An EPID based method for performing high accuracy calibration between an optical external marker tracking device and the LINAC reference frame

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(Received 29 December 2011; revised 12 March 2012; accepted for publication 30 March 2012; published 24 April 2012)

Purpose: With the increasing use of external 3D optical tracking cameras to guide modern radiation therapy procedures, it has become vitally important to have an accurate camera to linear accelerator (LINAC) reference frame calibration. To eliminate errors present in current calibration procedures based on the manual hand alignment of a device using the light field crosshairs and in room guidance lasers, a semiautomated quantitative calibration approach requiring only use of an electronic portal imaging device (EPID) was developed.

Methods: A phantom comprised of seven highly IR reflective plastic BBs was placed on the LINAC treatment couch and imaged with both a 3D stereoscopic IR imager and the on board megavoltage (MV) EPID imager. Having knowledge of the optically determined 3D positions and projected EPID images of the BBs, simulated annealing was used to optimize the location of the BBs in the LINAC frame using four different optimization functions. Singular value decomposition was then used to calculate the transformation matrix between the camera and LINAC reference frames. Results were then compared to a traditional camera calibration method for overall accuracy.

Results: Using modeled data, the simulated annealing process was able to determine the actual locations of the BBs with a RMSE of 0.23 mm. Using projection images acquired with an MV imager, the process was able to determine locations of BBs within .26 mm. The results depend on the choice of optimization function.

Conclusions: Results show that the method can be used to provide highly accurate spatial registration between an external 3D imaging reference frame and the LINAC frame. The experimental MV imager results, while not as precise as the simulated results, exceed 1 mm accuracy and the current accepted AAPM TG-142 standard of ≤ 2 mm positioning accuracy. © 2012 American Association of Physicists in Medicine. [http://dx.doi.org/10.1118/1.3703836]

Key words: EPID, stereoscopic camera, calibration, QA

I. INTRODUCTION

Infrared (IR) marker based systems with stereoscopic cameras can provide 3D motion tracking with a high degree of accuracy. As such, they have been used for guidance in frameless stereoscopic radio surgery (SRS),^{1–3} gated RT treatments,⁴ and SBRT.^{5,6} However, even with the sub-millimeter 3D positional resolution of current IR tracking devices, the accuracy of the image guided system as a whole will be limited by the accuracy of the calibration step used to define the coordinate transformation between the camera reference frame and the LINAC radiation field frame. Currently, IR marker based guidance systems are not calibrated directly against the LINAC radiation field, but rather indirectly through use of light field crosshairs and in room guidance lasers.¹ Here, the light field and lasers are used to set up a phantom designed with a well known spatial distribution of IR markers [Figs. 1(b) and 2(b)].

Several sources of error currently exist in the traditional camera to LINAC calibration step. First, the light field crosshairs and lasers, even if well calibrated, have a finite width that can suffer from user dependent variation in determining their center. Second, according to AAPM Task Group 142,⁷ for the purpose of annual quality assurance (QA), the light field and lasers have finite tolerances of 2 mm [1 mm for lasers

in intensity modulated radiation therapy (IMRT)] in terms of fidelity to radiation isocenter. Third, the actual LINAC isocenter defined by the light field cross hairs and lasers is not a well defined point like origin that the center axis (CAX) of the beam passes through for every gantry angle, but rather, due to gantry flex, is the approximate center of a 2 mm in diameter 3D sphere that the CAX must pass through at all beam angles. To overcome these limitations, we propose a novel quantitative camera calibration method that avoids use of light field crosshairs, in room lasers, and precise manual hand positioning of the calibration phantom but rather calibrates the camera system directly to the radiation treatment field.

The basic concept was inspired by a technique first presented by Mao *et al.*⁸ Here, Mao *et al.* acquired MV and kV image projections of a custom built phantom containing a well known 3D distribution of embedded ball bearings (BBs) and then used the data to solve for MV and kV imager QA parameters. To use this concept toward calibrating an external 3D imager to a LINAC reference frame, we have developed a new approach that differs from Mao *et al.* in several key ways. This includes removing the need for a precise prior knowledge of 3D BB positions, developing a new mathematical model that allows the LINAC frame of reference to be determined for each gantry angle based on

FIG. 1. (a) Illustration of simulated projections to determine correct position of BBs in the phantom. (b) LINAC frame geometry.

MV images, removing the need for a high degree of certainty in the setup of the calibration phantom with respect to the radiation coordinate system, and employing a stereoscopic based MV image cost function for increased accuracy. Since 3D BB positions are acquired during the actual calibration step by the 3D external camera through use of an open 3D ensemble of IR reflective BBs, potential issues with the associated complex manufacturing of a calibration phantom containing a precise rigid 3D distribution of BBs are also avoided. The method therefore has potential to be superior to current camera calibration techniques, since there are fewer opportunities for uncertainties to be introduced, as fewer assumptions are made about the geometry of the treatment setup.

II. MATERIALS/METHODS

II.A. Overview of technique

Given the 3D spatial location data for a set of reflective BBs placed in the camera reference frame, the task is to find the translation vector and rotation matrix required to register the camera frame with the LINAC reference frame. As shown in Fig. 1(a), the 2D MV projection image of a 3D ensemble of BBs is strongly dependent on the specific geometry of the MV beam and flat panel detector. This can therefore be formulated

as an inverse problem, where the task is to solve for the set of geometric LINAC parameters such that the MV EPID images correspond to the 3D stereoscopic camera image. There are two ways of defining a cost function for this problem. First, the comparison can be made by using the camera measured 3D BB distribution data to simulate projection MV images. The simulated projections are then compared to the actual MV images in the projection space. By optimizing the geometric LINAC parameters that result in the best match between simulated and measured 2D projections, a transformation matrix can be determined.

Alternatively, the MV EPID images may be back-projected to reconstruct a 3D image of the phantom. The 3D back-projection can then be compared to the 3D stereoscopic camera image in order to optimize the geometric LINAC parameters.

As shown in Fig. 1(b), the LINAC frame is defined with the origin at the radiation isocenter, which is the center of the ellipsoid created by the intersection of beam centers from different gantry angles. The x-axis is along left–right directions, the z-axis runs superior–inferior, and the y-axis is perpendicular to the ground. In this work, the three axes retain the same orientation, but the point origin of the LINAC frame can be different for each gantry angle due to mechanical flex. Therefore, at any gantry angle, we define the origin of the LINAC frame as the intersection point of its radiation central axis (CAX)

FIG. 2. (a) Novel calibration phantom setup on LINAC couch. (b) Manufacturer supplied phantom setup on LINAC couch, with level adjusted and aligned with lasers. (c) Novel calibration phantom in the Polaris coordinate system.

with the CAX of a nearby gantry angle. Here, we make the assumption that gantry angles $\leq 10^\circ$ apart do not undergo mechanical distortion and thus only differ by gantry angle.

The imaging system was characterized by the parameters of Table I. Through optimization, these parameters are solved to determine a transformation between the LINAC coordinate system and the stereoscopic camera coordinate system. ΔX , ΔY , ΔZ , γ , δ , and ε are the translations and rotations of the transformation matrix. ΔR , ΔF , α , and β are parameters that describe the geometry of the MV image acquisition, where SAD is the source to central axis distance, and SDD is the source to detector distance.

II.B. Forward and back projection calculations

II.B.1. Calculation of projection space images

As shown by Mao *et al.*,⁸ if the coordinates of the radiation source in the LINAC frame can be given in the cylindrical coordinates (R, φ, y_B) , the coordinates of a BB in the LINAC frame can be given in the cylindrical coordinates as (r, θ, y) , with SAD R and SDD F , then the coordinates of the BB in the projection image are given as

$$u = \frac{-Fr \cos(\theta - \varphi)}{R + r \sin(\theta - \varphi)}, \quad v = \frac{F(y_B - y)}{R + r \sin(\theta - \varphi)}.$$

Taking into account the fact that the detector may be tilted by β degrees along an axis rotated by an angle α , there exists a transformation into a rotated detector

$$\begin{pmatrix} p \\ q \end{pmatrix} = \begin{pmatrix} \cos \alpha & \sin \alpha \\ -\sin \alpha & \cos \alpha \end{pmatrix} \begin{pmatrix} u \\ v \end{pmatrix},$$

such that the projected coordinates on a tilted detector are given

$$\begin{pmatrix} u_1 \\ v_1 \end{pmatrix} = \begin{pmatrix} u \\ v \end{pmatrix} + \left(\frac{q}{F} + \frac{\beta}{2} \right) q \beta \begin{pmatrix} -\sin \alpha \\ \cos \alpha \end{pmatrix}.$$

II.B.2. Calculation of back-projection of MV EPID images

A point object is imaged with two different gantry angles, a_1 and a_2 , and source to detector distance F and source to axis

distance R . At angle a_1 , the object has the coordinates (u_1, v_1) on the imaging plane, and at angle a_2 , the object has the coordinates (u_2, v_2) on the imaging plane. The detector may be shifted from the center of the beam line by an amount $(\Delta u, \Delta v)$. Then, we can define two lines in space from the source to the image at different gantry angles by the four points

$$P_1 = \begin{bmatrix} \cos a_1 & 0 & \sin a_1 \\ 0 & 1 & 0 \\ -\sin a_1 & 0 & \cos a_1 \end{bmatrix} (u_1 + \Delta u, v_1 + \Delta v, R - F),$$

$$P_2 = \begin{bmatrix} \cos a_1 & 0 & \sin a_1 \\ 0 & 1 & 0 \\ -\sin a_1 & 0 & \cos a_1 \end{bmatrix} (0, 0, R),$$

$$P_3 = \begin{bmatrix} \cos a_2 & 0 & \sin a_2 \\ 0 & 1 & 0 \\ -\sin a_2 & 0 & \cos a_2 \end{bmatrix} (u_2 + \Delta u, v_2 + \Delta v, R - F),$$

$$\text{and } P_4 = \begin{bmatrix} \cos a_2 & 0 & \sin a_2 \\ 0 & 1 & 0 \\ -\sin a_2 & 0 & \cos a_2 \end{bmatrix} (0, 0, R),$$

where P_1 and P_2 define one line, and P_3 and P_4 define a second line. Then, in the reconstructed 3D space, the object lies at the interception of these two lines. Because even a small error can result in two lines that do not intercept, we calculate the point of closest approach on one line.

One line can be defined as

$$L_1(m) = P_1 + m(P_2 - P_1),$$

and the other line can be defined as

$$L_2(n) = P_3 + n(P_4 - P_3).$$

To find the point of closest approach, we define

$$c_1 = (P_4 - P_3) \times (P_2 - P_1),$$

$$c_2 = (P_4 - P_1) \times (P_4 - P_3),$$

and

$$b = c_2/c_1,$$

where b is understood to be the solution in the least squares sense to the equation

$$bc_1 = c_2.$$

Then, the point of closest approach on the line L_1 is given as

$$p = P_1 + b(P_2 - P_1).$$

The forward and back-projection methods of Sec. II.B can be used to bring images from the EPID system and the stereoscopic IR camera into the same reference frame. Once in the same reference frame, the correspondence can be evaluated with a number of cost functions. Four different cost functions were investigated for parameter optimization. Each cost function required the optimization of a different subset of the parameters, as not every parameter affects the outcome of each cost function in a strong way.

TABLE I. List of parameters to be optimized. ΔX , ΔY , ΔZ , γ , δ , and ε are the translations and rotations of the transformation matrix. ΔR , ΔF , α , and β are parameters that describe the geometry of the MV image acquisition, where SAD is the source to central axis distance, and SDD is the source to detector distance.

ΔX	Translation in x-direction
ΔY	Translation in y-direction
ΔZ	Translation in z-direction
α	Orientation of detector tilt
β	Magnitude of detector tilt
γ	Rotation around z-axis
δ	Rotation around y-axis
ε	Rotation around x-axis
ΔR	Deviation from nominal SAD
ΔF	Deviation from nominal SDD

II.C. Cost functions

II.C.1. First cost function: projection space, relative distance

The first cost function minimizes the RMSE of the distance between any set of BBs on the MV projection

$$C = \text{RMSE} = \sqrt{\frac{2}{n(n-1)} \sum_{i=1}^{n-1} \sum_{j=i+1}^n (\hat{d}_{ij} - d_{ij})^2},$$

where $\hat{d}_{ij}$ is the estimate of the distance between the i th and j th BBs on the imager plane based on simulating projections from 3D data using the technique of Sec. II.B.1 and an estimate of transformation parameters, and d_{ij} is the actual distance between their projections from the MV image. In this and every cost function, n is the number of BBs.

This cost function does not depend on the rotation of the phantom along the beam axis; therefore, to optimize all of the necessary parameters, two projections separated by 10° must be used. The separation of 10° was chosen because it is sufficiently large to provide a stereoscopic pair of images, but sufficiently small so as not to introduce substantial changes in the characterization of gantry sag.

II.C.2. Second cost function: projection space, absolute distance

Similar to the first cost function, we compare the simulated projection to the actual projection. In this case, we use the actual difference in locations, rather than the collection of pairwise distances, to calculate cost

$$C = \sum_i^n (|u_i - \hat{u}_i| + |v_i - \hat{v}_i|),$$

where u_i is the u -location of the i th BB in the projection space, $\hat{u}_i$ is the estimate of the u -location of the i th BB in the projection space, and v follows similarly.

II.C.3. Third cost function: 3D space, relative distance

In the third cost function, we use the two MV images as a stereoscopic pair to produce a 3D reconstruction of the phantom. Then, as in the first cost function, we use the RMSE of the distance between any two sets of BBs; however, this time in 3D space. The cost function is defined as

$$C = \text{RMSE} = \sqrt{\frac{2}{n(n-1)} \sum_{i=1}^{n-1} \sum_{j=i+1}^n (\hat{d}_{ij} - d_{ij})^2},$$

where this time d is defined in 3D back projection space, with estimated distances $\hat{d}_{ij}$ based on the estimated back projection locations using the technique of Sec. II.B.2.

II.C.4. Fourth cost function: 3D space, absolute distance

Similar to third cost function (Sec. II.C.3), two MV images are used as a stereoscopic pair to produce a 3D

reconstruction of the phantom. This time, we minimize the absolute distance between BBs in the two reference frames. In this way, the cost function is given by

$$C = \sum_i^n (|x_i - \hat{x}_i| + |y_i - \hat{y}_i| + |z_i - \hat{z}_i|),$$

based on the difference between the actual location of BBs in the reconstruction frame and the transformed Polaris frame, rather than the collection of pairwise distances, with similar definitions as in the second cost function. x_i , y_i , and z_i are BB locations in the stereoscopic IR camera reference frame, and $\hat{x}_i$, $\hat{y}_i$, and $\hat{z}_i$ are the estimated back-projection locations using the technique of Sec. II.B.2, translated and rotated according to the estimated parameters.

II.D. Cost function optimization with simulated annealing

Due to the large size of the parameter space and the nature of the parameters selected, the cost functions were found to contain a large number of local minima. To avoid convergence on such local minima, simulated annealing was used.⁹ Simulated annealing is an optimization process in which at each step, the proposed solution moves in a random direction as specified by a generator function. The new solution is then either accepted or rejected (as a basis for the next guess) based on both whether the new solution provides a lower cost function, and the current “temperature” of the system. The temperature is a variable that is constantly decreasing with iterations according to a cooling schedule, and at higher temperatures, it is increasingly likely that a less optimal solution will be accepted, therefore allowing early iterations opportunities to escape from local minima, while ensuring that later iterations zero in on local minima. In theory, with a sufficiently slow (logarithmic) cooling schedule, the solution will settle on the true global minima; however, in practice, relatively faster cooling schedules are used to balance accuracy and computation time. In the MATLAB implementation of simulated annealing that we use, we use a slow linear cooling schedule. Our generator function makes a random choice of a single parameter to change by an amount given from a uniform distribution.

II.E. Calculating the transformation matrix between camera and LINAC reference frames

To transform between the camera and LINAC reference frames, singular value decomposition (SVD) is used with the coordinates of n BBs in either frame. At least four BBs are required, because it takes the locations of four BBs to absolutely fix a rigid body’s orientation in 3D space. The n points on the phantom are given by the coordinates

$$(\bar{x}_i, \bar{y}_i, \bar{z}_i)$$

in one coordinate system, and

$$(\bar{x}'_i, \bar{y}'_i, \bar{z}'_i)$$

in a second coordinate system. Two matrices are defined

$$G = \begin{bmatrix} x_1 & y_1 & z_1 \\ \vdots & \vdots & \vdots \\ x_n & y_n & z_n \end{bmatrix}, \quad H = \begin{bmatrix} x'_1 & y'_1 & z'_1 \\ \vdots & \vdots & \vdots \\ x'_n & y'_n & z'_n \end{bmatrix}$$

to describe the phantom in two different coordinate systems. The phantom has a center of mass

$$(\bar{x}, \bar{y}, \bar{z}) = \frac{1}{n} \left(\sum_i x_i, \sum_i y_i, \sum_i z_i \right),$$

$$(\bar{x}', \bar{y}', \bar{z}') = \frac{1}{n} \left(\sum_i x'_i, \sum_i y'_i, \sum_i z'_i \right),$$

in the two coordinate systems. New matrices may then be defined, describing the phantom in the same two coordinate systems, only shifted so that in either coordinate system, the phantom is centered on the origin

$$J = \begin{bmatrix} x_1 - \bar{x} & y_1 - \bar{y} & z_1 - \bar{z} \\ \vdots & \vdots & \vdots \\ x_n - \bar{x} & y_n - \bar{y} & z_n - \bar{z} \end{bmatrix}$$

$$K = \begin{bmatrix} x'_1 - \bar{x}' & y'_1 - \bar{y}' & z'_1 - \bar{z}' \\ \vdots & \vdots & \vdots \\ x'_n - \bar{x}' & y'_n - \bar{y}' & z'_n - \bar{z}' \end{bmatrix}.$$

For the purpose of SVD, a new matrix is introduced

$$E = K^T J = Q J^T J$$

in which we note that the matrix K is equal to the matrix J multiplied on the right by the transpose of a rotation matrix. Using SVD, it is possible to decompose the matrix E as the product

$$E = U S V^T,$$

where S is a diagonal matrix of shared eigenvalues to the eigenvectors making up the unitary matrices U, V .¹⁰ Then, Q may be written as

$$Q = U \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & \det(U \cdot V^T) \end{bmatrix} V^T.$$

Thus, the rotation matrix, Q , and the translation vector,

$$(\bar{x}, \bar{y}, \bar{z}) = Q(\bar{x}', \bar{y}', \bar{z}'),$$

can be used to transform from one coordinate system to the other.^{10,11}

II.F. IR stereoscopic camera

A stereoscopic infrared camera (Polaris, Northern Digital Instruments, Ontario, CA) was used to acquire a 3D image of the reflective BB phantom. According to manufacturer's specifications, a single frame captured by the camera has a resolution of 0.25 mm (Ref. 12) with approximately 30 frames per second. The image observed by the camera (in the camera's frame of reference) is given in Fig. 2(c). To improve spatial precision, 300 frames were captured as plotted in Fig. 3. For a single BB, the calculated statistical uncertainty in the camera's position estimation was $\sigma_x = 0.026$ mm, $\sigma_y = 0.008$ mm, and $\sigma_z = 0.053$ mm.

II.G. MV EPID images

MV images of the phantom were acquired using an MV EPID system (PortalVision aS1000, Varian Medical Systems, Palo Alto, CA) with an active surface area of 400×300 mm and 0.392 mm resolution (1024×768 pixel imaging array). Projection images of the phantom were acquired for 0° and 10° [Fig. 4(a)].

Once the images were acquired, they were analyzed to determine the location of each BB's projection on the imager. A template matching algorithm was used to quickly segment the BB projections in the projection data. The image of one projected BB was used as a template [Fig. 4(b)], and correlation score was calculated for every pixel in the image [Fig. 4(c)]. Pixels with correlation higher than 0.7 were selected and grouped into candidates, and the centroid of each candidate was calculated as the BB location [Fig. 4(d)].

FIG. 3. Histogram of statistical uncertainties in BB location from IR camera.

FIG. 4. (a) MV image, (b) template, (c) correlation scores, and (d) locations of BBs overlaid with MV image.

III. RESULTS

The performance of each of the four cost functions (Secs. II.C.1–II.C.4) was determined using tests on three classes of data, in two cases simulated, and in one case, experimentally acquired

1. The problem was modeled with noise free data. Here, a virtual phantom was constructed and used to create analytically determined projection images based on an arbitrary set of parameters. The four cost functions were then used to solve for the set of parameters from the virtual phantom data and simulated projection data.
2. The same virtual phantom was used, but random Gaussian distributed noise with half millimeter standard deviation was added to model the noise characteristics of the system. The cost functions were once again used to solve for the optimal parameterization.
3. Real images from the EPID system and stereoscopic camera were used for parameter optimization. These images

are of the same nature as any images that would be used in a clinical setting. Once again, the four cost functions were used to solve for the optimal parameterization.

The first two tests use simulated data. Therefore, the truth is known exactly, and we can evaluate the transformation associated with each cost function by the RMSE of the distance between the locations of the BBs after using the transformation, and the BBs known locations. When random error was introduced in the simulated data, experiments were repeated 100 times with different random error, in order to produce a distribution of results as the outcome could depend heavily on the particular characteristics of the noise introduced in a single trial. For the third test of each cost function, all data are experimentally acquired, and because the task is to determine an accurate transformation matrix, there is no *absolute* gold standard against which to compare. Therefore, when using real data, we evaluate the accuracy of the transformation based the RMSE between the locations of

FIG. 5. Comparison of gold standard BB locations (filled) to calculated BB locations (empty) for the four cost functions investigated, when using data from EPID imager and Polaris camera, rather than computer modeled data.

the BBs after our transformation and the locations of the BBs when using a transformation matrix derived from the manufacturer supplied calibration phantom [Fig. 2(b)]. The manufacturer supplied phantom was aligned using lasers calibrated as according to the Winston-Lutz technique,¹³ and the transformation was determined with SVD, treating the manufacturer phantom BB locations in the LINAC frame as known.

Figure 5 compares the BB locations from the four different empirically derived transformations to their gold stand-

ard locations. Table II shows the RMSE for each cost function and each method of evaluation.

IV. DISCUSSION

With increasing use of external optical tracking cameras to guide modern radiation therapy procedures, it has become vitally important to have an accurate camera to LINAC frame of reference calibration. To eliminate potential errors introduced through using light field

TABLE II. RMSE of cost functions in various tests. Uncertainty in simulated results is based on 100 trials with different noise characteristics. Empirical result data are the result of imaging the phantom with the EPID system and the Polaris camera under conditions that would be seen in clinical preparation.

	2D relative	2D absolute	3D relative	3D absolute
Simulation, no error	0.112 mm	0.3 mm	4.5×10^{-4} mm	1.2×10^{-5} mm
Simulation, 0.5 mm error	8.46 ± 8.25 mm	1.164 ± 0.458 mm	0.234 ± 0.070 mm	0.249 ± 0.072 mm
Empirical results	4.068 mm	1.021 mm	0.259 mm	1.052 mm

crosshairs, in room guidance lasers, and manual hand positioning, we have developed a semiautomated quantitative calibration approach that only requires use of the MV treatment beam. Here, calibration is made directly against the actual MV treatment beam that fundamentally is the true reference standard in terms of patient positioning. Since the method can define a new LINAC frame of reference for each gantry angle, it can be used to take into account gantry flex and to define a well-defined point like isocenter. The method was found to meet the requirements for calibration set forth by AAPM TG 142, and does so while determining numerous other LINAC parameters that can be used for other QA purposes. Furthermore, the phantom does not require precise machining and is relatively inexpensive to manufacture.

On investigating four different cost functions, several techniques were shown to provide a registration between the camera reference frame and the radiation reference frame on the order of 0.25 mm when using simulated sources of error, and between 0.25 and 1 mm when using empirically collected data. As shown in Table II, the RMSE errors were found to be much larger for both the simulated with noise and empirical experiments when using a 2D MV image method. Because cost function 3 (3D space, relative positions) performed best with both simulated and real data, we recommend the use of cost function 3 for real cases. The error may be larger when using 2D cost functions because the noise can be smoothed out in the back-projection step of the 3D cost function. Additionally, requirements of cost function 1 are not as stringent as the requirements of cost function 2, so the increased spread in results may be a result of a degeneracy of adequate matches when using noisy data and cost function 1. Additionally, we note that in this case much of this 2D error was found to occur along MV beam axis. This is consequence of the close relationship between the axial position, and a number of other parameters, in the calculation of back projections and forward projections. For example, when considering the data from only a single gantry angle, small shifts in axial position can be equivalently considered as shifts in magnification from altering the SDD. Finally, errors in the calculation of detector tilt will alter the apparent magnification of distances between BBs on the projection plane. Errors in magnification will result in errors in calculation of axial position.

It is envisioned that the method will be incorporated into the clinical workflow in the following way. Before start of a radiation therapy procedure that involves use of the optical camera system, the user will place the phantom [Fig. 2(a)] on the treatment couch and loosely align with the in room lasers. The user will then exit the treatment room and acquire MV projections at two closely spaced gantry angles while simultaneously imaging the phantom with the optical camera system. The user will then run the algorithm, which takes as input the 3D BB locations from the optical camera and the two MV images to compute the transformation between the LINAC frame and the camera frame. The calculated transformation will then be compared to the last used camera transformation to ensure

consistency. If the newly calculated transformation and the prior used transformation are found to agree within a small millimeter level tolerance, the new transformation will then be used for patient tracking during the radiation therapy session. If the new and old transformations differ significantly, the user will be notified and prompted to perform further investigation. This could indicate that the optical camera system has physically shifted or that the algorithm has malfunctioned.

This protocol has the potential to improve current image guided radiation therapy procedures that use external marker tracking in several ways. First, it improves camera-to-LINAC frame transformation accuracy since the calibration is performed directly against the treatment beam, thus removing the need of intermediary devices such as the light field crosshairs or in-room lasers, which can suffer from their own inaccuracies. This will likely have the most impact in radiotherapy procedures that require extremely accurate patient positioning such as SRS or SBRT.¹⁵ Second, the method is completely quantitative in nature, with the accuracy limited by the EPID and camera detection resolutions. This allows for fewer opportunities for human introduced error than previous methods, which are generally qualitative in nature and depend strongly on the individual user's hand to eye coordination in aligning the calibration stage to the light field crosshairs or in-room lasers. Third, laser calibration is known to be a compromise for a different set of ideal calibrations for different gantry angles. The true isocenter has been shown to be a volume of approximately 1 mm,¹⁶ so any single transformation may be off by as much as 1 mm for some gantry angles. This method allows for a different calibration unique to each gantry angle, therefore, removing that 1 mm of error. Finally, since the method is largely automated, it can be performed within minutes, offering significant time savings that may help to improving clinical work flow by reducing patient setup time.

Though the results show that this method meets the goals of TG142, with further refinement in algorithm and image segmentation, the technique has the potential to be used as a replacement for the Winston-Lutz test. The Winston-Lutz test only provides a calibration of a single point at the nominal isocenter, whereas the new method provides a seven point calibration, ensuring that the six dimensional transformation is valid for a large volume, rather than a single point. The ability to replace the Winston-Lutz test with this method is currently under investigation.

The overall accuracy of the method is fundamentally limited by the center of BB detection precision of the optical camera and EPID imager. Here, the major sources of inaccuracies are the intrinsic ~ 0.25 mm error in the Polaris camera and the errors involved in the EPID image acquisition and subsequent image segmentation. There are a number of ways to reduce the uncertainty in the model. First, the MV image segmentation can be improved with a more adaptive template, or with use of an entirely different technique, such as the Hough transformation, which has been demonstrated to register the location of spherical

markers with accuracy on the order of 0.02 mm.¹⁴ Additionally, the MV imaging system can be more accurately modeled to eliminate some sources of uncertainty in preprocessing of images. Additionally, over time, it is expected that improvements in optical tracking cameras will result in higher 3D accuracies.

V. CONCLUSIONS

It has been shown for the first time that a 3D arrangement of markers, viewed simultaneously by both the optical tracking camera and MV imager, can be used to accurately calibrate the camera to the LINAC frame of reference. Unlike previous external camera calibration techniques, the presented method does not require precise phantom positioning based on light field crosshairs and in room guidance lasers, thereby eliminating several potential sources of error.

ACKNOWLEDGMENTS

Financial support for this work was provided by NIH Grant No. T32 EB002103 and the Cancer Research Foundation.

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