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► **To cite this version:**

Arif Badrou, Nicolas Tardif, Philippe Chaudet, Nathan Lescanne, Jérôme Szewczyk, et al.. Simulation of multi-curve active catheterization for endovascular navigation to complex targets. *Journal of Biomechanics*, 2022, 140, pp.111147. 10.1016/j.jbiomech.2022.111147 . hal-03755618

HAL Id: hal-03755618

<https://hal.science/hal-03755618>

Submitted on 22 Aug 2022

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Simulation of multi-curve active catheterization for endovascular navigation to complex targets

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Abstract

The recent development of endovascular therapies has been accompanied by increasingly accurate navigation simulations to assist surgeons in decision making processes or to produce training tools. However, they have been focused mostly on targets within the aortic vasculature. In order to reach complex targets such as cerebral arteries by endovascular navigation, an active guidewire made of a Shape Memory Alloy (SMA) was recently proposed. The active part becomes deformed by the Joule effect and this deformation induces a bending of the guidewire. This setup is particularly suited for facilitating the access to Supra-Aortic Trunks (SATs) and, in our case, especially the left carotid artery. A complete characterization of the endovascular active

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navigation was conducted. In this framework, a test bench was developed to obtain an order of magnitude of the velocities applied on the guidewire as well as on the passive catheter going along with it in endovascular navigation. A numerical model was developed and validated in the case of navigation in a complex phantom aorta. We succeeded in representing crucial phenomena observed experimentally: snapping, active curvatures, interactions between the tools. In the last part of this study, it was demonstrated that adapting the guidewire design made it possible to hook the left carotid on three complex aortas.

Keywords: Aorta, Endovascular navigation, Shape Memory Alloy, Catheter, Modeling

1. Introduction

There has been a development of endovascular treatments in recent years. The introduction of minimally invasive surgery has led to reduce mortality and morbidity rates compared with open procedures (Verhage et al. (2009)). The development of such techniques has been accompanied by more and more numerical simulations for decision support (Gindre et al. (2016); Perrin et al. (2014); Auricchio et al. (2013) or as training tools (Westwood (2005); Nesbitt et al. (2016); Alderliesten et al. (2004); See et al. (2016)). Among them, few are related to the navigation of tools, i.e., guidewires and catheters, in the aorta to reach the target to treat. In (Mouktadiri et al. (2013); Gindre et al. (2016); Mohammadi et al. (2018); Kaladji et al. (2013); Roy et al. (2014)) models were developed to predict arterial deformations during surgical tools navigation in the framework of aortic aneurysm. In Menut (2017) the simu-

14 lation of navigation tools was adapted to the thoracic aorta. Vy et al. (2018)
15 developed a patient-specific model to predict guidewire deformation on the
16 first stage of a transcatheter aortic valve implantation.

17 These works involve passive tools where their curvature only depends
18 on their interactions with the aortic wall. The use of such passive devices
19 is adequate in most cases, but in 20% of endovascular therapies, the areas
20 to treat are so-called *complex* due to large tortuosities, e.g., at the Supra-
21 Aortic Trunks (SATs) level, and the navigation may fail (Madhwal et al.
22 (2008); Macdonald et al. (2009); Lam et al. (2007)).

23 Recently, in order to reach complex targets, active devices have been
24 developed. These steerable devices can be made of Shape Memory Alloys
25 (SMAs) (Wayman and Duerig (1990); Ali et al. (2020); Haga et al. (1998)),
26 but other magnetic or electrostrictive materials are also used to curve the
27 guidewires (Ganet et al. (2015); Hwang et al. (2020)).

28 An active guidewire composed of a SMA was designed to facilitate the
29 access to complex areas including SAT in an interventional neuroradiology
30 context (Couture and Szewczyk (2017); Szewczyk et al. (2011); Szewczyk
31 (2011)). It was made of a long steel shaft and a blade-shaped distal tip on
32 which a Nitinol wire was attached. The combination of wire and blade is
33 called an active part. By applying an electric current, the wire heats up by
34 the Joule effect. The Shape Memory Effect (SME) inherent to SMAs thus
35 causes the wire to shrink (Lagoudas (2008); Maynadier et al. (2011)) forcing
36 the distal tip to bend. This controlled curvature facilitates the navigation
37 through tortuous endovascular paths.

38 The insertion of the guidewire is followed by the insertion of a Distal Ac-

39 cess Catheter (DAC) (Spiotta et al. (2011)) that generally exhibits a rigidity
40 gradient along its length to ease its navigation (Gindre et al. (2016); Mouk-
41 tadiri et al. (2013)). The success of navigation is driven by the mechanical
42 behavior of the inserted tools and their co-manipulation.

43 The aim of the present study was to develop a numerical simulation of
44 the navigation of the active guidewire and DAC towards complex targets.
45 Our success criterion is the ability to *catheterize* the left carotid. The model
46 was confronted with phantom experiments on three complex patient-specific
47 cases. Based on existing studies on endovascular navigation, the main con-
48 tributions of this work concerned: (i) active navigation to complex targets
49 leading to original navigation scenarios, including the simulation of snapping,
50 and (ii) measurements of clinical gestures thanks to an original bench test.
51 Herein, we first detail the navigation features observed and measured on a
52 dedicated set up. Then, the numerical model of this navigation is described.
53 Finally, results are presented and discussed.

54 **2. Characterization of the endovascular navigation on phantom** 55 **aorta**

56 Navigation towards complex targets is an elaborated combination of ges-
57 tures and co-manipulations of tools. To get a better insight in this procedure
58 and build the navigation model, this section presents observations and mea-
59 surements of navigation features performed on a test bench.

60 *2.1. The test bench*

61 We aimed at simulating the navigation of the devices in the same condi-
62 tions as those on the test bench at the French company BaseCamp Vascular

63 (BCV) that developed the active guidewire. The test bench consisted of a
64 module with two boxes: one containing the aortic arch, the other comprising
65 a cylinder representing the descending aorta (see Fig.1). Any phantom of
66 aortic arch can be mounted in the first box.

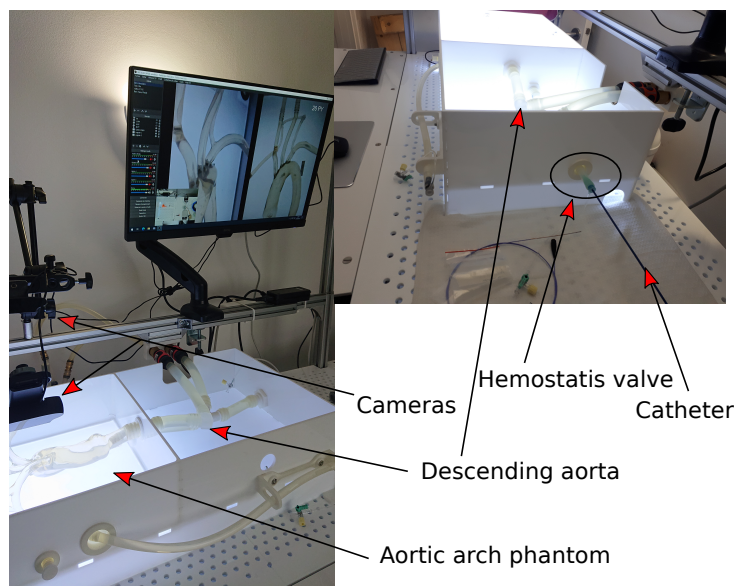


Figure 1: Overview of the phantom aorta test bench used at BCV.

67 The vascular structure was immersed in a mixture of washing liquid and
68 water. A flow of 250L/h was imposed inside the aorta to represent the
69 blood stream, which is rather low and constant compared to the pulsed flow
70 occurring in reality (Hashimoto and Ito (2010); Sochi (2013)).

71 *2.2. Active navigation and snapping effect*

72 The navigation goal was to enter the aortic arch through the descending
73 aorta (assuming femoral access) and to hook the left carotid artery which
74 leads to the brain. The aforementioned active guidewire was used with the

75 combination of the Navien A+ intracranial catheter and the introducer Neu-
76 ron MAX. Fig.2 displays the guidewire with two activated moduli showing
77 the S-curvature used to facilitate the navigation through aortas.

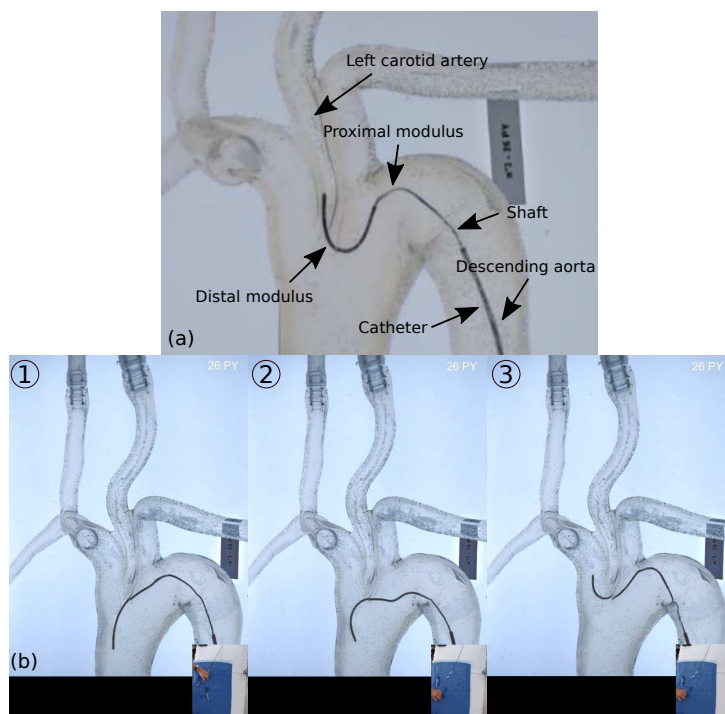


Figure 2: Focus on the activation of the guidewire and illustration of the snapping effect: (a) activation of a so-called double-deck guidewire: two active parts are visible in the figure, the wires of the active parts are arranged on both sides of the blade so that the double activation tends to snake the guide, (b) illustration of the snapping effect: 1) The guidewire is in contact with the aorta distally 2) Both moduli are activated and a rotation is imposed at the shaft 3) The snapping occurs and the distal part of the guidewire points to the opposite side.

78 Active navigation involves different movements including:

- 79 1. Pushing, pulling and rotating the guidewire;
- 80 2. Activating the distal active flexions;

81 3. Pushing the DAC to reach the target area.

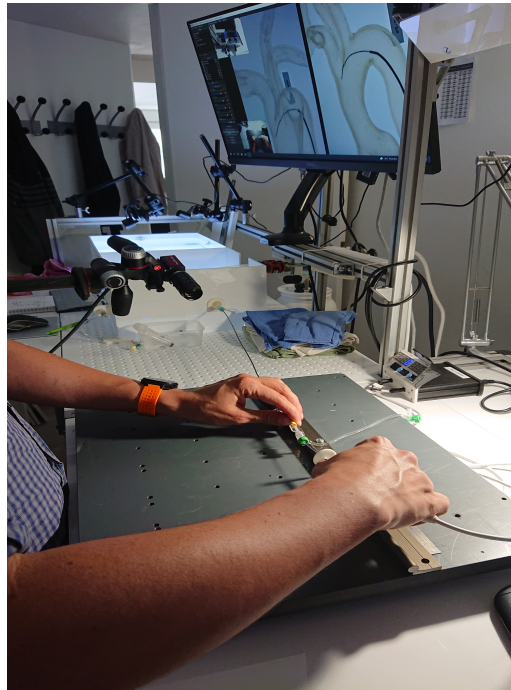
82 During navigation, clinicians often use a buckling phenomenon called
83 *snapping*, which can be described as follows: when the distal part of the
84 guidewire is in contact with the aorta, the activation of both the distal and
85 the proximal active parts, accompanied by a rotation gesture operated by
86 the handler, leads to a sudden rotation of the entire guidewire around its
87 own axis. Thus, the hook formed by the guidewire changes direction and
88 points in the opposite one, as illustrated in Fig.2. This technique, relying
89 on a S-shaped active double curve at the distal tip, appears to be an effi-
90 cient maneuver to stabilize the device within the aortic arch and point into
91 a selected SAT ostium.

92 2.3. Gestures by the clinician

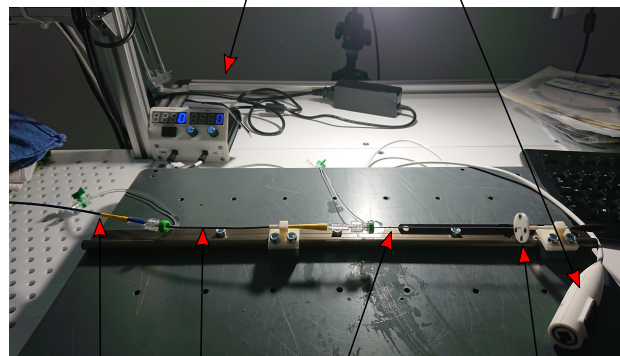
93 As explained, in the case of navigation with active devices towards com-
94 plex targets, clinicians perform many movements including rotations in ad-
95 dition to distal tip actuation. To correctly simulate the insertion of surgical
96 tools, we aimed at determining an order of magnitude of the velocities applied
97 to the devices.

98 These velocities were established thanks to a set-up added to the existing
99 test bench presented in Fig.3.

100 It consisted of two bearings sliding on a 500 mm rail. One of them
101 was attached to the catheter, and the other to the proximal part of the
102 guidewire. Targets were stuck to the bearings and a camera was able to track
103 the movements on both the catheter and the guidewire. For the rotation of
104 the guidewire, a stained token was attached to the shaft with adhesive tap.



Current generator for distal and proximal part
Remote control to activate the guidewire



Introducer
Catheter
Active guidewire
Token for rotations

Figure 3: Additional set-up to measure clinician movements during navigation: instrumentation of the BCV bench.

105 An other camera tracked the rotations. The frequency of the two cameras
 106 was 10Hz. Two additional cameras recorded the navigation through the
 107 aorta. As the activation of the guidewire consisted of applying a current, a
 108 last camera focused on an LCD screen displaying the electric current injected
 109 in the proximal and the distal moduli as well as the hands of the clinicians.

110 Several navigations with the goal of reaching the left carotid artery were
 111 filmed. Sequences were sorted and we computed an average velocity for the
 112 various phases as presented in Tab.1. All the measurements were performed
 113 on the same phantom (named PY aorta for the anatomy of the patient PY,
 114 see Fig.7).

Table 1: Mean velocities of the clinician gestures during active navigation.

Movement	Speed values (mm/s and °/s)	Description (see Fig.5)
Pushing guidewire	21.4 ± 5.1	PushG
Pushing catheter	-13.5 ± 4.2	PushC
Pulling guidewire	12.3 ± 4.6	PullG
Rotating guidewire	210.9 ± 80.4	RotG

115 3. Methods

116 Once the main features of navigation as performed by surgeons were char-
 117 acterized, we built a numerical model based on the Finite Element Method
 118 (FEM) to simulate this navigation and validate it against observations on
 119 the test bench.

120 3.1. Tools modeling

121 *Geometry.* The model includes the active guidewire, a Navien catheter and
 122 a Neuron introducer. Fig.4 describes their dimensions.

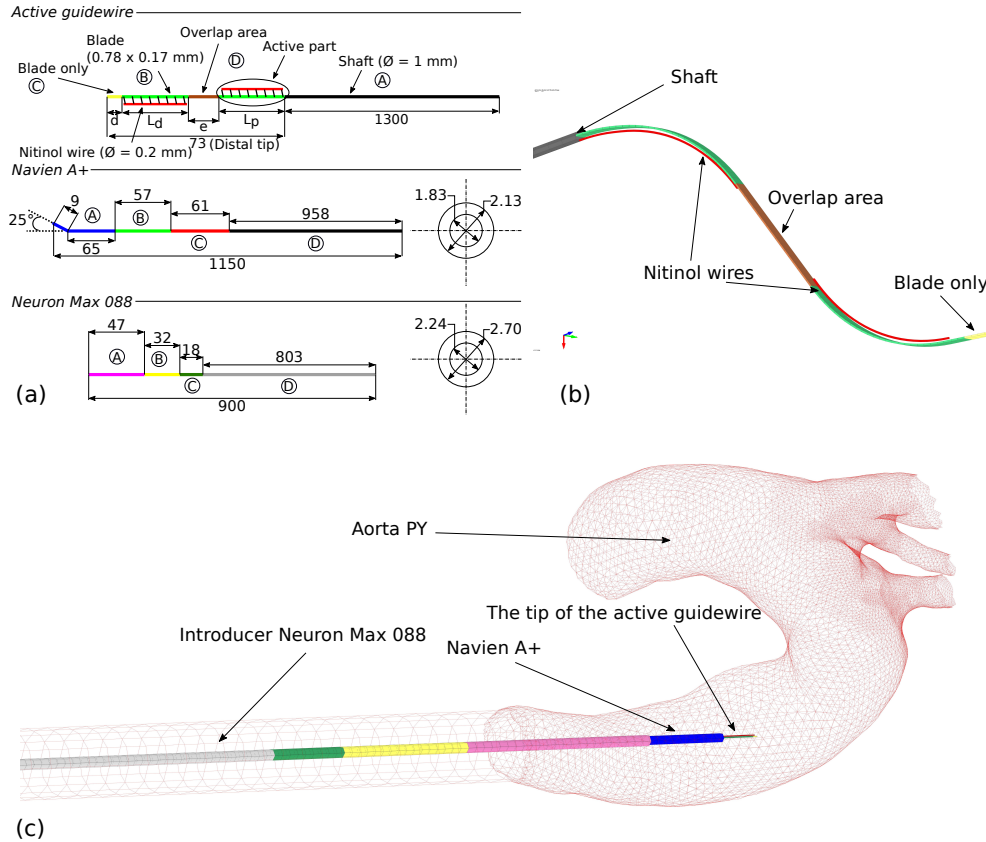


Figure 4: (a) Devices dimensions of the model: active guidewire and catheters used in our case. For each tool, the colors describe various portions having their different mechanical properties. Each portion is identified by a letter (from A to D) which refers to Tab.2 presenting the properties of the devices. (b) Visualization of the guidewire distal tip when activated: when the Nitinol wires are heated, they shrink making the blade (green parts) bend. (c) Assembly of the different tools (introducer + catheter + guidewire) in the PY aorta for the endovascular navigation.

123 The lengths d , L_p , L_d and the distance e between the two active parts
124 could be changed in the guidewire design. The total length of the distal part
125 was 73 mm. The distal tip of the guidewire included active parts where the
126 Nitinol wires were connected to the blade by rigid links. The gap between
127 the blade and the Nitinol wires was set to 0.285 mm. The stiffness of the
128 blade varied depending on the area. The three different areas are highlighted
129 in Fig.4. When the Nitinol wires are present on both sides of the blade (in
130 the so-called overlap area), the area is considered stiffer than the other parts
131 and inactive. The mechanical properties of the different portions of the blade
132 were determined in Badrou et al. (2022). The overlap area was characterized
133 using 3-point bending tests. The catheters are made of several portions of
134 varying stiffnesses that have been characterized in a previous study through
135 3-point bending tests (Badrou et al. (2022)). All the mechanical properties
136 of the tools are summarized in Tab.2.

137 The full length of the different tools was not modelled. Indeed, due to a
138 narrow entrance of the descending aorta visible in Fig.1, we assumed in the
139 case of simulating the experimental navigation, that the length outside the
140 box had no effect on the navigation itself. A sufficient length was adjusted
141 to allow the navigation to the left carotid.

142 *Mesh.* The FEM was chosen to simulate the navigation. We used Hughes-Liu
143 beam elements in Ls-Dyna (Ansys / LST, CA., USA, 1976) for the guidewire.
144 The wire and the blade (distal tip) were meshed with a 1-mm beam. Beam
145 elements of 2 mm were chosen for the shaft. The mesh size for the tools
146 and the aorta was determined based on a convergence analysis on aorta PY,
147 see Supplementary Material. One integration point was used for the Nitinol

148 wires in order for them to act as bar elements and the overall rigidity of
149 active parts was borne by the blade in green.

150 The catheters and the introducer were meshed with fully integrated shell
151 elements to avoid Hourglass modes. Belytschko-Tsay shells which are of
152 the type Reissner-Midlin were preferred for the introducer. The number
153 of integration points was fixed to three which was clearly enough for the
154 thickness of the elements.

Material law. Because of the complex composition of the catheters, some parts (mainly the distal one) exhibited a viscoelastic behaviour modeled by a generalized Maxwell formula with one element (Herrmann and Peterson (1968)):

$$G(t) = G_{\infty} + (G_0 - G_{\infty})e^{-\beta t}$$

155

156 Here, G_{∞} and G_0 were respectively the long and short time shear modu-
157 lus, and the parameter β was a constant expressed per unit of time.

158 The other parts were assumed to be elastic. Tab.2 lists the material
159 parameters for the Navien catheter, the introducer and the guidewire com-
160 ponents depending on their elastic or viscoelastic behavior.

Table 2: Mechanical properties along the surgical tools: areas A to D are related to those in Fig.4.

	A	B	C	D
Active guidewire	$E = 8047.8 \text{ MPa}$ $\nu = 0.3$	$E = 71130.0 \text{ MPa}$ $\nu = 0.3$	$E = 52720.0 \text{ MPa}$ $\nu = 0.3$	$E = 175000.0 \text{ MPa}$ $\nu = 0.3$
Navien	$\beta = 0.22 \text{ s}^{-1}$ $G_0 = 62.0 \text{ MPa}$ $G_\infty = 15.9 \text{ MPa}$ $B = 82.9 \text{ MPa}$	$\beta = 0.16 \text{ s}^{-1}$ $G_0 = 94.1 \text{ MPa}$ $G_\infty = 26.6 \text{ MPa}$ $B = 152.3 \text{ MPa}$	$\beta = 0.18 \text{ s}^{-1}$ $G_0 = 448.1 \text{ MPa}$ $G_\infty = 152.2 \text{ MPa}$ $B = 694.6 \text{ MPa}$	$E = 1519 \text{ MPa}$
Neuron	$\beta = 0.26 \text{ s}^{-1}$ $G_0 = 74.0 \text{ MPa}$ $G_\infty = 17.0 \text{ MPa}$ $B = 122.1 \text{ MPa}$	$E = 250 \text{ MPa}$	$E = 180 \text{ MPa}$	$E = 1701 \text{ MPa}$

161 As discussed previously, the guidewire was an active steerable device,
162 and its curvature could be controlled thanks to the SME. To model this
163 particular behavior, we implemented a 1D law described by Tanaka et al.
164 (1986) in Ls-Dyna, following studies previously conducted by Ianucci et al.
165 (2017). To represent the actuation of the Nitinol wires, we implemented
166 a 1D law described by Tanaka et al. (1986) in Ls-Dyna, following studies
167 previously conducted by Ianucci et al. (2017). The description of the law
168 is given in Appendix. We used this law in a simplified way: we were inter-
169 ested in the guidewire curvature (corresponding to the maximum recoverable
170 strain of the wire ϵ_L) as well as activation / deactivation times (t_a and t_d
171 respectively). Therefore, we identified parameters ϵ_L , t_a and t_d using the
172 experiments presented in Badrou et al. (2022). Then, in the simulation, we
173 imposed to the Nitinol wires a temperature load T as a linear function of

174 time between 37°C and 65°C. The time intervals from one temperature to
175 the other were defined to respect t_a and t_d . The wire model provided a
176 longitudinal strain leading to the device bending, until reaching ϵ_L . This
177 way of using Tanaka’s law is equivalent to drive the wire strain by displace-
178 ment; however, we used Tanaka’s law so that our model is easily adaptable
179 to a full description of SME. From Badrou et al. (2022), we determined that
180 $\epsilon_L = 0.0347$ and $t_a = t_d = 1.7\text{s}$ for the guidewire design tested in this paper.
181 We indeed observed experimentally that, due to the blade elasticity which
182 brings the wire back to its initial configuration at deactivation, the activation
183 and deactivation times are very similar. ϵ_L is a design parameter that can
184 be tuned to adapt to the considered patient as we will see in section 4.2.

185 *3.2. Aortic phantoms modeling*

186 *Geometry.* Three phantoms were available for the test bench, based on pa-
187 tient cases (patients PY, LP and FM). The same geometries were used in the
188 model. They were classified by surgeons in our group as ”complex” because
189 the insertion point in order to reach the left carotid was difficult to access
190 (see Fig.7). These are typically complex anatomical cases in which naviga-
191 tion with passive tools would most probably fail. Indeed, for the aorta PY,
192 the anchor point for hooking the carotid artery is low compared to the dome
193 of the aortic arch. The aorta FM has a bovine arch: the innominate artery
194 shares its origin with the left carotid. The origin of the carotid in the aorta
195 LP is narrow and is not aligned with the insertion points of neighbouring
196 arteries.

197 *Mesh.* The different geometries were meshed based on a mesh sensitivity
198 analysis. We chose to use Belytschko-Tsay triangular shell elements. Since
199 we were interested in hooking the left carotid we decided to finely mesh
200 the areas near the SAT to catch the tortuosities. The chosen mesh size
201 was 0.5 mm on these areas. A transition area with a mesh size of 1 mm
202 was applied and the rest of the aorta was meshed with 2-mm elements for
203 contact management. The thickness of the aortas varied from 2 to 4 mm on
204 the BCV test bench. We decided to fix an average thickness for the overall
205 aortas of 3 mm.

206 *Material law.* The aortas were assumed to be rigid.

207 3.3. Boundary conditions and numerical parameters

208 Fig.5 depicts the boundary conditions at the entrance of the descending
209 aorta model for the surgical tools.

210 Nodes of the catheter and introducer in area 5 were fixed in displace-
211 ments in X and Y and in rotation around the X and Y axis. These particular
212 conditions were due to the narrow entrance (see Fig.1) that constrained the
213 devices. The proximal end of the catheter (nodes of areas 3) was constrained
214 in all directions when the guidewire was moving. Nodes of area 4 were fixed.
215 Indeed these ends were assumed to be held by the handler. The end of the
216 guidewire (area 1) was presumed to move only along the Z axis and rotate
217 around it. A rotation speed was applied on nodes of area 2 and a linear
218 velocity command on those of area 1, their amplitude is given in Tab.1. Re-
219 garding the contact between the materials, frictional forces were assumed to
220 be present both between the tools themselves and between the tools and the

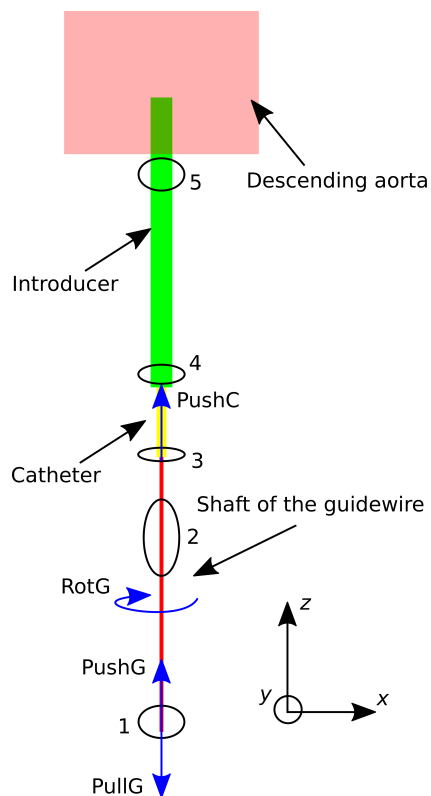


Figure 5: Boundary conditions applied on the devices during endovascular navigation. PushC, PushG, PullG and RotG are related to Tab.1 defining the different movements and their value.

221 aorta. We used a penalty method to represent the contact and a friction co-
 222 efficient of 0.2 was set by default except between the guidewire and the aorta
 223 and between the catheter and the introducer where the friction coefficient
 224 was set to 0.3 and 0.1, respectively. Mass scaling was employed with a given
 225 time step of $5 \cdot 10^{-6}$ s. To stabilize the active guidewire during the navigation,
 226 Rayleigh damping was used by playing of the mass proportional damping
 227 coefficient. We see in Fig.7 that damping energy as kinetic energy are low
 228 compared to internal energy which confirms the validity of the damping and

229 mass scaling levels adopted for this study.

230 4. Results

231 4.1. Validation

232 In order to validate the simulation of the endovascular navigation through
233 phantom aortas, we simulated the insertion of the surgical tools into the aorta
234 PY with regard to the experimental navigation. The inputs followed the
235 path-based gestures of the clinicians. Validation was conducted on passive
236 sequences (back and forth movements were applied to the guidewire without
237 any activation), and on active sequences in which the device was actively
238 bent. Attention was paid to several actions and phenomena occurring in
239 reality:

- 240 • Passive navigation: by pushing, pulling or rotating the devices without
241 any activation;
- 242 • Activation of the active parts: the distal tip of the guidewire curves in
243 a multi-curve way;
- 244 • Snapping.

245 We tried to correctly represent these phenomena in simulation. Fig.6 com-
246 pares the guidewire position between simulation and real navigation on the
247 test bench during both passive and active sequences. For active sequences,
248 we also followed the tip of the guidewire and recorded its coordinates with the
249 camera plane of the test bench. The results show good agreement between
250 the two datasets, a slight position deviation can be seen on the guidewire
251 just after snapping but it did not affect the final position.

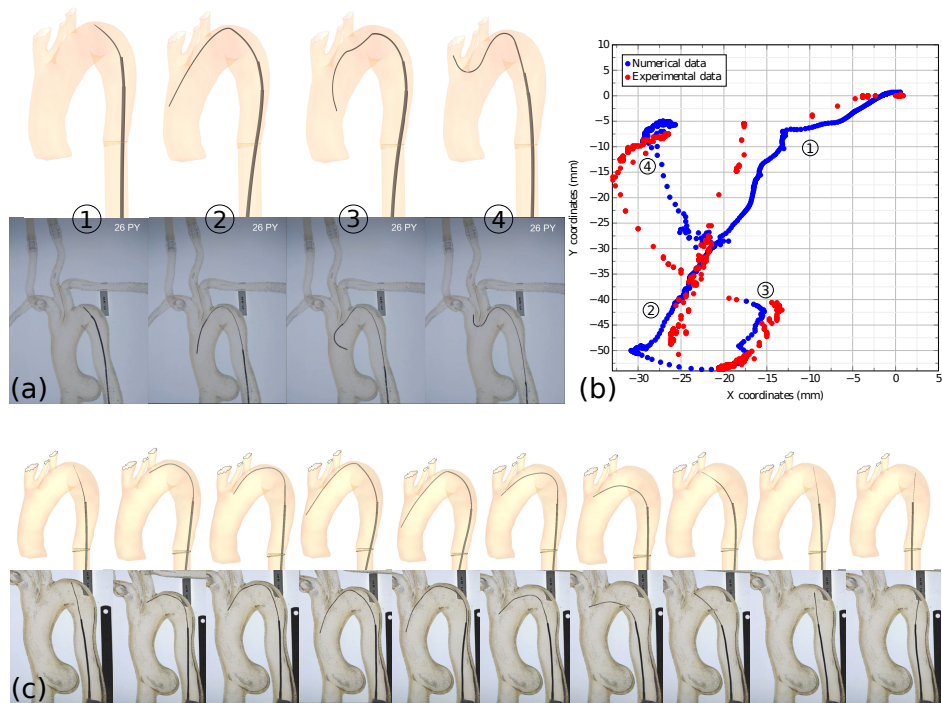


Figure 6: (a) Validation of the active navigation into the phantom aorta PY at different stages. (b) Guidewire tip coordinates during endovascular navigation in the camera plane: comparison between numerical (top) and experimental (bottom) data. The guidewire navigation begins at the top of the aortic arch in ① until it contacts the descending aorta in phase ②. In step ③ the guidewire is activated and "snaps" to reach the SATs in ④. (c) Passive navigation is compared between numerical (top) and experimental (bottom) data in the PY aorta.

252 The model parameterization made it possible to reproduce the various
 253 phenomena observed experimentally: snapping, active curvatures, interac-
 254 tions between the tools. Now that navigation within a phantom could be
 255 reduced, we wanted to make use of this simulation as a design assistance
 256 tool to find a combination of guidewire design parameters that would make
 257 it possible to hook the left carotid in different anatomies.

258 *4.2. Application: simulation of endovascular navigation in two additional*
259 *complex aortas*

260 For the PY aorta, "clinical" gestures were measured and combinations
261 of sequences could be reproduced. For the two other aortas, we relied on
262 the previously validated simulator to find appropriate navigation parameters
263 and guidewire design specifications to hook the left carotid.

264 The parameters on which we acted were the lengths L_p , L_d , e and the
265 maximum recoverable strain ϵ_L . The latter parameter concerned the ability
266 of the Nitinol wire to contract exhibiting a change in the current intensity or
267 in the grade of SMA. The values used to achieve hooking were determined
268 by trial and error.

269 For the navigation into the PY and FM aorta, the boundary conditions
270 were quite similar: the guidewire was pushed until the distal tip was in the
271 ascending aorta, the parts were activated and the guidewire was snapped.
272 The device was then pulled in order to hook the left carotid. Regarding
273 the LP aorta, the active parts did not overlap (blade only) and the distance
274 between them, e , was long. No snapping was used in this case. The proxi-
275 mal modulus helped to curve the distal part of the guidewire and the distal
276 modulus made it possible to point to the entrance of the carotid.

277 Fig.7 shows that we were able to hook the left carotid for all cases despite
278 their complexity, demonstrating the potential of the simulation to provide
279 guidance into navigation gestures and device design.

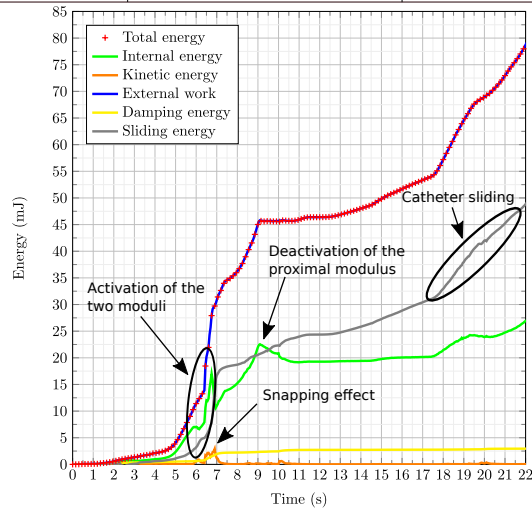
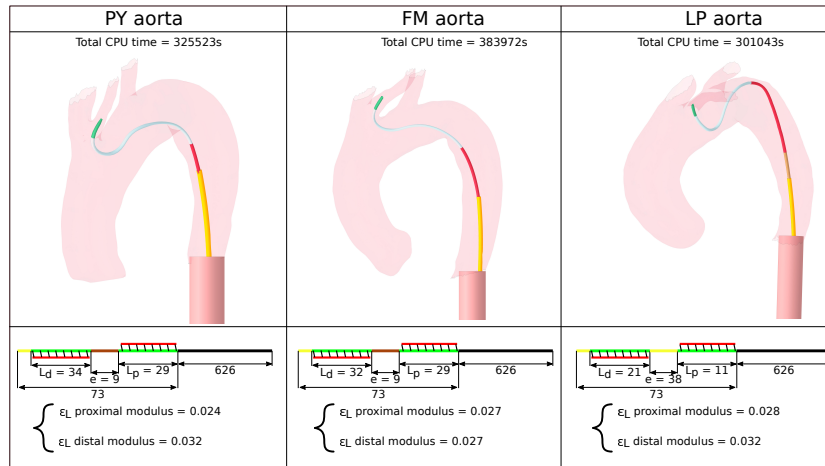


Figure 7: Hooking of three complex aortas with details of the guidewire configuration used in each case. ϵ_L is the maximal recoverable strain. CPU times using six Xeon 2.30 GHz cores are indicated. Graph of energies are plotted for the navigation into the PY aorta: the active guidewire first crosses the aortic arch until it is located nearby the left carotid artery. The two moduli are activated and the snapping effect occurs (kinetic energy peak). The active guidewire is pulled and the proximal modulus is deactivated. The left carotid is hooked and the Navien catheter slides on the guidewire to reach the carotid. The navigation sequences were similar for the other aortas.

280 **5. Discussion**

281 *5.1. Main results*

282 This article describes the development of a numerical model of endovas-
283 cular navigation involving an active guidewire and catheters to cerebral tar-
284 gets. To the best of our knowledge, this is the first simulation combining
285 an active steerable device, a catheter and their interactions with the vascu-
286 lar environment. In order to create the model, a complete characterization
287 of the navigation was conducted. This work enabled (i) a better overview
288 of the effects occurring during active navigation, such as snapping, and (ii)
289 measurements of the velocities imposed by the clinicians to the surgical tools
290 thanks to an original set-up.

291 Indeed, during endovascular navigation many movements are imposed on
292 the tools. It is thus crucial to have an order of magnitude of the veloci-
293 ties for pulling, pushing and rotating the tools as well as for the sequences.
294 Thus, an additional set-up was added to the phantom aorta test bench and
295 several navigations from experienced handlers were recorded. In particular,
296 experimental analysis was carried out of the snapping phenomenon. Through
297 observation and simulation we investigated the reasons of this buckling in ro-
298 tation and we found that snapping occurs when the proximal active modulus
299 is directly connected to the shaft and activated after the guidewire contacts
300 the aorta distally.

301 Regarding the experimental point of view, we performed a 3-point bend-
302 ing test on a portion of the phantom aorta and found a Young modulus of 4.3
303 MPa. The idea was to confirm the hypothesis of a rigid aorta in our model.
304 Therefore, we compared the final position of the guidewire's distal part when

305 navigating into the PY aorta between a rigid vs deformable aorta (phantom).
306 The difference in position was computed using the Modified Hausdorff Dis-
307 tance (MHD) (Dubuisson and Jain (1994)). A value of MHD less than 1 mm
308 was obtained and the assumption of a rigid aorta was therefore considered
309 acceptable. Of course, this assumption can be questioned when it comes to
310 real patient data and a probably less stiff aortic wall, but this present work
311 only focused on simulating a phantom aorta test bench.

312 We could thus create the endovascular navigation model based on this
313 result and the experimental characterization. The model was validated with
314 regard to the experimental navigation into a complex aorta and particular at-
315 tention was paid to the representation of experimentally observed phenomena
316 such as the activation of the guidewire, passive navigation or the snapping
317 effect.

318 Once our model was set-up, we demonstrated that hooking the left carotid
319 was partially driven by the guidewire design. Among the various parame-
320 ters, we acted on five parameters specific to the guidewire. Thanks to differ-
321 ent configurations we succeeded in hooking three highly complex anatomies.
322 Among these design parameters, ϵ_L is probably the most challenging as it
323 reflects both material (Nitinol grade) and activation (current intensity) fea-
324 tures. It is also affected by fatigue. Hence, preliminary tests would probably
325 be necessary to reach the targeted value of ϵ_L during design.

326 *5.2. Study limits*

327 The blood flow in the aortic arch is pulsed and considered to be turbulent
328 (Menut (2017)). This particular flow may have an impact on navigation. The
329 flow applied to the test bench was low compared to real conditions and in

330 simulation and the bloodstream was not taken into consideration at all in
331 the simulation. Moreover, navigation tests on animals were conducted by
332 the team and it was observed that blood flow and aortic wall deformations
333 could probably be neglected because of their small influence on the guidewire
334 position.

335 Concerning the geometries of the surgical tools involved in the simulation,
336 we did not represent the total lengths of the devices. We can consider that
337 it did not have any consequences related to the gestures of the clinician:
338 during experimental navigation, handlers navigate near the hemostatis valve
339 to reduce the risk of buckling when it comes to linear movements. But the
340 rotations are applied close to the proximal end of the shaft whereas in our
341 model they are applied near the valve. This difference resulted in a shorter
342 turnaround time in simulation but did not affect the navigation itself.

343 Regarding the experimental results on the measurement of the clinician
344 gestures, the uncertainties were large due to a low amount of data and to
345 movements that were difficult to repeat (especially the rotations). However,
346 the success of navigation was not limited to a few speed values but rather
347 allowed for a wide range of speeds. For our simulations, we considered average
348 velocities.

349 The simulation was performed on the same patient-specific geometries as
350 the ones used to create the phantoms for validation purposes. However, the
351 geometries slightly differ. Indeed, the Supra-Aortic Trunks (SATs) and the
352 ascending aorta are shorter in the numerical model. Manufacturing phan-
353 toms also generates small inaccuracies, that can be visible at the junction
354 between the SATs and the aortic arc and in the descending aorta for instance.

355 However, we consider that the experimental and simulation geometries are
356 close enough to ensure a satisfactory validation of the navigation related to
357 this study.

358 A rigid aorta clearly differs from real aorta modeling (Labrosse et al.
359 (2009)). The simulations were therefore not transferable to a clinical situa-
360 tion. But in reality, the aortic arch is quite rigid, thus, the position of the
361 guidewire should not be too far from the one obtained with a simulation on
362 a rigid aorta. Nevertheless, this requires further validation against patient
363 data.

364 Validation was performed in different configuration based on displace-
365 ments only. It showed that the passive behaviour of the guidewire is me-
366 chanically reasonable as the guidewire deformation due to contact with the
367 aortic walls is accurately simulated. However, concerning activation as pre-
368 viously discussed, the constitutive law used to represent the wire behaviour
369 is equivalent to a displacement-driven one. Therefore, in the simulation,
370 the curved guidewire would impose its motion to its environment and gen-
371 erate very high reaction forces; this is obviously not realistic and unfit for
372 validation. We anyway used this model as it was shown both on phantom
373 experiments and animal experiments that the activated guidewire is indeed
374 very stiff compared to the surrounding tissue and tools.

375 **6. Conclusion**

376 This paper presents a numerical model of active endovascular tools navi-
377 gating towards complex targets. It is based on patient specific anatomies and
378 experimental characterization of both the tools and the praticians's gestures.

379 Now that it has been shown that complex anatomies can be successfully
380 navigated on the test bench and in simulation, two main perspectives can be
381 drawn:

- 382 1. For design purposes and surgeons training, the numerical model could
383 be generalized to a larger cohort of anatomy using for example learning
384 or reduced order modeling approaches (Lu et al. (2017a); Lu et al.
385 (2017b));
- 386 2. For clinical application purposes, the model could be adapted to real
387 patient cases provided some assumptions regarding for example ma-
388 terial models, boundary conditions or effect of blood flow are further
389 questioned.

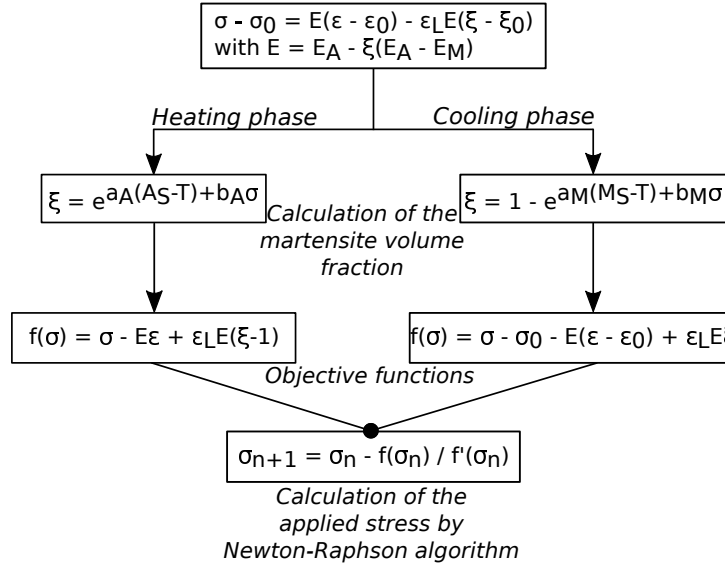
390 **7. Acknowledgements**

391 The French National Research Agency (ANR) partially supported this
392 work through the DEEP project: Devices for augmEnted Endovascular nav-
393 igation in complex Pathways (grant n°ANR-18-CE19-0027-01).

394 **8. Conflict of interest statement**

395 The authors declare that there are no conflicts of interest.

396 9. Appendix. Diagram of the Tanaka's Law



Parameter	Description	Value
E_A	Austenite Young modulus	50000 MPa
E_M	Martensite Young modulus	33320 MPa
ϵ_L	Maximum recoverable strain	0.0347
$M_{s/f}$	Martensite starting / finishing temperature	47 / 43°C
$A_{s/f}$	Austenite starting / finishing temperature	60 / 65°C
$T_{(de)activation}$	Time required to bend the guidewire	1.7 s
$a_A ; a_M ; b_A ; b_M$	Material constants	0.92 ; -1.15 ; 0.11 ; -0.14 (/°C)

Figure 8: Overview diagram of Tanaka's law and parameters used in our case. In the diagram, the parameters $(-)_0$ are related to the initial conditions. The applied stress σ was determined by Hooke's law added to a temperature dependent term. Depending on whether the material was heated or cooled, the martensite volume fraction ξ was computed using the temperature T and material constants a_A, a_M, b_A, b_M which are function of the starting / ending temperatures of the different phases. Finally, objective functions were created and a Newton-Raphson algorithm was used to compute the stress at the next timestep.

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Part	Configuration	Number of elements	CPU time (h)	Mean Hausdorff Distance (mm)
Aorta	Reference (fine mesh)	17103	35	
	Configuration 1	4714	40	0.65
	Configuration 2	7392	20	4.24
	Configuration 3	5226	17	4.29
Catheters	Reference (fine mesh)	17312	188	
	Configuration 1 (course mesh)	6132	93	0.39
Guidewire	Reference (fine mesh)	1373	202	
	Configuration 1	723	195	0.74
	Configuration 2	686	190	0.82
	Configuration 3	506	181	2.02

Figure 9: Results of the mesh convergence analysis for the catheters, the aorta and the guidewire. For these three parts, distinct studies were conducted involving endovascular navigation. The position of the guidewire was compared between the different configurations and a reference configuration (fine mesh) using Mean Hausdorff Distance. For the aorta and the active guide, 4 different mesh sizes were investigated. As the dimensions of the catheters were small, only two configurations were considered for the catheters as the mesh intended to preserve the tools geometry.