

MEDICAL ROBOTS

Dexterous helical magnetic robot for improved endovascular access

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Treating vascular diseases in the brain requires access to the affected region inside the body. This is usually accomplished through a minimally invasive technique that involves the use of long, thin devices, such as wires and tubes, that are manually maneuvered by a clinician within the bloodstream. By pushing, pulling, and twisting, these devices are navigated through the tortuous pathways of the blood vessels. The outcome of the procedure heavily relies on the clinician's skill and the device's ability to navigate to the affected target region in the bloodstream, which is often inhibited by tortuous blood vessels. Sharp turns require high flexibility, but this flexibility inhibits translation of proximal insertion to distal tip advancement. We present a highly dexterous, magnetically steered continuum robot that overcomes pushability limitations through rotation. A helical protrusion on the device's surface engages with the vessel wall and translates rotation to forward motion at every point of contact. An articulating magnetic tip allows for active steerability, enabling navigation from the aortic arch to millimeter-sized arteries of the brain. The effectiveness of the magnetic continuum robot has been demonstrated through successful navigation in models of the human vasculature and in blood vessels of a live pig.

INTRODUCTION

Among the most serious medical conditions are those that affect the blood vessels deep within the brain. These include acute ischemic stroke (AIS), where a blood clot blocks the blood flow to the brain (see Fig. 1C), and aneurysms, which can burst, leading to uncontrolled bleeding inside the brain. AIS alone is the second leading cause of death and long-term disability, affecting more than 13 million people annually (1).

The treatment of these vascular diseases requires accessing the affected areas within the brain. This is usually accomplished through a minimally invasive technique that uses thin wires inserted into the patient's blood vessels. These devices, measuring more than a meter in length, are carefully navigated by applying external manipulation—pushing, pulling, and twisting—to steer the curved tip on the other end deep within the body.

The devices are typically inserted through either the femoral artery in the leg or the radial artery in the arm and navigated toward the cerebral arteries via the aortic arch. The femoral approach is the conventional choice because of its relatively straight path from the insertion point to the arteries of the neck. The trans-radial approach has gained recognition as a feasible alternative because of the lower complication rate, but it presents additional challenges given that it requires navigating sharp turns that are difficult to overcome using traditional devices (2).

Mechanical thrombectomy (MT) is a procedure in which the occluding blood clot is physically grasped and removed (3, 4). MT is now the standard of care in large vessel occlusions in patients with AIS and has increasingly replaced intra-arterial thrombolysis in which a thrombolytic agent, such as urokinase and tissue plasminogen activator, is injected in large quantities into the vessels (5, 6). MT

clinical trials have helped establish it as the most effective treatment for AIS in many cases.

Although this minimally invasive endovascular treatment has revolutionized the way AIS is treated and markedly improved patients' chance of recovery, it can be challenging for the surgeon to perform. Navigating tortuous vessels often requires multiple attempts, repeatedly stressing the vessel walls and thereby increasing the risk of vasospasm and vessel dissections (7). For example, one of the leading causes of failed reperfusion, the restoration of blood flow in occluded vessels, is the inability to reach occluded regions (8). Particularly challenging are highly tortuous vessels with sharp turns, loops, and kinks (7). In such anatomies, pushability becomes challenging because of the high friction that the devices experience distally (see Fig. 1D). Too much space and a lack of proximal support cause the devices to buckle when pushed (7, 9). If this occurs in a bifurcation, then devices can buckle into the wrong vessel. In this case, proximal advancement is not translated to the tip, and a large loop is formed (10, 11) (see Fig. 1C). Friction can also prevent the guidewire-catheter system from moving forward. Potential energy accumulates, stretching and stressing the vessels. In such conditions, it is difficult to anticipate the release of the accumulated energy, which may occur ballistically, leading to vessel perforation (12). Friction and pushability problems can be mitigated by applying a hydrophilic coating on the outer surface of the device (13) and by optimizing the bending stiffness along the device (14). Increasing the stiffness comes with the drawback of making the devices too stiff to bend through certain anatomies, which can lead to substantial forces on the vasculature. Recent technical advances in catheter and guidewire design have mostly been based on improving stiffness and friction properties to improve pushability.

Robotically assisted endovascular interventions promise to revolutionize the operating room, improving clinical outcomes and reducing workplace-related hazards for the operator. These interventions are achieved through added sensing, software functionalities, and automation algorithms and have the potential to reduce ionized radiation exposure for both patients and operators, reduce procedural times, and improve surgical outcomes (15–17). In addition, being able to operate remotely in a telerobotic setting has great potential to make

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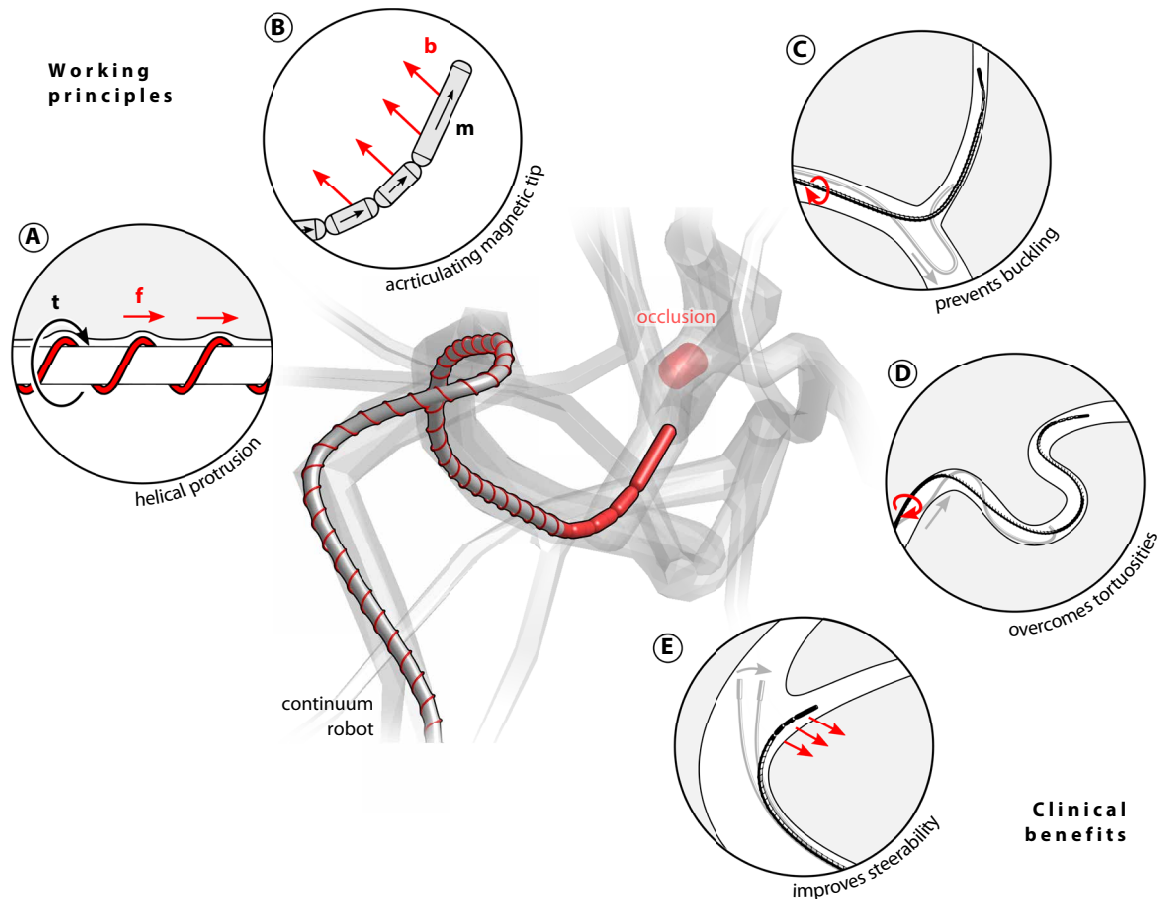


Fig. 1. Helical mCR with articulated magnetic tip. (A) The helical protrusion on the outer surface of the continuum robot, upon rotation, pushes the robot tip forward. (B) The segmented articulating magnetic tip increases the magnetic volume and decreases bending stiffness for improved steerability. (C) The helical robot prevents buckling and (D) overcomes tortuous vessels. (E) The articulating tip allows for large tip deflections.

this life-saving intervention accessible to more people around the globe, particularly to those living in rural regions. Endovascular robots first made their way into the operating room in 2012 with the launch of the Magellan robotic platform (Johnson & Johnson Inc., Redwood City, CA, USA) for peripheral vascular interventions (18, 19). These types of robotic systems have experienced a recent renaissance with the CorPath GRX (Siemens Healthineers International AG, Waltham, MA, USA) and the R-one (Robocath Inc., Rouen, France) systems (20). Despite the advantages offered by these robotic platforms, they primarily concentrate on steering traditional passive preshaped guidewires and catheters, using motorized advancer units that mimic human actuation. Although this approach adds remote steering capability, it does not address the inherent limitations related to friction, buckling, and steerability found in traditional manual devices.

Substantial efforts have been dedicated to the development of actively steerable and deflectable devices to enhance device steerability. Different technical solutions for distal tip steering have been proposed, with the common objective of increasing device maneuverability. The underlying technologies range from hydraulic (21–25), acoustic (26), and shape memory alloy (27–31) solutions to tendon-driven (32, 33) continuum robots. Although these solutions offer active steerability, they require lengthy tubing and cabling to connect the actuators and pumps to the device tip, which can be susceptible

to issues like hysteresis and nonlinear backlash (34). In contrast, magnetically steered continuum robots, another class of actively steerable devices (35), do not suffer from the shortcomings.

A magnetic continuum robot (mCR) is a device with embedded magnetic material that is steered through remote magnetic navigation (RMN) (36). In RMN, the mCR's distal tip is wirelessly steered by an external magnetic field that applies a magnetic torque on the embedded magnetic material. The magnetic fields, which are usually lower than 150 mT, are generated by a magnetic navigation system (MNS), a machine specifically designed to generate external magnetic fields for mCR steering (35). In general, an MNS uses either large permanent magnets that are moved on a robotic arm (37) or stationary electromagnets, in which case the system is called an electromagnetic navigation system (eMNS) (38). RMN has been of increasing interest for minimally invasive surgeries, including cardiac ablations (39) and neurovascular interventions (40, 41), because of its potential benefits in speed and accuracy (17). Both eMNSs and permanent MNSs have been commercialized (42, 43). In conventional catheterization, guidewire selection and shape are key factors for success (44) and are critical challenges facing clinicians when preparing for an intervention (14). Magnetic navigation offers the advantage of adaptive reshaping of the magnetic distal tip during the procedure.

We present a highly dexterous robotic device designed to navigate the dense and tortuous arterial structure of the brain (45). We used a locomotion modality that overcomes pushability and trackability limitations by coupling insertion with rotation and by adding a helical protrusion on the outer surface of the device. The protrusion engages with the vessel wall, analogous to a flexible screw, and pulls the device forward when rotated (see Fig. 1A). The rotation is induced at the proximal part of the device and transmitted toward the distal part along its torque-transmitting body. This actuation principle, combined with magnetic steering, enables the dexterity and navigability necessary to access hard-to-reach target locations in the vascular system. The segmented articulating magnetic tip design makes the device tip soft and atraumatic yet maximizes the actuatable tip deflection angle (see Fig. 1B). The helical device's effectiveness is demonstrated through successful navigation experiments conducted in silicone models of human vasculature, in an ex vivo human placenta, and in an in vivo porcine model.

RESULTS

Robotic system design

Our robotic platform consists of a helical mCR that is steered by external magnetic fields generated by an eMNS (Fig. 2C). The eMNS consists of three electromagnets in a triangular arrangement (see Fig. 2B and the “Electromagnetic navigation system” section), positioned on the cranial side of the patient. The height of the electromagnet array can be moved to match the height of the patient's bed. The mCR is advanced by a motorized advancer unit mounted on a surgical arm attached to the patient's bed (Fig. 2D). The operator

steers the mCR via a dedicated user console (46) either beside the patient or from a remote location in a teleoperation setting (Fig. 2A). Visual feedback is provided through x-ray images generated by a monoplanar fluoroscope imaging device (Fig. 2E). The subunits are interfaced with the Robot Operating System (47) because of its high degree of modularity and simple multi-threading capabilities.

Helical magnetic microcatheter and guidewire design

We have developed two implementations of the helical mCR design concepts that are tailored to navigate in the millimeter-sized cerebral arteries, such as the M1 [mean inner diameter (ID) of 3.1 mm] and the more distal M2 (mean ID of 2.4 mm) segments of the middle cerebral artery (MCA) (48). Both implementations comprise a torque-transmitting backbone with a helical protrusion on its outer surface [outer diameter (OD) of 0.7 mm with helix] and an articulating magnetic tip (OD of 0.8 mm). The first implementation is a steerable helical guidewire (see Fig. 3A), and the second is a steerable helical microcatheter (see Fig. 3B).

The guidewire version has a tapered nitinol thread core for tip stiffness optimization. An optimized stiffness gradient is crucial for navigating extremely tortuous anatomies (14). In the case of the microcatheter version, the inner lumen was left empty to enable fluid injection. An in vivo fluid injection can be seen in movie S3. The devices are sufficiently small to be deployed using both a trans-radial [French (Fr) of 4 to 6] and a transfemoral (5 to 9 Fr) approach through a conventional intermediate catheter. For an effective implementation of the rotating locomotion principle, torque transmission along the mCR is crucial. If the device is too torsionally elastic, then the rotation applied to the proximal end may not be effectively transmitted

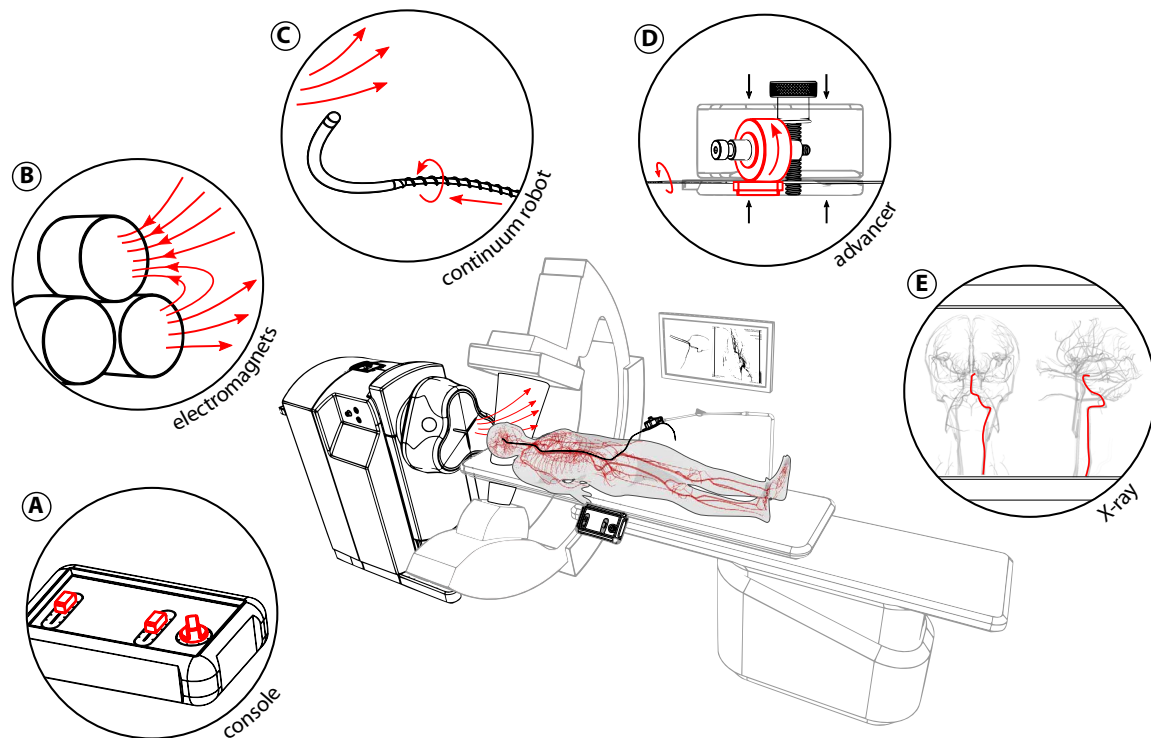


Fig. 2. Robotic system overview. (A) The console takes user inputs to control the magnetic field direction and insertion speed. (B) The electromagnets of the eMNS generate the external magnetic fields used to steer the magnetic devices inside the patient. (C) The helical magnetic endovascular device. (D) The motorized advancer unit pushes and rotates the helical magnetic devices. (E) The x-ray image provides visual feedback of the vasculature and the magnetic device.

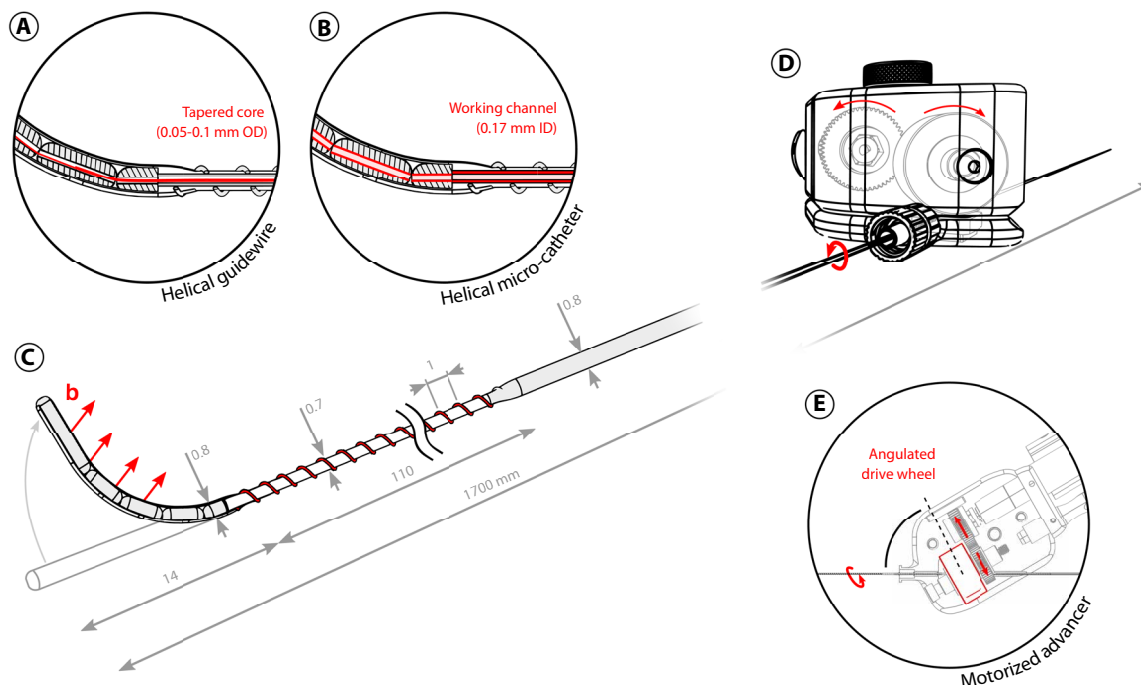


Fig. 3. Helical mCR and advancer design. (A) A magnified view of the guidewire helical mCR with a tapered nitinol core (red) for stiffness optimization. (B) A magnified view of the microcatheter helical mCR with working channel (red) for fluid injection. (C) The helical mCR has a flexible, magnetic tip with articulating links and a helical protrusion on its outer surface. (D) The advancer unit actuates the helical mCR by translation and rotation. (E) The roller axis of the motorized advancer is tilted to push and rotate the helical mCR.

to the distal tip. This can result in imprecise tip control, similar to backlash behavior, and, in severe cases, sudden releases of the accumulated energy. To meet the torquability and flexibility requirements, a multi-filar stainless steel torque coil was used as a torque-transmitting backbone. The nondestructive bending radius of the device's most distal section is less than 5 mm.

Advancer unit design

Advancing helical devices requires not only linear translation but also rotation, in contrast to traditional mCRs (41, 49). To address this requirement, we developed a remotely controlled, motorized advancer unit that pushes and rotates helical mCRs (see Fig. 3C). It can accommodate devices of different diameters and pitch angles and has unlimited driving length because of a friction-roller-drive mechanism (20). The helical mCR is clamped against the motorized roller by a grooved guiding insert at an angle such that the mCR is simultaneously translated and rotated (see Fig. 2D). The pitch angle can be adjusted by replacing the plate that holds the guiding insert (see fig. S1). To retract the helical device, the friction roller of the advancer is actuated in the inverse direction, causing a backward translational and rotational motion of the device. For more adaptability during surgery, the advancer was mounted on an adjustable surgical arm attached to the surgical bed. The passive components are autoclavable, and a sterile barrier in the form of a plastic drape is used to protect the motor.

Reachable workspace

To determine the behavior of the mCR tip in open spaces, we investigated its reachable workspace experimentally by performing a parameter sweep of the insertion length and magnetic field orientation

(see Fig. 4, A and B). The reachable workspace of an mCR is tightly coupled to the feasible workspace of the eMNS [for definition, see (50)].

To define the workspace for a given robotic task, the mCR and eMNS were considered as one system. Boehler *et al.* (50) have provided a systematic methodology and a toolbox to determine the workspace of eMNSs. The magnetic-feasible workspace of the three-coil eMNS at z height 0 mm is illustrated in Fig. 4B given the desired set of magnetic tasks D

$$D = \{\mathbf{b} \in \mathbb{R}^3 \mid 5 \cdot 10^{-3} T \geq \mathbf{b}_2\} \quad (1)$$

where $\mathbf{b} \in \mathbb{R}^3$ is the magnetic field generated by the eMNS.

The large magnetic volume and the flexible articulating links of the mCR tip enable tip deflections more than 90° and steerability even at small fields of 5 mT (see Fig. 4A). This allows for active tip steering from the aortic arch to the cerebral arteries. The reachable tip workspace with an overlay of the vascular anatomy is shown in Fig. 4B. Figure 4A shows that increasing the external magnetic field strength above 35 mT does not substantially affect tip reachability.

Helix effectiveness

To validate the effectiveness of the helical protrusion at transforming rotation into translation, the mCR was tested in a flat vessel model with multiple consecutive turns and with interchangeable lumen materials (see Fig. 4C). The devices described in this section were actuated manually by hand either by combining translation and rotation both clockwise or counterclockwise or pure translation.

First, the performance of the helical mCR was evaluated by comparison with the same device with no helix. The helix-free device is actuated by both rotation and translation as well as pure translation, whereas the helical device is actuated by clockwise rotation and

translation (Fig. 4D). The devices were inserted in a silicone lumen of shore hardness 60A. Silicone is commonly used in the fabrication of vascular models and has similar mechanical properties to cranial blood vessels (51). Figure 4D shows that the helical mCR could advance 14 turns before friction became too high to further advance the device, compared with the mCR with no helix, which advanced four turns.

Furthermore, the results in Fig. 4D indicate that rotation, in itself, improves advancement. The helix-free device, which is rotated and translated, advanced further than the one that was purely translated (four turns compared with two turns). This can be explained by the fact that a rotating device can be maintained in a dynamic friction

regime, even when it is not advancing. In contrast, a purely translated device must overcome static friction every time it is not advancing.

To further evaluate the effectiveness of the helix, the mCR was inserted into the model with clockwise rotation and translation, counterclockwise rotation and translation, and pure translation (see Fig. 4E). When rotated counterclockwise, the helix counteracted the forward advancement. The results shown in Fig. 4E indicate the effectiveness of the helix, because the clockwise-actuated mCR advanced seven times farther than the mCR that was actuated in a counterclockwise direction or translated without rotation.

To demonstrate that the helix engages with the soft lumen wall, the helical mCR was inserted in lumens made from materials with

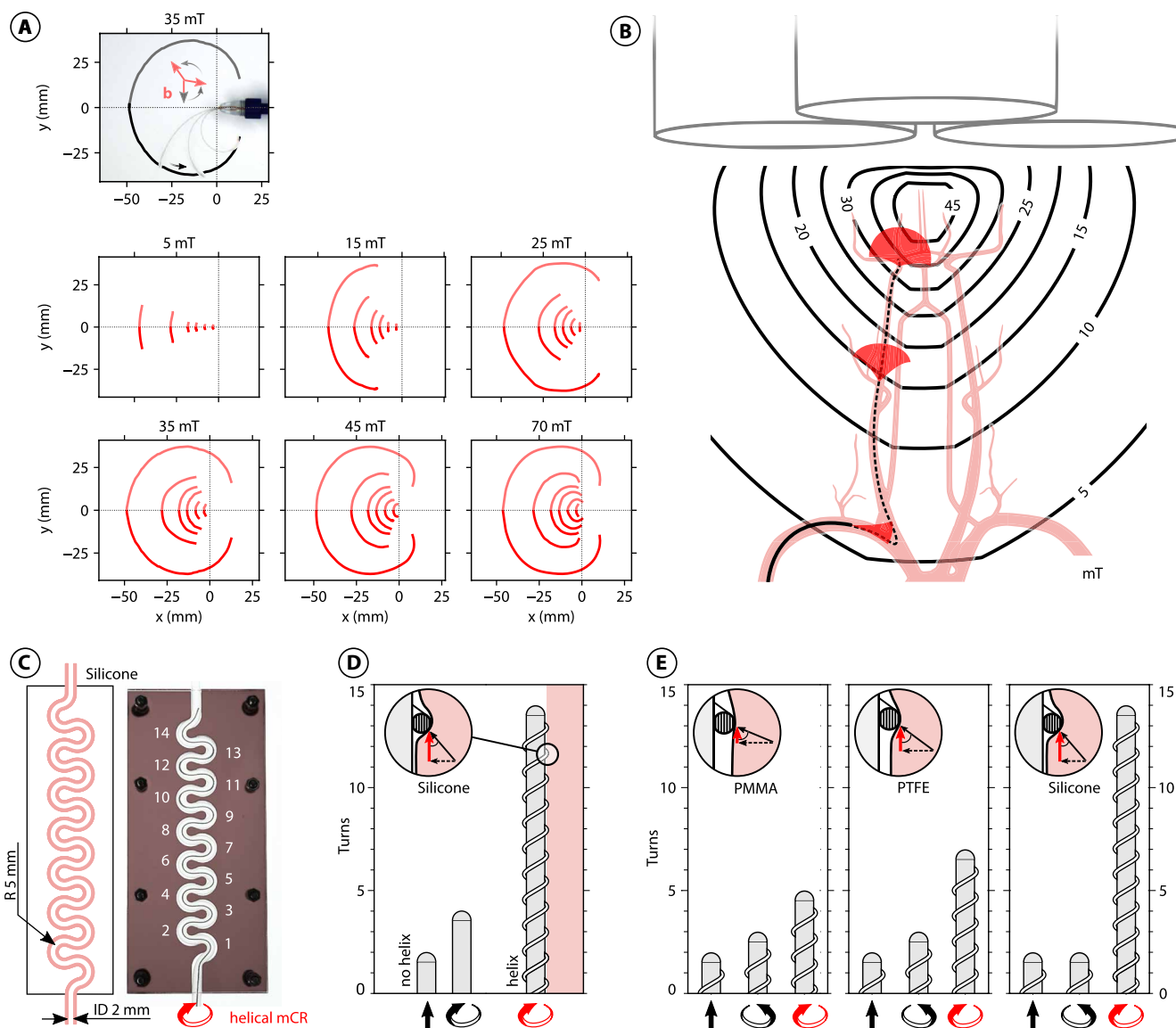


Fig. 4. Workspace and helix effectiveness measurement of the helical mCR. (A) A sweep of parameters and tip tracking reveal the reachable workspace at different magnetic field magnitudes. (B) The mCR reachable workspace when navigating the supraortic arteries and actuated by a three-coiled eMNS. (C) The flat vessel model with interchangeable lumen material: PMMA, PTFE, and silicone. (D) The comparison between a helical device actuated by rotation in a clockwise direction and a device without helical protrusion, actuated by translation and rotation. The helix engages with the soft material and, when rotated, generates a forward pushing force. (E) The helix effectiveness: The continuum robot is inserted as far as possible into three tortuous vessel models with three different materials for different engagement levels (PMMA, PTFE, and silicone). For each model, the mCR is inserted in three ways, first by pushing only, then while rotating counterclockwise, and lastly while rotating clockwise.

increasing hardness [polytetrafluorethylene (PTFE) and polymethyl methacrylate (PMMA)]. Hard surfaces reduce the engagement of the helix and, thus, its forward pushing effect. Figure 4E shows that the forward-rotated helical mCR managed fewer turns with increasing lumen hardness. In contrast, the inversely rotated and pushed mCRs advanced by almost the same amount, regardless of the lumen hardness.

In vitro navigation in human vascular model

Accessing the target site in the brain using the conventional manual technique involves multiple steps. The process begins by inserting a guiding catheter through either the leg (transfemoral approach) or the arm (trans-radial approach) and advancing it beyond the aortic arch into the neck arteries. The guiding catheter is then used as an anchor point to deploy and stabilize additional catheters and guidewires. Typically, the guidewires lead the way and act as a rail over which the flexible intermediate catheters are deployed. Depending on the depth and size of the target vessel, additional sets of micro wires and microcatheters are inserted. Mastering this procedure requires a high level of skill and may involve multiple attempts, often requiring the exchange of devices. Inserting multiple devices prolongs procedure time and increases the risk of arterial endothelial injury. Reducing the number of catheters and guidewires increases clinical efficacy and procedure speed while reducing fluoroscopy time and mitigating the risk of adverse events (16, 17).

Here, we demonstrated that our helical mCR can be used as an all-in-one device and is capable of navigating through complex anatomies. The device was navigated in a silicone model of human vasculature. Four navigation tasks were investigated that are challenging to perform with conventional manual devices. The first task consisted of navigating to distal millimeter-sized anterior cerebral vessels (MCA-M2 segments) through the trans-radial access path. Although the transfemoral approach is commonly preferred because of the relatively straight path from the leg to the neck, the trans-radial approach has gained recognition as a feasible alternative because of the lower rate of local vascular complications. However, the trans-radial approach is less commonly used because of the more challenging anatomical features (2). The magnetic field strength in the region of the aortic arch ranges from 5 to 15 mT and in the cerebral arteries from 20 to 30 mT (see Fig. 4B).

Figure 5A depicts the navigation of the sharp turn between the subclavian artery (SA) and common carotid artery (CCA). As can be seen in movie S2, the mCR did not buckle at the SA-CCA bifurcation and showed high steerability in the tortuous cerebral arteries.

In the second task, the helical guidewire was used as a delivery tool to guide with a 1.17-mm ID and 1.67-mm OD (5-Fr) intermediate catheter to the cerebral arteries. In the treatment of stroke, intermediate catheters are commonly used for decannulation to aspirate blood clots or to deliver MT devices. The deployment of the catheter is depicted in movie S2.

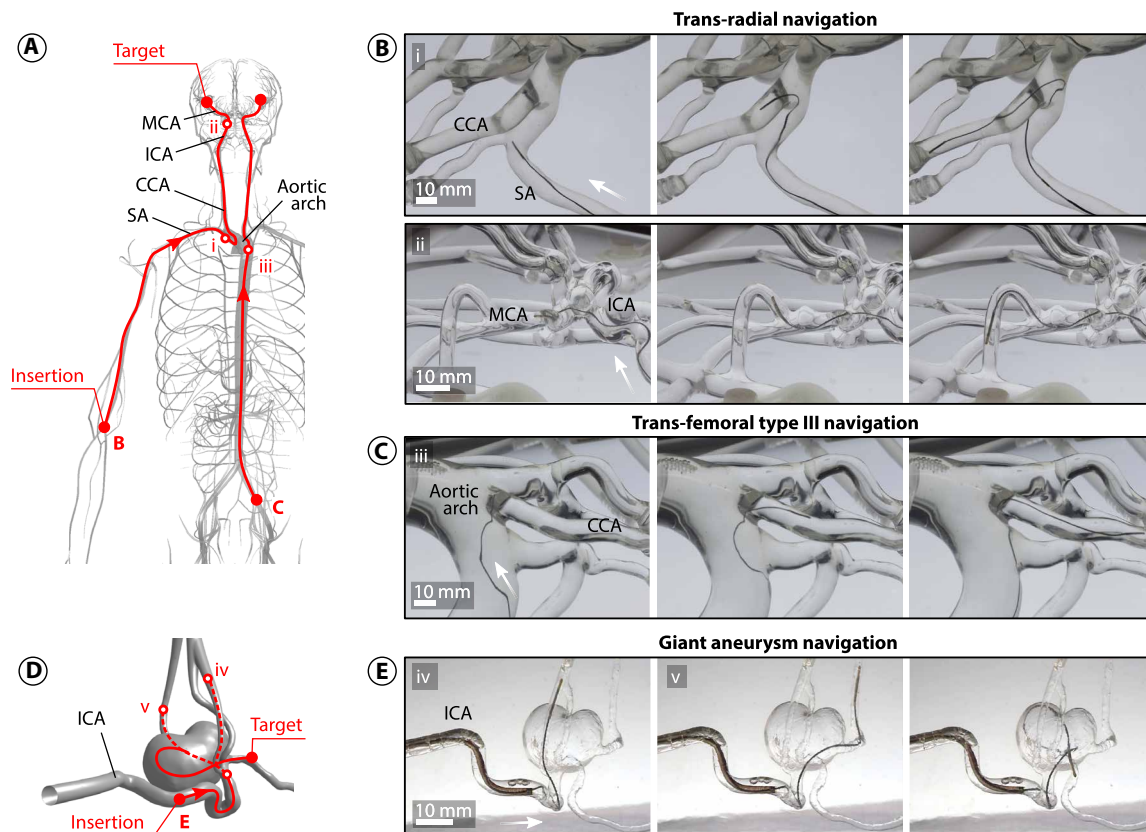


Fig. 5. In vitro navigation. (A) An overview of the human vasculature with insertion points in the radial (B) and femoral arteries (C) and target in the cerebral arteries. (B) The navigation in distal cerebral vessels through the trans-radial approach in a silicone model, starting from the SA, into the CCA, across the ICA, to the MCA-M2 segment. (C) The navigation across a type III aortic arch through the transfemoral approach access of a silicone model. (D) An overview of the navigation in a fusiform giant aneurysm with insertion through a catheter in the ICA (E). (E) The navigation along the parent vessels in a fusiform giant aneurysm situation in a silicone model.

In task three (Fig. 5B and movie S2), the helical mCR was navigated across a morphological type III aortic arch through the transfemoral approach without buckling. Type III aortic arches are particularly difficult anatomic variations of the aortic arch and are related to a higher complication rate in cranial reperfusion (7). The variation occurs in 20% of the stroke patients eligible for MT (52).

The fourth task consisted of crossing a fusiform giant aneurysm by navigating the helical mCR from the internal carotid artery (ICA) to the distal parent artery of the fusiform giant aneurysm (Fig. 5C). Fusiform giant aneurysms have a dome diameter of more than 25 mm and are particularly challenging to treat for manually steered guidewires. These guidewires often lack the necessary shape and sufficient lateral support, leading to buckling, which may necessitate multiple attempts, reshaping of the tip, and/or the use of additional catheters (53). In contrast, the magnetic tip of our device provides three-dimensional (3D) tip control that facilitates targeting the parent arteries, and the helix can push itself forward inside the aneurysm, preventing loop formations, as shown in movie S2.

During navigation, we observed some slight vibration of the tip. We can attribute this behavior to the device's overall geometry. The 1700-mm-long device may not always be perfectly straight, especially if it is not stored in a straight tube. This slight curvature could result in a small rotational imbalance. However, we have yet to encounter navigation issues caused by this imbalance, and we believe that more controlled manufacturing and storage methods could eliminate the effect.

Invasiveness study in ex vivo human placenta model

To evaluate the potential mechanical impact on the vessel wall, we conducted an invasiveness study using a helical mCR in an ex vivo human placenta (see Fig. 6A). Placentas are particularly suited for endovascular device testing because they are similar in size and curvature to human intracranial arteries (54, 55). We chose the placenta veins because they carry oxygenated blood, similar to the cerebral arteries (55), and are simpler to access and characterize (54). In this experiment, we collected a placenta immediately after birth to enable testing in blood vessels that have intact endothelial cells, which enabled the assessment of the condition of the individual vessel wall layers at the histological level.

The helical magnetic guidewire was inserted and retracted 20 times at an insertion speed of 5 mm/s and a corresponding rotation speed of five turns per second in one of the placenta veins (see Fig. 6B). For comparison, a nonmagnetic commercial guidewire (0.035 Glidewire, Terumo Medical Corporation, New Jersey, USA) was also inserted and retracted 20 times in a separate blood vessel. To establish upper and lower reference points for physical impact assessment, both positive and negative controls were implemented (see Fig. 6, B and C). The positive control was established by intentionally inducing damage to the cell layers using a sharp metal wire tool (see Fig. 6C, "positive control"). The negative control refers to the blood vessel into which no device had been inserted, leaving it unharmed. After the experiments, the tested placenta immediately underwent fixation by 4% formalin for 48 hours, followed by a histological analysis specifically at the levels of the areas of insertion of the different devices to assess the condition of the vessel wall layers. The analysis was performed by an expert pathologist who received the samples in a blind test.

The histological analysis of the negative control displayed an intact endothelial cell layer and vessel wall (see Fig. 6D, "negative control"), whereas the positive control demonstrated notable disruption of the

inner vessel wall, including complete separation and missing portions of the endothelium (see Fig. 6D, positive control). These controls effectively established a lower and upper reference for evaluating the extent of damage.

The analysis revealed that the helical guidewire had a minor physical impact on the vessel wall, as indicated by minimal separation of areas of the endothelium from the subendothelial stroma (see Fig. 6D, "helical guidewire"). These cell layer detachments can be attributed to friction between the device and the vessel wall, resulting in shear forces that led to the detachment of the cell layers. Similarly, the commercial guidewire displayed minor but slightly more damage in several locations in the form of endothelial detachment and small stromal tears, indicating slightly higher shear forces (see Fig. 6D, "commercial guidewire").

In vivo navigation in a porcine model

To demonstrate the validity and efficacy of the technology in a clinical setting, an in vivo feasibility study in a porcine model was performed. Porcine vessels are commonly used in research as models for human vessels because of their comparable size, anatomy, and general function (56). They enable the simulation of complete cranial interventions, starting from the placement of larger guiding catheters through transfemoral access across the aortic arch to the navigation within small-sized arteries. Such experiments are a first and important step toward applications in humans.

The first navigation showcases the device's ability to achieve distal access by navigating through narrow and tortuous arteries. The device was inserted through the femoral artery via a commercial 5-Fr vertebral catheter (Impress Diagnostic Peripheral Catheters, Merit Medical Systems Inc., Utah, USA) into the external carotid artery (ECA) and navigated to the millimeter-sized ethmoidal artery (EA) (see Fig. 7). To highlight the effectiveness of the working channel, a fluid injection was performed. One prominent application scenario is angiography, in which a liquid contrast medium is injected to enhance the visualization of blood vessels during x-ray imaging. This technique allows otherwise invisible blood vessels to be seen. In addition, fluid injection through the working channel can also be used for therapeutic interventions. For example, the administration of anticoagulants can support the dissolution of blood clots during stroke treatment, whereas embolization agents are injected to deliberately block blood vessels in the management of arteriovenous malformations. After performing angiography in the EA, the device was fully retracted and navigated to the facial artery (FA), where so-called super-selective angiography was performed. Super-selective angiography is the local injection of contrast agent to reveal small blood vessels.

After retraction back into the ECA, the magnetic device was navigated into the caudal auricular artery (CAA) to demonstrate the device's high steerability in low magnetic fields. Notably, the CAA presented a challenging scenario with a sharp bifurcation angle exceeding 90°. At the bifurcation between the CCA, the maxillary artery (MA), and the CAA, the magnetic catheter tip was 320 mm away from the magnetic coils of the eMNS. At the given distance, the magnetic field was 10 mT. Three attempts were needed to enter the CAA. The main challenge lay in the CAA orientation that was pointing out of plane. Note that what appears to be the catheter exiting the blood vessel in Fig. 7, fig. S2, and the corresponding movie S3 is, in reality, a shift between the roadmap and the vessels observed in real time.

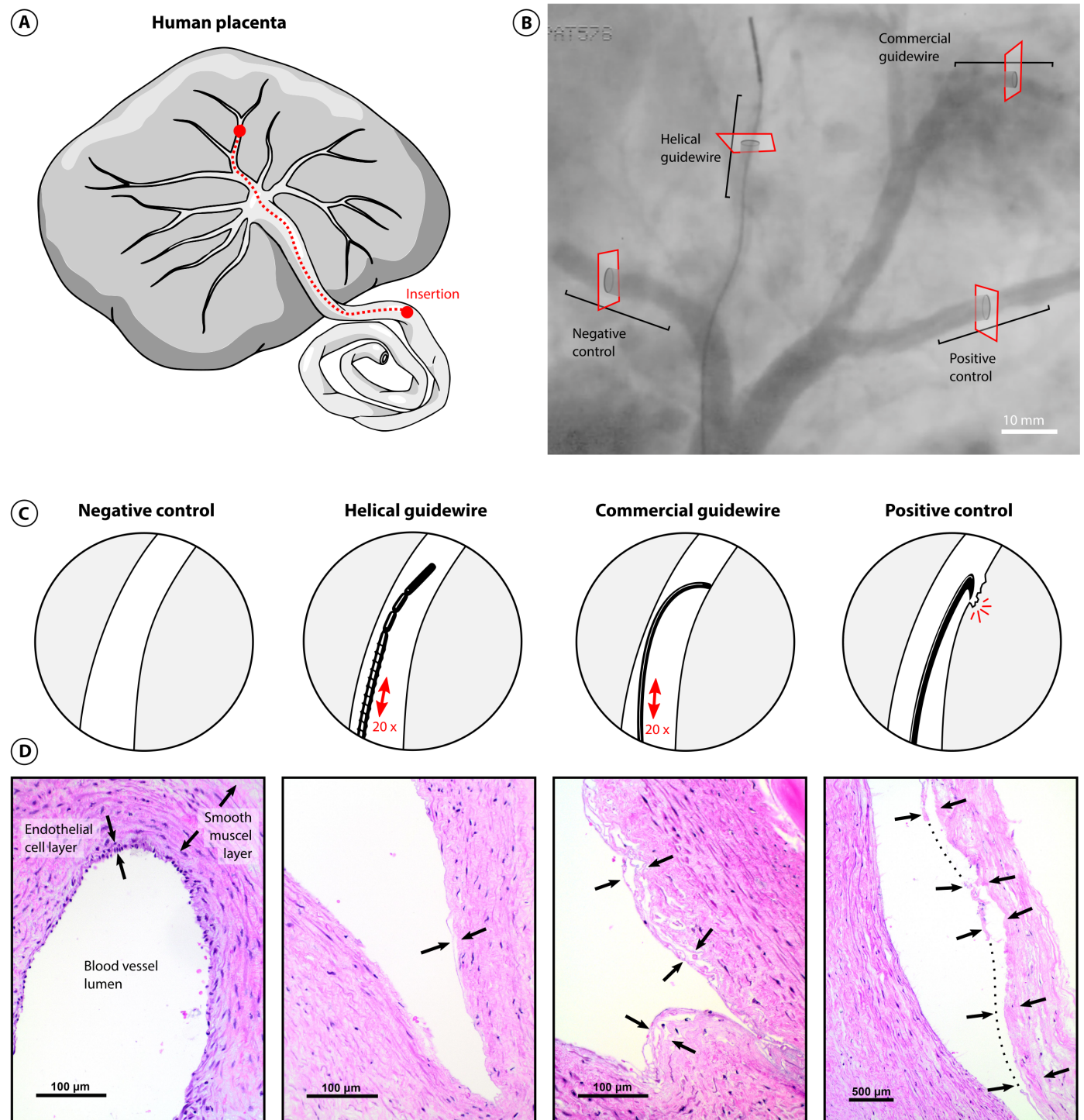


Fig. 6. Histological analysis of human placenta blood vessels. (A) An illustration of a human placenta with device insertion point. Ex vivo placenta veins were used for endoluminal device testing because they are similar to human brain arteries. (B) For comparison, different devices (negative and positive controls, as well as helical and commercial guidewires) were inserted into individual blood vessels. After navigation, histological analysis was performed on multiple sections (example slices are indicated in red), which were extracted from samples taken from various locations (examples are indicated by black lines). (C) The negative control is a blood vessel that was left untouched, whereas the positive control consists of a vessel intentionally damaged using a sharp wire tool. The helical and commercial guidewires were inserted and retracted 20 times. (D) Example images of the histological analysis reveal the physical impact on the endothelial cell layer and inner vessel wall. The black arrows indicate minor cell layer detachments (helical and commercial guidewire) and more extensive laceration of the endothelial cell layer (positive control). The black dotted line indicates the torn and missing cell layer.

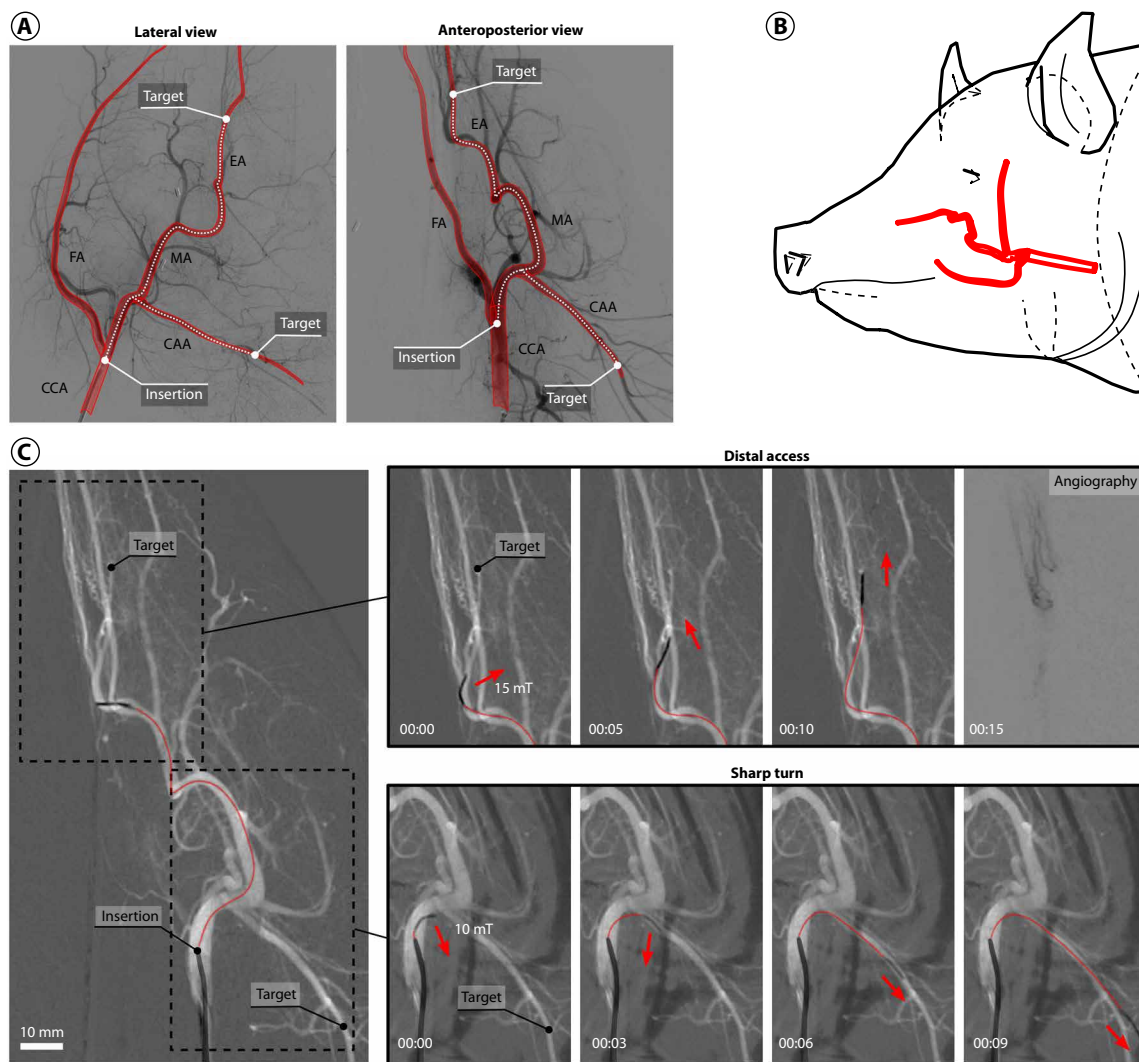


Fig. 7. In vivo navigation. (A) Lateral and anteroposterior view of the pig's vascular anatomy. (B) 3D representation of the pig's vascular anatomy. (C) In vivo navigation in millimeter-sized vessels with injection of contrast agent (angiography) and steering at low magnetic fields (10 mT), with more than 90° turn.

Postinterventional analysis demonstrated that, in the first two attempts, the device was oversteered with an offset angle (angle between the external magnetic field and the magnetic tip) of 108° and 164°, whereas in the successful attempt, the offset angle was 74° (see fig. S2). One advantage of robotically assisted interventions is the ability to record and analyze such events.

Simulated navigation of the helical mCR

Robotically assisted interventions facilitate data recording for post interventional analysis, as demonstrated in the previous section and illustrated in fig. S2. These interventions also provide the means to analyze, plan, and practice interventions on digital test benches. This requires high-performance simulators with a minimal gap between simulation and reality. To demonstrate the feasibility of realistic simulations, we developed a model of our helical mCR using the simulation framework described in (57) and enhanced it with a simple analytical model outlined in the “Analytical formulation of helix pushing force” section of the Supplementary Materials. Details

of the simulator can be found in the “Simulation of the helical magnetic continuum robot” section of the Supplementary Materials.

We validated the simulator by comparing digital navigations (in silico) with the real in vivo navigation presented in the “In vivo navigation in a porcine model” section, as well as an in vitro navigation performed on a 3D printed model. The navigations exhibited high similarities in terms of dynamics and qualitative behavior. For a side-by-side comparison between the real and simulated navigations, see Fig. 8 and movie S4.

DISCUSSION

The locomotion paradigm that we introduce is based on a helical protrusion on the device's outer surface that engages with the vessel wall (much like a flexible screw) and transforms a proximal rotation into a linear translation of its distal tip. When the helical mCR is at rest or stuck, rotation facilitates the transition from a static to a dynamic friction regime. For example, when the device starts to

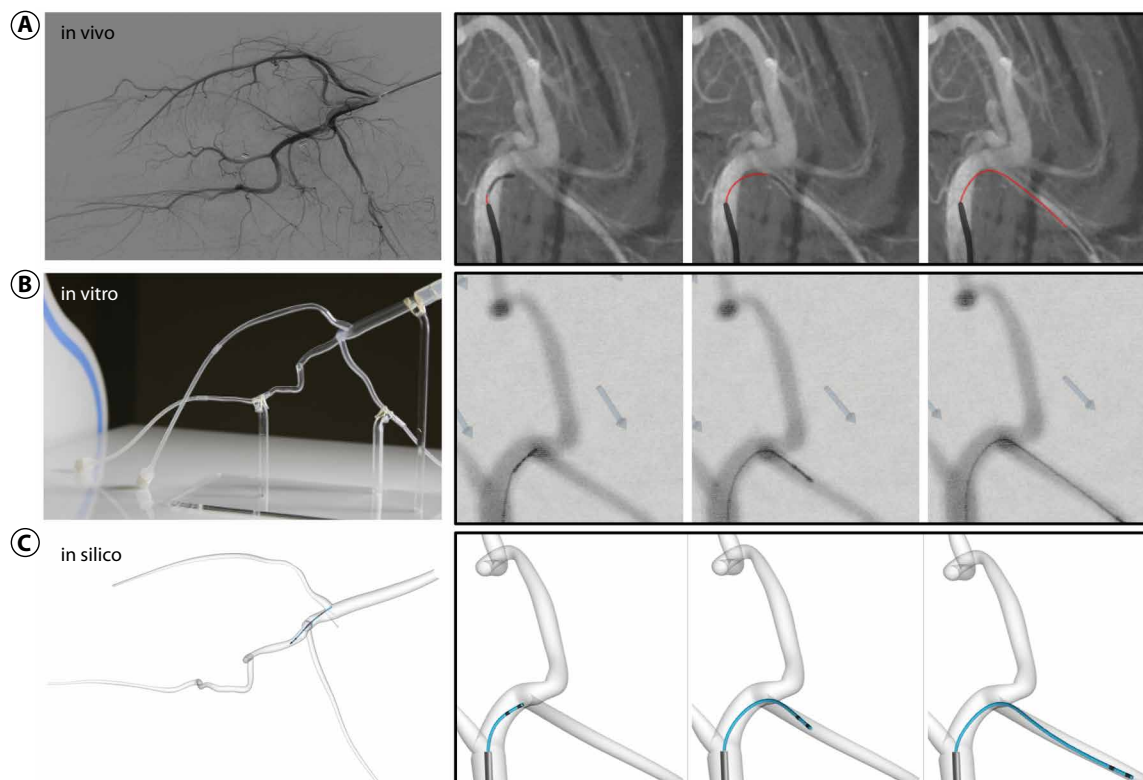


Fig. 8. Side-by-side comparison of in vivo, in vitro, and in silico navigation. (A) In vivo navigation. It serves as a ground truth for the simulated navigations. (B) In vitro navigation. (C) In silico navigation.

buckle, pushing will cause the device to collapse further and form a loop (see Fig. 1C). Thus, axial forces can no longer be transmitted, preventing distal tip advancement and risking vessel perforation or dissection. Rotation, on the other hand, can still be transmitted along the axis of the device without increasing the axial load. Another favorable effect that comes from rotation is the incremental release of elastic energy. Rotation promotes the propagation of small elastic deformations from back to front in a wave-like manner instead of releasing large deformations ballistically. Furthermore, the helical protrusion on the outer surface gently engages with the vessel wall and translates rotation to axial advancement. At every point of contact, the rotating helical continuum robot slightly pulls itself forward. This also facilitates the passage of bifurcations, where a conventional device is more likely to buckle (see Fig. 1C and movie S2).

In manual catheterization, the devices are primarily actuated by pushing, but rotation is used as well, mainly to redirect the preshaped device tip and occasionally to facilitate thrombus penetration (58). Rotation is tightly coupled with tip steering, and, therefore, its use for locomotion purposes is limited. Endovascular robots such as the CorPath GRX system (Corindus Inc., Massachusetts, USA) use motorized advancers to translate and rotate precurved endovascular devices, replacing the operator's hand motion with motorized actuators. The CorPath GRX has the same disadvantages as manual operation, where rotation is mainly dedicated to steering and cannot be independently used for locomotion. Although the CorPath GRX advancer is technically compatible with our devices, it lacks the necessary magnetic actuation required to use the rotational locomotion principle. A magnetically steered tip maintains a chosen tip direction even when the device is rotated around its longitudinal axis.

Nonmagnetic screw-tip continuum devices have been proposed but are designed either for large-scale gastroscopic applications (59) or to drill through thrombi in the vasculature (60, 61). Our device uses a helical protrusion along its entire length to translate contact points with the vessel wall into forward propulsion. Untethered helical magnetic devices have been proposed for vascular navigation (62–69) and for thrombi penetration (70–74). The devices promise a versatile range of applications because of their wireless nature. However, the functionalization of the devices is limited by their small size, and the technology has yet to prove itself in a clinical environment.

Tethered magnetic screw-tip devices have also been proposed. Yu and Kim (75) and Sperry *et al.* (76) demonstrated navigation of a magnetically actuated spiral guidewire for blood clot penetration and a magnetically steered flexible screw-tip needle for soft tissue navigation in the brain. In both approaches, the spiral structure is concentrated at a short and rigid screw-like tip. Although the rigid tip design and localized tissue engagement at the tip may be adequate for soft tissue drilling, the pulling forces are likely to be inadequate for locomotion in the vasculature. To generate effective pushing forces, the helix must have direct contact with the surrounding soft tissue. In applications such as liver or brain penetration, constant tissue contact allows the screw tip to generate enough pulling force to propel the rest of the device. However, in endovascular navigation, open spaces in blood vessels can cause loss of contact with the vessel wall. Spreading the helix along the device body establishes multiple contact points, ensuring continuous tissue contact and tip-independent pushing forces. Furthermore, by extending the helix along the body of the device, the local pushing forces are reduced, which, in turn, reduces the risk of tissue trauma, which our histology tests confirm. In the

“Invasiveness study in ex vivo human placenta model” section, we measured potential tissue trauma. The results of the invasiveness study indicate that our helical magnetic device causes minimal damage to the vessel wall, with less detectable damage compared with a commercial guidewire.

Although these findings are promising, they represent only the initial safety tests in a series of comprehensive evaluations required for regulatory approval before the device can be used in humans. Given the distinctive characteristics of the helical device, obtaining US Food and Drug Administration (FDA) registration will require clinical data from extensive in vivo studies involving multiple animals, including postmortem assessments, followed by subsequent human studies. In addition to these studies, comprehensive risk management will involve further investigation to demonstrate the safety and efficacy of the technology, including design, materials, manufacturing, and usability. Evidence of the system’s safe and intuitive operation will be required, as well as quantifying and comparing procedure times with those of legally marketed devices. The execution of in vivo trials and usability studies will require rigorous statistical analysis and a number of participants.

For conventional manual interventions, radiologists use their hands to manipulate guidewires and catheters, allowing them to perceive small variations in friction during the insertion process. Our system lacks this type of tactile and haptic feedback because of the absence of force sensing at the level of the advancer unit. Although advancer units with force sensing have been proposed (77), the reliable implementation of force data that radiologists can rely on remains a challenge because the forces generated at the device tip are small and are partially dissipated along the path to the sensor by device deformation, friction between insertion sheaths, hemostasis valves, advancer gears, and friction rollers. This is likely why commercial nonmagnetic robotic systems, such as the CorPath GTX, do not have integrated force sensing functionality. Furthermore, studies using commercial systems do not report any challenges or adverse events related to the absence of haptic feedback (78, 79). This can be attributed to the fact that radiologists primarily rely on the rich visual information provided by x-ray imaging, allowing them to detect subtle changes in device behavior and deformations. However, the ability to recognize hazardous situations on the basis of x-ray images requires years of experience. To provide assistance to the operator, sensor-less software algorithms based on x-ray images for the measurement of contact forces have been proposed, which could potentially eliminate the need for additional sensors (80).

The main focus in commercial magnetic actuation systems, such as Stereotaxis Genesis RNS, has been in increasing the magnetic field strength (0.08 to 0.12 T) (42) to achieve better device steerability. This compensates for the limited magnetization of the magnetic devices but requires powerful machines that have a large footprint and require permanent installations. As mobile and small footprint eMNSs with low magnetic fields in the 20 to 30 mT range move toward clinical application, the importance of magnetic device design becomes more relevant. Rather than increasing the external magnetic field magnitude, steerability can be improved by increasing the magnetic moment of the steerable device tip and decreasing its bending stiffness. Traditionally, mCRs are designed with one or more strong but rigid distal permanent magnets connected by flexible tubular sections (35, 36, 49, 81–83). This approach has an inherent trade-off: For a fixed length, more magnetic volume necessarily requires longer rigid magnetic sections and, therefore, less overall flexibility. Alternatively,

steerable mCR tips can be made from soft polymeric materials doped with magnetized particles (84, 85). Kim *et al.* demonstrated the effectiveness of this approach for neurovascular applications. Although these mCRs can be made soft and flexible, the magnetic moment of the magnetized polymers is considerably lower than rigid permanent magnets, requiring larger external magnetic fields (30 to 80 mT) (41). In contrast, our magnetic device, which incorporates articulating permanent magnets, operates effectively at low magnetic fields below 30 mT and offers steerability even at fields as low as 5 mT (see Fig. 4A).

Related magnetic tip designs using spherical magnetic chains for magnetic actuation were modeled and analyzed by O’Donoghue and Cantillon-Murphy (86), and Hong *et al.* (87) showed the application of a single ball joint magnetic tip designed for flexible needle steering in deep brain stimulation. Furthermore, Pittiglio *et al.* (88) presented workspace characterizations and closed-form kinematic modeling of a highly steerable ball chain robot. Here, we introduce a submillimeter device that incorporates a usable working channel. The design consists of interconnected hollow cylinders with rounded surfaces that function as articulations. The working channel of the device serves multiple purposes, including fluid injection and the delivery of various endovascular devices. Furthermore, it can be used to create a stiffness gradient along the magnetic tip, enabling improved navigation through highly tortuous blood vessels. Combining this with our helical actuation principle enhances the maneuverability and effectiveness of the device for navigation in complex vasculature.

The MNS used in this work is an eMNS with a small footprint. Because of their compact size, light weight, and mobility, these systems offer higher acceptance and easier integration into hospital workflows. Lightweight permanent magnet MNSs also exist (37, 41, 89). Kim *et al.* (41) proposed a platform for neurovascular applications that used a single cylindrical neodymium-iron-boron (NdFeB) magnet mounted on a 7-df KUKA robot arm. However, unlike permanent magnet systems, eMNSs have the advantage of having no moving parts. This eliminates the need for complex robot arm motion planning to avoid collisions with equipment and personnel in the operating room and/or obstruction of the x-ray view. In addition, the ability to deactivate the electromagnets can be a critical safety feature during interventions and when the system is not in use.

Despite the advantages of eMNSs, the helical mCR presented in this article is compatible with any MNS and, therefore, could be used in combination with the system described by Kim *et al.* (41). In this setup, the permanent magnet needs to be positioned close to the patient and must move with the mCR to generate sufficiently high magnetic torques and forces at the tip of the device. However, this restricts the maneuverability of the robot arm and makes motion planning more complex. The low-field steering capability of our device would increase the workspace of the permanent magnet, thereby expanding its range of motion. Furthermore, Kim *et al.* (41) reported a tendency for device buckling when navigating through artificial blood clots. The helical locomotion principle used by our device could potentially enhance mCR navigation in such challenging situations.

Simulators, such as the one we presented (see Fig. 8 and movie S4), are an ideal platform for the training of neural networks because they can produce large amounts of simulated training data at a relatively low cost (90, 91). Simulators also can extract otherwise inaccessible data, such as contact forces. Alternatively, robotic systems like ours can be used to record intraoperative data (see the “In vivo navigation

in a porcine model” section and fig. S2) to build large databases for analysis and the development of machine learning algorithms. Machine learning-based motion planning algorithms have been proposed for autonomous device navigation (90, 92, 93). However, to translate these simulation-based algorithms to the real world, it is essential to minimize the gap between simulation and reality. Although the in vivo and simulated navigations presented in Fig. 8 and movie S4 exhibit high similarity, a limitation of our simulation framework lies in the lack of fine details, such as small side branches in which the device could get caught and the lack of blood flow modeling. Capturing these details in the simulation will be essential to reduce scenarios in which the system does not behave as required. These unforeseen scenarios are often referred to as “edge cases.” In some artificial intelligence applications, edge cases might be acceptable. However, when it comes to automating navigation tasks in medical applications, small problems can often have large consequences. Medical robots interact with the physical world, and, therefore, there is the potential for real damage to be inflicted.

In the case of commercial neurovascular robots, decision-making and motion control are still exclusively in the realm of the surgeon [autonomy level 0 in the classification proposed by Yang *et al.* (94)]. However, these robotic platforms offer US FDA-approved software features, such as automated guidewire rotation, to assist the crossing of thrombi (20). Progressing to higher autonomy levels will require more time, primarily because of numerous unsolved regulatory and liability questions (95). Another challenge with advancing autonomy in neurovascular robotics into clinical practice is the restricted access to live x-ray images that could be used for online image processing. Many fluoroscope device manufacturers withhold permission for third-party processing of their x-ray images for safety and proprietary reasons. Nonetheless, research laboratories have demonstrated the exciting perspectives and benefits of autonomous device navigation (95). Early adoption of autonomous features may be applicable in high-risk, high-reward applications where alternative options are limited. One particularly promising application is the teleoperated treatment of stroke, where increased autonomy could potentially provide patients in rural areas with better access to this life-saving treatment (96).

MATERIALS AND METHODS

Remote magnetic navigation

The eMNS was modeled with the linear mapping equation

$$\mathbf{b}(\mathbf{p}) = \mathbf{A}(\mathbf{p})\mathbf{i} \quad (2)$$

which enabled computation of the electromagnetic currents $\mathbf{i} \in \mathbb{R}^n$ needed to generate a desired magnetic field $\mathbf{b}(\mathbf{p}) \in \mathbb{R}^3$ any position \mathbf{p} in the eMNS’s workspace. The actuation matrix $\mathbf{A} \in \mathbb{R}^{3 \times n}$ is system dependent and was calibrated experimentally using the method described in (97), where n is the number of electromagnets. In the case of the eMNS used in this work, n was 3. The external magnetic field induced a magnetic torque $\mathbf{t}_{\text{mag}} \in \mathbb{R}^3$ on the mCR’s magnets

$$\mathbf{t}_{\text{mag}} = \mathbf{m} \times \mathbf{b}(\mathbf{p}) \quad (3)$$

This equation shows that the magnetic torque scales with the misalignment between the magnetic field direction $\mathbf{b}(\mathbf{p})$ and the magnetic dipole moment \mathbf{m} . The magnetic forces induced by the magnetic field gradient had a minimal effect on the mCR tip and were therefore neglected.

Electromagnetic navigation system

The eMNS used in this work is the Navion system, a preclinical device developed by the Multi-Scale Robotics Lab (ETH Zurich, Zurich, Switzerland) and commercialized by Nanoflex Robotics AG (Nanoflex Robotics AG, Zurich, Switzerland). It is based on three electromagnets arranged in a triangular pattern. The electromagnets are mounted on a triangular yoke with a side length of 550 mm. The eMNS can generate a continuous current of up to ± 35 A for each electromagnet and has a workspace of 200 mm by 200 mm by 400 mm in which a magnetic field of 5 mT can be directed in any direction. The magnetic field strength 200 mm from the surface of the electromagnets is 25 mT. The eMNS has a footprint of 1200 mm by 720 mm and has been designed with wheels for easy mobility. In addition, the height of the electromagnets can be adjusted to match the height of the patient’s bed.

Torquable backbone

The torque-transmitting backbone consists of a custom double-layered, multi-filar torque coil (Asahi Intecc Co. Ltd., Aichi, Japan). Each layer comprises eight 0.06-mm 316 stainless steel filars wound tightly adjacent to one another. Adjacent layers are wound in opposite directions. The construction is similar to that of a hollow flexible shaft. The helix has a pitch length of 1 mm and is tightly wound on the outer surface of the torque coil and laser-welded at the ends. To reduce its stiffness and diameter, the most distal 100 mm was left partially uncovered [sleeve body design (44)]. To reduce the risk of kinking and buckling and to improve pushability, the device was reinforced proximally with 50- μm thermoelastic Pebax jackets to achieve the desired stiffness gradient along the central axis of the device. The distalmost jacket is 10 mm long with a shore hardness of 40D, followed by another 10-mm 63D jacket, and a 72D jacket covers the remaining proximal section. The jacket was fused on the backbone coil using reflow lamination techniques. At the intersection of the Pebax layers, the polymers melt together, producing a seamless transition between the sections. To further reinforce the most proximal part, a 38- μm polyester layer was added on top of the 72D Pebax layer up to 30 mm from the 60D/72D intersection.

Magnetic tip

There are two magnetic tip versions (see Fig. 3): one for the helical microcatheter and one for the magnetic guidewire. Both articulating magnetic tips comprise multiple hollow cylindrical magnets with adjacent convex surfaces. The segments have an OD of 0.7 mm and an ID of 0.17 mm, are made from gold-coated NdFeB, and are encapsulated in a 50- μm 35D Pebax jacket.

In the microcatheter version, the inner lumen was left empty (see Fig. 3B), whereas, in the guidewire version, the lumen was used to add a stiffness gradient along the magnetic tip (see Fig. 3A). A smooth gradient between the device tip and the body can be important when navigating sharp turns. If the transition between different stiffnesses is too abrupt, then the magnetic tip can kink, making it impossible to further insert the device. The stiffness gradient was achieved by incorporating three 0.05-mm nitinol wires with different lengths, resulting in a tapered core shape with an OD of 0.1 mm at its thickest and 0.05 mm at its thinnest end.

Advancer unit

The advancer consists of three parts: the main block, holding all the moving parts such as the friction drive wheel, gears, shafts, and the like; the clamping plate, containing the grooved guiding insert; and

the motor and surgical arm mount. The mCR is clamped between the friction roller and the guiding insert. Clamping can be adjusted by tightening a large thumb screw, which adjusts the distance between the main block and the clamping plate. The friction roller is made from silicone rubber and the guiding insert from transparent polycarbonate. The guiding insert has a groove that keeps the device centered and in contact with the drive. The guiding insert is held at an angle to the drive wheel corresponding to the pitch angle. A transparent material for the guiding insert was chosen so that the alignment of the mCR and the contact with the roller can be visually inspected and verified. The angle between the guiding insert and the drive wheel was adjusted by replacing the entire clamping plate with a clamping plate containing a guiding groove with a different pitch angle. Hence, the pitch angle can be adjusted when the mCR is exchanged, but not during operation. The main block can be detached from the motor, allowing the motor to be covered with a sterile drape, and the main block and clamping plate can be removed and autoclaved for sterilization. The device inlet and outlet are compatible with Luer-Locks, which allows the attachment of connectors and guiding tubes. The advancer unit was attached to a surgical arm (FISSO, Zurich, Switzerland) and directly mounted on the patient bed. A Maxon DCX 12 motor (Maxon Group, Sachseln, Switzerland) was used to drive the roller, controlled with an EPOS 4 (Maxon Group, Sachseln, Switzerland) position controller, and powered by a 12-V power supply. Gear reduction from the driving motor shaft to the roller shaft was 159:1. With a roller diameter of 25.4 mm and an mCR diameter of 0.8 mm, the total motor to mCR reduction is 1:5.

Reachable workspace measurement

The workspace measurement shown in Fig. 4A was performed in an eight-coiled eMNS called the CardioMag eMNS (50). The CardioMag can generate arbitrary 3D homogenous magnetic fields in a workspace of 100 mm by 100 mm by 100 mm. The mCR was inserted through a fixed insertion point at discrete steps of 5, 10, 15, 20, 30, and 50 mm, and the magnetic field was rotated at every insertion length from 0° to the singularity point—the point at which the mCR flips to a more stable equilibrium position. At insertion lengths above 20 mm, a small upward-pointing field component was applied for gravity compensation. The mCR distal tip was tracked with an optical color marker tracker. To this end, green and red markers were painted on the mCR tip and filmed with a Basler acA1920-48gc GigE camera (Basler AG, Ahrensburg, Germany). The image processing pipeline used for tip detection consists of an HSV threshold for green and red color, followed by a maximal distance limit threshold to discard false detections. The tracked tip path was filtered in postprocessing using a first-order Butterworth filter with a cutoff frequency of 0.024, normalized to the Nyquist frequency.

Setup of the helix effectiveness experiment

The experimental setup shown in Fig. 4C is made from three PMMA plates that have been screwed together. The middle plate has cutouts that hold and guide a plastic lumen made from materials with different hardnesses that are filled with a water-soap solution, reducing vessel wall friction (17, 98). The first lumen is a silicone tube with shore hardness of 60A, the second is made from PTFE with an intermediate shore hardness of 60D, and the last and hardest lumen is made from PMMA with a shore hardness of 90D and a square lumen cross section for easier manufacturing. Both the circular and square lumens have a diameter and a lateral length of 2 mm. The setups were backlit

through a light panel and filmed with a Canon EOS 6D (Canon Inc., Tokyo, Japan) camera. The commercial guidewire used as reference was 0.035-inch Glidewire with hydrophilic coating (Glidewire, Terumo Medical Corporation, New Jersey, USA).

In vitro models of the human vasculature

The helical guidewire was navigated in a silicone model (Trandomed 3D Inc., Zhejiang, China) containing the aorta, cerebral arteries, and major arteries of the lower and upper limbs. Furthermore, a silicon model was fabricated containing the ICS, a fusiform giant aneurysm with a dome diameter of 25 mm, and the distal parent arteries. The commercial catheter used in the experiment is a 5-Fr intermediate/aspiration catheter (AXS Catalyst 5 aspiration catheter, Stryker Corporation, Michigan, USA). Given that the friction between real blood vessels and endovascular devices is generally very low [friction coefficient < 0.017 (98)], the addition of soap to the water has proven effective in reducing the friction between the devices and the artificial vessel wall made from silicone rubber, thereby bringing the friction closer to reality. This practice is widely adopted (17, 99) and is also recommended by the manufacturer of the vascular model.

In vivo experiment

A 45-kg female domestic pig (*Sus scrofa domestica*; breed, Swiss large white; ~4 months of age) was placed under general anesthesia with continuous vital parameters monitored by a veterinarian team throughout the procedure. At the end of the procedure, the pig was euthanized. The animal study was approved by the local Committee for Animal Experimental Research (Cantonal Veterinary Office Zurich, Switzerland) under license number ZH213/2019. The right femoral artery was catheterized with an introducer sheath (Avanti + Introducer, 6F; Cordis, Miami Lakes FL, USA) placed percutaneously under ultrasound guidance using the Seldinger method. To reach the ECA, a guiding catheter commercial 5-Fr vertebral catheter (Impress Diagnostic Peripheral Catheters, Merit Medical Systems Inc., Utah, USA) was placed in the introducer sheath through which the mCR was inserted. A prophylactic dose of heparinized saline was applied (activated clotting time, 104 s). Imaging was performed with a monoplane angiography system (Allura Xper FD20 fluoroscope, Philips N.V., Amsterdam, The Netherlands). The horizontal distance between the fluoroscope center and the surface of the eMNS was 340 mm. At the given distance, the magnetic field in the fluoroscope's field of view ranged from 5 mT at the lower edge to 25 mT at the upper edge of the image. The helical magnetic microcatheter described in the "Helical magnetic microcatheter and guidewire design" section was used and advanced with the advancer unit presented in the "Advancer unit design" section. The intervention was performed by an experienced interventional neuroradiologist. Navigation was done at the bedside. All system inputs and outputs were recorded for analysis. The mCR navigation was performed under fluoroscopic control using roadmaps in different extracranial pig vessels. Angio runs were performed using contrast agent (Ultravist, Iopromid, Bayer Vital GmbH, Germany) saline solution.

Supplementary Materials

This PDF file includes:

Methods

Figs. S1 to S3

Table S1

Other Supplementary Material for this manuscript includes the following:

Movies S1 to S4

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body” covering the presented helical device. R.D., S.L., and B.J.N. are authors on patent application EP22177147 “Steerable device for use inside of a mammalian body” describing the magnetic articulating tip technology. C.C. is chief technology officer and cofounder, and B.J.N. is mentor and cofounder of the company Nanoflex Robotics AG, Zurich, Switzerland. B.J.N. is the chief scientific advisor of *Science Robotics*. The other authors declare that they have no competing interests. **Data availability statement:** All data needed to support the conclusions of this manuscript are included in the main text or Supplementary Materials. The data and code for this study have been deposited in the database 10.3929/ethz-b-000651465.

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Dexterous helical magnetic robot for improved endovascular access

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Editor's summary

Acute ischemic stroke can lead to long-term debilitating conditions and death, but, if patients are treated promptly, then their chances of survival and recovery increase exponentially. The challenge for clinicians is accessing the affected blood vessel in the brain. Because of the complex network and architecture of these blood vessels, there is still an unmet need for approaches that enable effective navigation of catheters to remove the blood clot. Dreyfus *et al.* report on a highly dexterous articulated continuum robot that is magnetically steered to improve the navigation of catheters in tortuous blood vessels. The device enabled successful endovascular navigation from the aorta to millimeter-sized cranial arteries in vivo, demonstrating its potential for atraumatic access to occluded vessels in the brain. —Amos Matsiko

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