A hardware investigation of robotic SPECT for functional and molecular imaging onboard radiation therapy systems

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(Received 26 May 2014; revised 6 September 2014; accepted for publication 1 October 2014; published 21 October 2014)

Purpose: To construct a robotic SPECT system and to demonstrate its capability to image a thorax phantom on a radiation therapy flat-top couch, as a step toward onboard functional and molecular imaging in radiation therapy.

Methods: A robotic SPECT imaging system was constructed utilizing a gamma camera detector (Digirad 2020tc) and a robot (KUKA KR150 L110 robot). An imaging study was performed with a phantom (PET CT PhantomTM), which includes five spheres of 10, 13, 17, 22, and 28 mm diameters. The phantom was placed on a flat-top couch. SPECT projections were acquired either with a parallel-hole collimator or a single-pinhole collimator, both without background in the phantom and with background at 1/10th the sphere activity concentration. The imaging trajectories of parallel-hole and pinhole collimated detectors spanned 180° and 228°, respectively. The pinhole detector viewed an off-centered spherical common volume which encompassed the 28 and 22 mm spheres. The common volume for parallel-hole system was centered at the phantom which encompassed all five spheres in the phantom. The maneuverability of the robotic system was tested by navigating the detector to trace the phantom and flat-top table while avoiding collision and maintaining the closest possible proximity to the common volume. The robot base and tool coordinates were used for image reconstruction.

Results: The robotic SPECT system was able to maneuver parallel-hole and pinhole collimated SPECT detectors in close proximity to the phantom, minimizing impact of the flat-top couch on detector radius of rotation. Without background, all five spheres were visible in the reconstructed parallel-hole image, while four spheres, all except the smallest one, were visible in the reconstructed pinhole image. With background, three spheres of 17, 22, and 28 mm diameters were readily observed with the parallel-hole imaging, and the targeted spheres (22 and 28 mm diameters) were readily observed in the pinhole region-of-interest imaging.

Conclusions: Onboard SPECT could be achieved by a robot maneuvering a SPECT detector about patients in position for radiation therapy on a flat-top couch. The robot inherent coordinate frames could be an effective means to estimate detector pose for use in SPECT image reconstruction.

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Key words: robotic, SPECT, region-of-interest (ROI), onboard, radiation therapy

1. INTRODUCTION

Onboard, or in-room, robotic SPECT has been proposed to provide functional and molecular information about tumor and nearby tissue while the patient is on the radiation therapy treatment table.^{1,2} SPECT is already utilized, prior to treatment and in combination with computed tomography

(CT),³ for assessing function, for characterizing tumor, and for treatment planning. For example, SPECT lung perfusion images have been used to modify the radiation beam to avoid healthy lung regions.⁴ Similarly, SPECT imaging of active bone marrow has been used in conjunction with intensity modulated radiation therapy (IMRT) to spare organs-at-risk (OARs).⁵ SPECT imaging of hypoxia, in the tumor region,

could be utilized to modify dose.⁶ SPECT tracers have been investigated for imaging cancers associated with different anatomical sites, including the breast, lung, and prostate.^{7–11} In general, SPECT and its nuclear-medicine cousin, positron emission tomography, have powerful molecular imaging capabilities in terms of depth, sensitivity to molecular probes, and corresponding minimal requirements on tracer concentration.¹² Onboard, or in-room, SPECT could bring these functional and molecular imaging capabilities to treatment delivery tasks such as patient alignment, time-gated delivery, and real-time treatment planning.¹³ However, currently there is no SPECT system for onboard or in-room imaging. Here, we present hardware studies aimed toward such a system.

There are several advantages to utilizing a robot for onboard SPECT. One is that the robot can bring the SPECT detector to the patient in position for radiation therapy and, after imaging, retract the detector to enable normal operation of the radiation therapy system. A second advantage is that the robot can maneuver the SPECT detector around an offset center of rotation, enabling region-of-interest (ROI) imaging at various anatomical sites. A third advantage is that a robotic SPECT system could be versatile in navigating the SPECT detector near the radiation-therapy flat-top couch so as to minimize the distance between the detector and the ROI. In our previous work, computer-simulation studies have demonstrated the imaging potential of robotic SPECT.¹

In the present study, we integrate an industrial robot with a gamma camera to construct a hardware robotic SPECT system. This robotic SPECT system for onboard imaging is utilized herein for hardware investigations into the versatility of robotic SPECT in maneuvering a gamma camera about a thorax phantom on a widely used, flat-top radiation-therapy treatment table. Trajectories investigated include parallel-hole imaging of a larger ROI and pinhole imaging of a comparatively small ROI. The two ROIs were centered at different locations to demonstrate the imaging versatility of robotic SPECT.

Robot technology has been used in medicine for decades, such as for robotic surgery.¹⁴ In radiation therapy, robot applications include the CyberKnife¹⁵ radiation therapy machine and various patient positioning systems.^{16,17} In imaging, robot applications include the Siemens Artis Zeego C-arm x-ray system¹⁸ and the Philips SKYLIGHT,¹⁹ which is a robotic gamma camera, though not one in a robot or detector configuration that is designed for onboard SPECT imaging. Considered here is a configuration closer to the Artis Zeego intraoperative x-ray system, but designed for onboard gamma camera imaging, and studied here for the close camera-to-patient proximity required in gamma camera imaging.

For typical gantry-based SPECT systems, detector pose is characterized by one rotation angle and an angle-dependent radius of rotation (ROR). SPECT image reconstruction can be performed by obtaining the angles and RORs from the gantry system. Here, for robotic SPECT, image reconstruction is performed with detector pose obtained from “base” and “tool” coordinates provided by the robot system. Utilization of

these coordinates demonstrates the applicability of robot technology, highly developed for precise industrial applications, to SPECT imaging.

2. MATERIALS AND METHODS

2.A. Robotic SPECT system

A robotic SPECT system was constructed (Fig. 1) utilizing a Digirad 2020tc detector (Digirad Corporation Poway, CA) and a KUKA KR150 L110 robot (KUKA Robotics, Aktiengesellschaft, Augsburg, Germany). The 2020tc detector consists of a CsI(Tl) scintillator with 64×64 pixels on a 3.25 mm pitch. The imaging area is 20.8×20.8 cm². Attachable to the 2020tc detector are a low-energy medium-resolution parallel-hole collimator and a pinhole collimator. The low energy all purpose (LEAP) parallel-hole collimator has hexagonal holes on a hexagonal grid, with a hole length of 22 mm, a flat-to-flat hole diameter of 1.5 mm, and a septal thickness of 0.2 mm. The pinhole collimator has a tungsten pinhole with 3 mm diameter, a 70° full pinhole opening angle, and a 175 mm focal length.

2.B. Robot coordinates and imaging trajectories

The robot has six axes of rotation (Fig. 1), which combine to determine the position and orientation of the detector. The robot controller reports position and orientation by several coordinate frames, which include base coordinate frames and tool coordinate frames (Fig. 2). The robotic system allows for user calibration of tool and base coordinate frames. Here, the tool corresponds to either the pinhole-collimated or parallel-hole-collimated detector. The purpose of tool calibration is to inform the robotic system of the origin and orientation of the tool coordinate frame. The origin was calibrated to be either the pinhole or the center of the phantom-side surface of the parallel-hole collimator. The base calibration establishes an

FIG. 1. Robotic SPECT system, including a KUKA KR 150 L110 robot and an attached pinhole-collimated SPECT detector, imaging a ROI on the phantom side distal to the robot with the phantom on a radiation-therapy flat-top patient table. The robot tool coordinate frame is centered at the pinhole. The robot base coordinate frame is centered on the phantom. The robotic arm has six axes of rotation, labeled A1–A6.

FIG. 2. (a) Robot tool coordinates ($x_{\text{tool_par}}$, $y_{\text{tool_par}}$, $z_{\text{tool_par}}$) and base coordinates ($x_{\text{base_par}}$, $y_{\text{base_par}}$, $z_{\text{base_par}}$) for the parallel-hole collimated detector. The tool coordinate origin is at the center of phantom-side collimator surface. The base coordinate origin is at the center of the phantom. (b) Robot tool coordinates ($x_{\text{tool_pin}}$, $y_{\text{tool_pin}}$, $z_{\text{tool_pin}}$) and robot base coordinates ($x_{\text{base_pin}}$, $y_{\text{base_pin}}$, $z_{\text{base_pin}}$) for the pinhole collimated detector. The tool coordinate origin is at the center of the pinhole. The base coordinate origin is at the center of the pinhole CV, which encompasses the two largest spheres in the phantom.

additional user-specified coordinate frame. This is done by touching the tool coordinate origin to the user-intended base coordinate origin, then moving the tool coordinate origin along the user-intended base coordinate x -axis, and finally touching the tool coordinate origin to a point in the base coordinate x - y plane. This process is performed with tool y and z axes parallel to the base x and y axes, respectively, such that the orientation between the tool and base coordinate frames is established. By touching the two origins, the displacement between the two coordinate frames is also established. Here, one user-determined base coordinate frame was set with axes parallel and perpendicular to the surface and edges of the flat-top. The CT image of the phantom was then used to determine the shift needed to position another parallel base coordinate frame at a user-specified location in the phantom, as shown in Fig. 2(a) for parallel-hole imaging and Fig. 2(b) for pinhole imaging. At each detector stop, the robot reports the position and location of the tool coordinate frame relative to the base coordinate frame, and this provides the information needed for image reconstruction.

For broad cross-section parallel-hole imaging, the center of the thorax phantom was the center-of-rotation (COR), and the common volume (CV) was a 20.8-cm-diameter cylinder which covers all five spheres [Fig. 3(c)]. For pinhole imaging, the COR was at the center of a 14.7-cm-diameter spherical CV that covered the two largest spheres in the phantom [Fig. 3(a)]. For parallel-hole collimation, the detector rotation

angle θ spanned 180° in 3° steps. The pinhole trajectory spanned 228° in 3° steps.

2.C. Phantom and imaging

A PET CT PhantomTM (Data Spectrum, Durham, NC), including five spheres of 10, 13, 17, 22, and 28 mm diameters, was positioned on a radiation therapy flat-top table, as shown in Figs. 1, 2, and 4. The filled volume of the phantom excluding the spheres was 10 000 ml.

To provide the attenuation map for SPECT image reconstruction, a CT scan of the phantom was acquired (0.6 mm slice thickness and 0.64 mm/pixel size). The CT image was segmented into water, air, and polymethyl methacrylate (PMMA), and attenuation coefficients of 0.15, 0.00017, and 0.18 cm^{-1} were assigned, respectively.

Four imaging acquisitions were performed using the robotic SPECT system: parallel-hole collimated and pinhole collimated scans of the phantom with water and radioactivity in the background region (par-background and pin-background), and parallel-hole collimated and pinhole collimated scans with air in the background region (par-no-background and pin-no-background).

The Tc-99m activity concentration in each sphere at the start time of image acquisition and the total acquisition time was ($3.5 \mu\text{Ci/ml}$, 26 min) and ($2.5 \mu\text{Ci/ml}$, 37 min) for pin-no-background and par-no-background, respectively. The Tc-99m activity concentration in each sphere at the start time of image acquisition and the total acquisition time was ($2.0 \mu\text{Ci/ml}$, 45 min) and ($1.6 \mu\text{Ci/ml}$, 61 min) for pin-background and par-background, respectively. Sphere-to-background ratios were 10:1 in the two with-background cases. The pin-no-background acquisition was used as a reference such that the acquisition times of other scans were adjusted for radioactivity decay to maintain the equivalent of a 26-min scan with a sphere activity concentration of $3.5 \mu\text{Ci/ml}$. The product of total acquisition time and sphere activity concentration was approximately similar for each of the four acquisitions.

2.D. Reconstruction and image analysis

SPECT images were reconstructed from the projection data by up to 20 iterations of maximum-likelihood expectation-maximization (MLEM). The reconstruction modeled attenuation in the phantom, but it did not model the photon scatter.

FIG. 3. (a) ROIs used to calculate contrast of the spheres relative to background. c_s and c_b are the reconstructed activity concentrations in the sphere and background ROIs. The two 3D ROIs for c_s are indicated by the bright white regions inside the two spheres. The background ROIs are the two 3D shells. S1 is the larger sphere and S2 is the smaller sphere. (b) The pinhole noise ROI [white region in (b)] is the pinhole common volume except excluding voxels that are within the inner surfaces of the two background shells. (c) The parallel-hole noise ROI [white region in (c)] is the pinhole noise ROI except excluding voxels that are not contained in the parallel-hole CV. Pinhole and parallel-hole CVs are indicated by intermediate gray levels in (a) and (c), respectively.

FIG. 4. Trajectories for robotic SPECT with (a) parallel-hole collimation and [(b) and (c)] pinhole collimation. The phantom is on a radiation-therapy flat-top table. Four of the 61 parallel-hole and 8 of the 77 pinhole detector stops are shown. (c), in particular, zooms in to show the robotic pinhole SPECT system tracing closely along the flat-top so as to achieve best-possible proximity to the phantom.

Profiles of the spheres were plotted in the reconstructed images. The contrast of the spheres relative to background was measured as

$$C = \frac{c_s - c_b}{c_b},$$

where c_s and c_b are the reconstructed activity concentrations in the sphere and background ROIs, as shown in Fig. 3(a). The spherical ROI is inside the reconstructed sphere and is placed off center to avoid the air bubble at the top of the sphere. The background ROI is a 3D shell that is concentric with the corresponding sphere.

Noise was calculated as the standard deviation of voxel values for $\mu\text{Ci/ml}$

$$\text{Noise} = \sqrt{\frac{1}{n-1} \sum_{i=1}^n (\lambda_i - \bar{\lambda})^2},$$

where n is the number of voxels in the noise ROI, λ_i is the value at voxel i , and $\bar{\lambda}$ is the average voxel value in the ROI. The noise ROI excludes voxels that are within the inner surfaces of the two background shells. The noise ROI for pinhole imaging is shown in Fig. 3(b) as the bright region. The noise ROI for parallel-hole imaging is shown in Fig. 3(c) as the bright region given by the pinhole noise ROI except excluding voxels not contained in the parallel-hole common volume. Contrast versus iteration, noise versus iteration, and contrast versus noise were plotted for pinhole and parallel-hole images.

The contrast to noise ratio (CNR) of spheres was calculated as

$$\text{CNR} = \frac{C}{\text{Noise}},$$

where C , the contrast of spheres and Noise were calculated from the above equation.

RORs for pinhole and parallel-hole imaging trajectories were plotted versus detector angle. The geometric sensitivity and resolution corresponding to these parallel-hole and pinhole collimators were calculated based on the equations, as shown below²⁰

$$S_{\text{pin}} = \frac{d_{\text{pin}}^2}{16 \times \text{ROR}_{\text{pin}}^2},$$

$$R_{\text{pin}} = \sqrt{R_{\text{intrinsic}}^2 + \left(d_{\text{pin}} \left(1 + \frac{1}{\frac{f}{\text{ROR}_{\text{pin}}}} \right) \right)^2},$$

$$S_{\text{par}} = K^2 \frac{d_{\text{par}}^2}{(l)^2} \left(\frac{d_{\text{par}}}{d_{\text{par}} + t} \right)^2,$$

$$R_{\text{par}} = \sqrt{R_{\text{intrinsic}}^2 + \left(\frac{d_{\text{par}}(\text{ROR}_{\text{par}} + l)}{l} \right)^2},$$

where d_{pin} is the pinhole diameter, f is the pinhole focal length, d_{par} is the parallel hole diameter, K is a constant for hexagonal parallel-hole shape, l is the parallel-hole length, t is the parallel-hole septal thickness, and $R_{\text{intrinsic}}$ is the detector intrinsic resolution of 3.25 mm. Log-sensitivity versus resolution was plotted for all detector stops of both systems.

FIG. 5. (a) The CT image of the PET-CT phantom with five spheres of 10, 13, 17, 22, and 28 mm diameters. The black spots at the top of spheres are air bubbles. (b) The first (top left) and the last (bottom right) stops of the pinhole detector trajectory. The dotted line connecting the two pinhole stops illustrates that the CV, shown by the bright circle inside the phantom, is contained within the convex hull of that trajectory.

In addition to the pinhole and parallel-hole collimators used in the experiment, sensitivity and resolution were analytically calculated for a 4 mm pinhole collimator and a low energy high resolution (LEHR) parallel-hole collimator. The LEHR parallel-hole collimator has hexagonal holes on a hexagonal grid with a hole length of 32 mm, a flat-to-flat hole diameter of 1.5 mm, and a septal thickness of 0.2 mm.

3. RESULTS

Shown in Fig. 4(a) are four stops from the 180° trajectory of the parallel-hole collimated detector. Shown in Figs. 4(b) and 4(c) are eight stops from the 228° trajectory of the pinhole

collimated system. With both collimations, the robotic SPECT system traced along the flat-top and phantom surfaces so as to maintain the closest possible proximity to the ROI, while keeping the ROI in the field-of-view (FOV) and avoiding collision.

The CT image and the convex hull of the pinhole trajectory, which contain the entire pinhole ROI, are shown in Figs. 5(a) and 5(b). The CV of the pinhole imaging system is shown as the bright circle with a diameter of 14.7 cm. This CV is centered at the top of the 28-mm-diameter phantom sphere. It covers the 28- and 22-mm spheres. The dashed line connects the first and last stops of the pinhole trajectory and illustrates that the CV is contained within the convex hull of that trajectory.

Figure 6 shows transverse views and profiles for images reconstructed from parallel-hole and pinhole collimated SPECT, without and with background. In the reconstructed images without background, all five spheres are visible for the parallel-hole system and four spheres are visible for the pinhole system. In the reconstructed images with background, three of the spheres are visible for the parallel-hole system. For the pinhole system, the two spheres in the pinhole common volume are visible, while the three spheres outside the pinhole common volume are not. The image fusion between CT and SPECT shows good overlay of the spheres. The profiles show good sphere-to-background contrast recovery.

Plotted in Fig. 7 are contrast versus iteration, CNR versus iteration, and contrast versus noise. At each iteration, the pinhole image has better contrast [Fig. 7(a)] but less CNR [Fig. 7(b)] than the parallel-hole image. For pinhole imaging, sphere S2 has higher contrast than sphere S1. (On average, the pinhole trajectory is closer to S2 than S1.) For parallel-hole imaging,

FIG. 6. SPECT images obtained without (columns 1 and 2) and with (columns 3–5) background activity. The parallel-hole and pinhole images are from iteration 20 of MLEM. The pinhole image in column 5 was smoothed with a 3D Gaussian kernel with $\sigma = 0.256$ cm. Shown in row 1 are images for the entire phantom volume. In row 2, these SPECT images are fused with the CT image. Row 3 shows the SPECT images only over the CV of that acquisition, which is 20.8 cm in diameter for parallel-hole (columns 1 and 3) and for pinhole is 14.7 cm in diameter and shifted leftward and anteriorly (columns 2, 4, and 5). Row 4 shows profiles through the largest sphere in the row 3 images.

FIG. 7. (a) Contrast vs iteration, (b) CNR vs iteration, and (c) contrast vs noise for pinhole and parallel-hole images. In (c), the iteration number increases from left to right. Sphere labels S1 and S2 are defined in Fig. 3.

sphere S1 has higher contrast than sphere S2. (S1 is a larger sphere than S2.) Figure 7(c) shows that parallel-hole imaging has less noise at a given contrast, but that pinhole imaging achieves a greater maximum contrast than parallel-hole imaging, e.g., 4.3 versus 4.0 for S1 and 5.3 versus 3.6 for S2 at iteration 20.

Figure 8 shows the RORs at every detector angle for pin-hole and parallel-hole detectors. Plotted in Fig. 9 is log-sensitivity versus resolution over these RORs for the 3-mm-diameter pinhole and LEAP parallel-hole systems used in the experiment (solid data points) and for two additional systems: a 4-mm-diameter pinhole and a LEHR parallel-hole collimator (open data points). For the trajectories used in this paper, the pinhole detector has better resolution but lower sensitivity.

4. DISCUSSION

The experiments of this paper demonstrate a robot maneuvering a gamma camera about a phantom that is on a radiation-

FIG. 8. ROR vs detector angle for pinhole and parallel-hole systems. The zero degree angle is at the left side of the phantom. RORs are from the COR to either the pinhole or the phantom-side surface of the parallel-hole collimator.

therapy flat-top table. The experiments support the proposal that robotic SPECT could provide functional and molecular imaging as patients are in position for radiation therapy. As shown in Fig. 4, the robot was versatile in positioning the detector system close to, but not in collision with, the patient table, so as to minimize the distance between the detector collimator and the detector center of rotation. The robot implemented two cases of ROI imaging—a broader ROI for parallel-hole imaging and a smaller ROI for pinhole imaging. As shown in our computer-simulation studies¹ and other studies,^{21–23} ROI imaging can be effective at improving image quality within the ROI. Notably, ROI imaging may be particularly relevant to radiation therapy, especially onboard SPECT imaging, since the approximate tumor location is known from the planning CT or CBCT.

FIG. 9. Log-sensitivity vs resolution for pinhole and parallel-hole detectors over the range of RORs utilized during the SPECT scans. Each point corresponds to an ROR at which data were acquired. RORs increase from left to right. The smallest and largest RORs for the pinhole detector were 10.5 and 35.5 cm, and those for the parallel-hole detector were 12 and 18.5 cm, respectively. The solid points are for data acquired with collimators available in the lab. Open points are analytic calculations for collimators that were not available.

In this study, with the robot bolted to the floor, detector maneuverability was versatile enough for pinhole ROI imaging and broad cross-section parallel-hole imaging. Even for a ROI on the phantom side distal to the robot, the robot could maneuver the pinhole detector so as to enclose the ROI central plane in the convex hull of the pinhole trajectory. That noted, as shown in Fig. 4, the first stop of the pinhole detector on the distal side of the phantom is at the mechanical limit of the robot, and the inability to achieve additional pinhole positions on the distal side of the phantom necessitates the large-ROR pinhole positions on the proximal phantom side, shown in Fig. 4(c). Hence, while the robot shows substantial capability, limitations in detector maneuverability are also apparent. In the previous computer-aided design (CAD) work, we have shown that a robot mounted on a rail could provide more versatile detector motion, as illustrated in Figs. 2 and 1 of Refs. 1 and 2, respectively.

Robots comparable to the one considered here are already in use in many hospitals. These robots include the CyberKnife¹⁵ radiation therapy machine and the Siemens Artis Zeego C-arm x-ray system.¹⁸ Thus, the feasibility of using such robots in hospitals has been established. That noted, the robots do present structural requirements. For example, with the robot installation employed here, additional concrete in the floor was necessary in order to support the torque produced by the robot. The concrete floor requirements for these robots may exceed those of some radiation therapy rooms. Robots such as that investigated here, and employed for the CyberKnife and Siemens Artis Zeego, are designed to provide highly stable and reliable position when installation requirements are met.

Retail costs for the robot used in this study are in the approximate range of \$100 000. A company producing robotic SPECT would also have development costs in robot programming. Some of this expense would be offset in that a gantry would no longer be needed.

The pinhole diameter used here, 3 mm, was the largest available, and better pinhole imaging results might have been obtained with a larger pinhole. The intrinsic detector resolution is about 3.25 mm and the average magnification was about 1. Given the strong contribution of intrinsic detector resolution to total gamma camera resolution, use of a larger pinhole may have improved pinhole efficiency more substantially than it degraded total system spatial resolution. The poor sensitivity of the 3 mm pinhole is reflected in the much greater noise levels seen in the pinhole images and noise measurements, and this poor sensitivity is confirmed by the analytic calculations of Fig. 9. The analytic calculation in Fig. 9 for a 4 mm pinhole shows improved sensitivity but at the cost of worse spatial resolution. For the parallel-hole detector, similarly, Fig. 6 shows poor resolution mainly because the collimator available in the lab, a LEAP parallel-hole collimator, is a relatively poor-resolution collimator. The analytic calculation in Fig. 9 for a higher-resolution LEHR collimator shows improved resolution, albeit at the cost of less sensitivity at a given resolution.

The results demonstrate the ability of a robotic SPECT system to achieve the π -plus-fan-angle pinhole short-scan conditions²⁴ for acquiring all line integrals through the ROI central

plane, accomplishing this for an off-center ROI and in the presence of a radiation therapy flat-top. The robot is also demonstrated to be effective at maneuvering a compact parallel-hole collimated detector in the vicinity of the flat-top. The images, profiles, and contrast and noise curves reflect this SPECT imaging capability of the robotic approach.

In conventional gantry-based SPECT, detector pose might be given by one rotation angle and an angle-dependent ROR. These angles and RORs would be provided by hardware components of the SPECT gantry and would be utilized for SPECT image reconstruction. Commercial robots typically report a more complex set of coordinates, obtained by hardware components of the robot. The study here shows that these coordinates can be harnessed for SPECT image reconstruction. Because of their wide use in industry for precision tasks, economies of scale and market competition have made robots versatile and precise while also moderately priced. Commercial robots thus present a cost-effective opportunity for expanding the capabilities of SPECT imaging. It is pertinent then that the present study demonstrates the ability of general-purpose commercial robots to provide coordinates that can be transformed into those needed for SPECT image reconstruction.

Here, in the absence of a CBCT or Linac in the robot laboratory, the initial base coordinate frame was calibrated to align with the flat-top table surface and edges. A CT scan of the phantom was used to determine the shift needed to define a new base coordinate frame with origin at a user-intended location in the phantom. For robotic SPECT imaging at a Linac with CBCT, the initial base coordinate frame could be calibrated to align with that of the Linac and CBCT, and onboard CBCT images could be used to determine the shift needed to define a new base coordinate frame with origin at the clinician-intended location in the patient.

Before and after imaging, the robot retracted the SPECT system fully away from the flat-top, as it would also when installed in a Linac vault, thereby allowing for normal, unhampered operation of the radiation therapy system.

5. CONCLUSIONS

Robotic SPECT was investigated by integrating a gamma camera with a commercial robot. The experiments demonstrated the abilities of a robotic arm to achieve needed detector trajectory range and ROI coverage while maintaining close proximity to the phantom and avoiding collisions. It was shown that inherent robot coordinate frames can be utilized to estimate detector pose for use in SPECT image reconstruction. This study supports the proposal that robotic SPECT could provide onboard functional and molecular imaging for patients in position for radiation therapy on a standard flat-top table.

ACKNOWLEDGMENTS

This work was supported by PHS/NIH/NCI Grant No. R21-CA156390-01A1. The authors would like to thank KUKA Robotics Corporation (Shelby Township, MI) and Digirad

Corporation (Poway, CA) for technical support of the robot and detector; the Duke University Medical Center Radiopharmacy for providing Tc-99m radiotracer for the experiment; and Bernie Jelinek and Richard Nappi from the Duke University Physics Instrument Shop for constructing the connection joint between the detector and the robotic arm.

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