the technology advancement of PET/CT, propose imaging techniques that will help improve registration of the PET and CT data in the thorax, and emphasize the particular requirements of PET/CT and for radiation oncology.

II. PET DETECTOR

The PET of a PET/CT scanner today is distinguished by the scintillator materials used for the detection of 511 keV annihilation photons. The major materials are bismuth germinate oxide (BGO), gadolinium oxyorthosilicate (GSO), lutetium oxyorthosilicate (LSO), and LYSO (LSO doped with a small amount of yttrium). BGO has a slow decay time and a low light output leading to relatively poor timing and energy resolution. It has a high stopping power to capture most of the photons reaching the detector. The new materials of LSO and LYSO have a fast decay time and a high light output leading to improved timing and energy resolution. Their stopping power is slightly inferior to that of BGO. Intermediate to both is GSO, which has a higher light output and a faster decay time than BGO and yet is not as efficient as BGO or LSO/LYSO in stopping the 511 keV photons. Table I lists the attributes of different detector material used in commercially available PET/CT scanners. It is clear that LSO/LYSO has and will become the detector of choice in the future PET/CT as manufacturers have chosen either LSO or LYSO as their new PET scintillation material.

III. SPATIAL RESOLUTION OF PET

The limit spatial resolution of clinical PET scanners with a ring diameter of 80 cm is about 2 mm. There are basically five factors that determine the spatial resolution: (1) positron range: the distance between the point of emission of the positron to the point of annihilation of the positron with the electron. The higher the kinetic energy of the positron, the longer is the distance to the site of annihilation. This distance is depend on the isotope; (2) noncolinearity: deviation from the assumed emissions of two 511 keV photons at 180 deg from the point of annihilation. The larger the ring diameter is; the more non-colinearity becomes; (3) detector geometry: smaller detector element of size d leads to better spatial resolution of size d/2; (4) photon interaction in the detector: Compton scattering is the dominate interaction between the 511 keV photon and the detector. Multiple interactions are required to stop the photon in the detector. Some interactions may even spread over to adjacent detector elements. Typical detector elements are in the size of 4–6 mm square and 2–3 cm thick. The result is some uncertainty for identifying the first interaction of the photon and the detector; and (5) reconstruction algorithm that models the degradation from emission of the positron to detection of the photons and that is capable of deconvolving the degradation to improve the resolution. Noncolinearity and detector size are the predominant factors contributing to the degradation of spatial resolution for the clinical PET scanner. Some pre-clinical PET scanners have been shown to achieve spatial resolution of 0.4 mm with a ring diameter of 2 cm and a detector size of 0.25 mm. The spatial resolution of the clinical PET scanner is about 5–8 mm, and can still be improved in the future. The impact of improving spatial resolution may be more significant in diagnosis than in RT. This is mainly due to the clinical target volume and planning target volume margins that are added to the GTV thereby reducing the advantage of PET imaging resolution.

IV. TIME-OF-FLIGHT PET

Conventional reconstruction of PET images assumes that the annihilation event of a positron and an electron detected by a pair of detector elements 180-deg apart, is equally likely to occur at any point along the line defined by this pair of detector elements. This assumption has served well for many years when detectors with short decay time, high light output, and fast electronics were not available. The short decay time of new detector material such as LSO and LYSO combined with fast electronics has enabled time-of-flight (TOF) PET. The timing resolution to locate an annihilation event to within 5 mm of PET resolution is approximately 30 ps. The current commercial TOF PET has the timing resolution of about 600 ps (Ref. 100) and can locate an annihilation event to within 18 cm which in turn improves contrast. The gain in signal to noise ratio in PET images with TOF is proportional to the object size and inversely proportional to the timing resolution. Current technology of 600 ps timing resolution shows improvement of signal to noise ratio for large patients. The future improvement of timing resolution will further reduce noise and improve the signal to noise ratio for all patients. This benefit can also be translated into shorter scan time or reduced injected radiopharmaceutical dose to the patient and reduced radiation exposure to the technologist.

V. TWO-DIMENSIONAL AND THREE-DIMENSIONAL PET IMAGING

PET data can be acquired in two-dimensional (2D) or three-dimensional (3D) mode. In 2D mode, a collimation (septa) is positioned between detector rings in the axial di-
rect. The collimation minimizes the scatter and provides a uniform sensitivity profile along the cranial-caudal direction. An overlap of only 1 cm is needed to compensate for the lower sensitivity of the edge slices between two bed positions in order to provide a uniform sensitivity along the axial direction in a whole body PET/CT scan. On the other hand, 3D PET imaging does not utilize septa during data acquisition. It has higher sensitivity but more randoms and scatters than 2D PET. Due to its nonuniform sensitivity profile (high at the center of the detector with a linear drop to the edge of the detector), it normally requires up to 50% detector overlap between two bed positions. With the improvement of 3D reconstruction techniques and better modeling of the randoms and scatters, it has been shown that 3D PET results in better image quality than 2D PET with less injection of 18F-FDG. This has resulted in lower radiation exposure to the patients, nursing staff and technologists. Therefore, 3D PET imaging is bound to become the standard mode of imaging in the future. Some manufacturers of PET/CT scanners provide 3D only capabilities on their systems.

VI. FOUR-DIMENSIONAL PET IMAGING

Four-dimensional (4D) PET was first developed for cardiac imaging to assess myocardial motion and to obtain ejection fraction. It has been adopted for tumor imaging of the thorax in the last several years. When data are acquired in 4D or gated mode, the data are split into several exclusive bins. For example, there can be 8 bins of 500 ms for an average respiratory cycle of 4 s. Because of insufficient statistics of photons obtained in PET imaging of 3–6 min for the 8 bins in each bed location, the duration of a 4D PET scan is normally prolonged to over 10 min to compensate for the fewer photons recorded in each bin. Image reconstruction is performed on the data of each bin and the result is a set of 3D PET images over a respiratory cycle for the assessment of tumor motion and quantification. Even though, the number of photons in each bin is small, resulting in higher noise in the 4D images; the 4D images can potentially be used for accurate assessment of FDG uptake. 4D PET has not taken off due to its complexity and limited workflow in which 4D PET had to be performed in a separate session after a routine PET/CT scan on most of the PET/CT scanners, prolonging the acquisition time and impacting patient comfort. In our opinion, an ideal PET/CT imaging session should take less than 20 min. Any duration beyond that will increase the potential for patient motion.

PET/CT scanners have recently been equipped with list-mode data acquisition whereby events from each coincidence pair of 511 keV photons are stored in a list stream for later reconstruction. It has been demonstrated that the list-mode data acquisition can be performed with either cardiac or respiratory triggering during a normal static image acquisition. This functionality offers the capability of retrospectively sorting the coincidence events into multiple phases/bins for the reconstruction of 4D PET images. Current PET scanners can be configured to acquire data which can produce both static and 4D PET data to freeze tumor motion. 4D CT may be needed for accurate quantification of 4D PET data as each phase of PET data may need its own CT data for attenuation correction. Although, it is still cumbersome to acquire 4D PET data due to its prolonged acquisition time, it is expected that with the new LSO/LYSO detector, large detector coverage for higher sensitivity, and the advent of 3D image reconstruction to better cope with the increased noise from randoms and scatters in 3D data acquisition, the limitation in acquisition time will be minimized and 4D PET will become a clinically feasible solution to improving the quantification accuracy of tumor in motion.

VII. LARGE BORE PET/CT

Most current commercially available PET/CT scanners have a 70 cm bore size for both PET and CT scanners. Early models had a smaller bore size in PET than in CT. However, it is clear that the bore size will have to increase to accommodate PET/CT simulation of the breast patients and some large patients. The CT scanner of 85 cm bore size was introduced in 2000 to address this issue. Today most of the PET/CT scanners have 50 cm standard field of view in CT image reconstruction with up to 60–65 cm extended field of view reconstruction. This was accomplished by extrapolating the truncated projection data in CT normally in the anterior-posterior views to match the total attenuation measured in the none-truncated projection data in the lateral views. This approach may not provide accurate CT numbers for RT. A correct solution to the truncation problem would be to increase the detector fan angle to avoid truncation by increasing the number of detector elements. One manufacturer has increased the detector size to better control the truncation problem up to 70 cm, and one manufacturer has offered a PET/CT with 85 cm bore size for both PET and CT. It is expected that bore size of 85 cm should become a standard and the CT detector size should be increased to accommodate up to 70 cm CT image reconstruction without truncation for RT.

VIII. MULTISLICE CT

PET needs CT data for tumor localization, attenuation correction, and quantification. PET acquires data over 3–6 min for a 15-cm superior-inferior coverage. With a clinical whole-body scan from base of the skull to midthigh of about 100 cm, the acquisition time for PET data is about 18–21 min for 3 min/bed in 2D data acquisition. To improve the CT image quality of the thorax and the abdomen, the patient is normally asked to raise his/her arms over the head during data acquisition of the CT and the PET data. In this position, an average person may not be able to remain stationary for over 20 min. Figure 1 demonstrates an example of patient motion in the middle of a scan. If patient motion is not accounted for, the PET images may not be reliable for diagnosis. Incorporation of CT has greatly improved the throughput of PET/CT scan and patient comfort when a CT scan of 100 cm can be performed in less than 20 s on a modern 16-slice PET/CT with good image quality.
The first prototype PET/CT was with a single-slice CT (SSCT), which was the major CT scanner in diagnostic radiology at the time. A CT scan would take 5–10 min with this prototype PET/CT. Today almost all the PET/CT scanners are with multislice CT (MSCT), and the number of detector channels or slices has increased from 2 to 64 in the last several years. In general, 16-slice CT is sufficient for tumor imaging and 64-slice CT is necessary for coronary artery CT imaging of the heart. A 64-slice CT can provide high spatial resolution of 0.5 mm in 3D, fast temporal resolution of 100–200 ms at a gantry rotation cycle time of less than 0.35 s, and short breath-hold time of less than 10 s, critical to coronary artery imaging. One vendor has made available 320-slice CT of 16 cm coverage to image the heart in a single gantry rotation. Perfusion CT, which requires a cine scan of over 30–50 s at the same location for assessing the function of tumor by analyzing the dynamics of injected iodine contrast in and out of the tumor, may require more than 64 slices for its expanded coverage of more than 4 cm. A 16-slice CT covers only 2–2.4 cm per gantry rotation. Incorporation of MSCT in PET/CT has and will have a profound impact on improving the registration between the CT and the PET data, and the assessment of tumor motion in 4D CT imaging. Both applications are critical to RT.

IX. AVERAGE CT

The issue of potential misalignment between the CT and the PET data in the thorax and the abdomen was discovered soon after PET/CT was introduced, and has become the most widely addressed issue in PET/CT. MSCT normally acquires data at a very high temporal resolution of less than 1 s, while PET acquires data in several min. Mismatch in temporal resolution leads to a potential mismatch of the tumor positions between the CT and the PET data, and compromises the quantification of the PET data and the determination of PET GTV. Current design of PET/CT only matched the spatial resolutions of CT and PET by blurring the CT images to match the PET images in spatial resolution. No attempt has been made to match the temporal resolutions of CT and PET. Since the PET data is time averaged, it was recognized that PET data can be a more accurate representation of the 3D volume encompassing the tumor motion. However, quantification of the PET data relies on the CT data. If the CT data are not matched with the PET data in position, quan-
tification of the PET data will be compromised. The impact becomes severe when the tumor is near the diaphragm such as some lung, liver, and esophageal tumors.

Mismatch between the CT and PET data can be identified by a curvelinear cold artifact paralleling the dome of the diaphragm at the lung bases or photopenic region in the PET images. We normally spend more time in exhale than in inhale. The PET data averaged over several minutes is closer to the end-expiration than end-inspiration phase. If the CT data are acquired in or near the end-inspiration phase, the inflated lung due to inspiration will be bigger than the deflated lung. The larger area of the inflated lungs in CT renders less attenuation correction in the reconstruction of the PET data resulting in a photopenic region identified as the misaligned region. Several measures have been proposed to mitigate this problem. Coaching patient to hold-breath at midexpiration during CT acquisition was suggested, and the outcomes were mixed because coaching patient to hold breath at certain state may not be reliable both from the patient and the technologist operating the PET/CT scanner perspectives. Another approach to improve registration between the CT and the PET data is to bring the temporal resolution of the CT images to that of the PET data. Recognizing the fact that PET is averaged over many breath cycles, an average CT (ACT) image of about one breath cycle has been shown to improve registration between the CT and the PET data. Figures 2 and 3 show two examples of misalignment between the CT and the PET data corrected by ACT to improve the assessment of lymph node $^{18}$F-FDG uptake and the accuracy of tumor targeting for RT. This concept has also been shown effective in improving registration between the CT and the PET data in cardiac PET/CT imaging.

Acquiring ACT can be accomplished by scanning at the same slice location over a breath cycle at a very high speed gantry rotation for a subsecond temporal resolution. The CT images, almost free of motion artifacts, in a respiratory cycle are averaged for ACT. The conventional approach with slow-scan CT for ACT is ineffective and should be discouraged. Figure 4 shows a clinic example of the same patient scanned with slow-scan CT of 4 s and ACT of fast gantry rotation of 0.5 s for 4 s. Cine-CT and low-pitch helical CT (pitch < 0.1) scans can be adopted to obtain ACT and both have been utilized in 4D CT imaging. However, it is not clear whether the setup of 4D CT imaging for the assessment of tumor motion is ideal for obtaining ACT when the majority of PET/CT scanners are in nuclear medicine and are without a respiratory monitoring device for 4D CT. A practical approach is to acquire ACT without a respiratory monitoring device to improve quantification of the PET data. The additional radiation dose for ACT was estimated to be from 5 to 50 mGy and the additional scan time is less than a couple of minutes and will not significantly impact the overall scan time of a PET/CT procedure. ACT can also be derived from 4D CT, incorporation of ACT for attenuation correction of the PET data should be made available in the future. ACT can also be used for dose calculation due to its time-averaging of CT numbers. Selection of optimum parameters for acquiring ACT with cine-CT or low-pitch helical CT has also been investigated.

MSCT can acquire the ACT data over the chest and the abdomen in less than a couple of minutes due to its large detector coverage over SSCT. It can also provide a sequence of almost motion-free cine CT images at the same slice location due to its subsecond gantry rotation. The marriage of MSCT and PET in PET/CT thus has provided a platform for
registration of the CT and the PET data, directly impacting RT.

**X. MAXIMUM INTENSITY PROJECTION CT**

Maximum intensity projection (MIP) CT images can be derived by finding the maximum pixel value at each pixel location from all the phases of 4D CT (Ref. 123) or cine CT images. It has been shown that MIP CT images are effective in depicting the extent of tumor motion. For peripheral lung tumors (surrounded by the lower density air in the lungs), MIP CT images can be used to assist the determination of the tumor target volume and to avoid ambiguity in using a threshold of SUV to determine the target volume in PET. Any 18F-FDG uptake in the lungs should be supported by tissues with or without motion. Any PET GTV determined with a threshold should not exceed the boundary in the MIP images. Both MIP and ACT images can be derived without gating. Their application for treatment planning for stereotactic RT has been demonstrated. Their applications for PET/CT in RT are expected to become important as they can be applicable to most of the PET/CT scanners without the hardware setup for 4D CT. MIP CT can help determine the tumor volume in the thorax. Figure 5 shows an example of determining the extent of PET GTV with MIP CT. A similar concept has been attempted with regular helical CT data for peripheral NSCLC by Biehl et al., when the border of the tumor can be identified by the helical CT data.

**XI. 4D CT IMAGING**

4D CT (Refs. 122–124) has found its acceptance in RT for providing the gated CT images of multiple phases over a respiratory cycle to assist contouring the extent of tumor motion. 4D CT takes less time in acquisition than 4D PET does. It normally takes less than 2 min (like ACT) to cover 40 cm in the superior-inferior direction on an 8- or 16-slice CT. Two data acquisition modes have been used in 4D CT. One is cine CT and the other one is low-pitch helical CT. Both acquisitions scan any point in space for over one breath cycle plus one (or 2/3) gantry rotation cycle. Cine CT uses less radiation and generates thinner slices than low-pitch helical CT. On the other hand, low-pitch helical CT has an advantage over cine CT in acquisition time. The cine CT based 4D CT is available on 4, 8, 16, and 64-slice CT scanners, and the low pitch helical CT based 4D CT is only available on newer 16- and 64-slice CT scanners. Multislice CT has significantly improved the speed of 4D CT data acquisition and made 4D CT clinically acceptable. It is expected that 4D CT will become an integral part of PET/CT imaging of the thorax and 4D CT will be incorporated in the quantification of 4D PET as each phase of gated PET image needs its own gated CT image for attenuation correction. The ACT and MIP images can be derived from 4D CT or directly from cine CT.

**XII. DETERMINATION OF TUMOR CONTOUR IN PET**

Determination of tumor volume in PET image is normally performed with SUV threshold(s). This is an important yet
controversial issue, in particular for the lung tumors impacted by the respiratory motion. To determine the gross tumor volume in PET, it is generally based on the visual interpretation by the experienced nuclear medicine physician. Some suggested an absolute threshold of 2.5, or a threshold between 15% and 50% of the maximum SUV \( mSUV \). Recent studies by Biehl et al. and Nestle et al. suggested that no single SUV can reliably be used for segmentation of the PET tumor volume, and that an individualized threshold should be derived in consideration of the size, location, nonuniform distribution of \(^{18}\text{F}-\text{FDG} \) activity of the tumor. Most of the discussions have been with the data of stand-alone PET. The issue may be further complicated in PET/CT. Any misalignment can further compromise the quantification of the PET data. Erdi et al. reported that different phases of 4D CT image can cause up to 24% variation of mSUV in the PET data. Pan et al. observed a variation of more than 50% mSUV between the PET data with ACT and the same PET data with helical CT. Chi et al. investigated the effects of misalignment to the determination of PET GTV and found that GTV centroid location could shift by 2.4 mm and GTV volume change could be as high as 154% for the tumors of less than 50 cm\(^3\).

Partial volume also impacts the determination of SUV. Partial volume is caused by the limited spatial resolution of PET scanners (5–8 mm), and renders a blurry look of the PET image. Partial volume mostly affects smaller tumor sizes of less than 1–1.6 cm or 2 times the PET resolution. In addition, studies have shown that SUV is a function of uptake time between injection of \(^{18}\text{F}-\text{FDG} \) and data acquisition. Care should be taken to ensure the same uptake time for the same patient imaged in a multiple of PET/CT sessions during the course of diagnosis, treatment, and staging. Tumor volume determined by PET may be more indicative of the GTV plus motion since PET data are acquired over several min. PET GTV delineation has been and will be a very active area of research. Input from MIP and 4D CT images will be very important and complementary to this endeavor. Contouring a tumor in PET/CT images should be an integral process with complementary inputs of both PET and CT images.

XIII. WORKFLOW

PET/CT simulation of the chest and abdomen poses a challenge for registration of the CT and the PET data. We can acquire the list mode PET data with the physiologic triggers of respiratory or cardiac nature. Two sets of PET data can be reconstructed: one to generate the static PET data for an initial whole-body diagnosis; and the other to generate a gated or 4D PET data over a certain anatomical region using the list mode data and physiologic triggers. Static PET data should be attenuation corrected with ACT. 4D PET data should be corrected with 4D CT data. The images for radiation therapy should be free of misalignment and truncation artifacts which can be accomplished using a large bore PET/CT and a large detector CT, respectively. All the data acquisition should be performed in 20 min to improve patient comfort and avoid potential patient motion.

XIV. SUMMARY

There has been a tremendous growth in the use of PET/CT scanners over the past several years. The combination of faster detector of LSO/LYSO and faster electronics has made the TOF PET/CT feasible in improving the image quality of PET. The improvement in 3D reconstruction with accurate modeling of randoms and scatters has also made 3D PET more attractive than 2D PET. The benefits of 3D include the reduction of both the injected radiopharmaceutical dose to the patient and the radiation exposure to the nursing staff and the technologists. The large bore PET/CT and the large detector CT will make the positioning of a large patient in
the scanner much easier and minimize the truncation of CT images, respectively. Application of PET/CT for radiation therapy is expected to grow albeit there are some challenges. Reimbursement would be feasible when more clinical data to support PET/CT simulation become available. A similar challenge was experienced during the adaptation of CT simulation in the 1990s when the radiation oncologists were not familiar with GTV delineation on the CT images. With the development of 4D CT, the difficulties in the simulation of the thorax and abdomen have largely been minimized. Further improvement in the registration of the PET and CT data by ACT, MIP, 4D CT, and 4D PET will improve the registration in the thorax and the abdomen. Furthermore, data acquisition should be less than 20 min to simulate the patient condition in a treatment session.

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