Image-Guided Robotic Radiosurgery

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Abstract: We describe a radiosurgical system based on a novel medical linear accelerator of particularly light weight, a six degree-of-freedom robotic arm and an online x-ray vision system. The system introduces a class of radiosurgical procedures, termed *non-stereotactic*, or *image-guided* radiosurgery. A treatment planning subsystem based on new paradigms for inverse planning was developed in an interdisciplinary project at Stanford University in conjunction with Accuray Inc., Santa Clara. Two patients have been treated with a fully integrated system prototype at Stanford University Medical Center. The new system has the potential for revolutionary changes in radiosurgery by allowing for high accuracy targeting, fractionated treatment and highly conformal radiosurgical treatment of non-spherical lesions. Furthermore, it is expected that image-guided procedures will replace radiotherapeutical methods for many extra-cranial lesions.

1 Introduction

Over the past 20 years there has been a revolution within medical imaging. To date these changes have had their greatest effect on diagnosis. However, in the second phase, new methods in medical imaging are having a growing impact on treatment. Since there is now seldom a need for surgeons to visually inspect an area of pathology, their efforts have increasingly been directed towards planning and performing the least invasive procedure. Imaging is being used to guide such operations. One of the least invasive procedures of all is called stereotactic radiosurgery.

Stereotactic radiosurgery is a surgical technique which destroys small brain tumors and other lesions through the closed skull with a single large, necrosing dose of high energy radiation [2, 3, 5]. The current procedure for stereotactic radiosurgery begins by screwing a frame into the patient's head (figure 1). The frame provides reference to localize the tumor from either CAT or MRI scans (accurate to 1 mm). In addition, the frame holds the patient's head rigidly during treatment. Using high-speed workstations, imaging studies are processed thereby permitting a tumor's spatial coordinates and a radiation treatment plan to be calculated. With the frame holding the patient's head securely, a high-energy linear accelerator destroys the tumor by cross-firing from a large number of directions. The actual radiation time for typical radiosurgical treatment is 30 minutes, but the frame must remain in place for 5-7 hours. This is necessary because the frame must must be visible on the CAT/MRI images. After treatment the frame is removed, and in most cases the patient can be released from the hospital the same day.

Although the treatment is associated with few risks and is highly effective, there are limitations to what can be accomplished. For example, when treating tumors larger than 3.5 cm in diameter, the complication rate increases. Ideally, such larger tumors should be treated over 2 or 3 successive days. However, the frame which is screwed to the patient's head is painful, and it is difficult to keep it in place for that length of time. Pain is also the reason that children who undergo radiosurgery must be given a general anesthesia throughout the 5 to 7 hours it takes to do the procedure. Lastly, because current methods of targeting require a stereotactic frame, it has not been possible to treat tumors outside the head with radiosurgical methods; this is especially unfortunate since the number of patients afflicted with such tumors is much greater than that with brain lesions.

In response to the limitations of present day radiosurgical instrumentation, we have developed a novel robotic system for stereotactic radiosurgery. This technology, termed image-guided radiosurgery targets a tumor precisely without a frame attached to the patient. Instead, a new system of localization has been developed which uses digital x-ray images "correlated" via computer with the 3D anatomy of the skull determined by a previous tomographic scan. Furthermore, a new radiation source for radiosurgery is being used. This radiation source is a 6 MV linear accelerator with particularly light weight. Due to the low weight of the linear accelerator, it becomes possible to move the radiation beam with a 6-dof robot. This device not only addresses the shortcomings of present radiosurgical systems, but with its robotic targeting, introduces a degree of flexibility that promises to revolutionize radiosurgery.

To date, two patients have been treated with a fully integrated prototype installed at Stanford University Medical Center.

2 Frameless Localization

The system utilizes the bony skeleton as a reference system whose spatial relationship is fixed with respect to the radiosurgical target. The unique structure of the skeleton is used to automatically determine the orientation of the skull during treatment. This process is rapid enough to permit real-time tracking of the head during radiosurgery, obviating the need for rigid head fixation with a stereotactic frame.

The vision system matches a 3D model of the skull with the location of the skull in "treatment space." The 3D model of the skull is constructed from the CT data and provides information on the position of the target with respect to the skull. The location of the skull in treatment space is determined from two virtual real-time X-ray images taken along linearly independent (nearly orthogonal) viewing directions. For each X-ray image, source position, as well as image plane orientation and position in 3D space are fixed with respect to the isocenter of the radiosurgical system. Prior to treatment radiographs are synthesized from the CT by integrating through the imaged volume along directions which approximate the viewing directions of the X-ray cameras. Four hundred such radiographs are constructed, each of which represents a slightly different viewing direction and displacement.

During treatment, the real-time X-ray images are correlated with the pre-computed radiographs. Error signals are computed from these correlations which yield lesion position with submillimeter accuracy. These error signals are used to position the linear accelerator. Instead of rigidly immobilizing the patient, the vision system tracks at short intervals the movements made by the awake yet cooperative patient. Assuming that head movement is restrained to a cubic volume of ± 1 cm (through the use of molded head support and Velcro straps), the system is capable of

Figure 1: For conventional radiosurgery, the patient's head must be immobilized with a frame via holes drilled directly into the scull. Image-guided radiosurgery obviates the need for rigid head fixation.

dynamic (real-time) guidance. Because many other regions of the body remain fixed relative to the skeleton, this approach should be applicable to a variety of extracranial tumors.

3 High Energy Radiation Source

The radiation source is a novel low weight 6 MV linear accelerator producing a photon beam. Because this device runs at a much higher ("X-band") frequency, it has been possible to reduce significantly the size and weight of critical components. While conventional "S-band" linear accelerators used in medicine weigh 3-4,000 pounds, the new accelerator weighs only 280 pounds. Despite the small size of this X-band radiation source, its performance is comparable to conventional S-band machines. For small fields (≤ 7 cm) the radiation output is 300 rad/min, similar to conventional linear accelerators. Unlike current radiosurgical systems, the capability of being turned on and off instantly allows the system to avoid particularly radiation sensitive healthy structures such as the optic nerves.

The system currently installed uses a set of cylindrical lead *collimators* for focusing the beam. The beam is cylindrical, with fixed diameter. The beam diameter can be set before treatment between 5 and 60 mm by inserting a collimator into the beam path.

4 Robotic Gantry

The system uses a 6-dof GMF robotic manipulator arm to position and orient the radiation beam. The manipulator is capable of positioning the linear accelerator at any point over the full treatment sphere at a distance of 60-100 cm with an absolute precision of ± 0.5 mm in position and ± 0.0001 radians in orientation. Calibration methods described below allow for further improvements of this accuracy. The system has the potential to outperform conventional radiosurgery with respect to accuracy, which is limited by the mechanical flex inherent to all stereotactic frames. Targeting inaccuracies of up to 4 mm have been reported for standard radiosurgical equipment [1]. This is particularly problematic for small lesions with typical diameter of 10 mm or less. It is likely that such inaccuracies are a major source of complications. For research purposes, such inaccuracies have limited the value of clinical studies comparing different radiosurgical treatment procedures.

The path of the radiation source can be altered in response to small patient movements. After patient movement is recorded by the online imaging system, a change in isocenter is calculated and the new coordinates are sent to the manipulator. This feedback is used to dynamically adjust the robot's reference coordinate system. This allows the manipulator to track the motion of a lesion without having to alter pre-determined treatment plans. In the prototype installed, the manipulator travels at a speed of approximately 1.5 cm/sec and head movement updates are provided at 1 second intervals.

5 Treatment Planning

A beam placement together with a pointing direction is called a *beam configuration*. During treatment the beam is activated at a series of configurations. The duration of beam activation at such a configuration determines an intensity or *weight* of the configuration.

Forward treatment planning computes and visualizes a dose distribution given beam configurations and weights. *Inverse treatment planning* aims at finding appropriate beam configurations and weights (i.e. an appropriate *treatment plan*) to treat a given condition. By region of high dose (RHD) we denote the region absorbing 80% or more or the maximum radiation dose. An appropriate dose distribution for a given tumor should satisfy the following constraints:

1. Critical and/or radiation-sensitive healthy structures near the tumor (i.e. nerves, blood vessels, sensitive brain structures) should receive very low dose.
2. For certain cases, the dose inside the tumor should be highly uniform, i.e. the dose in the tumor should not only be lower-bounded but also upper-bounded. Inhomogeneities, even inside the tumor can cause severe complications.
3. The RHD should cover the entire tumor and should match the tumor shape.
4. The drop-off around the RHD should be as sharp as possible, i.e. the (generalized) distance between the 80% isodosic surface and isodosic surfaces of lower values should be small.

Several paradigms for evaluating radiosurgical and radiotherapeutical dose distributions are in clinical use. The main paradigms are *dose-volume histograms* [3, 4, 5] and color representations of distributions in 2D overlaid upon tomographic images. A distribution can be evaluated with respect to constraints 1, 2, and 4 by dose-volume histograms. In addition to dose-volume histograms and 2D color representations, the system uses 3D isodosic surfaces overlaid on reconstructions of anatomic regions, tissue-volume ratios [4], and integrals over dose-volume histograms for distribution evaluation. These paradigms refer to dose distributions computed in simulation. In addition, calibrated photographic film phantoms (CFP) are used for the new system. Means for electronic processing of such phantoms have been developed. Thus dose-volume histograms can be computed directly from actual radiation exposure and not from simulation (see section 6).

For current radiosurgical systems a treatment plan is determined manually by the surgeon for each individual patient. In many cases, appropriate dose distributions require a large set of beam directions and/or complex motions of the radiation source.

Only simple motions of the radiation source can be executed with conventional radiosurgical systems. To achieve additional flexibility, conventional procedures manually move the patient couch during treatment in addition to moving the radiation source. The accuracy of such procedures is limited. It is generally not

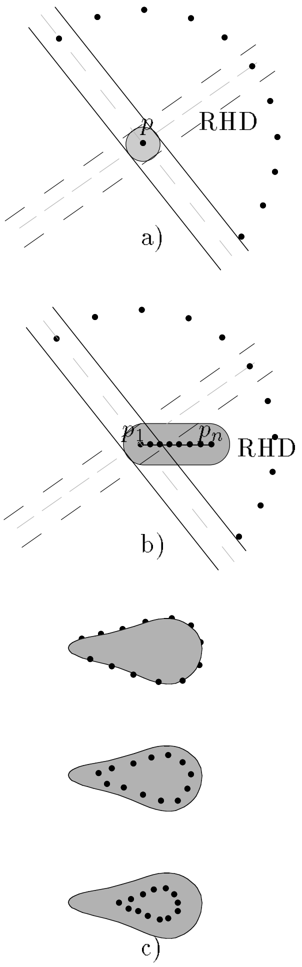

Figure 2: Beam selection paradigms

practical to generate complex beam paths with conventional radiosurgical equipment.

The new system has six degrees of freedom, and thus offers full kinematic flexibility for placing and directing the radiation beam in three dimensions. To make full use of this added flexibility, treatment planning methods based on geometric algorithms have been developed in an interdisciplinary project.

The following planning approach, placing emphasis on inverse planning is used. Treatment planning is divided into three subtasks.

1. Selection of beam configurations
2. Motion planning (computation of a collision-free path connecting beam configurations)
3. Computation of beam weights.

Each subtask is regarded as entirely *geometric* planning task. Thus the constraints for appropriate dose distributions are expressed by geometric requirements. Steps 1 and 3 are based on a geometric model of the physical characteristics of the photon beams produced by the accelerator subsystem. Furthermore, each of the three subtasks is considered separately in the planning process. Thus the path connecting the beam configurations does not influence the distribution. However, the inverse planning component must ensure that a path is executable and does not cause collisions or occlusion of the on-line vision system.

Fully computerized optimization of distributions is problematic and is not used for the following reasons:

- The constraints describing an adequate dose distribution are conflicting. As an example, in many cases it is difficult to achieve accurate shape matching (of tumor and region of high dose) *and* sharp dose drop-off around the RHD.

- The described planning problem is non-linear. In general, optimization methods will only yield a local optimum, with no information about global optima.

- The results of an optimization depend on settings of internal system parameters. Small changes in these values can yield largely diverging results. It is difficult to make the influence of such parameter settings and trade-offs transparent to the system user.

For similar reasons we do not include radiobiological considerations or complication probabilities into algorithmic planning.

Complete methods are used for steps 2 and 3. Thus the correctness of output results given current user input settings can be guaranteed based on algorithmic principles. Furthermore bounds for the required computing time can be given in advance. For step 1 complete algorithmic solutions are problematic for combinatorical reasons. Therefore an extended set of standard paradigms for selecting beam configurations has been developed. These paradigms are direct extensions of paradigms in clinical use with earlier systems. The extensions are based on the fact that the beam can be rotated *and* translated along arbitrary paths in space.

To plan a treatment, the user outlines the tumor and anatomical structures close to the tumor on the console computer screen. Figure 2 illustrates paradigms for beam selection. The beam used in the current system is cylindrical, i.e. the beam cross-section is circular. For sphere-shaped tumors, a spherical RHD is most adequate. Such an RHD is generated by a series of beams in which the axes of all beams cross a single point p (figure 2-a). p is called

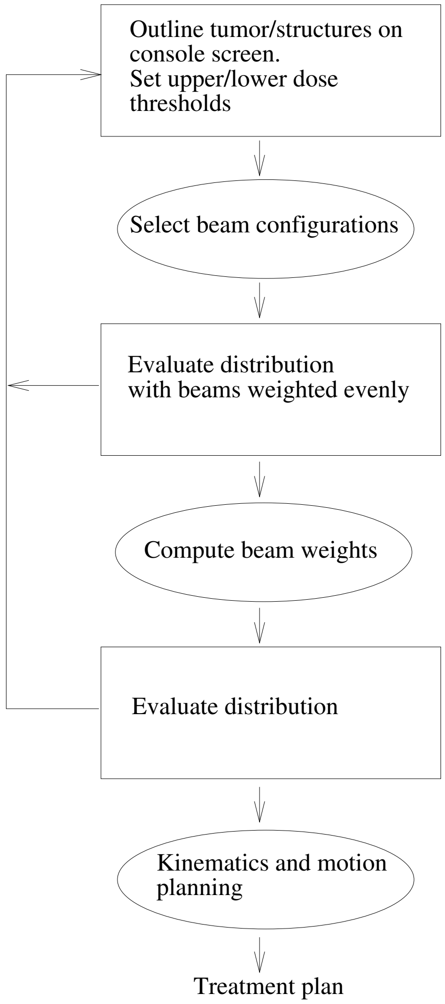

Figure 3: Planning steps

isocenter. The RHD is spherical, if all beams have even weights. This procedure is called single-isocenter procedure. A first generalization of this procedure is shown in figure 2-b. A series of points $p_1, \ldots, p_n$ is placed evenly spaced along a line segment. As in the single-isocenter procedure, a set of beam directions is computed for each point. We thus call the points $p_1, \ldots, p_n$ *isocenter points*. The resulting RHD is a generalized cylinder (Figure 2-b). Spheres and generalized cylinders are adequate for many tumors. Case c) shows beam configurations for treating more general shapes. In this case the system computes a representation of the tumor surface. A grid of isocenter points on the tumor surface is then computed. The grid can be retracted or expanded interactively. Each isocenter point is resolved into a series of beam configurations as in the single-isocenter procedure. Note that automatic beam weighting methods are necessary to homogenize the RHD in this case.

Figure 4: Robotic radiosurgical system with set-up for determining beam targeting accuracy. Linear accelerator mounted to 6 dof robot arm.

In the weighting phase, the user can set upper and lower dose thresholds for the tumor and healthy structures in the anatomy. Based on geometric algorithms, it can be decided whether the distribution thus specified can be achieved [6]. If the distribution can be achieved, appropriate beam weights are computed, and the distribution is displayed and evaluated. The beam configurations are linked to an executable path for the gantry (see section 6). Figure 3 illustrates interactive planning.

Note that the system kinematics and planning paradigms allow for all treatment procedures used by classical radiosurgical systems. Thus distribution characteristics achievable with earlier systems are also achievable with the new system. Results of a test series are reported in [6].

6 Safety and Quality Assurance

The distance between the radiation source and the target is large when compared to the target diameter. Therefore we can use standardized *template paths* for moving the radiation source during the execution of treatment plans. A template path consists of a sequence of pre-computed and hard-coded beam configurations, called *nodes*. An executable treatment plan consists of a series of beam configurations together with assigned weights, and a path connecting the configurations. Each treatment beam configuration is equal to a node in terms of *beam position*, i.e. position of the robotic wrist tool center point. However, treatment beams can differ from nodes in terms of *beam pointing directions*, i.e. orientation of the robotic wrist.

Due to the distance/size relationships we can bound the angular deviation between treatment beams and nodes. Thus an executable treatment path is obtained by adapting a template path. Only beam pointing directions, but not beam placements are adapted.

The following set-up was used to determine the accuracy of beam targeting. An optical crystal of diameter 1mm was placed at the work-space origin (Figure 4). A thin laser beam pointing along the radiation beam axis was placed into the accelerator subsystem. The optical crystal is activated by the laser beam. Measures of activation can be read from the crystal. A template path consisting of a series of activation points was executed with the crystal and laser beam in place. The deviations in pointing accuracy were measured by reading the activation degree of the crystal at each configuration.

The above set-up is also used to *calibrate* template paths. Calibration compensates for repeating inaccuracies at each node. For each node an offset (difference between commanded laser beam position and actual laser beam position) is determined. Treatment beam configurations are linked to a node. As noted above, the treatment beam position is the same as the node position. The treatment beam pointing angles are very close to the node pointing angles, and inaccuarcies stem in large part from gross-motion repeating inaccuracies. Offsets computed for nodes are thus used for treatment beam configurations as well. An inaccuracy range of length 0.3 mm was determined experimentally with this calibration scheme.

The overall system accuracy was determined with a numercially controlled machining table (NC-table). The following set-up is used (figure 5). A photographic phantom consisting of film slices is placed inside a plexiglass reconstruction of a human skull. The skull reconstruction is mounted to an NC-table and filled with water. The NC-table can be moved under computer control, so that the actual path travelled by the phantom is known. A planned treatment is then executed. The photographic film phantom is then *electronically* processed, i.e. the film slices are scanned, and dose distributions computed in simulation are compared to the actual radiation distribution. Test cases considered with this set-up were actual cases treated with classical radiosurgical systems. A very close correspondence between the phantoms and dose distributions obtained by forward planning was established. An evaluation based on dose volume-histograms computed directly from film phantoms for 10 cases to be treated with the new system is in preparation.

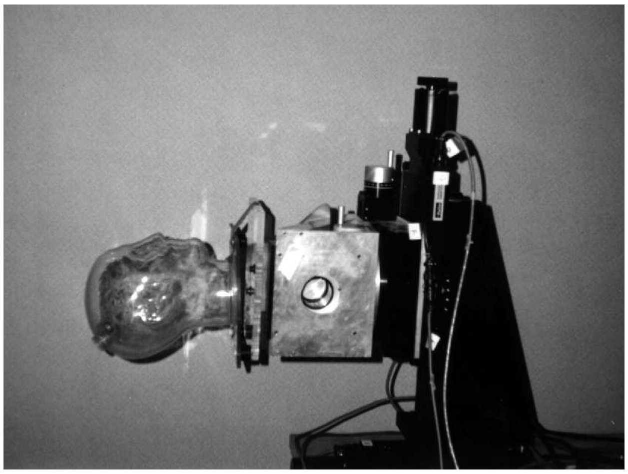

Figure 5: NC-table set-up with photographic film phantom

7 Conclusion

Two patients have been treated with a fully integrated prototype at Stanford University Medical Center. The six degree-of-freedom manipulator in conjunction with a unique compact X-band linear accelerator provide a system with the capability of irradiating homogeneously tumors of any shape and delivering fractionated treatment.

To fully utilize the system's capabilities, a new treatment planning method has been developed. Planning is divided into three subtasks. Each subtask is regarded as entirely geometric planning task. This approach allows for a particularly uniform and transparent planning method. To verify results, photographic phantom studies can be processed electronically.

Improved patient comfort, decreased treatment time and the potential to treat extracranial targets are further benefits of this approach to radiosurgery.

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