A practical approach to the design and control of active endoscopes

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Abstract

Actual endoscopes and boroscopes, widely used in industry and in minimal invasive surgery, have considerable limitations, mainly due to their low number of degrees of freedom and their manual operation. Two different solutions for the electrical actuation of articulated endoscopes are presented in this paper. The technical constraints for this kind of application are very limited space for the actuators and high performance in terms of torque and angular reach. The first solution classically consists in a 2 d.o.f. structure steered by two pairs of antagonist shape memory alloy (SMA) wires. The sizing and preload determination for those actuators follow an original analytical approach. The second solution consists in a multi-d.o.f. structure actuated by thin NiTi springs mounted in an antagonist configuration and directly integrated in the structure of the endoscope. The geometry of the springs is obtained by optimization through genetic algorithms and finite elements method. Experiments show good adequacy between real behaviour and numerical model and also validate the approach.

This study is also enhanced by a control scheme specifically developed for SMA actuators in an antagonist configuration. It is based on a first order sliding mode scheme, which has the advantage of a great structural simplicity. The experimental results show that this solution can reach a good compromise between the dynamic behaviour of the actuator, its energy consumption and the structural lifetime of the endoscope.

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1. Introduction

An endoscope is a long thin tubular device for non-invasive inspection of motors or bodies interior cavities, canals, vessels, etc. inserted through a natural or surgically produced orifice. A typical outer diameter of endoscopes is 10 mm and their length varies from 70 to 180 mm. The endoscope body contains several light guides (typically 2), tool channels (biopsy grippers, snare, cytology brush) and optics or electronics for the image transmission. Endoscopes can be rigid or articulated. In the later case, they present at most 2 degrees of freedom, controlled manually by cables and levers. These manually commanded mobilities allow to bend the distal extremity of the device for route selection or changing the field of vision. Fig. 1 shows a 1 degree of freedom (d.o.f.) endoscope used in gastro-intestinal inspection.

Actual endoscopes have considerable limitations, which results in difficulties in minimal invasive observations and interventions:

1. They are unfit for the exploration of small and difficult to access areas and cavities, as for example in the head and neck surgery. This is mainly due to their lack of mobilities and their oversize diameter.

2. They are controlled manually, and should be hold continuously by the operator. In consequence, he often has to keep an uncomfortable position for a prolonged lapse of time.

3. Considering surgical applications, they are not completely safe and can cause harm by perforating contacting tissues. This is mainly due to a total lack of force feedback, as the integrated mechanisms are often cable based and thus irreversible.

It appears clearly from above that there is a consequent need for a minimal invasive exploration tool for industry or surgery which is of reduced size, easily and intuitively controllable and with a high number of mobilities.

A well-suited solution to this problem consists in realizing active endoscopes driven through shape memory alloy (SMA) integrated micro-actuators. The principle of SMA actuation, along with a preliminary approach to their design and control is presented in our previous works [1,2]. In this paper, we describe and validate an original approach of these concepts. Its benefits lie particularly in geometrical optimization of the SMA actuators and the definition of an appropriate control scheme. Two types of endoscope structures are addressed: the classical 2 d.o.f. endoscopes structure and the “snake-like” multi-d.o.f. endoscope structure.

The next section is devoted to a brief state of the art in the field of active endoscopy. Sections 3 and 4 present two new methods for
the optimal design of, respectively 2 d.o.f. and multi-d.o.f. SMA driven endoscopes. In Section 5, we propose a new approach for the tuning of general 1st order sliding mode controller applied to SMA driven endoscopes. Finally, conclusion and perspectives are drawn in Section 6.

2. State of the art

There are several solutions found in the literature for an actively controlled endoscope. Most of them address the field of minimal invasive surgery. Generally, these systems are dedicated to a particular type of medical intervention. For example, in [3], authors propose an autonomous poly-articulated endoscope for a coronary anastomosis, where the kinematic and the actuation of the device are optimized for needle insertion. An other active endoscope, especially designed for the treatment of an aortic aneurysm, is presented in [4]. It has a serial/parallel structure of pneumatic actuators for precise positioning of an endovascular stent. These systems are obviously geared towards very specific operations and therefore cannot be used for general endoscopy.

Other systems are designed for automatic progression in the body (arteries, intestines). For example, in [5], authors present a catheter with three inflatable balloons able to move itself inside a tube in peristalsis mode. In [6], a rolling stents systems mounted on the head allows the progression through the intestines. [7] presents different systems for “inchworm” locomotion alongside with in vivo experiments on pigs. These solutions do not resolve the progression and curvature control problems in case of cavity exploration where the device cannot rely on the environment.

On the contrary, several works deal with active structures, capable of maintaining their configuration without contacting the environment. Several 1 or 2 degrees of freedom devices are found in that category. Some of them are actuated through electrical motors that pull the cables in place of the operator [8,9]. But these solutions are somehow cumbersome and are not well-suited for portable instruments. The other 3 d.o.f. active endoscopes that can be found in the literature are driven using SMA wires [10,11], or helicoidal springs [12]. In this case, the SMA wires or springs are embedded into the instruments body and disposed antagonist two by two so that a pair of actuators can control the device movement in a given flexion plane. However, the maximal bending angles these devices can reach are generally significantly below the angles the existing manually controlled endoscopes reach. We show in Section 3 that this problem can be solved by a consistent choice of the SMA wires dimensions and preload.

Fig. 1. An existing 1 d.o.f. endoscope and its application to stomach wall inspection.

Fig. 2. Existing prototypes of active endoscopes.
However according to the needs stated above, and especially to allow to work in areas difficult to access, several “snake-like” devices with a great number of mobilities, each controlled independently, were also developed. Most of these systems also rely on Shape Memory Alloys for their actuation (Fig. 2). In [13], the endoscope presented has 13 mm in diameter and is comprised of five segments actuated by thin SMA solenoids. The radius of curvature of the device is interesting though the generated forces are weak. A comparable system, having 2 mm in diameter is presented in [14]. A spine-like structure is presented in [15]. It is comprised of binary actuated simple elements, using transverse SMA blades. An important number of elements is needed for a decent curvature. Finally, in [16] authors present a thin structure (0.9 mm in diameter), actuated by NiTi plate springs. This solution allows for a better integration of the structure but the produced motion is of weak amplitude. Moreover, in most of these prototypes, the internal diameter for passing tools or electrical connexions or optical fibers is generally too small. In this paper, we present an approach combining genetic algorithm and finite elements method for the design of an original distributed actuation system suited for multi-d.o.f. endoscopes actuation (Section 4).

Finally, control of SMA based active endoscopes requires a particular approach taking into account the specificities of SMA materials as well as the severe constraints encountered in endoscopy, especially in the case of medical applications:

1. SMA springs or wires that are mounted in an antagonist configuration can sometime pull again each others and generate internal efforts without producing any motion. This is the case when they are activated simultaneously or simply because they are identically pre-stressed. Unnecessary internal efforts have to be kept small in order to avoid important structural constraints and energy waste.

2. SMA based actuators have very poor efficiency. The major part of the injected energy is released as heat and contributes to temperature elevation. The energy consumption has thus to be minimized by limiting unnecessary or chaotic activation of SMA actuators.

3. In common endoscopic applications, manually driven endoscopes can bend their field of vision from 0° (frontal vision) to 90° (lateral vision) in less than 1 s. An SMA driven endoscope should exhibit a comparable bandwidth although a well known drawback of SMA based actuators is their lack of rapidity especially when their cooling relies on natural convection only.

4. The large number of joint axis to be controlled in case of “snake-like” devices implies to consider only simple control solutions, easily repeatable, with as few sensor equipment and tuning parameters as possible.

Various types of control schemes for SMA based actuators have been proposed in the literature (for a large survey of existing solutions, one should refer to [17]). They are generally of three types: (1) Linear controllers [18,19], which are simple but poorly compatible with the non-linear and hysteretic behaviour of SMA. (2) Model based controllers [20,21], which require a difficult off-line identification of parameters such as the convection coefficient, the specific heat. (3) Variable structure controllers such as on–off controllers [22] or sliding mode controllers [23,24], which combine parametric simplicity and relative insensitivity in respect to model uncertainties. In Section 5, an original efficient approach for the tuning of 1st order sliding mode controllers applied to SMA based actuation systems is proposed and then experimentally validated on a typical 2 d.o.f. active endoscope.

3. Design of a 2 d.o.f. SMA based active endoscope

3.1. System description

This section describes the structural design and dimensioning of a 2 d.o.f. active endoscope actuated by four antagonist SMA wires. This device (Fig. 3) is composed of an articulated head (made by Fort Imaging Systems Company), a 32 cm long tubular body containing the SMA wires, a stretcher for the preloading of the SMA wires and a handle.

The mobile head of the endoscope is composed of a serial arrangement of six stainless steel rings whose external diameter is 7 mm (Fig. 4). The rings are linked to each other through rotoid joints whose axes are alternatively oriented by 90°. The internal diameter of the rings is 5.6 mm. The maximal bending angle is about 100° in all directions and the minimal curvature radius is less than 20 mm.

The mobile head is classically actuated by stretching on four cables mounted in a two by two antagonist configuration (Fig. 4). Hence, the left and right (resp. up and down) cables produce bending in the horizontal (resp. vertical) plane. Stretching force is created by contracting SMA wires attached to the cables. The contraction of the SMA wires is obtained by increasing their

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Fig. 3. Two d.o.f. active endoscope.

Fig. 4. Mobile head of the 2 d.o.f. endoscope.
temperature. Power is supplied to the SMA wires by electric current and is regulated through Pulse Wide Modulation.

The present solution has the major advantage to be applicable on existing endoscopes without significant modifications on their mechanical structure and dimensions. Moreover, this solution has the particularity to save space for passing various materials (optical wires, electric connections, tool channels) through the body of the endoscope.

3.2. Actuators sizing

Fig. 5 shows the required forces to be applied on the left and right cables to bend the endoscope from $-90^\circ$ to $90^\circ$ in the horizontal plane. At least 1500 g are required to bend the endoscope to its maximal extend.

The SMA wires are made of a Nickel–Titan alloy (NiTinol™) supporting up to 500 MPa of continuous stress and whose thermo-mechanical characteristics (both martensite and austenite) are depicted in Fig. 6. The wires were chosen with a diameter of 250 µm and are able to deliver a continuous force up to 2400 g.

Fig. 7 shows the right cable displacement while the endoscope is bent both ways. Accordingly, a 5 mm forward displacement (left flexion) and 6 mm withdraw (right flexion) are required. The NiTinol™ alloy has a maximal strain of about 6%. Thus, SMA wires must be at least 185 mm long to produce the required 11 mm of total deformation. In the following, 300 mm long SMA wires are considered. This length is the maximum allowed regarding the length of the device.

3.3. Prestrain

The maximal bending angle that the device can achieve depends on the prestrain of SMA actuators. This prestrain stand for the strain that the SMA wires undergo when the mechanical structure to be actuated is in its original position and the SMA wires are not activated.

As an example, Fig. 8 shows the forces the right and left SMA wires would apply to the structure (with respect to the structure bending angle) for three different values of prestrain. These prestrain values are, respectively 1%, 4% and 7% (corresponding to points A, B and C in Fig. 9).

For a given prestrain of the right and left SMA wires, the maximal bending angle the device can achieve (when bent to the right in the horizontal plane for example), is determined by the intersection of the austenite characteristics of the activated right SMA wire and the total resistive load (Fig. 10). This total resistive load is obtained by adding the loads from the stretched left SMA wire and the bent mechanical structure (deduced from Fig. 5).

According to that, the maximal achievable bending angles for different prestrain values ranging from 3% to 7% were estimated for the endoscope of Fig. 3 and for the selected SMA wires dimensions (Fig. 11). Accordingly, the optimal initial strain is 5.0%, leading to a maximal achievable bending angle of $80^\circ$.

These results confirm the experimental maximum bending angles obtained on the setup depicted in Fig. 12. These angles are
comprised between 75° and 80° for both horizontal and vertical bending planes.

4. Design of a multi-d.o.f. SMA based active endoscope

The above described active endoscope has only two independently controllable degrees of freedom. It cannot produce complex curves in 3D which would require that each joint of the structure be actuated independently from each others. To reach this goal, the preceding actuation system made of four SMA wires should be replaced by an actuation system made of independent actuators located at each joint of the mechanical structure (see Fig. 13a). These local actuators should moreover be placed on the rings itself in order to preserve the inner space of the endoscope.

Fig. 13b shows a pair of articulated rings that has been modified accordingly. Two diametrically opposite cavities are reserved on each ring to hold the actuators. As we can see, the volume reserved for the actuators is small (about $2 \times 2 \times 3.5 \text{ mm}^3$) and curved.

There are several actual micro-actuation technologies adapted to these small spaces [25]. In our case, Shape Memory Alloys seem to be the better choice, as they present a good compromise between the weigh and the power when compared to for example...
Among the different forms of SMA actuators, the best solution for such a compact volume are plate springs (see Fig. 14). In order to adopt the curvature of the rings, it would be preferable to use a pair of springs for every housing, as shown in the rightmost part of Fig. 14. Compared to SMA wires or coil springs, this geometry offers a better compromise between the force and displacement [28,29]. Moreover, its higher surface over mass ratio contributes to improve the thermal dissipation and thus increases the bandwidth of the actuator.

4.1. Mechanical specifications

An experimentation was conducted in order to measure the magnitude of the joint stiffness of the endoscope already used in Section 3. Different loads were applied at the extremity of this endoscope and resulting displacements were measured (Fig. 15). By using inverse geometric and static models of the structure, experimental values for joint torques were obtained. These values are interpolated to obtain a nominal loading line as shown in Fig. 16.

Moreover, the minimal radius of curvature is estimated at about 50 mm, which is in accordance with the commercially available endoscopes. In the case of a 80 mm long structure, including about 10 rings (a joint in a given plane every 8 mm), this radius of curvature is obtained for a rotation of $9^\circ$ at each joint. This constraint, according to the graph of Fig. 15, gives $T_{\text{min}} = 6.4 \text{ mN m}$ for the value of minimum joint torque.

The SMA material used in the manufacture of the springs is NiTi 50–50, produced by Memory Metalle GmbH (Germany). Transition temperatures of this material were measured using Differential Scanning Calorimetry (at ESPC Paris). $A_S$ and $M_S$ are between $50^\circ C$ and $60^\circ C$ and then compatible for intra-body use. These temperatures are proven sufficiently stable after an intensive thermo-mechanical cycling. As shown in Fig. 17, the ratio between the stiffness of austenite and martensite phases (respectively $E_a$ and $E_m$) is 2.5. The ratio between the start stress of martensite reorientation for each phase (respectively $\sigma_a$ and $\sigma_m$) is three. The maximum acceptable stress for this material in alternative loading is given by the constructor as $\sigma_{\text{max}} = 550 \text{ MPa}$.

**Fig. 12.** Experimental validation.

**Fig. 13.** The mechanical structure of the poly-articulated endoscope.
4.2. Optimization of the spring geometry

In order to define the SMA springs geometric parameters which correspond to an optimal actuator structure a genetic algorithm based method was used.

Fig. 18 shows the geometric parameters of an SMA spring. These properties were chosen as the alleles characterising the populations of actuators, each actuator being constituted of four identical SMA springs in an antagonist configuration two by two. At each generation, the population of actuators is renewed by keeping the most fitting individuals and creating new ones by crossover and mutation.

The software used for this part of the work is based on GALib 2.45, a freeware library available at http://www.lancet.mit.edu/
Genetic parameters were established using the Steady State Method [30].

The evaluation of each individual (candidate actuator) from a given generation is accomplished using an external finite elements module (Fig. 19), developed with Castem 3000 software (available at http://www-cast3m.cea.fr/cast3m). In addition to the kinematics properties of the endoscope structure, this module takes into account the non-linear mechanical specifications of the NiTi for both martensite and austenite phases as shown in Fig. 17.

The fitness $F$ of each candidate is calculated by multiplying two dimensionless fitness' $F_T$ and $F_r$, describing the quality of the candidate in terms of torque ($F_T$) and internal stress $F = F_T \times F_r$.

The fitness $F_T$ is calculated as follows (see also Fig. 20):

\[
\text{If } T \leq T_{\text{min}} \text{ then } F_T = \sup \left( 0, \frac{T}{T_{\text{min}}} \right) \\
\text{Else } F_T = 1 + 0.1 \times \frac{T - T_{\text{min}}}{T_{\text{min}}}
\]

where $T$ represent the torque the candidate can produce at the desired configuration $\theta_{\text{des}} = 9^\circ$ and $T_{\text{min}}$ is the expected external load at this configuration $T_{\text{min}} = 6.4 \text{ mN m}$). The fitness $F_r$ is calculated as follows (see also Fig. 20):

\[
\text{If } \sigma < \sigma_{\text{max}} \text{ then } F_r = 1 + 0.1 \times \frac{\sigma - \sigma_{\text{max}}}{\sigma_{\text{max}}} \\
\text{Else } F_r = \sup \left( 0.1 - 10 \times \frac{\sigma - \sigma_{\text{max}}}{\sigma_{\text{max}}} \right)
\]

where $\sigma$ is the largest stress value observed in the pair of springs which is in its austenite phase at the desired configuration (i.e.: the pair of springs that produce the rotation) and $\sigma_{\text{max}}$ is the yield stress of the spring material in the corresponding phase.

This approach has the advantage to produce lower values of the overall fitness $F$ for individuals with a certain disequilibrium between $F_T$ and $F_r$ hence to privilege individual with equally balanced performances in each criteria. Moreover, $F$ will be null for any individual presenting in the desired configuration either a negative output torque or an internal stress exceeding by more than 10% the yield point $\sigma_{\text{max}}$.

Fig. 21 shows the evolution of the overall fitness $F$ plotted vs. the generation. A satisfactory convergence is reached at the 40th generation. Past this point, the best individual has a fitness greater than one, $F = 1.1$. Table 1 shows the geometric properties for this individual. The evolution of each parameter’s final value is plotted separately in Fig. 22. Each parameter is clearly inside the domain of exploration (except CA, which was bound to be positive). This behaviour shows that the retained optimal solution is independent from the choice of the limits of the domain of exploration.

Theoretical performances of an actuator fitted with optimized NiTi SMA springs are listed in Table 2. The maximum internal stress is less than $\sigma_{\text{max}}$ The joint torque at $9^\circ$ is largely superior to the minimum requested to actuate the structure. Fig. 23 shows the angle/torque relationship of the optimized actuator along with

Table 1

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Final value (mm)</th>
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<tr>
<td>LA</td>
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</tr>
<tr>
<td>HA</td>
<td>2.38</td>
</tr>
<tr>
<td>AT</td>
<td>0.22</td>
</tr>
<tr>
<td>RA</td>
<td>0.055</td>
</tr>
<tr>
<td>SO</td>
<td>0.27</td>
</tr>
<tr>
<td>CA</td>
<td>0</td>
</tr>
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</table>
4.3. Experimental validation

A prototype actuator was produced as shown in Fig. 24. This prototype is comprised of two articulated rings and four integrated SMA springs. The springs are mounted two by two on each side of the joint axis. They are positioned and insulated from the metallic structure by four bases made of peek material. Different electrical connexions complete the assembly.

From the electrical point of view, springs situated at the same side of the link are serially connected to each other using flexible copper blades, which also supply the connection to external wires. Epoxy resin, with a high shear strength is used for fixing all parts. Note that included springs do not come from the optimization presented above but were designed for a preliminary evaluation (Fig. 25). These springs were produced using laser cutting. A microscopic observation of the produced springs has shown a good surface quality (Fig. 26).

The performances of this actuator were experimentally studied. As shown in Fig. 27, the maximum rotation of the unloaded prototype is 10.5° in both directions. The setup shown in Fig. 28 was used to evaluate its load capacity. The torque measured at 0° rotation is 17 mN m. Table 3 shows a comparison between these measured performances and the results of a numerical simulation for this actuator. As we can see, the differences are relatively small: 5.5% for the torque and 16% for the rotation. Hence, the simulation tool is validated and theoretical results obtained for the optimized springs can be considered trustworthy.

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**Table 2**

<table>
<thead>
<tr>
<th>Specs</th>
<th>Value</th>
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</thead>
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<tr>
<td>Torque at 0°</td>
<td>30.5 mN m</td>
</tr>
<tr>
<td>Torque at 9°</td>
<td>10.1 mN m</td>
</tr>
<tr>
<td>Maximum stress</td>
<td>543 MPa</td>
</tr>
<tr>
<td>Rotation range (loaded)</td>
<td>11.5°</td>
</tr>
<tr>
<td>Rotation range (unloaded)</td>
<td>13.5°</td>
</tr>
</tbody>
</table>
5. Control solution for SMA based active endoscopes

5.1. Principle

Hereafter, we present a first order sliding mode controller suited for the position control of any kind of actuator constituted by two SMA elements in antagonistic configuration. The following notations were used (see also Fig. 29):

- $\theta$ real joint angle
- $\dot{\theta}$ desired joint angle
- $e = \dot{\theta} - \theta$ joint angle error
- $\dot{e} = \frac{\dot{\theta}}{C_0} (\dot{\theta} - \theta) = \dot{\theta} - \dot{\theta}$ joint angular velocity error

A sliding curve equation was also defined in the plane $\dot{e}$ vs. $e$:

$$S(e, \dot{e}) = 0 \quad \text{with} \quad S(e, \dot{e}) = \dot{e} + Ke \quad \text{and} \quad K \text{ a strictly positive real value}$$

As shown in Fig. 30, the joint angle error $e$ tends to zero as the system trajectory in the plane $\dot{e}$ vs. $e$ remains close to this sliding line [31]. Ideally, one would have:

$$S(e, \dot{e}) = \dot{e} + Ke = 0, \quad \forall t \Rightarrow e(t) \propto e^{-Kt} \Rightarrow e(t) \to 0_{t \to 0}$$

Additionally, in order to impose the system trajectory to converge to the sliding line, the following control law was implemented:

- Fig. 24. Two ring structure fitted with SMA springs.
- Fig. 25. Laser cutting produced spring.
- Fig. 26. Surface quality of the produced spring.
- Fig. 27. Experimental setup for joint rotation.
1. If $e > 0$ and $\dot{e} > 0$ then left SMA is heated (point A in Fig. 32)
2. If $e > 0$ and $\dot{e} < 0$ then
   (a) If $S(e, \dot{e}) > 0$ then left SMA is heated (point B in Fig. 32)
   (b) If $S(e, \dot{e}) < 0$ then nothing (point C in Fig. 32)
3. If $e < 0$ and $\dot{e} < 0$ then right SMA is heated (point D in Fig. 32)
4. If $e < 0$ and $\dot{e} > 0$ then
   (a) If $S(e, \dot{e}) < 0$ then right SMA is heated (point E in Fig. 32)
   (b) If $S(e, \dot{e}) > 0$ then nothing (point F in Fig. 32)

In the area containing points A and B, the joint angle error is positive thus, the left SMA actuator is activated in order to produce a positive joint torque and a joint acceleration in the same sense. This ensures that $\dot{e}$ (in case of area A) or $e$ (in case of area B) decreases (curves a and b, respectively in Fig. 33). For the same reason, the right SMA actuator is activated as the system is located in the areas of points D and E.

In the two areas corresponding to points C and F, $\dot{e}$ and $e$ have opposite signs, resulting in a decrease in magnitude of $e$. Moreover,
decreasing velocity specified by the slope $K$. In this case, it is then wiser to let the system evolve freely, without activating any SMA actuator until another area of the $\dot{\varepsilon} \times \varepsilon$ plane is reached. Avoiding any SMA activation in these areas allows to conserve energy and to minimize the stress on the structure.

Practically, trajectories in these areas can be of two types as illustrated by the curves $c_1$ and $c_2$ in Fig. 33. In the first case, the system's passivity and the loss of kinetic energy (mainly due to friction) make the velocity error $\dot{\varepsilon}$ decrease in magnitude rapidly enough to reach the sliding line before the error $\varepsilon$ is nullified. In the second case, the system enters another area before reaching the sliding line.

This control solution satisfies the last requirement mentioned in Section 2, as it has a very simple structure with only two parameters to be fixed:

1. The minimal supplied electric power necessary to activate an SMA actuator. It has to be sufficient to ensure an increase of the material (NiTi) temperature up to a complete austenite transformation. This parameter is fixed to complete the martensite to austenite transformation in a finite time in regards with the desired actuator dynamics. It is determined off-line and only depends on the actuator geometry and surrounding convection conditions.

2. The slope $K$ of the sliding line in the $\dot{\varepsilon} \times \varepsilon$ plane. In the following, an experimental method is presented to fix this parameter at a value allowing a satisfying compromise between the first three requirements: velocity, energy consumption and internal efforts.

5.2. Experimental implementation

We tested the proposed sliding mode controller on the endoscope driven by SMA wires of Section 3. In the system of Fig. 3, the structure deflection was measured using an external potentiometer and the SMA wire's traction forces were measured using four force sensors mounted in serial with the wires. The energy consumption was evaluated regarding the delivered current and the mean electrical resistance of the wires.

Several tests were performed, consisting in commanding different step displacements ranging from $\theta = \pm 15^\circ$ to $\theta = \pm 70^\circ$ and for seven different values of the sliding line slope, from $K = 0.8 \, \text{s}^{-1}$ to $K = 100 \, \text{s}^{-1}$ (Fig. 31).

5.3. Experimental results

The smallest value, $K = 0.8 \, \text{s}^{-1}$ always lead to a very slow trajectory during which the sliding line was crossed very often (see Fig. 34 corresponding to $\theta = 70^\circ$).

On the contrary, for $K = 100 \, \text{s}^{-1}$ the trajectory in the $\varepsilon \times \dot{\varepsilon}$ plane was a spiral which converged rapidly to the zero point (Fig. 35). This trajectory shown large oscillations around the desired position ($\varepsilon = 0$). This result is consistent with the fact that for very large values of $K$, the C and F neutral areas in the $\dot{\varepsilon} \times \varepsilon$ plane nearly vanish, leading to an elementary two states controller and imposing the actuator to be continuously activated with respect to the sign of the position error $\varepsilon$.

The result for an intermediate value $K = 4 \, \text{s}^{-1}$ is shown in Fig. 36. In this case, the convergence toward the zero point was more direct and the obtained trajectory in the $\dot{\varepsilon} \times \varepsilon$ plane is interestingly close to the fixed sliding line.

Note that for any value of $K$, a limit cycle around the desired angular position appeared. However, its magnitude remained always below $1^\circ$. It is found that this limit cycle can be almost completely suppressed by adding a dead zone to the proposed control scheme, like depicted in Fig. 37.

The influence of the parameter $K$ is explored through the measurements presented in Fig. 38. The evolution of rising times, falling times, the energy consumption and maximal internal efforts...
with respect to the parameter $K$ on four different controlled steps of displacement are shown.

For $K < 4 \text{s}^{-1}$, the rising times increase significantly and rapidly exceed a delay of 1s which constitutes a reference value as stated in Section 2. A similar threshold at $K = 5 \text{s}^{-1}$ exists for the falling times (Fig. 38b). Besides, on Fig. 38c and d it can be noticed that the energy consumption and internal efforts increase monotonously with $K$ but remain almost constant beyond $K = 20 \text{s}^{-1}$ and $K = 10 \text{s}^{-1}$, respectively.

On one hand, the results on the settling times indicate that the slope $K$ should be fixed at least equal to $5 \text{s}^{-1}$ to ensure a reasonable bandwidth. On the other hand, the slope $K$ should be chosen as low as possible in order to minimize the energy consumption.

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**Fig. 35.** Experimental results: $K = 100 \text{s}^{-1}$.

**Fig. 36.** Experimental results: $K = 4 \text{s}^{-1}$.

**Fig. 37.** SMA activation strategy with a dead zone.

**Fig. 38.** Influence of parameter $K$. 

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and the induced stress. A compromise choice for \( K \), in the particular case of this SMA wire driven endoscope, is obviously \( K = 5 \, \text{s}^{-1} \).

6. Conclusion

As presented earlier in this paper, solutions for active endoscopy in the literature are numerous and of various types but always suffer performance limitations. Hence, in this paper we presented two original approaches for the design of integrated actuation systems for a 2 d.o.f. endoscope on one hand and for a poly-articulated endoscope on the other hand. In each case, we detailed the method for designing the actuation system in an optimal way.

Particularly for a poly-articulated endoscope, the design method makes use of finite elements and genetic algorithms. This approach takes into account desired mechanical performances and volumetric constraints. A prototype actuator was then constructed and experimented. The experimental results are in good accordance with the results obtained by simulation, in terms of maximum joint rotation and torque.

Additionally, a control scheme was developed specifically for this kind of application based on Shape Memory Alloys. The proposed solution is a first order sliding mode type variable structure controller. The main advantage of this solution comes from its simplicity and intrinsic insensitivity to modelling inaccuracy. This control scheme was experimentally validated on a real 2 d.o.f. endoscope driven by SMA wires. It has been shown that a satisfying compromise between speed, low energy consumption and low internal stress on the structure could be reached by appropriately choosing the sliding line in the phase plane.

A prototype with optimized SMA springs is actually in the works. In addition to the actuation system, it will also integrate a Hall effect linear position sensor. Moreover, a molded plastic polymer model of the mechanical structure will be realized in order to facilitate the electrical integration and lower cost production of the proposed actuator. In long term, a full endoscope with 12 actuated rings will be produced in order to conduct in vivo experiments.

References


