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Development of an Automated Steering Mechanism for Bladder Urothelium Surveillance

W. Jong Yoon,

Department of Mechanical Engineering, University of Washington, Seattle, WA 98195,
wjyoon@u.washington.edu

Sangtae Park,

Section of Urology, Department of Surgery, Pritzker School of Medicine, University of Chicago,
Chicago, IL 60637

Per G. Reinhall, and

Department of Mechanical Engineering, University of Washington, Seattle, WA 98195

Eric J. Seibel

Department of Mechanical Engineering, University of Washington, Seattle, WA 98195

Abstract

Given the advantages of cystoscopic exams compared with other procedures available for bladder surveillance, it would be beneficial to develop an improved automated cystoscope. We develop and propose an active programmable remote steering mechanism and an efficient motion sequence for bladder cancer detection and postoperative surveillance. The continuous and optimal path of the imaging probe can enable a medical practitioner to readily ensure that images are produced for the entire surface of the bladder in a controlled and uniform manner. Shape memory alloy (SMA) based segmented actuators disposed adjacent to the distal end of the imaging probe are selectively activated to bend the shaft to assist in positioning and orienting the imaging probe at a plurality of points selected to image all the interior of the distended bladder volume. The bending arc, insertion depth, and rotational position of the imaging probe are automatically controlled based on patient-specific data. The initial prototype is tested on a 3D plastic phantom bladder, which is used as a proof-of-concept in vitro model and an electromagnetic motion tracker. The 3D tracked tip trajectory results ensure that the motion sequencing program and the steering mechanism efficiently move the image probe to scan the entire inner tissue layer of the bladder. The compared experimental results shows 5.1% tip positioning error to the designed trajectory given by the simulation tool. The authors believe that further development of this concept will help guarantee that a tumor or other characteristic of the bladder surface is not overlooked during the automated cystoscopic procedure due to a failure to image it.

Keywords

active; programmable remote steering; optimal path; bladder surveillance; shape memory alloy; segmented actuators

1 Introduction

1.1 Background

Bladder cancer is the fourth most prevalent cancer for males in 2007. The estimated number of new urinary bladder cancer cases and the estimated deaths for both sexes in the United States population during 2007 are 67,160 and 13,750, respectively [1]. Bladder cancer has also been reported as the most expensive cancer from diagnosis to death, due to the need for lifelong surveillance and postoperative monitoring [2].

Bladder tumors have a considerable risk for multiple tumor recurrences with respect to space and time, e.g., another bladder tumor will reappear out of 60% of all fully treated bladder cancer cases [3]. Patients need careful follow-up surveillance of superficial bladder tumor after surgery to avoid the high risk of recurrence and development to a higher stage of the disease. The current practice is that patients with high-grade tumors of any stage should undergo cystoscopy and bladder wash urine cytology every three months during the first year of surveillance. If any cancer does not recur during the first year of surveillance, follow-up monitoring continues every 6 months during the second year. Annual cystoscopic surveillance will follow after two years of intensive follow-up monitoring [4].

The gold standard examination of the lower urinary tract consists of an optical examination of the bladder urothelium, the tissue layer lining the interior of the urinary tract, or cystoscopy. Bladder tumors/malignancies are also detected via urine cytology and/or via upper urinary tract studies such as computed tomography (CT) urogram, ultrasound, magnetic resonance imaging (MRI), intravenous pyelography (IVP), or virtual cystoscopy. Urine tumor markers and urine cytology are cost-effective for diagnosis of bladder tumors, especially for a high-risk population, and are noninvasive [2]. However, its sensitivity is low at 55.7% [2] compared with CT/MRI cystoscopy at 88.9–100% [5] and conventional CT images at 76–80% [6]. Although CT has a sensitivity of around 90%, it does not readily distinguish low volume tumors [7]. For example, 10% of the bladder lesions smaller than 5 mm that were detected by the direct visualization, cystoscopy, were not detected by CT images and virtual cystoscopy [8]. IVP is also a minimally invasive diagnostic procedure and uses contrast material with an X-ray. This procedure thoroughly examines the urinary tract for presence of neoplasia, but tumors smaller than 3–5 mm in size can be missed because the natural folds of the bladder prevent precise examination of the inner lining by conventional X-rays. All virtual endoscopies, most recent advances, can be criticized because of insufficient information on the color and texture of the urothelium and mucosa and the absence of biopsies [5].

Long-term endoscopic follow-up in the bladder after treatment is crucial to guarantee early detection and treatment of cancer recurrence. However, while cystoscopy is the “gold standard” for a routine surveillance exam, it remains imperfect, with various experimental studies demonstrating 4–10% missed bladder tumors by conventional cystoscopy [9–11]. Cystoscopes are positioned manually by a clinician and provide only a narrow field of vision. Thus, the completeness of the physician’s scan cannot be verified while being skill and method dependent. Urologists prefer the rigid cystoscopes so that he/she can easily biopsy anything he/she finds. However, current cystoscopes range in size from 6 Fr (French, 1 mm=3 French) to 27 Fr, including the flexible cystoscope (6–14 Fr), and rigid cystoscope (17–27 Fr). According to Dryhurst and Fowler [12], a thinner flexible scope is beneficial for significantly increasing patients’ comfort if it provides comparable image quality to rigid cystoscopes, reasonably priced cost, and better usability. The paper of Dryhurst and Fowler [12] suggests that a flexible small-caliber instrument should be used for most diagnostic cystoscopies and larger cystoscopes should be reserved for operative intervention.

Due to the limitations of currently available technology, there is a growing need for efficient and size-effective postoperative surveillance and preventive screening tools for all patients at high risk for bladder cancer. The specific goal of this paper is to propose a steering mechanism and software algorithm that will help the imaging probe automatically scan the entire inner surface of the bladder by means of a remotely controlled scanning path. This approach will eliminate the critical dependence on the physician's skill and memory. Three basic features, which will constitute the final surveillance system, are introduced below.

1.2 Scanning Fiber Endoscope

The novel design of scanning fiber endoscope (SFE) being developed at the University of Washington promises an ultrathin optical imaging and tumor detection tool without sacrificing image quality [13]. The small diameter of the SFE may reduce the use of anesthesia and dilation for the endoscopic procedures. Recently a 1.2 mm outer diameter SFE having high color imaging quality at video rate has been developed and successfully tested [14]. With a rigid 9 mm long micro-optical scanner at the distal tip, this new design produces full color and higher pixel-count images than fiber-optic imaging bundles of comparable size. The scanner of the SFE contains a tubular piezoelectric actuator around the single-mode optical fiber with the distal tip vibrating in resonance to scan laser illumination at about 10 KHz resulting in 500 line images at 30 Hz frame rate.

1.3 Multisegmented SMA Active Tip Bending Mechanism

A multisegmented SMA active tip bending/navigation mechanism can be integrated to the SFE. Each SMA segment is activated by providing an electrical current to heat the segment. The activated portion contracts and produces a mechanical force that deflects the distal end of the imaging probe. When deactivated, the contracted portion of the SMA returns to its original shape and length by the restoring force of the shaft elasticity. By controlling the individual segments, it is possible to manage different configurations of the SFE shaft. A single (or multi-) axis multistep graded bending motion of the SFE can be demonstrated by a straight electrically multitapped SMA wire (or sinusoid or helix), which is attached to one (or multiple) side(s) of the SFE probe tip. Initial selection of the parameters such as length/number of segments and the stiffness of the shaft is important in the prediction of the position of the SFE tip in its approach to a desired point in the workspace.

Unlike competing smart materials such as elastomeric polymers and some electroactive polymer actuators, the force generation of a SMA material is very high, typically 10–100 times than those of polymer based actuators. This in combination with a capability of undergoing large strain was the reason that SMA was chosen as the “best choice” material for making the scanning mechanism of the ultrathin and flexible cystoscope due to the large force generation and strain.

1.4 Automated Optimal Trajectory

The automated continuous and optimal path of the imaging probe, unlike manual cystoscopic observation by clinicians, may help insure that each local internal surface of the bladder is inspected in a controlled and uniform manner. The occurrence of a tumor on a small portion of the bladder surface that is not imaged and is thus overlooked can have serious and even life threatening consequences to the patient. Thus, imaging of the surface of a bladder to detect problems such as cancer must be complete and include substantially the entire surface. An automated controller can be programmed to position and orient the imaging probe by controlling at least one of the following: (a) the insertion depth of the elongated optical imager, (b) the rotation of the elongated optical imager about its longitudinal axis, and (c) the bending of the optical imager. Using a three-dimensional (3D) model of the bladder, a simple but crucial

test is performed to check whether the scanning tip can be located at the designated points to ensure that all portions of the bladder surface are imaged completely.

2 Materials and Methods

In order to validate, optimize, and manage the segmented active tip steering mechanism, we need to know the exact tip location and orientation in a patient-specific bladder environment. The basic principle of our design is to use open-loop control as far as possible, thereby minimizing closed-loop or complex control laws that would require sensor feedback. Selden et al. [15] presented a segmented linear SMA actuator. They obtained the global linear displacement by the combination of small displacements of each segment. In this paper, we focus on two new aspects: applying modules of independently controllable series of SMA segments to achieve large global motion and introducing a patient-specific trajectory design for a bladder surveillance application.

2.1 Modeling Tip Bending

For a simple 2D cantilever bending configuration, the relationship between the bend radii and the strain of the innermost part is

$$\varepsilon = \frac{(R - r_n)\varphi}{r_n\varphi} = \frac{t/2}{r + (t/2)} \quad (1)$$

where ε indicates the strain of the innermost side and t denotes the diameter of the object being bent. R , r_n , and r are the radii of the outmost, neutral, and innermost axes, respectively, and φ is the bending angle. Equation (1) indicates that strain and thickness are the only determining factors in the bend radius with the same shaft stiffness. Therefore, as expected, adding activated length only increases the bending angle, while the bend radius remains the same (See Fig. 1).

For a multisegmented bending mechanism, the positioning is mainly determined by the degrees of freedom (DOFs) of each segment and number of segments. A multimodule structure, where a module is defined as an independently controllable group of segments (Fig. 2), covers wider locations in space than a single module system with the same number of segments [16]. The number of tip location is a function of the number of module and segment, the number of longitudinal base shift (NBS) for the shaft insertion depth, and the degree of freedom of each segment. The total number of tip locations (NTL) can be expressed, as follows:

$$\text{NTL} = (s+1)^m \times \text{NBS} \times \text{DOF} \quad (2)$$

where s is the number of segment per module, and m refers to the number of module. To simplify the modeling, the interface circuit, and the manufacturing process, we make the following assumptions: First, each segment bends to only one side (positive deflection, one degree of freedom). Second, graded bending motion always starts from the distal side. Third, each segment has the same length (L) and deflection angle (θ). By controlling the arc through which the distal end of the imaging probe is deflected, the insertion depth, and the rotation of the shaft about its longitudinal axis, it is possible to position and orient the imaging tip in the bladder to image the urothelium.

In order to simulate the bending of the distal end of the imaging probe and its changing position and orientation, a forward kinematic model of a multilink planar manipulator is introduced and programmed in MATLAB® (MathWorks, Inc., Natick, MA). The basic relationship between the bending radius (r), the bending angle (Φ), and the deflection angle (θ) is illustrated in Fig.

2 for a model having two modules with two segments. The simulation tool predicts the location and orientation of the tip with a multiple module/segment structure based on the input information such as bend radius, number of modules, number of segments, segment length, and segment activation order.

2.2 Design of Optimal Trajectory

Completion of the tip bending model allows us to determine an optimal trajectory within a specific bladder volume. Figure 3 illustrates one exemplary approach for manipulating the position and orientation of the imaging probe to ensure that the entire inner surface of a sphere-assumed bladder is scanned to produce images. In this simplified two-dimensional view, a plurality of points is designated at spaced-apart intervals around the inner surface. Figure 3 (a) shows an imaging probe that is inserted into the distended bladder and positioned at the point selected to produce an image of portion A of the inner surface. In this initial position, the distal end of the imaging probe is not deflected. Next, an actuator is selectively activated to produce a deflection of the distal end of the shaft in a prescribed configuration. An image of a portion B is then produced with the imaging probe slightly positioned to the upside of the initial position. A succession of additional images B' is then produced by rotating the imaging probe about the longitudinal axis of the scope. Figure 3(b) illustrates an imaging probe configuration after it has been deflected and retracted by Δd_1 to take an image of C and C'. Then the image probe is reinserted by Δd_2 and is bent through a substantially greater arc compared with that shown in Figs. 3(a) and 3(b) to see D and D', around 180 deg behind the bladder neck (Fig. 3 (c)).

There are multiple ways of approaching the prearranged points with a prescribed orientation varying segmental information, in other words, a shaft arc, as well as the insertion depth. The simulation tool helps to find an optimal setting, which satisfies basic requirements. The designer selects input parameters and the filtering process sorts out the only appropriate tip locations, as described in the flowchart in Fig. 4. A data set that contains the geometric coordinate information of the bladder surface wherein each point may be described as a reference point of the contour is imported for the filtering process. The tip positions are chosen such that they are spaced apart from each other and at a defined distance from the bladder surface with minimal variation to produce stable imaging. The entire surface of the bladder surface must be optically scanned (100% coverage) including the vicinity of the urethra entrance. It may be desirable to select the position so that discrete images of the surface produced at adjacent points slightly overlap, to facilitate creating an overall composite image of the surface. Finally, it is recommended that the number of segments and modules are minimized to avoid too much complexity.

As a tip location is chosen, a constrained region is generated for simulating the region of viewing based on the field of view (FOV) and depth of focus (DEPTH_FOCUS). If $m+1$ points from the bladder contour are included in the region (n through $n+m$), the simulation tool checks each data point to see whether the point is included in the area of interest. The dot product of two vectors at the tip is

$$TP \cdot TB_{n+k} = |TP| \cdot |TB_{n+k}| \cos \theta_{n+k} \quad (3)$$

where TP denotes the vector starting from the imaging tip T extended by the FOV with the same direction, TB_{n+k} represents the vector between the imaging tip and each contour point, B_{n+k} , and θ_{n+k} refers to the angle between TP and TB_{n+k} .

Two parameters, which determine the area-of-interest check, are as follows:

$$\text{Safety distance} < |TB_{n+k}| < \text{max marginal distance} \quad (4)$$

$$\theta_{n+k} < \frac{\text{FOV}}{2} \quad (5)$$

Similarly, orthogonality of the tip to the bladder surface is checked comparing the angle between TP and the vector $B_n B_{n+m}$. If the tip location satisfies the orthogonality criteria shown below, it passes.

$$TP \cdot B_n B_{n+m} = |TP| \cdot |B_n B_{n+m}| \cos \alpha \quad (6)$$

$$\text{Pass if } \alpha > \text{orthogonality criteria angle} \quad (7)$$

where B_n and B_{n+m} are the first and last contour points included in the area of interest, respectively, and θ denotes the sharp angle between TP and $B_n B_{n+m}$.

Figure 5 shows a schematic of an exemplary tip location with the area-of-interest and orthogonality check. The simulation tool returns validating information such as the number of tip locations chosen, tip location/orientation, bladder wall imaging coverage (percent), and total overlapped portion of the imaged area as well as the current input setup.

2.3 Shaft Preparation

A two-module-three-segment prototype was built for experimental validation. The system parameters, such as number of modules/segments and segment length, were determined by the simulation tool described in the modeling and design sections. The shaft preparation procedure, shown in Fig. 6, can be summarized as follows.

The original tubular design of the Teflon sheath for the SFE revealed some difficulty undergoing large bending deflections with the limited force generated by the contracting actuators. In the tube configuration, bending is inhibited by the outer tube sheathing because the outer tube is located far from the neutral axis of the shaft and needs to be extended when the SMA actuator is contracted. The flexural rigidity (EI , where E is the modulus of elasticity and I is the second moment of inertia of the shaft) of the original SFE was measured as $1.16 \times 10^{-4} \text{ N m}^2$. In order to reduce the stiffness of the shaft in the bending direction and obtain small bending radius, a medical grade microurethane tubing (85 Durometer Polyurethane, $1.02 \times 1.40 \text{ mm}$ ID and OD, Scientific Commodities, Inc.) was used. This microurethane tubing has service temperature ranging up to 80°C , which fits well for this application with the 70°C activation temperature of the SMA helix.

In order to make the shaft quickly spring back to its original shape when the bending stress is removed, a straight superelastic nitinol wire (0.25 mm OD, NDC Co., Fremont, CA) was added to the inside tubing along the optical fibers. In this design, the shaft components of the SFE including the nitinol wire work as a restoring force mechanism. The superelastic alloy wire can be deformed eight to ten times more than an ordinary stainless steel wire without permanent plastic deformation and snap back to its “preset shape” with a considerable force. In addition, the modulus of elasticity of nitinol is substantially lower than that of stainless steel, 20–83 GPa versus 200 GPa. After changing the tubing material from Teflon to polyurethane, the flexural

rigidity value dropped down to less than one-half of the original value, $5.17 \times 10^{-5} \text{ N m}^2$ including a nitinol wire.

The BioMetal Helix (0.150 mm OD, Toki Co. Tokyo, Japan), a commercial microcoil SMA actuator, was chosen for the prototyping. The advantage of this helical shape actuator over the standard SMA nitinol wire exists in its large contracting strain, 100% versus 5%. This actuator consumes 200–300 mA of current when it contracts and can generate a force around 3 N. The partially plastic-deformed SMA helix actuator, which is elongated over its limit length from its initial setting, 250%, no longer returns to its original length although it still contracts when activated. In that case, it shows more force generation and less strain than the original helix, which is elongated within its limit length. For example, when stretched to 500% of its original length, it generates a 5 N recovery force with 50% stroke of the extended length. As we needed more contraction force, the 500% stretched SMA helix was used for the prototyping.

Eight electric connection taps made out of eyelet tube terminals (Mark Eyelet Inc. Watertown, CT) were evenly distributed (8.5mm apart) and assembled to the partially plastic-deformed SMA helix actuator (500% prestretched) by crimping. The resistance between taps is approximately 1.5 Ω . The electrical lead wires were soldered to the connector taps as numbered and the preassembled SMA-lead wires were attached to the SFE shaft using a fast setting adhesive (Loctite® 401™ Prism®). The shaft components are labeled and displayed in Fig. 6 along with major components used during shaft assembly and preparation.

2.4 Phantom Bladder

A phantom bladder was fabricated and used as a proof-of-concept in vitro model to test the trajectory for optimal bladder imaging. A computed tomography (CT) cystogram of a 59 year old female patient's bladder filled with 300 cc radiographic contrast medium was used to construct a 3D plastic phantom bladder. The patient was then scanned from top to bottom (cranial to caudal) using 0.6 mm increment. A series of total of 138 image files of the bladder was acquired and each edge of the sliced bladder was traced. Finally, the edge-marked images were reviewed by an experienced urologist on a slice-by-slice basis. A customized LABVIEW (National Instruments Corporation, Austin, TX) program was used to calculate the coordinates of all marked edge points. The point cloud information of the edge with x , y , z coordinates (4800 points) was converted into a solid object in stereolithography (STL) format which is exported to a 3D printer, Dimension BST 768 (Stratasys, Inc. Eden Prairie, MN). A plastic urethra (10 mm OD) and two ureter (4 mm OD) tubes used as cavities in the mold were attached to the rapid prototyped bladder model. Captured images of an exemplary procedure are introduced in Fig. 7. The 3D printed plastic bladder was used as a core of a mold casting for a polydimethylsiloxane (PDMS) elastomer. The hollow PDMS mold represents a 300 cc bladder, as shown in Fig. 8.

2.5 Electronics, Hardware Setup, and Control User Interface

A single-axis bending motion of the shaft assembly was demonstrated by constructing a customized electronic switch circuit (Fig. 8 and Fig.9). When being heated above the transition temperature by the electrical current flowing through the SMA actuators, the controlled portion of each module contracts according to the number of actuated sections. Two eight-channel CMOS analog demultiplexers (model DG408DJ™, Maxim Integrated Products, Sunnyvale, CA) were used for the two-module-three segment prototype. They decode one-of-eight lines and modulate metal oxide semiconductor field effect transistors (MOSFETs) to control the electrical current to the SMA segments.

In order to position the imaging probe efficiently within a volume with minimal circuit complexity, it is essential to control the insertion depth and rotational position of the imaging

probe. The drive station includes a translational linear actuator, CMA-25CCCL (Newport, Irvine, CA), that controls the insertion depth of the shaft in the x -direction to advance or retract the imaging probe. A rotational stage, SR50 (Newport, Irvine, CA), rotates the shaft in either direction about its longitudinal axis, x -axis. Analog output voltages are delivered by a National Instruments analog output board, PIC-6722 (8 Channel). A LABVIEW application was developed for translational and rotational motion controls and a standard RS-232-c communication protocol was used between LABVIEW and the motion controller, ESP 300 (Newport, Irvine, CA) (Fig. 9).

A virtual instrument control panel in LABVIEW was designed to allow the user to preprogram an activation sequence of the segments in each module. The actual sequence used was computed using the geometry of the bladder. In addition, the translational position steps and rotational angle per each position can be preprogrammed. Using this configuration, the automated controller can position and orient the imaging probe by controlling all three parameters, the insertion depth, the rotation, and the arc through which the distal end of the imaging probe is deflected.

2.6 Tip Tracking

We used a state-of-the-art commercial motion tracker to measure the position and orientation information of the imaging probe, microBIRD (Ascension Technology Co., Milton, VT), a 6DOF dc magnetic tracker with a sensor diameter of 1.3 mm. This system provides 0.5 mm resolution and instantaneous tip localization technology to track the tip location with 1.4 mm rms accuracy and 60 measurements per second update rate. Three orthogonal receiving coils at the tip sequentially sense orthogonal magnetic fields that are generated by a flat type transmitter located under the phantom. A utility software program, PCIBIRD CUBES (Ascension Technology Co., Milton, VT), records the receiver position and orientation in space with respect to the origin of the transmitter.

3 Results and Discussions

3.1 Design Setup and Simulation Results

Two independently controllable modules of three segment actuators were finally chosen for the prototyping considering the physical volume size of the bladder, complexity of the control circuit, and design requirements such as imaging coverage, overlap (20–30%), and distance from the wall (10–15 mm). Table 1 shows the finalized prototype design setup. From the total 400 possible tip locations that are controllable with the setup shown in Table 1, the simulation tool generated 11 optimal tip positions and a manipulation procedure for imaging the specific bladder (Fig. 10). It should be noted that the nonsymmetric bladder wall with respect to the rotation axis may produce some difficulties in precise tip positioning. However, the large range of focus of the SFE, 2–55 mm, and the large overlap would overcome the effects caused by the irregular bladder shape.

3.2 Testing of Prototype

An active steering system consisting of two independently controllable series of three SMA segment actuators has been fabricated and tested without an imaging probe. This prototype with the setup given by the simulation tool provided a minimum bend radius of 16.1 mm with more than 180 deg of tip bending (Fig. 11). This is comparable to commercial flexible wire-pulling ureteroscopes, which has a bend radius of 20 mm or bigger. The multimodule prototype has at least two independent modules and each has its own gradual bending which results in more selection of tip locations for orthogonal viewing. For example, as shown in the right four exemplary images of Fig. 11(b), this two-module bending mechanism provides some shaft configurations that are not possible from the conventional wire pulling bending system.

The 2D coordinate data from the electromagnetic tracking system were compared with simulation results. We obtained a 5.1% tip positioning error compared with the desired trajectory (Fig. 12). The tip positioning error was calculated based on the average distance between each measured and simulated location divided by the shaft length. A few factors could explain the difference between the experimental and analytical results. First, the prototype had a little fabrication misalignment between the modules resulting in a discrepancy between the measured and simulated tip location. Second, more thorough control of each SMA segmental actuator is recommended while generating prestretched deformation. Nonetheless, since the configuration was set to provide around 20–30% overlapping visual field, the proposed preprogrammed steering of the prototype shaft covered the entire bladder contour. To verify full coverage of the bladder surface, the 3D tracked tip trajectory of the imaging probe is illustrated in Fig. 13. The tracked results supports that the proposed configuration covered the entire surface of the bladder.

A parametrically defined 3D imaging trajectory enables the accurate and systematic scanning of the entire surface of the bladder. The latitudinal circular scanning obtained by a 360 deg rotation of the shaft and discretely controlled vertical bending motion will sufficiently cover the whole surface of the bladder. A few different approaches can be used to control the location at which images of the surface are produced during the scanning process. One variation of the latitude trajectory is the spiral trajectory which also effectively covers the surface of the bladder. A continuous movement of the scanner tip along the spiral path from the farthest point from the urethra entrance to the proximate to the urethra entrance, bladder neck, will scan all regions of the inner sphere of the bladder.

Some alternative techniques will be considered for the next prototype for more precise trajectory control. Currently, the positioning technique is open loop. A closed loop control can be implemented by using the proposed trajectory as a rough guide with position and orientation of the scope tip actively sensed. To achieve a closed loop trajectory control, the feedback could be derived from image processing, acoustic sensors, or mechanical pressure, contact sensors. In addition, there is an important issue that should be implemented in the future research. Although this test was conducted in an open-air environment, the real cystoscopic procedure is usually performed in saline filled bladder. To support the clinical utility of these preliminary results, we have conducted a simple tip bending experiment in de-ionized water. When the actuators were activated in 20°C water encapsulated in a water-proof fabric shielding (GoreTex), 11–17% power consumption increase was necessary in order to have the same deflection angle obtained in air. The power consumption will vary depending on the heat transfer conditions such as the shielding material, external fluid forcing conditions, rate of probe motion, and fluid temperature. Thus, subsequent simulations and experiments need to consider transient heat transfer effects from the surrounding fluid for better prediction and control. It is also possible that the surveillance procedures can be conducted in an open-air environment filling the bladder with air. According to personal communication with urologists, distending the bladder with air is feasible and safe although the current standard procedure is performed using a saline fill.

4 Conclusions

Given the advantages of cystoscopic exams compared with other procedures available for detecting bladder cancers and other urological conditions, it would be desirable to develop an improved automated cystoscope that can be readily controlled to scan the entire inner region of a patient's bladder. Locating medical instruments in 3D space is a key requirement for image-guided procedures, especially, for scanning a surface of a volume that is accessed through an opening, where a cross-sectional dimension of the volume is substantially greater than a cross-sectional dimension of the opening and the passage. The ability to automatically provide a

whole map of the scanned organ indicating tumor location and size would also be advantageous for record keeping, continuity of care, and preoperative planning. We report an active, programmable distal steering mechanism prototype and an efficient motion sequence simulator tested on a phantom bladder. Preliminary experimental test results ensure that the motion sequencing program and the steering mechanism will efficiently move the image probe to scan the entire surface of the bladder.

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Fig. 1. Diagram of three different bending motions with the same bend radius, (a) undeflected state, (b)–(d) one, two, and three distal segments are deflected with the same bend radius r , respectively

Fig. 2. Definition of segment coordinates and trigonometric relationship between bending angle/radius and deflection of a multisegmented bending mechanism (for clarity, only two segments in Module No. 2 are activated out of the two module-two segment structure)

Fig. 3.

One exemplary approach for manipulating the position and orientation of the imaging probe in a sphere-assumed bladder using a shaft arc, insertion depth, and rotation (black portion in the shaft stands for rigid shaft). Shaded areas illustrate the portion being imaged.

Fig. 4.
Tip location decision making flowchart

Fig. 5.
A schematic of an exemplary area-of-interest and orthogonality check. Solid squares represent the reference data points, which describe bladder surface, and the shaded area denotes the area of interest.

Fig. 6.
Shaft building process

Fig. 7. (a) Original CT image; (b) edge detected, white dots, and calculated coordinates of edges by LABVIEW; (c) point cloud; (d) mesh object; (e) 3D printed bladder model with ureters and urethra tubes attached; and (f) the reference frame for the position and orientation of the imaging tip and CT images

Fig. 8.
Test setup with PDMS phantom bladder

Fig. 9. Schematic of the prototype and test setup (only one module/half circuit shown for clarity)

(a)

Base shift [mm]	Set up Info		Activated segment #		Tip Location		Orientation	
	# of module	# of segment	M1	M2	Z [mm]	Y [mm]	Z [mm]	Y [mm]
36	2	3	3	2	45.86	38.45	-6.06	5.82
29	2	3	0	2	74.40	11.54	4.14	7.31
29	2	3	3	3	28.58	32.19	-8.17	1.98
28	2	3	2	1	56.16	34.74	-0.11	8.40
25	2	3	3	2	34.86	38.45	-6.06	5.82
24	2	3	1	1	67.08	20.11	4.14	7.31
21	2	3	0	1	70.76	4.23	7.26	4.23
21	2	3	1	0	67.29	17.08	7.34	4.28
20	2	3	2	0	52.45	33.74	4.19	7.40
19	2	3	0	0	70.00	0.00	8.50	0.00
17	2	3	2	1	45.16	34.74	-0.11	8.40

(b)

Fig. 10.

Simulation results of the tip locator algorithm: (a) eleven tip locations (red circles) with bladder contour (black solid dots) are shown as the optimal scan trajectory. Radial lines (green) from the probe tip represent the imaging area, while a 45 mm radius circle (blue) is shown for reference. (b) Series of shaft manipulation procedural showing the expected tip coordinates/ orientation.

(a)

(b)

Fig. 11.
 (a) Captured image of bend radius measurement in a customized LABVIEW program, and
 (b) captured images of four exemplary shaft bending motions

(a)

	Simulation (S)		Experiment (E)	
	Z (mm)	Y (mm)	Z (mm)	Y (mm)
1900	70.00	0.00	70.60	0.00
2101	70.75	4.23	71.03	4.74
2110	67.29	17.08	66.62	19.16
2821	56.16	34.74	51.51	36.83
2933	28.58	32.19	23.13	27.76

(b)

Fig. 12.

Exemplary five simulated probe tip locations verified with experimental measurements (a) in graph, (b) detailed comparison data (e.g., S_2101 versus E_2101: Simulated and experimental base displacements at 21 mm with zero and one segment of modules 1 and 2 being activated, respectively)

Fig. 13.
3D tracked tip locations of the prototype

Table 1

Variable setup for prototyping

Variable	Value	Unit
Number of modules	2	N/A
Number of segments in a module	3	N/A
Length of each segment	8.5	mm
Length of rigid scanner at the tip	10	mm
Length of connection part between Modules 1 and 2	0	mm
Field of view, 70 deg	1.2	rad
Depth of focus (focal length)	30	mm
Max distance from wall	15	mm
Min safety distance from wall	10	mm
Min distance between two tips	11	mm
Min angle between two tip vectors	$\pi/6$	rad
Min angle between wall and tip vector for checking orthogonality	$\pi/4$	rad
Base displacement at start from bladder entrance (x-coordinate)	36	mm
Base displacement at finish from bladder entrance (x-coordinate)	11	mm
Translational shift increment	1	mm