Steerable catheters for minimally invasive surgery: a review and future directions

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ABSTRACT

The steerable catheter refers to the catheter that is manipulated by a mechanism which may be driven by operators or by actuators. The steerable catheter for minimally invasive surgery has rapidly become a rich and diverse area of research. Many important achievements in design, application and analysis of the steerable catheter have been made in the past decade. This paper aims to provide an overview of the state of the art of steerable catheters. Steerable catheters are classified into four main groups based on the actuation principle: (1) tendon driven catheters, (2) magnetic navigation catheters, (3) soft material driven catheters (shape memory effect catheters, steerable needles, concentric tubes, conducting polymer driven catheters and hydraulic pressure driven catheters), and (4) hybrid actuation catheters. The advantages and limitations of each of them are commented and discussed in this paper. The future directions of research are summarized.

KEYWORDS

Steerable catheter; Continuum manipulator; Design; Minimally invasive surgery

1. Introduction

Minimally invasive surgery (MIS) utilizes image-guided procedures to diagnose and treat diseases in nearly every organ system, and MIS has revolutionized surgery in the last two decades. By minimizing the physical trauma to the patient, MISs can reduce infection rate and recovery time considerably and it can also allow shortening a hospital stay of the patients. One of the common tools used in MIS, in particular, is the catheter which is a long, thin and flexible tube or wire. The catheter is inserted into the vascular system, gastrointestinal tract and airway for both diagnosis and treatment. Currently, the majority of MIS procedures are performed manually with the so-called deflectable or conventional catheters. Such catheters have a limited range of motion and flexibility and much rely on the operators’ skill and experience to maneuver the catheter tip to reach and interact the target site in a stable manner. The complexity of the anatomy of the pathway and target site and the lack of information of the contact force are responsible for some difficulty in the operation of catheters in terms of dexterity, safety and stability (inappropriately leading to failures of the operation).

The steerable catheter refers to the catheter that is manipulated by a mechanism which may be driven by operators or by actuators (i.e., automatically). The steerable catheter has thus many advantages over the conventional catheter. The deflection of the distal tip of the steerable catheter is more controllable with a possibility of remote control. A typical steerable catheter with its control unit is shown in Figure 1, in which the distal tip of the catheter can be deflected by a pair of tendons. By comparing the steerable catheter with the conventional catheter, the main advantages of the steerable catheter are the improved accessibility to the difficult anatomy, the improved catheter stability in operation, the then reduced fluoroscopy times and the then decreased total radiation exposure to both patients and physicians [1–4].

The technology for the steerable catheter has been advanced greatly in the past decade. The commercial products of steerable catheters and robotic catheters (i.e., more advanced steerable catheters) are available, such as Polaris XTM Steerable Diagnostic Catheters.
(Boston Scientific Inc., Marlborough, USA), Artisan Extend Control Catheter (Hansen Medical Inc., Mountain View, USA), and Niobe Magnetic Navigation System (Stereotaxis Inc., St. Louis, USA). A number of minimally invasive clinical applications benefit from steerable catheters, including cardiac surgery [5,6], vascular surgery [7–9], aneurysm surgery [10,11], neurosurgery [12,13], arthroscopy [14,15] and intrauterine fetal surgery [16,17] and so on. It is noted that each application of the steerable catheter is subjected to a different constraint but each plays the same function. The concept of the constraint and function is widely discussed in the engineering design research community and is thus referred to (Fan et al. 2015).

Steerable catheters can be viewed as a kind of hyper-redundant robots or manipulators and differ from the so-called continuum manipulator. A continuum manipulator refers to a continuously bending manipulator made by elastic materials, which does not contain any rigid link to a generic function – i.e., the motion and force. Note that the device that fulfils the motion and force transfer is called mechanism [18]. One example of the continuum manipulator in medical devices is the colonoscope [19,20]. The continuum manipulator is similar to but distinct from the hyper-redundant robot in that the former does not contain any rigid link while the latter contains rigid links (i.e., a hybrid mechanism [21–23] or system in a more general sense in engineering [24]). As such, continuum manipulators are inherently compliant and articulate. Details of the continuum manipulators may be referred to the review articles [25–27].

The steerable catheter needs to meet the following requirements: (1) The diameter of the catheter is strictly limited due to their working environments, e.g., vascular system, gastrointestinal tract, airway or man-made pathway (e.g., the inner pathway of the endoscopy); (2) The catheter is expected to have a high degree of dexterity in order to reach any intricate target site in the “channel”; (3) The catheter is expected to have a proper stiffness for resisting the force, coming from the channel (the high stiffness of the catheter may provoke injuries, while the low stiffness of the catheter may not be able to perform a force-compliant task well owing to an insufficient tip stability of the catheter); (4) The material of the catheter is expected to be biocompatible (in mechanical, thermal, and biochemical) [28] and sterilizable; (5) The magnetic compatibility is expected if the catheter is used in the magnetic resonance imaging (MRI) environment; (7) The safety and cost of the catheter are two important design constraints; (8) The catheter is expected to be intuitive and easy to operate.

This paper provides a review of the steerable catheter with the goal to identify a gap between what is required and the current state of the steerable catheter. The review differs from the one conducted by Ali et al. [29] in several aspects: (1) the scope of applications in their review only covers cardiology while our review assumes a broad scope of applications; (2) the criterion they employed to classify steerable catheters is how the tip force is generated, which makes them classify the steerable catheter into two categories, namely direct force generation and indirect force generation, while our review focuses on the working or actuation principle of the steerable catheter.

The remaining part of the paper is organized in the following. In Section 2, steerable catheters and their classification based on their actuation mechanism or principle are presented. In Section 3, a comparison of different types of steerable catheters in terms of the
safety, cost, performance, and size of catheters is presented. Conclusions along with discussion of the future research challenges of the steerable catheter are given finally in Section 4.

2. Classifications of steerable catheters

Steerable catheters can be classified in several ways. For instance, steerable catheters can be classified based on the actuation principle or driving principle into the tendon-driven catheter, magnetic navigation catheter, soft material driven catheter, and hybrid actuation catheter. Based on the number of sections where actuators stay, steerable catheters can be classified into the single-section and multi-sections. Based on the workspace, steerable catheters can be classified into the planar and spatial ones. It is noted that catheters can also be classified upon their intended medical applications, particularly characterized with their diameter, length, shape of the tip, hydrophilic coating, with or without lumen, and torquability/stiffness. The detailed information of the selected steerable catheters can be found in Table 1. In this paper, the classification of steerable catheters based on the actuation principle is focused on.

2.1. Tendon-driven catheters

One of the most popular actuation principles is that a set of antagonistic tendons is used to control the orientation of the distal tip of the steerable catheter (Figure 2). Most of this kind of steerable catheters have a single flexible backbone that supports the tendons. The backbone is made up of a variety of materials, such as springs, elastic tubes and braided polymer tubes and fabricated using the laser-machined slotting patterns. Among these materials, the super-elastic material nitinol (NiTi) alloy is often used. The tendons are placed at equally spaced intervals along with the backbone and the structure of the tendons makes the shape of the backbone approximate a circular arc. The tendons at the proximal part of the catheter are controlled by the actuation unit which is further controlled by operators (Figure 2).

2.1.1. Single-section

The single-section tendon-driven catheter can have one to four tendons. The locations of the tendons in the manipulators are shown in Figure 3. The simplest tendon-driven catheter is composed of one backbone and one or two tendons, e.g., Polaris X™ Steerable Diagnostic Catheter (single tendon) and SteeroCath-Dx™ Bidirectional Steerable Diagnostic Catheter (a pair of tendons) by Boston Scientific Inc. [5]. The similar design of the steerable catheter can be found in [35,36,38].

To the steerable catheter used for the applications such as diagnosis and radiofrequency ablation (RFA) of tumors [44,45], there is no central open lumen (i.e., inner channel) in the catheter system. While for other applications, there is a lumen inside the catheter. Kutzer et al. [15] and Murphy et al. [46] reported a tendon-driven cannula with a large open lumen used to remove the osteolysis formed behind the acetabular shell of the primary total hip arthroplasties (Figure 4, Left). It has a 5.99 mm-outerdiameter and 5 mm-inner-diameter nitinol tubes as a backbone that is cut into many interlocked slots. Two channels are cut axially through the outer wall of the backbone and spaced 180° apart. Two driving cables are threaded through the channels to bend the distal tip. Jung et al. [42] reported a prototype of the catheter that consists of four articulation tendons and a flexible Teflon backbone (Figure 4, Right). The tendons are enclosed in a polyester mesh sleeve to hold the control wires to the catheter body. The catheter is driven by a continuum robotic electromechanical system, which consists of four servo motors to independently control each tendon, and a linear actuator to control the insertion of the catheter.

One sophisticated and commercial available tendon-driven steerable catheter is the Sensei X Robotic Catheter System for cardiac surgery by Hansen Medical Inc. [47,48], and this catheter is an electro-mechanical master/slave system that remotely controls a steerable guide catheter positioning within the heart. The catheter is composed of three components: the physician’s workstation, remote catheter manipulator, and steerable guide catheter (Artisan® Extend Control Catheter). There are two force sensors in the distal tip of the catheter to get the force information to the physician. The steerable catheter was developed by [43,49]. It is a 3.8 mm diameter catheter for cardiovascular surgery. The cross-sectional view of catheter is shown in Figure 5. It has a super-elastic NiTi spine as the central backbone for the articulation inside the catheter. The catheter is controlled with four equally spaced tendons that lie approximately on its perimeter within the polyimide sheath to allow for low friction. The tendons are restrained by a braid of stainless steel flat wire. The inner and outer diameters of the catheter are lined with laminated plastics for medical use. Four tendons are controlled by four DC motors with encoder feedback, respectively.
| Author (year)                          | Actuation principle | Section | DOF | Bending angle | Sections coupling | Distal tip force | Application                  | Outer Diameter (mm) |
|--------------------------------------|---------------------|---------|-----|---------------|-------------------|------------------|-------------------------------|-------------------|
| Guo et al. (1995) [11]               | Conducting polymer  | 1 2     |     | 42.3°         | N/A               | ×                | Aneurysm surgery              | 2                 |
| Della Santa et al. (1996) [30]       | Conducting polymer  | 1 2     |     | 28°           | N/A               | ×                | General MIS                   | 0.8               |
| Lim et al. (1995) [31]; Lim et al. (1996) [32] | SMA                | 2 4     |     | 13°           | Co-placed         | ×                | General MIS                   | 2.8               |
| Haga et al. (1998) [33]; Haga et al. (2000) [34]; Haga et al. (2002) [71] | SMA                | 1 2     |     | 90°           | N/A               | Sensor           | General MIS                   | 1.4               |
| Park and Esashi (1999)               | SMA                | 2 4     |     | 51°           | Co-placed         | Sensor           | General MIS                   | 2                 |
| Takizawa et al. (1999) [66]          | SMA                | 1 2     |     | 45°           | N/A               | Sensor           | Cerebral surgery              | 1.5               |
| Jayender et al. (2009) [65]          | SMA                | 2 4     |     | 24.7°         | Co-placed         | Model            | Cardiac surgery               | 1.5               |
| Szewczyk et al. (2011) [13]          | SMA                | 1 2     |     | 70°           | N/A               | ×                | Neuroradiology                | 1.2               |
| Bailly et al. (2011) [10]            | Hydraulic          | 2 4     |     | 90°           | Independent       | ×                | Aneurysm surgery              | 4.9               |
| Bailly et al. (2011) [10]            | Hydraulic          | 1 2     |     | 90°           | N/A               | ×                | Aneurysm surgery              | 4.9               |
| Ikuta et al. (2012) [8]              | Hydraulic          | 2 2     |     | 170°          | Co-placed         | ×                | Endovascular surgery          | 3                 |
| Degani et al. (2006) [42]            | Tendon            | 1 3     |     | 90°           | Independent       | ×                | Cardiac surgery               | 12                |
| Harada et al. (2007) [16]            | Tendon            | 1 2     |     | 40°           | N/A               | ×                | Intrauterine fetal surgery    | 2.4               |
| Zhang et al. (2009) [17]             | Tendon            | 3 6     |     | 90°           | Co-placed         | Model            | General MIS                   | 7.5               |
| Xu and Simaan (2010) [45]            | Tendon            | 1 1     |     | 135°          | N/A               | ×                | Removal of osteolysis         | 5.99              |
| Kutzer et al. (2011) [15]            | Tendon            | 1 3     |     | 180°          | N/A               | ×                | Cardiac surgery               | 6.35              |
| Jung et al. (2011) [37]              | Tendon            | 2 4     |     | 90°           | Independent       | ×                | Cardiac surgery               | 4                 |
| Camarillo et al. (2009) [46]         | Magnetic          | 1 2     |     | 45°           | N/A               | ×                | Endovascular surgery          | 2                 |
| Jeon et al. (2012) [59]              | Magnetic          | 1 2     |     | 120°          | N/A               | Sensor           | Cardiac surgery               | 2.67              |
| Niobe II system (Stereotaxis 2016)   | Magnetic          | 1 2     |     | 120°          | N/A               | Sensor           | Cardiac surgery               | 2.67              |
| Webster et al. (2006) [73]           | Pre-bent concentric tube | 1 2 |     | N/A           | N/A               | ×                | General MIS                   | 0.7               |
| Webster et al. (2009) [83]           | Pre-bent concentric tube | 3 6 |     | N/A           | Co-placed         | ×                | General MIS                   | 0.8(d)/2.39(p)    |
| Dupont et al. (2010) [81]            | Pre-bent concentric tube | 3 5 |     | N/A           | Co-placed         | ×                | General MIS                   | 1.85(d)/2.41(m)/2.77(p) |
| Abayazid et al. (2013) [80]          | Pre-bent concentric tube | 1 2 |     | N/A           | N/A               | Model            | General MIS                   | 0.5               |
| Butler et al. (2012) [12]            | Tendon & Pre-bent concentric tube | 3 6 |     | –/–/100°      | Independent       | ×                | Neurosurgery                  | 1(d)/1.3(m)/3.7(p) |
| Magellan™ robotic catheter (Hansen Medical Inc 2016b) [7] | Tendon & Pre-bent concentric tube | 3 6 |     | –/–/180°      | Independent       | Sensors           | Peripheral vasculature         | 1(d)/2.13(m)/3.25(p) |
| Couture and Szewczyk (2017) [66]     | SMA                | 2 6     |     | 155           | N/A               | Sensor           | Cardiovascular system         | 2                 |

*x* means unavailable; (d) means the distal section; (m) means the middle section; (p) means the proximal section.
The aforementioned tendon-driven steerable catheters have one entire backbone, and its stiffness can be varied in its entirety during the operation. While steerable catheters with several backbones connected in series (called the discrete catheter) can vary their stiffness separately, which can facilitate the varying of the stiffness at the distal tip. In the discrete catheter, the backbone segments are connected by joints, e.g., spherical joint (Figure 6, Figure 7). Figure 6 shows a highly articulated robotic probe (HARP) for minimally invasive cardiac surgery developed by Degani et al. [40]. It consists of two concentric tubes. Both tubes consist of the rigid cylindrical links connected by the spherical joints which can rotate 15° range, and there are four cables in the catheter: three for the outer tube (120° apart) and one for the inner tube. It is noted that both the orientation and stiffness of the distal tip can be controlled with the cables. When the cables are relaxed, the tubes become limp, and vice versa. Figure 7 shows a bending manipulator of 2.4 mm in diameter with a centrally inserted laser fiber (0.7 mm in diameter) for the intrauterine fetal surgery, developed by Harada et al. [16]. The distal tip of the manipulator is composed of a series of cylindrical parts controlled with four wires and a series of spheres controlled with one fiber, see Figure 7 (Right). The entire manipulator is designed based on modular design; particularly one module includes a cylindrical part and a sphere part. The number of modules can be changed according to the stiffness of the centrally inserted tool. The four wires are controlled by two ultrasound motors (Figure 7).

The steerable catheter with a multi-backbone was designed by eliminating the backlash, enhancing the down-scalability, and improving the payload. Such a steerable catheter is typically composed of several parallel elastic rods or tubes. For instance, Figure 8 shows a steerable catheter, which has a primary backbone and two secondary backbones; the two secondary backbones function to control the deflection. Xu and Simaan [41] reported a single-section continuum robot for minimally invasive surgery. It has four super-elastic NiTi tubes as its backbones. One primary backbone is centrally located and is attached to the base disk and end disk. Three identical secondary backbones are equidistant from each other and serve as driven tendons. Each secondary backbone is actuated in push–pull mode by an actuation rod.

2.1.2. Multi-section

In order to provide a sufficient number of degrees of freedom (DOF) or a sufficient large workspace for tasks, the steerable catheter has been designed to have multiple sections (as each section serves as one link and the connection between two neighbouring links gives a more freedom of relative movement between the links). Each section is steered with a group of tendons. The number of tendons for the distal section may not be the same as the number of tendons for the proximal section. For the multi-section tendon-driven catheter, the way the sections are connected (or called the section coupling) is crucial to the dexterity and overall size of the catheter. Usually, there are two ways for the section coupling: co-placed and distributed (Figure 9). In the co-placed coupling of sections, the tendons for the proximal section share the same pathways with the tendons for the distal section; the examples can be found in [50,51]. In the distributed coupling of sections, the tendons for each section have their independent pathways; the examples can be found in [47,52,53]. The co-placed section coupling may render to a smaller diameter catheter while the distributed section coupling has more
Camarillo et al. [52] designed a catheter with two articulated sections and two tendons per section arranged antagonistically from the distal to proximal. The way the sections are coupled is the distributed one. Carlson and Barbagli [47] presented a two-section catheter. Each section is controlled with four orthogonal tendons. The type of the section coupling is a distributed one. Xu and Simaan [51] presented a three section multi-backbone continuum robot for minimally invasive surgery. The structure of each section is the same as their single section multi-backbone continuum robot [41]. The type of the section coupling is a co-placed one. The backbones of the proximal and middle sections are concentric NiTi tubes, while the backbone of the distal section is the NiTi beam, which passes through the backbone tubes of the middle section.

Figure 4. Steerable catheters with an open lumen. (Left: The cannula driven by two tendons with an open lumen was designed by Kutzer et al. [15]. Reprinted with the permission from [15] @ 2011 IEEE. Right: The catheter driven by four tendons with an open lumen was designed by Jung et al. [42]. Reprinted with the permission from [42] @ 2011 IEEE.

Figure 5. Cross-sectional view of the tendon driven steerable catheter [43].

Tendon-driven catheters have simple structures and can be steered remotely easily. Due to the actuation principle, the structure is relatively bulky, causing the limit in terms of the small size of catheters (in particular to the multi-section catheter). The friction between the tendon and guide channel as well as the backlash in the joints of the tendon makes it difficult to control the catheter.

2.2. Magnetic navigation catheters

In the magnetic navigation technique, the catheter with the magnetic distal tip is steered within the patient body. Several large magnets are placed on either side of the patient, which generates a magnetic field around the patient. Physicians can control the distribution of the magnetic field to deflect the distal tip of the catheter within the patient body to the desired direction.

The commercially available products for magnetic navigation catheters include Niobe® ES magnetic navigation system (Stereotaxis, St. Louis, MO, USA) (Stereotaxis 2016) and Catheter Guidance, Control and Imaging-Maxwell (CGCI) (Magnetecs, Inglewood, CA, USA) (Magnetecs 2016). Niobe® ES magnetic navigation system is designed to make the treatment of complex cardiac arrhythmias safe and effective (Figure 10) (Stereotaxis 2016), [54]. There are two permanent magnets to generate a uniform magnetic field (0.08 T) in this system. The distal tip of the catheter is embedded with a small permanent magnet that interacts with the magnetic field. The intensity of the magnetic field is controlled by adjusting the relative position of the two permanent magnets. The insertion or retraction of the catheter is controlled by a catheter
manipulator. Operators steer the catheter fully automatically by a joystick or mouse based on the image feedbacks from a user interface system. The system proved a safe and effective tool in the treatment of supraventricular and ventricular arrhythmias based on the clinical feedbacks [55–60], (Arya et al. 2008). The main advantages with the remote magnetic navigation are the reduced fluoroscopy time and less complication. The drawback is, however, the increased total procedural time.

The magnetic field can also be generated by the electromagnets in addition to the one generated by the permanent magnets Nguyen et al. [9,61–63]. Stereotaxia Inc. designed a similar magnetic navigation system as Niobe® ES magnetic navigation system in the early stage named Telstar [61], in which the magnetic field is generated by three orthogonal electromagnets. Directional catheter navigation is accomplished by generating a desired magnetic field vector. Nguyen et al. [63] presented the CGCI system for cardiac surgery. The system consists of eight coil-core electromagnets, arranged semi-spherically around the chest on a standard fluoroscopy table. They can generate approximately 15 cm³ 3D region with a maximal uniform magnetic field strength of 0.14 T and the maximal perpendicular force (25 g) at the distal tip of catheter in the Niobe magnetic navigation system. The precision of the magnetic field both in its magnitude and direction is greatly improved, and the nearly real time control of the catheter tip can be achieved. Jeon and Jang [9] developed a catheter system to steer and unclog the catheter by the magnetic torque and force. The catheter is composed of a flexible tube with a rotatable drill tip at the distal tip, which is a permanent magnet. The magnetic field is generated by two types of magnetic coils: uniform coils and gradient coils. The magnetic torque and force can be independently controlled. To compare with the magnetic fields generated by permanent magnets, they have two advantages: (1) they can increase the stability and contact force of the distal tip by increasing the strength of the magnetic fields; (2) The system can
continuously and rapidly shape and reshape the magnetic fields, which provides instantaneously changes to the tip of the magnetized catheter, leading to a near real-time remote navigation.

Another way for magnetic navigation is to guide the catheter inside a magnetic resonance imaging (MRI) system. Gosselin et al. [64] presented a catheter and a guidewire with the ferromagnetic heads, steered by applying magnetic gradients inside a MRI system. However, the two ferromagnetic spheres introduce undesired dipole-dipole interactions. Gudino et al. (Gudino et al. 2011) reported that a micro-catheter was steered inside a MRI system. The tip of the catheter was built by an array of steering coils. Similar studies can be found in [65,66]. Their work shows that the optimization of the catheter design as well as the coil activation current is necessary for the safe remote control of the navigation of the catheter [65].

The control of the magnetic field with the permanent magnets by changing their relative position but without a power switching mechanism may cause some safety problem. Indeed, the switching of the entire system (on and off) can easily be realized by the electromagnets [56,61,63]. The major drawback of the control with the electromagnets is that the magnets show a nonlinear behavior, especially hysteresis, which is a challenge to control. Yet, another drawback is that the control of horizontal bending and that of vertical bending are highly coupled. Finally, only one section structure can be realized by the magnetic technique due to the limitation that the magnetic field always serves as one source of driving.

2.3. Soft material driven catheters

Soft material driven catheters are made by so-called smart materials, including shape memory alloy (SMA), hydraulic bellows, and conducting polymers, and they can be bent continuously via their own structures rather than via other external mechanisms (i.e., tendon and magnetic field). SMAs, especially of the nitinol family of material, have super-plasticity, biocompatible, high recoverable strains, good kink resistance, good steer-ability and torque-ability. SMAs can be made in a very small diameter (under 1 mm) and this feature makes them an ideal candidate for the guide-wires or catheters for MIS [67,68]. The catheters built with SMAs can be classified into two categories based on how the effect of SMA is applied to manipulate the tip: (1) the catheter is entirely shaped with the SMA effect, and (2) the tip of the catheter is shaped with the SMA effect.

2.3.1. Shape memory effect catheters

Catheters of this type are actuated with the SMA effect only. The catheter bends and shrinks due to the force generated by SMA actuators when heated. The SMAs transforms between the austenite phase and the martensite phase to generate force and deformation, induced by the temperature. After being plastically deformed in its low temperature (martensite) phase, the SMA material can recover strains up to 8% when heated to austenite (Tung et al. 2008). The structure of SMAs can be wires, coils, carved tubes, and flat springs.

Dario et al. [14] designed a catheter tip actuated by four SMA wires (90° apart), which are further embedded into the wall of the catheter. Four optical fiber sensors are embedded in the wall in order to get feedback information. The SMA wires are heated and cooled by circulating fluid through the same lumens, in which the SMA wires are located. Takizawa et al. [69] presented a catheter actuated by three SMA wires (120° apart) with 1.5 mm diameter. Tactile sensors are designed and installed on the tip of the catheter. It takes around 2 seconds to achieve the maximum bending angle 45°. Jayender et al. [70,71] developed a catheter actuated by three SMA wires spaced at 120°. The wires are micro-welded to stainless steel pads, which are further glued to the central catheter to be a multi-section catheter to obtain varying bending angles. The catheter with SMA wires is packaged in the case of injury of the vessel wall by the heat. Similar catheters for neuroradiology can be found in [13,72].

Using SMA wires can facilitate the miniaturization of the catheter. It can produce a large force but only generate a small displacement. In order to obtain a large displacement (which can subsequently produce a large bending angle) and quick response, micro SMA coils are used. The catheters are composed of multi-links which can produce an enough bending angle. Lim et al. [31,32] developed a multi-link active
catheter with 2.8 mm diameter. It consists of several links and joints made of SMA actuators. Three SMA coils are located at nearly 120° apart and heated by electrical current. The SMA actuators can be heated directly. However, they have low resistance (10 Ω). They can produce an indirect heating method using a nickel thin film deposited on a parylene coated SMA actuator (200 Ω). The maximum bending angle is 13°. Haga et al. [33,34] presented three active catheters whose diameters are less than 2 mm, which are actuated by series of distributed SMA coils. The first catheter consists of many distributed link-joint units and the diameter is 1.2 mm. Three groups of SMA coils are fixed at equivalent angles between two links and one joint. When the SMA coils are heated with the electrical current above a certain transition temperature, it shrinks and bends the catheter. The maximum bending angle is 11° under 80 mA input current. The other two catheters eliminate the link-joint units and the SMA actuators are fixed on the inner tube with adhesive materials. One has the liner coil outside the SMA actuators, and the other has the liner coil inside the SMA actuators. The liner coil is used as an electrical common ground and both terminals of each joint are connected to this liner coil. Their bending angles can achieve 60° under 50 mA input current and 90° under 80 mA input current. They also designed a guide-wire with only 0.5 mm diameter [73]. It can bend to one direction using a meandering SMA actuator with a stainless coil spring, and it can bends over 60° under 50 mA input current.

Although SMA actuators offer a compact alternative to conventional actuators, SMA actuators have several drawbacks: (1) they have relatively low machinability and need a special manufacturing process, i.e. laser machining; (2) the active catheters usually need several linked SMAs, and this requires many lead wires to control the SMA actuators, making the system more complex; (3) compared with the tendon-driven catheters and magnetic catheters, the SMA-catheter generates a relatively low tip-force; (4) the heat that is needed for the SMA actuators to work may cause some problem in terms of safety (e.g., removal of the heat generated from human body is a challenge). A high temperature may cause the injury of cells or tissue; (5) they have a nonlinear behavior of the strain with respect to the electric current or temperature. SMAs have a high hysteresis characteristic, as a result of which their control can be difficult [70,71].

Efforts have been taken to overcome the shortcomings of the SMA-catheters. Regarding Problem (2), as mentioned above, Park and Esashi (Park and Esashi 1999) presented a multi-link active catheter with an integrated COMS interface circuit for communication and control (C&C IC) within a 2.0 mm diameter (Figure 11). The number of lead wires is reduced to 3. Three SMA actuators are fixed between the links 120° apart with 3% deformation strain. The catheter bends due to SMA actuators. The maximum bending angle is 51° under 100 mA input current.

Regarding Problem (4), Tung et al. [74], (Tung et al. 2008) developed an actuator made from the laser machined SMA tubes with 1.5 mm long and 1.27 mm diameter (Figure 12). It can produce force of 1–2 N at 20% elongation. The 180° turn can be achieved with 9 actuators (each with bending angle of 20°).

2.3.2. Steerable catheters with a partial shape memory effect

In this case, the super-elasticity and large strain recovery properties of SMA materials are exploited yet without any heating process. Such an actuation mechanism is usually applied to flexible needles that
directly interact with soft tissues [75,76], (Webster 2007). Abolhassani et al. [77] provided a survey regarding needle insertion into soft tissues. One way to steer such a kind of needle is to make the needle with a beveled tip and to steer it using the asymmetric forces on the beveled tip. The beveled tip of the flexible needles can be bent due to the asymmetry of the tip when the tip contacts the tissue during insertion. The beveled tip can be re-oriented by rotation and then subsequently pushed forward to the target (Figure 13). The flexible needle will follow the insertion path when retracted. It is noted that the steerable needle cannot be operated in a free space but a constrained space.

Webster et al. [75,76] presented a flexible needle made from Nitinol for the insertion of steerable needles. The beveled steering effect is enhanced with the flexible material. A robot actuates and controls the needle. According to Webster et al. [75,76], the depth of needle in the tissue is 22–25 cm, the deflection of needle is about 10% of the depth, and the velocity of the needle ranges from 0.5 to 2.5 cm/s. In Webster et al. [75,76], a kinematic model was used for control, which is limited in the control performance due to missing the knowledge of the needle–tissue interaction along the length of the needle. Alterovitz et al. [78,79] developed a 2D planning algorithm for the insertion of a flexible needle with a beveled tip into soft tissues with obstacles. The interaction between the needle and soft tissue was modeled with the finite element method, and the model was used for the planner to account for tissue deformation. Golzman and Shoham [80] presented a robotic system for steering the flexible needles in soft tissue in the real-time closed-loop control. The planned needle-tip trajectory avoids the obstacle and hits the targets. Duindam et al. [81] presented a constant-time 3D motion planning algorithm for flexible needles using a so-called geometrical inverse kinematic model. Abayazid et al. [82] presented an ultrasound image guided control system to steer a flexible needle made from nitinol with an asymmetric tip. They used both the kinematics model and kinetics model of the needle–tissue interaction to predict the needle deflection and to control the needle.

2.3.3. Concentric tubes

In another way to steer the catheter or needle, the concentric tubes (Webster 2007), [83–85] are developed, which are made of a set of flexible tubes embedded with each other. When a curved elastic tube (guide-wire or catheter) is inserted into a series of elastic tubes, their shapes are decided by the mutually resultant curvatures because they have the same stiffness. By translation and rotation of the tubes with respect to each other, their curvatures and overall length of tubes may vary, so as to serve as the purpose of manipulation. The tubes are typically made from Nitinol in its super-elastic phase. They can be made into desired shapes by the heat treatment before assembly. A review of the state of arts of the device of the concentric tube robot can be found in [86].

Specifically, Webster et al. (Webster 2007), [85] presented an active cannula made from nitinol and composed of three telescoping, concentric and pre-curved tubes (Figure 14). The diameter of the largest part of the tube is 2.4 mm and the diameter of the smallest part is 0.8 mm. Each tube has a translation and rotation degree of freedom. A unit with three tubes and six-DOFs controls the concentric tubes. Other important achievements regarding the design, modeling and control of the concentric tube system can be found in [86–90].
Dupont et al. [83] developed a concentric tube robot, composed of three pre-curved nitinol tubes, which is similar to the active cannula by Webster et al. [85]. The diameter of the largest part is 2.77 mm and the smallest part is 1.85 mm. A tele-operation system was developed to achieve the real-time position control of tubes based on the forward kinematics and inverse kinematics models. Gosline et al. [84] used this concentric tube robot to percutaneously access the right atrium and deploy a tissue approximation device to complete the intracardiac beating-heart surgery. Other important achievements regarding design, modeling and control of the concentric tube system can be found in [91–93]. Xu et al. [94,95] presented a concentric tube robot composed of two Nitinol tubes. The outer tube was fixed while the inner tube was rotated. The position of the distal tip can be obtained from an electromagnetic tracking system. A fast torsionally compliant kinematic model using the global variables and model based control method was developed to control the tubes.

Compared with other types of robotic catheters, the concentric tubes are more flexible and smaller in terms of diameters. They do not utilize the full shape memory effect (i.e., without the need of heating the material). Additional sections can be easily added by increasing the number of tubes, which is often a challenge with other types of steerable catheters. Furthermore, the lumen of the concentric tubes can provide addition tubes or control wires for the tools mounted at the distal tips. However, the concentric tube steerable catheter has several limitations: (1) the selection of the initial curvature of each tube is non-trivial task (Webster 2007); (2) the small radii of the pre-curvature may cause the tube damage; (3) instability, caused by the snapping; (4) low rigidity to the environment; (5) difficulty in control for achieving a high steering accuracy; and (6) the bulkiness of control unit outside of patient.

### 2.3.4. Conducting polymers driven catheters

To a class of polymers, they can change shape after being doped electrochemically. Particularly, when ions are driven into such polymers axially, the polymers change their volumes axially, and when ions are driven into such polymers transversally, the polymers bend. Such a polymer is called electro-active polymer (EAP). Catheters can be driven by EAP. Particularly, the tip of catheters is made of EAPs and put in an electrolyte environment. After the electrical current passes through the polymers, ions are driven into polymers to expend the polymers. Dops can be configured to make the tip bending. The advantage of the EAP tip of the catheter includes: (1) low voltate (under 2 V) and (2) ease with miniaturization [96].

Guo et al. [11] proposed three micro catheters with an active guide-wire that has two bending degrees of freedom, actuated by an Ionic Conducting Polymer Film (ICPF). The basic structure of the catheter is shown in Figure 15. The bending principle is that an ICPF fixed at the distal tip of a guide-wire can be bent under a voltage input. Two lead wires were used for supplying the electric energy to the ICPF. The diameters of the micro catheters were 1 mm, 1.3 mm and 2 mm, respectively. The ICPF actuator had fast response, driven by low voltage (2 V), compared with the SMA in the wet condition without electrolysis and heat, safety in body [97]. The maximum bending angle was 41° with the distal displacement of 10 mm. Della Santa et al. [30] proposed two active catheters with 0.8 mm diameters with the polypyrrole conducting polymer and the ionic conductor solid polyelectrolyte as an ionic reservoir. In one catheter, two groups of conducting polymer actuators in the form of thin strips with a solid polymer electrolyte (SPE), respectively, are inserted in the catheter wall. In the other catheter, the wall of the catheter is made of conducting polymer fibers (CPF) embedded in a SPE elastomeric matrix (50% SPE and 50% CPF). The maximum bending angles (with a distal placement of 7.4 mm) are 28° and 24°, respectively. One drawback with SPE is that its stiffness is too low. Sewa et al. [98] presented a gold chemically plated perfluorocarboxylic acid film for the catheter system. The polymer can be
bent 90° with the 8 mm displacement of the distal tip under 2 V input voltage in water. The polymer also shows durability more than 10 million cycles. Alici et al. [99] proposed a conducting polymer actuator, which has potential to be the bending of the tip of the steerable catheter. The electrolyte layer in the middle is the polyvinylidene fluoride, which is clamped by two polymer polypyrroles as the electroactive components. The thin layers of the platinum are coated onto the polypyrroles to increase the conductivity. The maximum bending angle for the actuator strip with the length of 10 mm, the width of 1 mm and the thickness of 0.21 mm can be over 90° under 1 V input voltage. However, the maximum output force at the tip is only 0.006 N under 1 V input voltage. Fang et al. [100] developed an active guide-wire for cardiac catheterization by using the ionic polymer metal composites (IPMCs). A pair of parallel IPMCs are fixed at the distal tip of guide-wire, in particular, one serving as an actuator and the other serving as a sensor which uses the inverse way. The control signal consists of high and low frequencies. The low frequency signal makes the IPMC to deform and change its surface electrical resistance, while the high frequency signal retains the deformation information. By utilizing a lock-in amplifier to demodulate the high frequency signal, the deformation can be measured. The main drawback of the conducting polymers is of relatively low response and the nonlinearity due to their actuation principle, i.e. hysteresis and back relaxation. It is a challenge to control such a kind of actuators.

2.3.5. Hydraulic pressure driven catheters

Another actuation principle is that the catheters are composed of one or several hydraulic bellows and the distal tips are steered by hydraulic pressure of bellows. The variation of hydraulic pressure inside the bellows modifies their lengths and, thus the bending of the catheter. Haga et al. (Haga et al. 2005) developed a catheter of 0.94 mm in diameter, actuated by the hydraulic suction mechanism for intravascular MIS. The catheter is made of the Nitinol tube covered by a silicone rubber tube (Figure 16), in which the nitinol tube is cut of a line of rings with the connecting beams. The catheter is filled with water and its bending angle is controlled by the suction of water.

Bailly et al. [10] proposed a two-section catheter called MALICA for the endovascular treatment of aortic aneurysm. The structure of catheter consists of two bases interconnected by three 120° positioned bellows. The actual prototype has a diameter of 4.9 mm, a length of 20 mm, and a working channel of 2 mm diameter and includes two spacer disks for each section (Figure 17). The pressure variation inside the bellows leads to the variation in their length, which then induces the bending of the distal tip. The section structure is modular, which easily builds a multi-section catheter. Bailly et al. [10] proposed another catheter, in which the distal tip of catheter is driven by three completely stress-free electrodeposited nickel bellows that are actuated by hydraulic pressure. To prevent bellows from buckling while maintaining a sufficient bending, two intermediate spacer disks are used. Ikeuchi and Ikuta [101] presented a micro active catheter with 0.3 mm diameter using a membrane micro emboss following excimer laser ablation process. A bellow at the tip is composed of a series of folded micro-chambers connected by micro-channels. The bellow can be bent on one side by hydraulic pressure within 0° to 180°. Ikuta et al. [8] proposed a two-section hydraulic pressure driven active catheter (Figure 18). They use two valves with a different range of pressures to control the two sections sharing common bellows. Each section has a bellows-shaped bending actuator and a micro valve to control the opening and closing the channels respectively, in particular, a low-pass valve (LPV) and a high-pass valve (HPV). Each section can be bent when the bellows is expanded by the normal saline which is not harmful for the patients in the event of the leak.

The catheter driven by hydraulic pressure is safer than other types of catheters and retreated quickly due to its actuation principle. Nevertheless, it shows some drawbacks. The output force is relatively small, which cannot complete some complex tasks. It needs bellows and micro-valves to control hydraulic pressure, which is bulky for catheters and needs relatively complex fabricating techniques. It may have the leakage problem and cause pressure losses (thus limit the efficiency of catheters).

2.4. Hybrid actuation catheters

Combination of any two actuation principles (Section 2.2 and Section 2.3, respectively) will result in a new
multi-section catheter, called hybrid actuation catheter, which is promising to achieve a relative compact structure with easy control. Details of the engineering hybridization principle can be found in [102,103].

Butler et al. [12] presented a robotic endoscopy for MIS, which is composed of a steerable concentric tube manipulator and an exoskeleton designed to manipulate a manual endoscope (Figure 19). The system consists of five segments. Part A is the initial portion of the scope neck with the rigid torsion. Part B is the bending section which is controlled by tendons. Part C is a short, straight and rigid section, which houses the tip optics and the steerable concentric tube (Part D). The concentric tube can be rotated and translated with its stiffness dominated by the scope. Part E is a straight mono-polar cautery wire device with its stiffness dominated by the concentric tube.

The Magellan™ robotic catheters, designed and manufactured by Hansen Medical Inc. [7], have three sections for the peripheral vascular intervention (Figure 20). Magellan™ robotic catheters have three different sizes (6 Fr, 9 Fr and 10 Fr), depending on the different diameters of the proximal part of the catheters. The middle section and proximal section are driven by a group of tendons, and the distal section is a pre-curved nitinol guide wire. The distal tip can be rotated and pre-curved by the middle section. By comparing with the robotic endoscopy by Butler et al. [12], the maneuverability of the Magellan™ robotic catheters is improved by incorporating the tendon-driven section into the pre-curved system. Chikhaoui et al. [104] proposed a steerable catheter by combining a concentric tube system and a micro-soft actuator, and the micro-soft actuator is still under development.

3. Discussion

Each type of steerable catheters has advantages and disadvantages. A comparison of different types of steerable catheters will be discussed in this section from the aspects of maneuverability, performance, size of catheter, safety, and cost in particular.

### 3.1. Manoeuvrability and performance

Manoeuvrability includes navigation and capability of being used for complex tasks. Navigation refers to the control of bending angle. The easier and faster to control the bending angle with a large amplitude, the better maneuverability. Catheters driven by tendons or hydraulic pressure can easily control the bending angle (the bending angle up to 180° can be achieved). The friction between the tendons and pathway, along with the backlash and slack of the tendons, can cause some difficulty in control of the tip of the tendon-driven catheter in terms of high accuracy. The magnetic catheter can achieve a large bending angle. The catheter driven by SMA actuators or by the conducting polymer can only achieve a relatively small bending angle compared with the foregoing three types of catheters. The maneuverability of pre-curved nitinol catheters is relatively poor, as catheters are not articulated in terms of their structure. Such a catheter usually needs to work in a semi-automatic manner (i.e., works with the human operator) – see the work of Couture and Szewczyk [72]. This semi-automated system concept is a basic reason that such catheters are usually simple in their structure and easy for manufacturing. For the capability that a catheter can be used for performing complex tasks, there are the following factors: the output force, accuracy of maneuver, DOF and workspace of catheters. If the output force is large and accurate with a sufficient number of DOFs and a sufficiently large workspace, the catheter can perform complex tasks. The tendon-driven catheters have a fast maneuver speed in comparison with the other types of catheters, and they can also generate a large force. The response of the magnetic navigation catheter system cannot reach too fast when the magnetic field is generated by permanent magnets. A delay occurs between designation of the magnetic field vector and catheter movement [61,63]. The hysteresis should be taken into account in control of this kind of catheters. The output force can be controlled by the strength of the magnetic field. The catheter driven by hydraulic pressure and a pre-curved nitinol catheter are inherently more compliant, which restricts their output force and stability. Catheters actuated by SMAs and conducting polymers rely on their materials. These materials show significant nonlinearities (i.e. hysteresis and time-varying mechanical characteristics), which is challenging to the control of such catheters. Accuracy of the maneuver is low with SMAs but the output force is high and the maneuver range is large. Catheters actuated by the conducting polymers have a fast response.
The DOFs and workspace are two basic requirements for the catheters. The magnetic catheters have a limited workspace and DOF due to the single-section structure. There is an approach to combining several catheters in operation to increase the DOFs and workspace. For instance, Simaan et al. [50] proposed a three-armed robot with three distal dexterity units (DDUs), which serve as three tendon-driven steerable catheters for MIS of the throat, in particular two for manipulation and one for suction. In the magnetic navigation catheter system, two or more catheters cannot be steered in the magnetic field simultaneously, because two or more catheter tips interfere with each other by the magnetic fields. However, the other types of catheters can avoid this problem.

3.2. Size of the catheter

For the MIS surgery, the small diameter of the catheter is one of the basic requirements. The size depends on the structure and fabricating process of catheters. The tendon-driven catheter has one or several tendons, which transmit the input force to the distal tip. The tendons must have a sufficient stiffness to overcome the friction between the tendons and channels. They cannot be made in very small diameters in terms of reliability. This drawback may be amplified with more numbers of sections. Although the hydraulic pressure driven catheters have one or several hydraulic bellows, the catheters can be made into very small ones. They can be made with multi-sections without increasing the hydraulic bellows by using different pressure valves. The steerable catheters driven by SMAs, conducting polymers share some common features, that is, the actuators are fixed at the distal tip and thin lead wires are used for motion and force transfer. The COMS interface circuit can reduce the lead wires. This kind of catheters can be made in very small size. The magnetic catheter embeds a small magnetic tip at the distal tip only. They do not contain any other wires or channels, which makes them with small diameters. The pre-curved flexible nitinol catheter is the same as the conventional catheter or micro-catheter in terms of the structure. So that, concentric tubes can be fabricated and assembled into very small diameters.

3.3. Safety

Safety is one of the important factors for medical devices. A catheter must be maneuvered without causing any complications, such as perforation. One factor that affects the safety is the stiffness of catheters. The tendon-driven catheters and SMA catheters are relatively rigid, and therefore they have a potential risk of perforation. To avoid this, the accurate information of forces between the catheter and tissues has to be available to the human operators or robots during the operation. Various force sensors are developed and

![Figure 17. The prototype of two-section catheter by Bailly et al. [10]. Reprinted with the permission from [10] © 2005 IEEE.](image1)

![Figure 18. Hydraulic pressure driven active catheter [8]. Reprinted with the permission from [8] © 2012 IEEE.](image2)

![Figure 19. Schematic diagram of robotic endoscopy developed by Butler et al. [12]. Reprinted with the permission from [12] © 2012 IEEE.](image3)
installed at the distal tips of catheters. However, the force sensor is not as reliable as the haptic feeling of physicians. Magnetic navigation catheters and conducting polymers catheters can be made by very soft and small in sizes with yet relatively rigid distal tips. These catheters as well as hydraulic pressure catheters are inherently more compliant, and they are unlikely to cause perforation. It is worth to mention that some clinic feedbacks [54,59] have shown that the magnetic navigation catheter systems (Niobe systems in particular) were conducive to reducing the complication. However, from other clinic feedbacks [2], the complications are not reduced with the Sensei X Robotic Catheter System, and the complications are even higher than manual operations. This phenomenon may be related to a relatively high rigidity of this kind of catheters.

The catheters actuated by shape memory effects which are further driven by electrical power need some attention to safety, as they may cause the over-heat problem. As opposed to such type of catheters, catheters driven by conducting polymers with small voltage (under 1V) are much safer. The fluid in the hydraulic pressure catheters is usually a kind of physiological saline, which is not harmful to patients. The magnetic navigation catheter system needs more attention, as movement of the permanent magnet may hit patients or physicians.

3.4. Cost

The cost of medical devices is an essential concern both for physicians and patients. Among all the steerable catheter systems, the most expensive one is the magnetic navigation catheter system. The cost includes the cost for both the robotic system and the operation environment. Installing a magnetic navigation catheter system may require that the catheter system be equipped with steel plates and specialized equipment to prevent the magnetic fields from interfering with other equipment. Such installations can be costly and time consuming. Other types of catheters do not have special operation environment requirements, compatible and portable to the existing operation rooms. The catheters actuated by shape memory effects, conducting polymers and hydraulic pressure require relatively complex fabrication techniques and thus may increase the cost. The tendon-driven catheters and pre-curved SMAs are inexpensive compared with the others.

4. Conclusions and future research challenges

This paper presented a review of steerable catheters in the past decade. The paper takes a view that the steerable catheter is a human-machine system, and the machine system consists of the actuator, mechanism, and end-effector, called the generalized mechanism [18]. Then, the classification of this generalized mechanism in the tissue environment was made based on the working principle of the mechanism. Specifically, the steerable catheter was classified into the following categories: (1) tendon driven, (2) magnetic field driven, (3) soft material driven, and (4) hybrid driven. To each of these categories, sub-categories were classified upon more specialized working principles. The classification of steerable catheters based on the tissue environment was also performed, specifically resulting in three categories: (i) the catheter (or flexible needle) is directly steered into the tissue, and (ii) the catheter is steered into channels in human body. After that, the classification along with discussion on the categories of steerable catheters was made upon the attributes such as: (a) navigation capability (or manipulability) and performance, (b) size (including the potential for miniaturization), (c) safety, and (d) cost.

There are several new findings generated from this review regarding the state of arts of steerable catheters along with identification of the future challenges. First, there have been many steerable catheters developed; however, there seems to be no comprehensive
evaluation system available in the literature. This situation can hinder the further advancement of the steerable catheter technology. A future effort is warranted on overcoming this shortcoming. The authors of this paper believe that a classification scheme based on function-behavior-structure (FBS) knowledge architecture [24,105] should be taken in future. A successful example of the classification of systems based on the FBS knowledge architecture can be found on pipe inspection robots (Li et al. 2015) and stick-skip actuator system [106]. Second, from the literature, there is little attention to the navigation of steerable catheters in various parts of human body. A misconception may occur such that steerable catheters are a fully automated system. In fact, the surgeon plays an important role in steering catheters. The entire system should be viewed as a human-machine (surgeon-catheter) system [19]. It is worth to mention that a promising work of Couture and Szewczyk [72] is in fact a human-device system – the rotation of the catheter requires a manual operation. Nevertheless, the operability of steerable catheters, collaborated the human physician, is not discussed in literature, which warrants a future work. Third, the control of steerable catheters, especially the control of the tip force, has less been discussed; for instance, the issue of how to get the information of the tip force is not well discussed. It may be true that this issue can be addressed by designing a sensor on the tip; however, such a steerable catheter may not be very reliable, apart from the difficulty of installing a small sensor on the tip. A future effort should be taken on the topic of the tip force control. Fourth, the design of steerable catheters in literature is not quite systematic and rational from the point of view of engineering design excellence (Fan et al. 2015); [102]. General design theories and methodologies in the engineering design community may further improve the design of the steerable catheter. A systematic and rational design process starts with the specification of design goals or functional requirements and constraints, namely in the case of the steerable catheter: (1) the size, (2) the workspace, (3) maneuverability and (4) stability. These goals may conflict with each other; in other words, the design of steerable catheters is a trade-off process. Further, each human individual is different, the steerable device may be desired to be patient-personalized. How to optimize such a trade-off process and to design adaptive catheters warrant some future attention. Design theories such as Axiomatic Design Theory (ADT) may need to be employed for the optimization of the design process, and the latest discussion of ADT is referred to (Fan et al. 2015). Finally, there seems to lack an effective analysis tool for simulating the dynamic behaviour of the steerable catheter, which is needed to optimize the design as well as to facilitate the development of an analytical model for the dynamic behaviour of the steerable catheter for the real time feedback control of the catheter. The previous studies in robotics, e.g., Bi et al. [107], Zhang et al. [108,109], can certainly be extendable to the steerable catheter.

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