GammaPod—A new device dedicated for stereotactic radiotherapy of breast cancer

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Purpose: This paper introduces a new external beam radiotherapy device named GammaPod that is dedicated for stereotactic radiotherapy of breast cancer.

Methods: The design goal of the GammaPod as a dedicated system for treating breast cancer is the ability to deliver ablative doses with sharp gradients under stereotactic image guidance. Stereotactic localization of the breast is achieved by a vacuum-assisted breast immobilization cup with built-in stereotactic frame. Highly focused radiation is achieved at the isocenter due to the cross-firing from 36 radiation arcs generated by rotating 36 individual Cobalt-60 beams. The dedicated treatment planning system optimizes an optimal path of the focal spot using an optimization algorithm borrowed from computational geometry such that the target can be covered by 90%–95% of the prescription dose and the doses to surrounding tissues are minimized. The treatment plan is intended to be delivered with continuous motion of the treatment couch. In this paper the authors described in detail the gamma radiation unit, stereotactic localization of the breast, and the treatment planning system of the GammaPod system.

Results: A prototype GammaPod system was installed at University of Maryland Medical Center and has gone through a thorough functional, geometric, and dosimetric testing. The mechanical and functional performances of the system all meet the functional specifications.

Conclusions: An image-guided breast stereotactic radiotherapy device, named GammaPod, has been developed to deliver highly focused and localized doses to a target in the breast under stereotactic image guidance. It is envisioned that the GammaPod technology has the potential to significantly shorten radiation treatments and even eliminate surgery by ablating the tumor and sterilizing the tumor bed simultaneously. © 2013 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4798961>]

Key words: breast cancer, stereotactic radiotherapy, SBRT, GammaPod

I. INTRODUCTION

According to the American Cancer Society, breast cancer is the most prevalent malignancy among women in the United States, with more than 220 000 new cases per year.¹ Awareness and regular screening have resulted in more patients being diagnosed in the early-stage of the disease. For stage I and II breast cancer, breast conserving therapy (BCT) has become an accepted treatment option. According to the standard of practice, BCT consists of a lumpectomy followed by 5–7 weeks of radiation therapy (RT), mainly irradiating the whole breast using external beam, which normally starts a

few weeks following lumpectomy to allow the swollen surgical cavity filled with seroma go through the initial healing process.

Compared with mastectomy, BCT provides comparable outcome with much superior cosmetic results and reduced psychological and emotional trauma.^{2–5} Most of the current EBRT techniques in BCT treat the whole breast in supine position using parallel-opposed tangential beams with 25–33 fractions over 5–7 weeks. Such long overall treatment duration of BCT significantly worsens the quality of life of the patients, and also discourages some women from choosing BCT over mastectomy despite the obvious cosmetic and emotional

advantages.^{6,7} This is especially true for the patients who are elderly or who live a significant distance from treatment centers. Acute and late toxicities of present therapy further compromise the quality of life for patients receiving BCT.⁸ Whelan *et al.* has reported their clinical trial with hypofractionation and demonstrated that for most patients receiving BCT, the overall treatment time can be shortened.⁹ Irradiation the breast in prone position has also been shown to improve dose uniformity, lower doses to the lung and the heart, and minimizing the effects of breathing induced motion, especially for large pendulous breasts.¹⁰ However, prone positioning is more difficult for both the patients and the therapists and is not widely adopted.

Breast cancer is multifocal by nature with the most common arrangement being a tumor foci clustering about a central mass.^{11,12} Research studies found that the accumulated probability of finding tumor foci decreases sharply with the distance from the edge of the central mass,¹² which is consistent with clinical findings that most of the local recurrences were within 20 mm from the surgical margin.^{13,14}

The benefit of irradiating the breast following lumpectomy is the sterilization effects of radiation on the microextensions of the tumor around the surgical bed. This is the main hypothesis of accelerated partial breast irradiation (APBI), where only the tissues surrounding the surgical bed is irradiated to sterilize the residual tumor foci.¹⁵ Some early results of APBI support the hypothesis that for a selected group of patients, whole breast irradiation is unnecessary.^{16–18} The recently published results of the targeted intraoperative radiotherapy (TARGIT-A) trial also added strength to this argument. In the TARGIT-A trial, a single fraction of intraoperative radiotherapy restricted to the tumor bed is compared with the conventional WB-EBRT.^{19,20} At 4 yr of follow-up of over 1100 patients in each cohort, a single dose of radiotherapy delivered at the time of surgery has shown to achieve equivalent local control to WB-EBRT delivered over several weeks.

There have been intensified efforts in recent years for delivering higher dose per fraction for postoperative²¹ and preoperative^{22,23} breast irradiation. The success of these trials points out not only the feasibility of a shorter treatment regimen, but also the strong likelihood that such shorter radiation treatment could be as effective as conventional fractionation schemes for a selected patient population.

However, there are obstacles for delivering a large dose precisely to a target in the breast using EBRT. First, unlike the brain, where the radiation beam can approach the intracranial lesion from more than 2π solid angle, the breast can only be approached from limited unobstructed angles by external radiation beams generated by a linear accelerator. Second, unlike the skull that can be securely fixated into a coordinate system so that the geometric discrepancy between imaging and treatment is minimal, such degree of immobilization and stereotactic localization has not been achieved with breast previously with EBRT.

To address these obstacles, a new EBRT device named GammaPodTM is developed. The GammaPod can be used for irradiating the breast under stereotactic image guidance. In

this paper, the gamma radiation unit, stereotactic localization of the breast, and the treatment planning system (TPS) of the GammaPod system are described. It is envisioned that GammaPod can be used for both accelerated postoperative radiotherapy and radiosurgery with concurrent tumor bed sterilization.

II. METHODS AND MATERIALS

The design goal of the GammaPod as a dedicated system for treating breast cancer is the ability to deliver ablative doses with sharp gradients under stereotactic image guidance. Stereotactic localization of the breast is achieved by a vacuum-assisted breast immobilization cup with built-in stereotactic frame. The use of vacuum-assisted breast cup allows the breast to be imaged at separate locations in the hospital analogous to use of head frames in brain radiosurgery. Highly focused radiation is achieved at the isocenter due to the cross-firing from 36 radiation arcs generated by rotating 36 individual Cobalt-60 beams. The dedicated treatment planning system determines the focal spot path using an optimization algorithm borrowed from computational geometry such that the target is covered uniformly by the prescription dose and the doses to surrounding tissues are minimized. The treatment plan is intended to be delivered with continuous motion of the treatment couch. Sections II.A–II.C describe the radiation unit, immobilization and stereotactic localization, and the treatment planning system.

II.A. Gamma radiation unit

The Gamma radiation unit focuses 36 rotating beams to the isocenter to achieve an intense focal spot with a sharp dose fall-off gradient. As shown in the front-view [Fig. 1(a)] and cross-section-view [Fig. 1(b)] of the GammaPod radiation unit, the main components of the unit include the top shielding door that is 81 cm from the floor level, the shielding body, radiation source carrier, collimator, and the patient support system.

II.A.1. Source carrier

All of the sources are contained in the hemispherical source carrier with each source located at 380 mm from the isocenter. To maximize the cross-fire effect, the source carrier rotates together with the selected collimator such that the beams form 36 noncoplanar and nonoverlapping arcs converging at the isocenter. The 36 sources are evenly spaced every 10° longitudinally and every 1° latitudinally from 18° to 53° in the source carrier to fully utilize the shielding power of the rotationally symmetric shielding body.

With the source to focus distance (SFD) of 380 mm, the dose rate to a small volume of water at the isocenter is calculated as

$$D = D_{\text{air}}^* (\mu_{\text{en}}/\rho)_{\text{air}}^{\text{water}} = 0.974 \Gamma^* A^* f/38^2, \quad (1)$$

FIG. 1. (a) The GammaPod radiation unit without the cover showing the treatment couch at its near upright position with the breast cup locked in the opening. Also shown are the treatment doors, the support columns, and the base frame the unit rests on; (b) break-up view of the central section of the radiation unit showing the major inner components including the collimator, the source carrier, and the shielding body, all concentrically mounted.

where $\Gamma = 13\,700\text{ R cm}^2/\text{Ci-h}$ is the dose rate constant for Cobalt-60, A is the activity in Ci, and f is the self-attenuation factor of the source, which is estimated to be 0.95.

To achieve an initial dose rate of 5 Gy/min inside a 7.0 cm radius breast phantom centered at the isocenter, the total initial activity of Cobalt-60 is estimated to be 4380 Ci using Eq. (1) assuming a tissue-air-ratio of 0.78. For the initial system used for testing, a total activity of 4320 Ci, or 120 Ci per source, was used. The Cobalt-60 source is double-encapsulated with the inner capsule packed tightly with cylindrical Cobalt-60 pellets of 1 mm in diameter and 1 mm long. The cylindrical shaped inner capsule of the source shell measures approximately 11 mm in height and 3.4 mm in diameter, which reflects the true radiation source size of a single source. The double-encapsulated source is placed in another stainless steel container which also contains a small primary collimator made of tungsten alloy.

The 36 beam orientations (latitudinal angle) span from 18° to 53° off the horizontal plane, with 1° latitude intervals. Large beam angles, i.e., beams shooting directly into the chest, should be avoided to minimize damage to the critical organs in the chest. For the beams to be able to reach lesions near the chest wall, the isocenter needs to be able to reach a couple of centimeters above the bowl surface so that when the treatment couch is lowered to the lowest allowable height, the center of the focal spot (i.e., the isocenter) is approximately 1 cm above the top surface of the couch. For the isocenter to reach higher (more chest wall coverage) with preferred smaller beam angles (exit beams avoid the lungs/heart), the sources need to be seated as shallow as possible. However, considering the limit on the leakage radiation from the top surface of the source carrier, the sources cannot be seated too shallow from the top surface of the bowl. To best achieve these conflicting goals, tungsten alloy is used at directly above

the shallowest six individual sources. Figure 2(a) shows the shielding around the uppermost source at 18° .

II.A.2. Collimator

Inside and concentric to the source carrier is the collimator carrier, also bowl-shaped with two sets of 36 collimators and one set of 36 tungsten rods. One set of the collimators has a projected opening of 15 mm at the isocenter and another set is 25 mm. Inside the collimator is a hemielliptical-sphere treatment space, within which the breast is positioned. Unlike radiosurgery of intracranial lesions, very small collimator sizes are generally not needed for breast cancer due to the typical tumor sizes and the need to sterilize the tissues surrounding the tumor owing to its multicentricity nature. If the source were infinitesimal as a point, the design of the collimator would be straightforward. Because the source has a physical dimension (cylinder of 11 mm in height and 3.4 mm in diameter), the apex of the collimator cone must be placed further away from the source such that the entire source is contained within the cone. Figure 2(b) illustrates the cone angle design for the 15 mm (i) and 25 mm (ii) collimators. The close-up view of the 25 mm collimator in the active source region (iii) shows the collimator cone encompasses the entire active source. Note that the apex of the conical collimator is located at different distances from the isocenter for different collimator sizes, further away for the smaller 15 mm collimator. Due to the complex contributing factors of the source shape and self-attenuation, neither the physical SFD nor the apex of the cone can be used as the virtual source location for the inverse-square radiation output calculation.

When the GammaPod is not in use, all sources are aligned to the solid tungsten rods on the collimator, which defines the “beam-off” position. A treatment starts with the opening of

FIG. 2. (a) Shielding around the uppermost source at 18° when the beam is at the “off” position; (b) the cone angles of the collimators are designed to encompass the entire active source inside the cone.

the shielding door on the top. The patient in prone position with her breast immobilized in the vacuum-assisted breast cups placed through an opening in the treatment couch and locked on to the couch is then lowered to align the isocenter with the starting position in the target. The 36 sources are then aligned with a selected size of collimator holes through a relative rotation between the collimator and the source carrier, which is referred to as one of two “beam-on” positions. The source carrier and the collimator carrier then rotate in synchrony forming 36 noncoplanar conical arcs. The rotations of the source carrier and the collimator carrier are independently driven by servo motors. There are redundant rotary sensors to ensure that the source and the collimator are rotating in synchrony during treatment. From the beam-off position (sources are aligned with tungsten blocks), a relative rotation of the collimator carrier and the source carrier by 20° allows the sources to be aligned with the 15 mm collimator holes. Another 20° relative rotation will align the sources with the 25 mm collimator holes. The two sets of collimators can be switched dynamically during treatment delivery. In case of power failure, an uninterrupted power supply (UPS) unit will provide enough power to rotate the sources to their blocked positions. The beams can also be manually “turn-off” with a hand crank if the UPS unit also fails.

II.A.3. Housing and shielding

Outside and concentric to the source carrier is the shielding body, which also serves as the housing of the entire unit. The shielding body is made of cast iron and its thicknesses in all dimensions ensure the radiation level behind the shielding is lower than the regulatory limits (average dose rate $20 \mu\text{Sv/h}$ and maximum dose rate of $100 \mu\text{Sv/h}$ measured at 1 m from the source) for the source housing configuration and the initial source activity of 4320 Ci. As shown in Fig. 1(a), there are two shielding doors on top of the gamma-radiation unit made of iron and lead to prevent leakage radiation entering the treatment space from further exiting to the treatment room when

not treating patients. During normal operation, both shielding doors are driven by servo motors but can also be manually closed by hand and if the power to the door is lost.

II.A.4. Dynamic treatment couch

The patient support system (treatment couch) is capable of dynamic motion in three dimensions during beam delivery. The radiation focal point (shot) can be optimally placed within and around the target dynamically, allowing the shot to continuously paint the dose distribution across the target volume, analogous to three-dimensional printing. The motion system including the control is borrowed from robotics and computer numeric control (CNC) machining, capable of tenth-millimeter accuracy. During the dynamic treatment delivery, the system is designed to make it physically impossible to harm the breast by collision or crushing. In addition to electronic sensors and software limits, every movement that could go out of the allowable range is physically blocked. For example if the shielding door is in the closed position, the couch cannot be lowered from its highest position and further downward movement is physically prevented by metal posts mounted on the shielding door, which is to avoid collision of the patient breast against shielding door surface. Likewise when the couch is in the treatment region, the sides of the couch carrier will physically block the inadvertent closing of the shielding doors. Upon power loss, a battery backup power will drive the collimator to turn off the beam. The operator is also able to manually turn off the beam if needed.

II.B. Breast immobilization and localization

There is no on-board imaging with the GammaPod irradiation unit. The clinical procedure for the GammaPod treatment closely mimics that of stereotactic radiosurgery of intracranial lesions. The patient will be imaged on a CT or MRI at a different location in the hospital wearing the vacuum-assisted breast cup and be transferred to the GammaPod for treatment.

FIG. 3. (a) Vacuum assisted breast cup, showing the outer cup, inner cup, silicone flange, the cup-locking feature, and the localization frame. (b) The stereotactic localization frame detected from a set of CT images (square dots) and the theoretical locations by design.

If multiple fractions are required, each fraction will require the same imaging, planning and treatment procedure. To accurately target the tumor identified on a three-dimensional image set (MRI/CT), two conditions must be met: (1) the breast's shape must be identical during imaging and treatment; and (2) the entire breast must be located in a common coordinate system during imaging and during radiation treatment.

To satisfy the first condition, a patented breast immobilization cup with a two-layer design using negative air pressure [see Fig. 3(a)] is used to immobilize the breast. The breast cup consists of an inner cup, made from polyethylene terephthalate (PET) (same as most clear drinking water bottles), an outer cup made from polycarbonate (same as modern lenses of eye glasses to be shatter-proof), and a silicone flange. The inner layer is perforated and generally conforms to the breast; and the outer layer is rigid and air-tight. They are tightly coupled with the silicone flange, which is taped on to the chest wall of the patient with a thin layer of silicone adhesive coated on the surface of the flange. There are 28 different sizes and shapes of breast cups to accommodate different patients. When a small negative pressure is applied to the space between the inner and the outer cups, the skin of the breast will be pressed against the inner cup, and the friction will prevent the sliding motion between the skin and the inner surface of the inner cup. The vacuum source automatically maintains a preset air pressure in between the inner and outer cups. When the patient is set up in prone position, the breast tissue tends to fall away from the chest wall. With the negative pressure, there will be additional mild pulling of the breast tissue away from the chest wall. To increase the security of the immobilization

and comfort, the base of the cup is also affixed to a wide strap, wrapping around the chest of the patient. To satisfy the second condition, a MRI or CT compatible stereotactic localization frame is affixed on the breast immobilization cup as shown in Fig. 3(a). The localization frame consists of a hollow plastic tube diagonally wrapping around the side of the outer cup. The hollow plastic tube is filled with either copper sulfide solution for MRI or copper wires for CT, and is glued to the machined groove on the outer cup. Making the stereotactic localization frame as part of the outer layer of the breast cup ensures that there is no geometric uncertainty of the location of the wire loop. Figure 3(b) shows an example of the detected frame locations from a set of CT images (square dots) and the actual locations of the wire (solid line).

To ease prone setup, enhance patient comfort and ensure geometric consistency between breast imaging and treatment, two motor-driven prone-setup couches of the same design and surface foam padding that lift and rotate the patient from standing position to prone position are employed for both imaging and treatment [Fig. 1(a)]. By adjusting the step height of the couch, the patient standing on the couch can place her breast immobilized in the vacuum-assisted breast cup into the opening on the couch. After the breast cup is locked on to the couch with a three-point locking feature, the patient will be secured and the couch will lift and rotate the patient to the horizontal prone position for either imaging or treatment. Because the breast immobilization cup is locked onto the couch, the cup-based coordinate system has a fixed relationship with the couch. Therefore, any region in the breast can be placed at the radiation focal spot by the movement of the couch. We enrolled 15 patients with metal

clips in their breasts in a IRB approved reproducibility study using the immobilization cups and the localization method. The coordinate mismatch between two CT scans separated by 20–30 min was 1.83 mm with 0.51 mm standard deviation.²⁶ Details of the reproducibility study will be covered in a separate publication.

II.C. Treatment planning system

A dedicated TPS has been developed for planning the GammaPod breast radiotherapy. Its basic functions include the capability of importing and processing imaging data, automatically registering stereotactic coordinate systems, automatically detecting the breast/body surfaces, allowing the user to delineate the gross tumor and other volumes of interest including normal structures; and allowing the clinicians to specify the doses to the different targets and dose limits to normal structures.

After the breast CT/MRI images are imported, the semi-automatic image registration function identifies points of the stereotactic frame, performs refinements using regression to match the known geometry of the frame, and places the breast image into this stereotactic coordinate system. After stereotactic frame registration, the planner is required to delineate the gross tumor volume (GTV) and its subclinical extensions. Multiple targets within a breast are allowed. When automatic expansions are used for obtaining clinical target volumes (CTVs) from an inner target, the contours are automatically truncated to maintain a set distance from the skin surface. After target delineation, the operator is required to prescribe doses to the targets. Different doses can be prescribed to different targets. The planner can also delineate organs-at-risk (OARs), such as skin, lung, heart, and ribs. Simple dose limit can be placed for each of the OARs.

Based on the user-defined targets and dose prescriptions, the system optimizes the dynamic paths of the two focal spots using an inverse planning algorithm. The user can also place dose limitations to nontarget structures. Once the dose prescription is provided, the minimum treatment times is set by the ratio of the target integral dose and the integral dose rate of the 25 mm focal spot, and the maximum treatment time is set by the ratio of the target integral dose and the integral dose rate of the 15 mm focal spots. i.e.,

$$T_{min} = \frac{\int D(r)dr, r \in \{target\}}{\int \dot{D}(r)dr, r \in \{25 \text{ mm Kernel}\}} \quad (2)$$

and

$$T_{max} = \frac{\int D(r)dr, r \in \{target\}}{\int \dot{D}(r)dr, r \in \{15 \text{ mm Kernel}\}}, \quad (3)$$

where $D(r)$ is the target prescription dose, $\dot{D}(r)$ is the kernel dose rate, and the 3D integrals are carried out through the volumes within the target (the numerator) and the kernels (the denominator). Equations (2) and (3) also provide a method for hand calculation. Given the total time durations of each collimator is used for the treatment from the treatment plans and

the target volume, the dose delivered to the target is

$$D_{target} \cong \frac{\dot{D}(t_{25}V_{25} + t_{15}V_{15}f)}{V_{target}}, \quad (4)$$

where $\dot{D}$ is the calibrated dose rate using the 25 mm collimator, V_{25} and V_{15} are the focal spot volumes of the two collimators, t_{15} and t_{25} are the planned beam-on times using the two sizes of collimators, and f is the relative output factor of the 15 mm shot.

The operator has the opportunity to set the desirable delivery time between the two extremes. The optimized dynamic paths of the focal spots are each defined as a series of traveling segments, with each segment having a different traveling length and velocity. During delivery, the switching between the two collimators is achieved dynamically but not frequently. A typical treatment plan uses both collimator sizes and involves two dynamic paths, one for each collimator size. The construction of the path employs the “traveling salesman” algorithm.²⁴ The treatment delivery is referred to as “dynamic dose painting.” As compared with “sphere packing” techniques commonly used in radiosurgery, dynamic dose painting creates much more uniform dose distribution in the target. It also allows the delivery of different dose levels to different targets.

After the optimization is finished, the system provides the user with visual display of the dose distributions in both 2D and 3D renderings and calculates dose volume matrix. Finally, it formats the treatment plan as series of control points and transfers the plan to the control console of GammaPod radiation unit for treatment delivery. Any two successive control points define a dynamic delivery segment.

Dose calculation is based on precomputed 3D dose distributions, commonly referred to as “kernels,” in a homogeneous breast phantom using Monte Carlo simulation. Dose distributions at different locations in the same breast and in different sizes of breasts are different. Due to the negative pressure applied between the inner and outer cups, the breast tissue fills the inner cup when the correct size of cup is used. This fact allows the precalculation of dose distributions at all locations in all 28 sizes of breast cups. Kernels on a rectilinear grid of locations with 5 mm grid size in all 28 breast cups of different sizes were precalculated and stored. Each breast cup is assumed to be filled with homogeneous medium with a density of 0.935 gm/cm³, and atomic composition of water. The kernel of each grid point is computed by translating the breast cup to place the grid point of interest at the isocenter of the GammaPod. Due the rotational symmetry of the breast cups, kernels are computed only on half of the transverse plane of the breast cup through its central axis. Dose distribution for a treatment plan is simply the superposition of kernels with different weights (dwelling time) at different locations interpolated between the control points. Therefore, for planning using either CT or MRI images, the same homogeneous breast density is assumed.

The GammaPod Cobalt-60 source and collimation details are modeled using BEAMnrc, an EGS4nrc user code dedicated for radiation source simulation,²⁵ with “bremstrahlung splitting” set to uniform and “splitting factor” of 20.

Gamma-rays of two energy levels, 1.17 MeV and 1.33 MeV, emit from the sources with same probability. The BEAMnrc code uses a series of component modules to model the exact shape of the source and collimator geometry and outputs phase-space data, which include all the particle information (i.e., the charge, position, direction, energy, and history tag for each particle). 360 static beams, with a longitudinal interval of 1° are used to simulate each of the 36 conical arcs. The resulting phase-space data are used as the input for the kernel calculation using the DOSXYZnrc code²⁵ with the cut-off energies of 700 keV for electrons (ECUT) and 10 keV for photons (PCUT).

A prototype GammaPod system was installed at University of Maryland Medical Center and has gone through a thorough functional, geometric, and dosimetric testing. The results of the initial dosimetric testing are detailed in the second paper of the series. Linear potentiometers with 0.1 mm resolution were used for all linear motion measurements. Because beam “on” and “off” and collimator size selection are achieved with relative rotations between the source carrier and the collimator, the source/collimator (S/C) rotation accuracy were measured indirectly by measuring the cord lengths of different rotating angles using a digital caliper. Since both the source and the collimator rotate together to create the arcs, their synchrony is crucial for dosimetric accuracy. The S/C synchrony was assessed by digitally recording the dynamic linear displacements between the two.

III. RESULTS

In the following, we only summarize the mechanical and functional testing results.

III.A. Functional verification of radiation unit

III.A.1. Mechanical accuracy of moving axes

Accuracy/precision and linearity of three axes of treatment couch were tested to verify the mechanical performance of the stereotactic couch system. The location of the couch was measured at different positions from both directions, for each of the three axes. For X- and Z-axes, the couch can move within a circle of 75 mm radius. For vertical (Y-axis) movement, the couch travels 310 mm downwards from its initial raised position before treatment to the lowest position. The accuracy and precision for all three motion axes over their ranges of travel are all better than 0.2 mm. The nonlinearity for each axis is tested over its full range of travel and is smaller than 0.2%.

The rotation accuracy and precision for both source and collimator are better than $\pm 0.02^\circ$. The S/C rotation synchrony during treatment is also tested and it is within 0.01° at all times.

III.A.2. Radiation shielding verification

Radiation survey measurements were taken at different surface points around the GammaPod with the source closed and shielding doors closed. The survey measurements are

taken using a calibrated pressurized ion chamber (Fluke Corporation, Model 451P) and averaged over three measurement points. With a source activity of approximately 4320 Ci, the highest dose rate on the unit surface is $109 \mu\text{Sv/h}$, recorded directly above the shallowest source (18° latitudinal angle) on the top surface of the shielding doors. The mean surface dose rate averaged for all the measurement points is less than $10 \mu\text{Sv/h}$. At 1 m from the GammaPod surface with the source closed and shielding door closed, $1.68 \mu\text{Sv/h}$ was measured at the lower back of the GammaPod. The mean exposure rate at 1 m averaged over all positions was $<1 \mu\text{Sv/h}$.

III.B. Stereotactic coordinate registration of TPS

Figure 3(b) shows the display after stereotactic frame registration, providing the planner a visual confirmation of the matching between the detected and the computer modeled frame locations. The dots representing semiautomatically detected frame pixels and continuous curve is the ideal location of the frame. After coordinate registration, the mean error and standard deviation of the distances between detected frame pixels and ideal curve are 0.20 mm and 0.15 mm, respectively.

III.C. Single shot dose distribution by Monte Carlo simulation

For a static 25 mm shot placed at the center of a breast phantom, the MC simulated dose distributions on the patient's axial and coronal planes through the isocenter are shown on the left and right in the top row of Fig. 4, respectively. The profiles in the patient PA and LR directions are shown on the left and right in the bottom row of Fig. 4, respectively.

As seen in Fig. 4, the single shot dose distribution in the vertical plane (patient's sagittal or axial planes) is butterfly shaped with lower side (toward the nipple) being hotter. This asymmetry is due to the fact that the lower side is closer to the sources and the beams are less attenuated by the breast tissue. For a single beam near the isocenter, the proximity attributes approximately 5%/cm and the attenuation attributes to about 3%/cm. When 36 conical arcs overlap from different angles, the geometric focusing plays a dominant role. From the isocenter toward the source, we see a fast dose fall-off instead of dose increase. In the opposite direction, the dose fall-off is also much faster than 8%/cm predicted for a single beam. The power of geometric focusing is the fundamental principle of SRS and SBRT. The dose distribution in the horizontal plane (patient coronal plane) is cylindrically symmetric, which is as expected because the beams rotate around the vertical axis through the isocenter.

III.D. Treatment plan results

GammaPod treatment plans are generated for three artificial clinical cases using patient images acquired in our earlier IRB approved reproducibility study with the vacuum-assisted breast cups locked on to the imaging couch in prone positions. Planning typically takes 15 min, with about 10 min spent on target delineation. Plan optimization typically takes less than

FIG. 4. Dose distribution of a single 25 mm shot around the isocenter. Upper panel shows the isodose distributions on the vertical and horizontal planes (patient’s axial and coronal planes). Lower panel depicts the vertical (patient’s PA direction) and lateral (patient’s LR direction) dose profiles.

2 min because doses to all positions are precalculated. In the first case, a relatively small tumor with 2.4 cm³ GTV is contoured near the center of the breast using a real patient CT image set. A uniform 16 Gy is prescribed to the target (CTV 1 = GTV). The dosimetric results generated by the inverse planning of GammaPod TPS are as following: with 95% of the target receiving prescription dose, the maximum dose in the target is 16.8 Gy, or 105% (of prescription); the maximum skin dose is 1.5 Gy, or 9% of prescription dose. The plan delivery parameters are as following: the treatment couch moving trajectory contains 66 control points; and it takes around

4 min to deliver the treatment assuming initial source strength of 4320 Ci. The 2D dose distributions in patient’s axial and coronal planes are shown in Fig. 5(a) and the DVHs for the target and the normal breast tissues are shown in Fig. 5(b). In the second case, a relatively large target measured 6.5 cm by 7.6 cm with 150.7 cm³ target volume is contoured on a set of CT images. A uniform 8 Gy is prescribed to the target. With 95% of the target receiving prescription dose, the maximum dose in the target is 9.94 Gy, or 124%; the maximum skin dose is 2.34 Gy, or 29%; and the 1.6 Gy or 20% isodose is contained inside the breast. The volume of normal

FIG. 5. (a) Planned dose distributions on axial and coronal planes of a GammaPod treatment delivering 16 Gy to a 2.4 cm³ target. (b) DVHs of the target and normal breast tissues.

FIG. 6. (a) Planned dose distributions on axial and coronal planes of a GammaPod APBI treatment delivering 8 Gy to a 151 cm³ target. (b) DVHs of the target and normal breast tissues.

breast tissue receiving greater than 4 Gy (V_{50}) is only 18%. The treatment contains 196 control points which takes around 19.5 min to deliver assuming initial source strength of 4320 Ci. The dose distributions in patient's axial and coronal planes are shown in Fig. 6(a) and the DVHs for the target and normal breast tissues are shown in Fig. 6(b).

In the third case, a 2.4 cm³ gross tumor is contoured and prescribed to receive 18 Gy and its 15 mm surrounding region, a hollow shell of 54.5 cm³ volume (tumor bed), is prescribed to receive 9 Gy to demonstrate the ability of the GammaPod of delivering ablative doses to the gross tumor and sterilizing doses to the tumor bed simultaneously. With 95% of the gross tumor receiving prescription dose of 18 Gy, 98% of the tumor bed is covered by 9 Gy; the maximum dose is 18.7 Gy, or 104%; the maximum skin dose is 2.1 Gy, or 11%; and the 1.8 Gy or 10% isodose is contained inside the breast. The treatment contains 99 control points which takes around 11.3 min to deliver assuming initial source strength of 4320 Ci. The dose distributions in patient's axial and coronal planes are shown in Fig. 7(a) and the DVHs for GTV, tumor bed, and normal breast tissues are shown in Fig. 7(b).

IV. DISCUSSIONS

The GammaPod breast radiotherapy or radiosurgery will be under MRI or CT image guidance as with stereotactic intracranial radiosurgery. The proposed vacuum-assisted

breast cup for stereotactic localization can provide millimeter accuracy.²⁶ Combined with MRI and/or CT to assess tumor extent, GammaPod has the potential to accurately deliver high radiation dose to the gross tumor and its micro extensions. Breathing motion should have little effect on the geometrical accuracy because the patient is treated in prone position¹⁰ and the skin of the breast is immobilized by the vacuum-assisted breast cup.

The detailed design of the GammaPod achieves a balance among conflicting requirements. In determining the range of latitudinal beam angles, the isocenter is limited to 10 mm from the top surface of the couch to allow the treatable region reaching the chest wall. Moving the isocenter further up to increase treatable region with larger beam angles would increase dose to the patient's thorax from the exit beams. To maximize the dose rate at the isocenter with fixed amount of total radioactivity, the source to isocenter/focus distance (SFD) should be as small as possible. However, the SFD, which equals to the summation of the length of the tungsten collimator and the radius of the collimator carrier, cannot be too short as the collimator needs to be thick enough to meet radiation shielding requirement and the radius of the collimator carrier needs to be large enough to accommodate large breasts.

The dose rate aimed to achieve and the number of individual sources again takes into account conflicting factors. On one hand, the manufacturing cost, source loading, and

FIG. 7. (a) Planned dose distributions on axial and coronal planes of a GammaPod treatment simultaneously delivering 18 Gy to a 2.4 cm³ GTV and 9 Gy to the tumor bed, a 3D expansion from the GTV by 15 mm. (b) DVHs for GTV, tumor bed, and normal breast tissues.

servicing processes are proportionally burdened by the number of individual sources used. However, reducing the number of sources would either lower the achievable dose rate or force the amount of activity packed in each source to be increased, which not only stresses the overall shielding requirement, but also results in physically larger active source volume, hence resulting in bigger geometric penumbra of the focal spot dose distribution, which is undesirable. Further increase of the number of sources to achieve a higher dose rate would require an increase in the diameter of the collimator carrier to gain the space needed for the additional collimators, which would push the SFD longer.

Unlike other radiosurgery devices that use the method of packing multiple static shots, also known as sphere packing, GammaPod employs dynamic dose painting with simultaneous irradiation and movement of the patient. As shown by the DVHs for the three planning examples, such dynamic delivery creates highly uniform dose distributions in the target. It also allows the delivery of different dose levels to different targets. The first two cases of the planning exercise illustrate that the GammaPod could be used for hypofractionated APBI, even with larger surgical cavities. There is evidence that APBI with EBRT is associated with adverse cosmetic outcome due to excess dose to the normal breast.^{27,28} It has been shown that V_{50} greater than 40% is associated with higher incidences of subcutaneous fibrosis and adverse cosmesis.^{27,28} With GammaPod, the use of the vacuum-assisted breast cup and stereotactic localization should improve setup accuracy and afford a reduction in the margins associated with setup uncertainty. The fast dose fall off beyond the target volume makes V_{50} substantially lower than what we commonly see with APBI using EBRT. For the large target size in the second case, the V_{50} is only 18%. The treatment time for the second case at the initial radioactivity is 19.5 min. For such large targets, conventional external beam would no longer be able to meet the dosimetric criteria for APBI. Using the GammaPod, excellent dose distributions are still achievable but the treatment time becomes much longer. The third case of the planning example demonstrated that it is dosimetrically possible to deliver 18 Gy uniform dose to the gross tumor while covering the surrounding region of 15 mm expansion with a lower dose of 9 Gy. Because the treatment procedure is noninvasive, such treatment can also be repeated in 2–3 fractions.

It is important to note that the dosimetric feasibility of GammaPod in delivering the SBRT and SRS dose distributions is different from its clinical feasibility, which has not been demonstrated. Clinical trials have been planned. The initial clinical results will take years to emerge. Nevertheless, the GammaPod system offers an opportunity to assess the efficacy of SBRT and SRS for breast cancer. If successful, it could bring positive changes in the management of early-stage breast cancer.

The current design of the GammaPod system does not include on-board imaging. An on-board imaging device may be developed as part of the GammaPod system to allow more logistically convenient workflow. The stereotactic frame and the process of transferring patient from imaging room to the treatment room could be eliminated, and the patient can be

imaged, planned, and treated while lying prone on the same treatment couch.

V. CONCLUSION

Advances in imaging and breast cancer awareness have allowed more breast cancers being diagnosed in the early stage and the importance of image guidance in cancer intervention is widely recognized. The GammaPod, an image-guided breast stereotactic radiotherapy device has been developed to take advantage of these trends. With the highly focused dose delivery and accurate stereotactic localization mechanism, it is envisioned that the GammaPod technology has the potential to significantly shorten radiation treatments and even eliminate surgery by ablating the tumor and sterilizing the tumor bed simultaneously.

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