

A portable and flexible intermediary patch for in vivo magnetic localization

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Safe operation in visually obstructed anatomy relies on accurate in vivo localization of medical devices, yet current magnetic systems struggle to adapt across device types and clinical environments. Here we present a portable and flexible patch that functions as an intermediary in a dual-stage localization paradigm of a source–sensors–source configuration, enabling three-dimensional localization in vivo with mean positional errors under 300 μm and orientation errors under 0.3°. The patch can be customized in shape and size to accommodate diverse medical scenarios requiring precise localization. We validate its flexibility and versatility through systematic characterization, and through in vitro and in vivo studies in vascular and gastrointestinal surgery, where the system maintained continuous magnetic capsule tracking for 6 hours. Our magnetic localization patch could potentially broaden the compatibility and adaptability of magnetic localization across various medical applications.

Accurate in vivo localization of medical devices is essential for many clinical procedures, especially in scenarios where visibility is obstructed by anatomy^{1–4}. Precise localization enables surgeons and clinicians to navigate complex environments, enhancing both the safety and efficacy of interventions. Traditional localization technologies such as computed tomography (CT)⁵, magnetic resonance imaging (MRI)⁶ and digital subtraction angiography (DSA)⁷ provide high-resolution views of internal structures but are limited by their static nature, exposure to ionizing radiation (for CT and DSA), two-dimensional positional information (for DSA), low real-time performance (for CT and MRI), and the need for large and immobile equipment⁸. Optical imaging, while offering high sensitivity in certain contexts, suffers from limited penetration depth, particularly in dense tissues^{9–11}. Magnetic localization, in contrast, offers a unique advantage by providing real-time, three-dimensional (3D) spatial tracking without the need for ionizing radiation or complex infrastructural requirements^{12–17}.

Despite these advantages, current magnetic localization systems face several limitations (Supplementary Note 1). Many depend on bulky external equipment and are prone to interference from surrounding magnetic materials, restricting their use in dynamic surgical environments¹⁸. In addition, the need for a controlled, non-magnetic setting often constrains the working area and reduces operational flexibility¹⁹. Furthermore, most systems use a one-stage localization principle between the magnetic source and receiver, where accuracy depends on source strength and the working area is constrained by receiver size^{19,20}. These limitations have hindered the widespread adoption of magnetic localization in routine clinical practice.

In this work, we present a portable and flexible intermediary patch capable of highly accurate, 3D in vivo magnetic localization (Fig. 1a–c). Inspired by the principle of an assisted global positioning system (A-GPS)²¹ (Fig. 1d), where the satellites determine the exact location of the handset in poor signal conditions by base stations (Fig. 1d, left),

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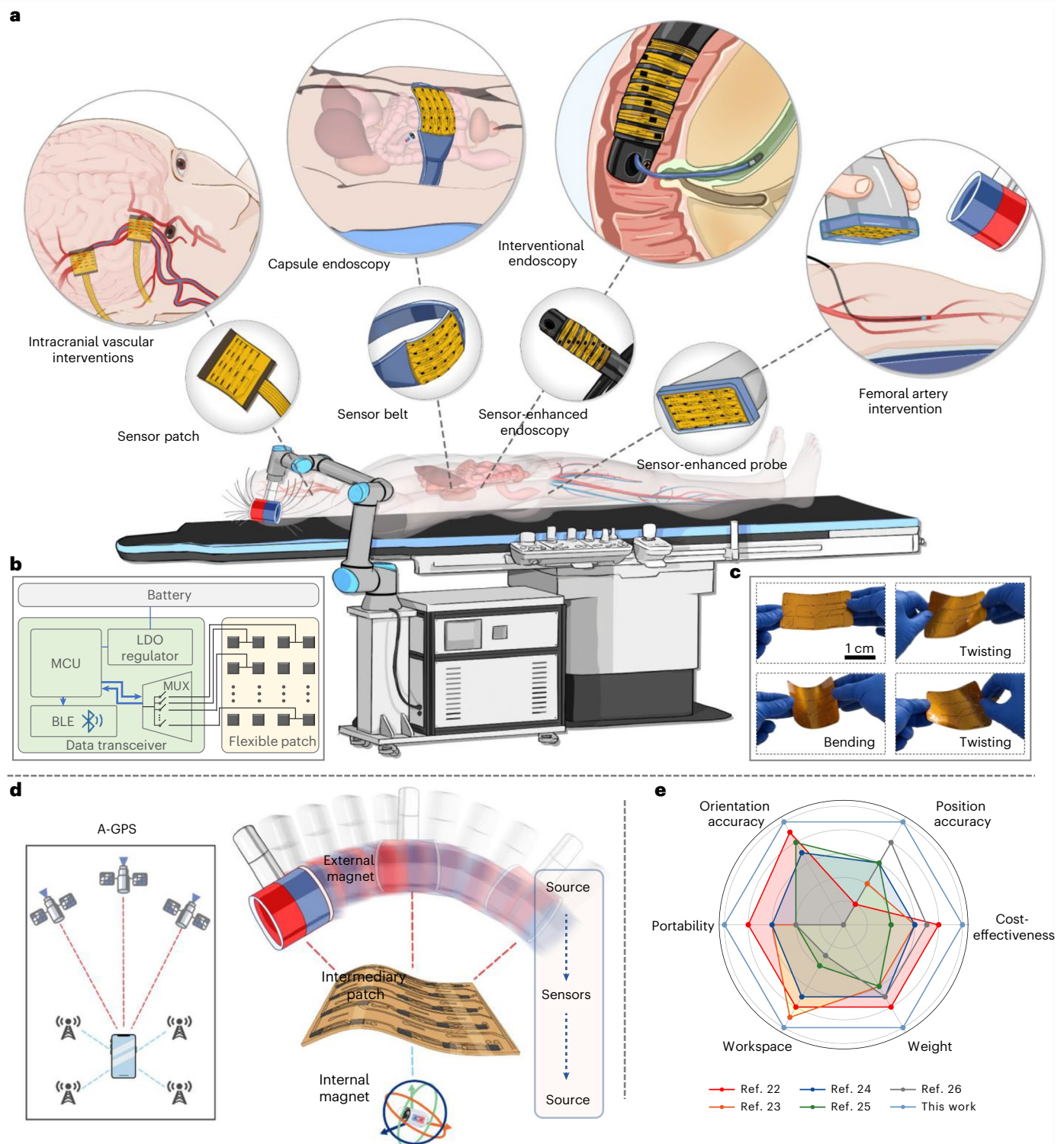


Fig. 1 | Overview of the flexible intermediary patch for dual-stage magnetic localization. a, Applications of the flexible intermediary patch for localization in diverse in vivo scenarios. **b**, Block diagram of the portable and flexible magnetic localization system. MCU, microcontroller unit; LDO regulator, low-dropout regulator; BLE, Bluetooth Low Energy; MUX, multiplexer. **c**, Demonstration of flexibility and deformability through bending and twisting. **d**, The dual-stage

magnetic localization paradigm, inspired by the working principle of A-GPS, features a source–sensors–source configuration. **e**, Comparison with other advanced magnetic localization techniques^{22–26} across six dimensions: position accuracy, orientation accuracy, portability, workspace, weight and cost-effectiveness.

the proposed dual-stage localization paradigm, also using a source–sensors–source configuration (Fig. 1d, right), dramatically expands the localization space and flexibility. The patch, designed with a Hall sensor array on a flexible printed circuit board (FPCB), can be customized to

various shapes (Supplementary Note 2 and Supplementary Fig. 1), accommodating a wide range of medical scenarios that require precise localization of the working end. This system eliminates adverse effects due to the sensor array's geometric size, enabling seamless integration

with existing operating room hardware. Furthermore, the cascaded localization paradigm enhances the flexibility and specificity of conventional magnetic localization, substantially expanding its potential applications across diverse surgical settings. Comparative results in Fig. 1e highlight the intermediary patch's superior performance in accuracy, workspace, cost-effectiveness, weight and portability relative to other advanced magnetic localization systems^{12–14,16,19,20,22–39} (Supplementary Table 1). Our innovative introduction of the flexible form of magnetic sensor array and dual-stage paradigm holds promise for further advancing magnetic localization technology as a high-precision and versatile solution for interventional medicine.

Results

Principle, design, and characterization of the intermediary patch

Drawing inspiration from the principle of A-GPS (Fig. 1d), which enables signal transmission over extended distances, we propose a dual-stage localization paradigm based on a source–sensors–source configuration to overcome the workspace limitations of current systems. The dual-stage magnetic localization algorithm operates in two phases. In the first phase (Supplementary Note 4), an external magnetic source determines the position and orientation of the intermediary patch within a global coordinate system. In the second phase (Supplementary Note 5), the intermediary patch, with its position and orientation established, locates the internal magnetic source at the working end of the medical instrument. Notably, the proposed dual-stage paradigm allows for the coexistence of two magnetic sources. The effects of both sources on the sensors can be theoretically decoupled, enabling the external source to serve simultaneously as an actuation source.

To accommodate diverse medical scenarios requiring precise localization, the intermediary patch must be lightweight, adaptable and compact, seamlessly conforming to anatomical or surgical instrument surfaces^{40–42}. Consequently, the patch is designed with only Hall sensors and filter capacitors on an FPCB, with the sensor number and density determined in advance through a reasonable trade-off between accuracy and computational speed (Supplementary Note 8). Benefiting from the flexible substrate and compact onboard chips, the patch endures bending strains beyond normal limits while preserving sensing stability, enabling robust dual-stage magnetic localization for diverse applications (Supplementary Note 9). Depending on the scenario, the patch can be tailored into various geometries: planar for guidewire navigation in interventional surgeries, arcuate for medical capsule monitoring, and spiral for precise selection during endoscopic procedures (Figs. 1a and 2a).

To ensure localization accuracy, both static and dynamic characterizations were performed (Supplementary Note 10). In the first phase (source–sensors), rectangular intermediary patches were fixed to the base with sinusoidal, planar and arc surfaces, as shown in Fig. 2b. The position and orientation of each Hall sensor were determined, yielding mean localization errors below 250 μm and 0.15° (Fig. 2c). Successful tracking of sinusoidal patterns validated the dynamic performance (Fig. 2d). In the second phase (sensors–source), the position errors of both the $\varnothing 1\text{ mm} \times 1\text{ mm}$ and $\varnothing 10\text{ mm} \times 10\text{ mm}$ internal magnets ($\varnothing$ denotes diameter) were maintained below 300 μm (Supplementary Fig. 13a,b), with successful localization of Lissajous patterns at varying frequencies further validating the second phase's dynamic performance (Fig. 2e). To validate the entire system's dynamic performance, all components (source–sensors–source) were moved along square, circular and triangular trajectories. As illustrated in Fig. 2f–h and Supplementary Fig. 13d, during continuous movement, the flexible patch achieved average localization errors of 120 μm and 0.18°, while the internal magnet maintained average localization errors of 170 μm and 0.26°.

To demonstrate the improved target localization enabled by flexible sensors, we designed the experiment in Supplementary Fig. 16,

where internal magnets ($\varnothing 1\text{ mm} \times 1\text{ mm}$, $\sim 3\text{ mm} \times 3\text{ mm}$) passed through a rigid circuit and a deformed flexible patch. Experimental results show that bending the patch substantially improves localization accuracy, reducing the localization error by more than 75%, enabling the detection of small-scale magnets, whereas the rigid array shows large errors and fails for the $\varnothing 1\text{ mm} \times 1\text{ mm}$ target. This is due to the flexible design allowing the array to be closer to the target, which is particularly valuable for medical micro-magnetic robots with weak magnetic fields and small magnetic moments, offering promising out-of-sight state feedback for such systems.

Consequently, tests confirmed that the proposed system, benefiting from the flexible design that allows closer placement to the target, achieves high accuracy across internal magnets of different sizes (1–10 mm), with average errors below 300 μm in position and 0.3° in orientation within their respective effective workspaces.

In vitro evaluation on vascular model

We further evaluated the localization capabilities through in vitro experiments simulating several common surgical scenarios (Supplementary Note 11). In a femoral artery intervention simulation, a $25 \times 15\text{ cm}$ phantom (Supplementary Fig. 18c), approximately 45-times larger than the magnetic sensor patch, was used. Under external magnetic-field actuation, the guidewire with a $\varnothing 2\text{ mm} \times 4\text{ mm}$ tip magnet showed flexible steering (Fig. 3a and Supplementary Fig. 18a,b), sequentially entering 6 branches of the deep femoral artery. Simultaneously, a probe equipped with the intermediary patch scanned 3 cm above the vascular model, tracking the guidewire tip. The reconstructed localization data (Fig. 3b) closely matched the guidewire's trajectory along all six pathways, indicating that the dynamic tracking capability substantially expands the working space and improves localization accuracy. In addition, the high consistency across the three repeated experiments further demonstrated the system's robustness (Supplementary Fig. 19). The comparison experiment with NDI Aurora further highlights the unique advantages of the proposed magnetic localization system in tracking magnetic guidewires and untethered robots (Supplementary Fig. 20).

In intracranial vascular interventions, where precise guidewire localization is critical to reducing the risks of brain vascular damage or infection⁴³, we tested the system in a human cerebral vasculature phantom (Fig. 3c). The intermediary patch was attached near the affected vascular region and was initially localized by an external magnet (Fig. 3d). Once mounted, the relative positions of the sensors within the array remain fixed. The guidewire, controlled through the internal carotid artery (ICA) and the vertebral artery (VA) into deeper cerebral vessels, was tracked simultaneously. Several simulated aneurysms were introduced, and the guidewire tip was directed into one for treatment—clearly reflected in the localization results (Fig. 3e). To verify the excellent consistency of the magnetic localization system, we conducted multiple repeated trials within the phantom (Supplementary Fig. 21). This highlights the system's versatility, providing a flexible and user-friendly approach to magnetic tracking.

In vitro evaluation on gastrointestinal model

To further demonstrate the system's versatility, we simulated a more technically demanding procedure—endoscopic retrograde cholangiopancreatography (ERCP)—on an in vitro model. ERCP typically requires an experienced physician to selectively cannulate the bile or pancreatic duct using continuous X-ray imaging⁴⁴, which provides only two-dimensional information, placing considerable demands on the physician's tactile precision and skill. With the compact and flexible intermediary patch design, we wrapped it around the tip of the endoscope (Fig. 4a), enabling it to serve as a localization device.

As demonstrated in Fig. 4d, the ERCP procedure consists of three sequential key stages. First, the endoscope is introduced through the oesophagus into the stomach, and upon further advancement into the

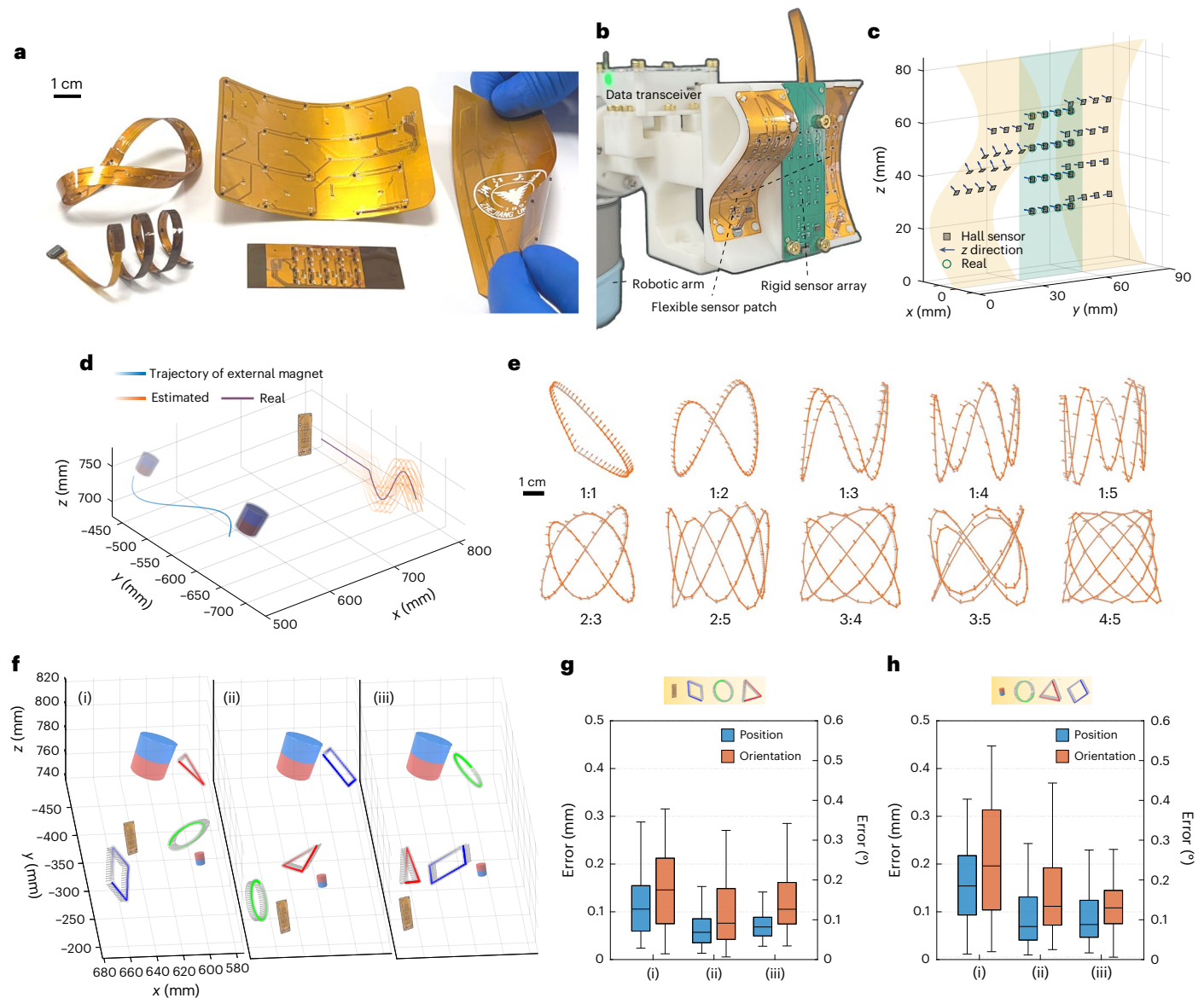


Fig. 2 | Characterization of the flexible intermediary patch. **a**, Different configurations of flexible intermediary patches. **b**, Hardware set-up of the sensors during the localization characterization experiment. **c**, Localization results of various configurations of intermediary patches (source–sensors). **d**, Dynamic localization of the patch while the patch and external magnetic source move simultaneously. **e**, Using a sensor patch to locate an internal magnet moving in Lissajous trajectories with varying frequency ratios (sensors–source). **f**, Magnetic-field separation and localization (source–sensors–source) as three components move along different trajectories simultaneously.

g, Position and orientation errors of the sensor patch along square, circular and triangular trajectories. The box edges represent the 25th and 75th percentiles, the horizontal line indicates the median, and the whiskers extend to the minimum and maximum errors. Each boxplot shows the distribution of position or orientation errors for $n = 60$ independent points along the corresponding trajectory. **h**, Position and orientation errors of the internal magnet along circular, triangular and square trajectories, containing $n = 60$ independent points along the corresponding trajectory for each boxplot. The data distribution of the boxplot is described in the same way as in **g**.

duodenum, it is positioned at the opening of the pancreatic and bile ducts. Next, the physician uses the endoscope to guide a guidewire into the target duct (either the bile duct or the pancreatic duct). Finally, the guidewire is precisely advanced within the duct to facilitate stent placement and the use of other therapeutic instruments. To validate the diverse and reliable localization capabilities enabled by the proposed flexible patch, we simulated these three critical steps in an in vitro model undergoing respiratory movements and achieved satisfactory results at each stage.

The endoscope, wrapped with the flexible patch, first entered the gastric model through the cardia and was then steered under tendon-driven control towards the pylorus, while an external magnetic source with a known position was applied. The relative position of the sensor had already been determined during the initial static positioning phase,

so during dynamic positioning, the sensor itself could also serve as a localized object, indicating the endoscope's current position and the path it has traversed. In this process, the externally fixed magnetic source was placed nearby to improve the signal-to-noise ratio and reduce the influence of Earth's magnetic field due to the endoscope's movement. As shown in Fig. 4c, regardless of whether in a respiratory or rest state, the localization results of the endoscope's tip within the gastric model closely matched the actual trajectory.

The endoscope was then advanced to the duodenum, and once it remained stable, the dynamic localization task of the sensors was terminated. The magnetic guidewire equipped with a $\varnothing 2 \text{ mm} \times 4 \text{ mm}$ tip magnet was successfully navigated into both the bile and pancreatic ducts (Fig. 4e) under external magnetic-field guidance. In this experiment, the rhythmic respiration caused periodic variations in

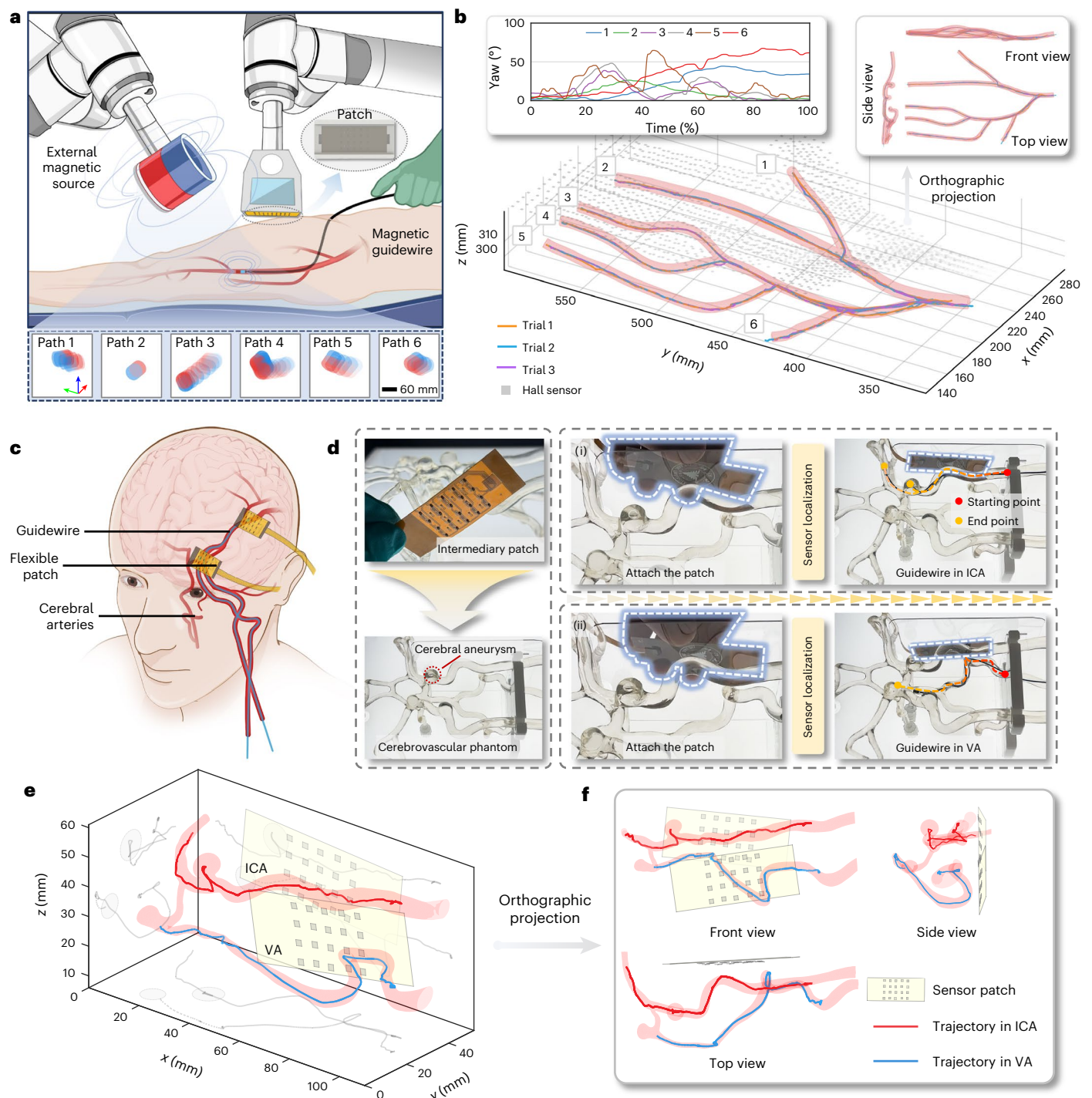


Fig. 3 | In vitro evaluation on vascular model. **a**, Schematic diagram of the intervention operation in the femoral artery and the trajectory of the external magnetic source steering the guidewire along different branches. **b**, The trajectories of the guidewire tip in three repeated experiments tracked by the probe with an intermediary patch. The top-left inset shows the variation in the

yaw angle of the guidewire tip at each vascular bifurcation. **c**, Schematic diagram of guidewire localization in the cerebral artery intervention. **d**, The intermediary patch adheres to the adjacent surface to localize the magnetic guidewire advancing in the ICA and the VA. **e, f**, Localization results of the trajectory in the ICA and the VA, with the guidewire passing through an aneurysm during the ICA stage.

the magnetic-field measurements. After identifying the respiratory frequency, we were able to eliminate this bias, enabling the magnetic localization system to perform well in the task of guidewire navigation even during respiration. The 3D localization results (Fig. 4f) clearly distinguished between the ducts, offering more intuitive and safer navigational assistance. In addition, the guidewire's movement within the digestive tract's optimal localization range, enabled by the sensors wrapped around the endoscope, ensured precise localization.

To visually demonstrate the localization of the guidewire during retrograde insertion into the bile or pancreatic duct, we designed a transparent model with nine curved, narrow channels that simulate the physiological structure of the human bile and pancreatic ducts. The endoscope, wrapped with a flexible patch, was placed in the centre of the model, while the guidewire was guided into different channels under the influence of an external magnetic source, advancing retrogradely (Fig. 4g). The results shown in Fig. 4h indicated that the

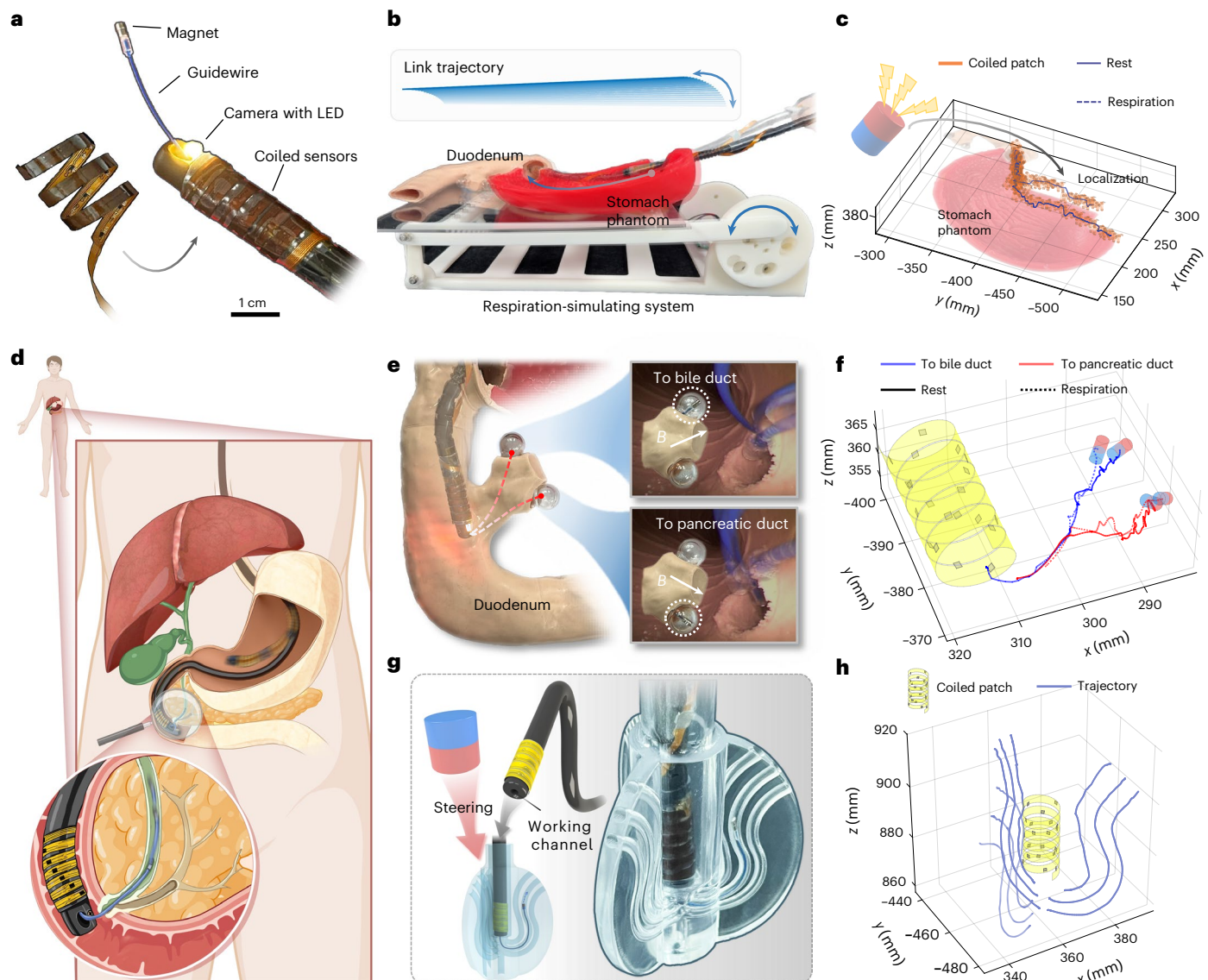


Fig. 4 | In vitro evaluation on gastrointestinal model. **a**, Endoscope wrapped with a flexible intermediary patch. LED, light-emitting diode. **b**, The endoscope is advanced into a stomach phantom that performs rhythmic movements to simulate respiration. **c**, Localization result of the endoscope's tip with coiled patches under respiration and rest conditions. **d**, Schematic diagram of the full ERCP procedure. **e**, The guidewire guided by the external magnetic field enters the bile and pancreatic ducts. *B* refers to the external magnetic field for guidewire

steering. **f**, Under respiration and rest conditions, the trajectory of the guidewire advancing into different ducts localized by the intermediary patch helically wrapped around the endoscope. **g**, Guidewire localization in a transparent model simulating retrograde bile and pancreatic ducts. **h**, The patch's position is determined using an external magnetic source, allowing the guidewire to move along the narrow channel of the model and be subsequently localized. The localized trajectories align closely with the actual model.

movement trajectory of the guidewire near the optimal working space was tracked with high accuracy.

In vivo demonstration on live pig

To validate the generalizability and accuracy of our magnetic localization system under realistic physiological conditions, we conducted in vivo experiments on two Bama pigs (weight, 30 kg; age, 8 months; Supplementary Note 12 and Supplementary Fig. 22). The procedures used in the study were reviewed and approved by the Institutional Animal Care and Use Committee at the Shenzhen Advanced Medical Services (AMS E2407008P).

First, we simulated retrograde guidewire localization in the small intestine of a pig (Fig. 5a). Under physician guidance, an endoscope wrapped with the flexible intermediary patch was inserted into the small intestine, and an external magnetic source localized the patch (Fig. 5b). When encountering sharp bends, a soft guidewire with the

same tip magnet as in the in vitro experiments was deployed, advancing smoothly along the intestinal wall into deeper regions. The guidewire's trajectory (Fig. 5c) closely matches DSA results (Fig. 5d and Supplementary Fig. 25a). Hall sensor measurements (Supplementary Fig. 23a) showed step-like patterns during X-ray imaging pauses, and the system's high accuracy enabled the detection of breathing-induced fluctuations while the pig was under anaesthesia.

We then performed a vascular intervention on the pig's femoral artery (Fig. 5e,f). The guidewire was inserted into the artery while the patch was adhered near the groin. The guidewire was intermittently advanced and DSA captured its position. Owing to the artery's straight segment, displacement was used to quantitatively assess movement (Fig. 5g). Over 240 s, the guidewire advanced 21 mm, with each plateau in the displacement curve corresponding to a DSA capture (Fig. 5h). Periodic fluctuations, with an amplitude of 0.19 mm and frequency of 12 cycles per minute, were observed during magnetic-field plateau

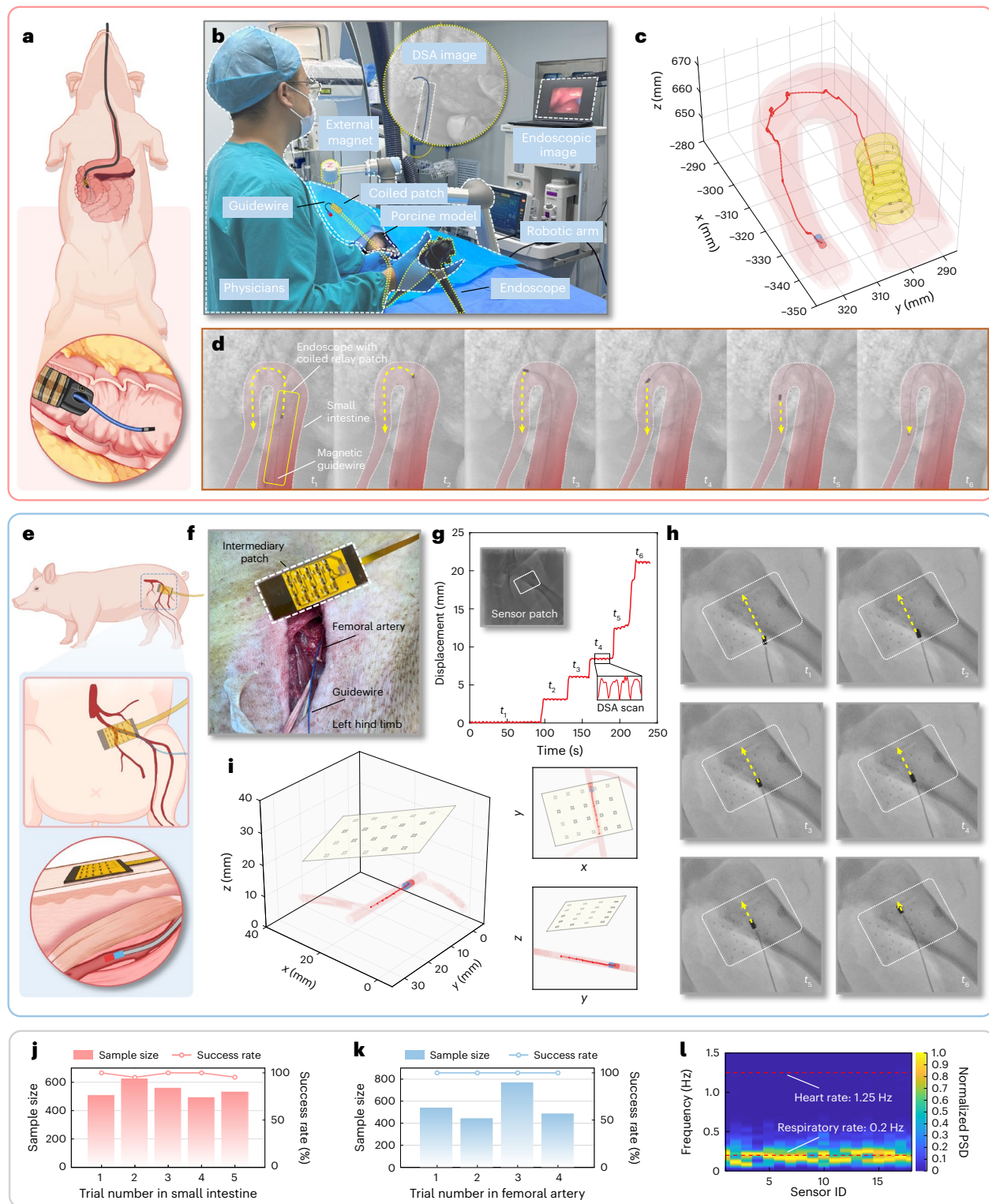


Fig. 5 | In vivo demonstration on live pig. **a**, Schematic diagram of the guidewire extended from the endoscope navigating a pig's gastrointestinal tract, in which the guidewire is localized by a coiled intermediary patch on the endoscope's tip. **b**, Image of the experimental set-up for guidewire exploration and localization within the intestine. **c**, The trajectory of the guidewire localized by the helical intermediary patch. **d**, The endoscope tip and the trajectory of the guidewire under DSA. t_1 – t_6 indicate successive time points within the experiment. **e**, Schematic of the guidewire inserted through the femoral artery in pigs, with a skin-mounted intermediary patch used for guidewire localization. **f**, The patch applied to the skin over the femoral artery in pigs. **g**, The displacement of the guidewire tip over time, and the upper-left panel shows the patch under X-ray

imaging. The plateau of the curve corresponds to the time when X-ray scans are performed, while the fluctuations in the curve reflect the pig's breathing during anaesthesia. **h**, DSA image of vascular intervention in the pig's femoral artery. The white dashed rectangles delineate the patch, and the yellow arrows indicate the trajectory of the guidewire tip. **i**, The localization results and spatial relationship between the patch and the guidewire trajectory. **j,k**, The sample size and success rate of multiple localization in the pig's small intestine (**j**) and femoral artery (**k**), with success rate determined based on whether the localization results showed distortion. **l**, The PSD of signals from multiple sensors. The frequency corresponding to the peak matched the respiratory rate, with no notable signal detected in the frequency range of the heart rate.

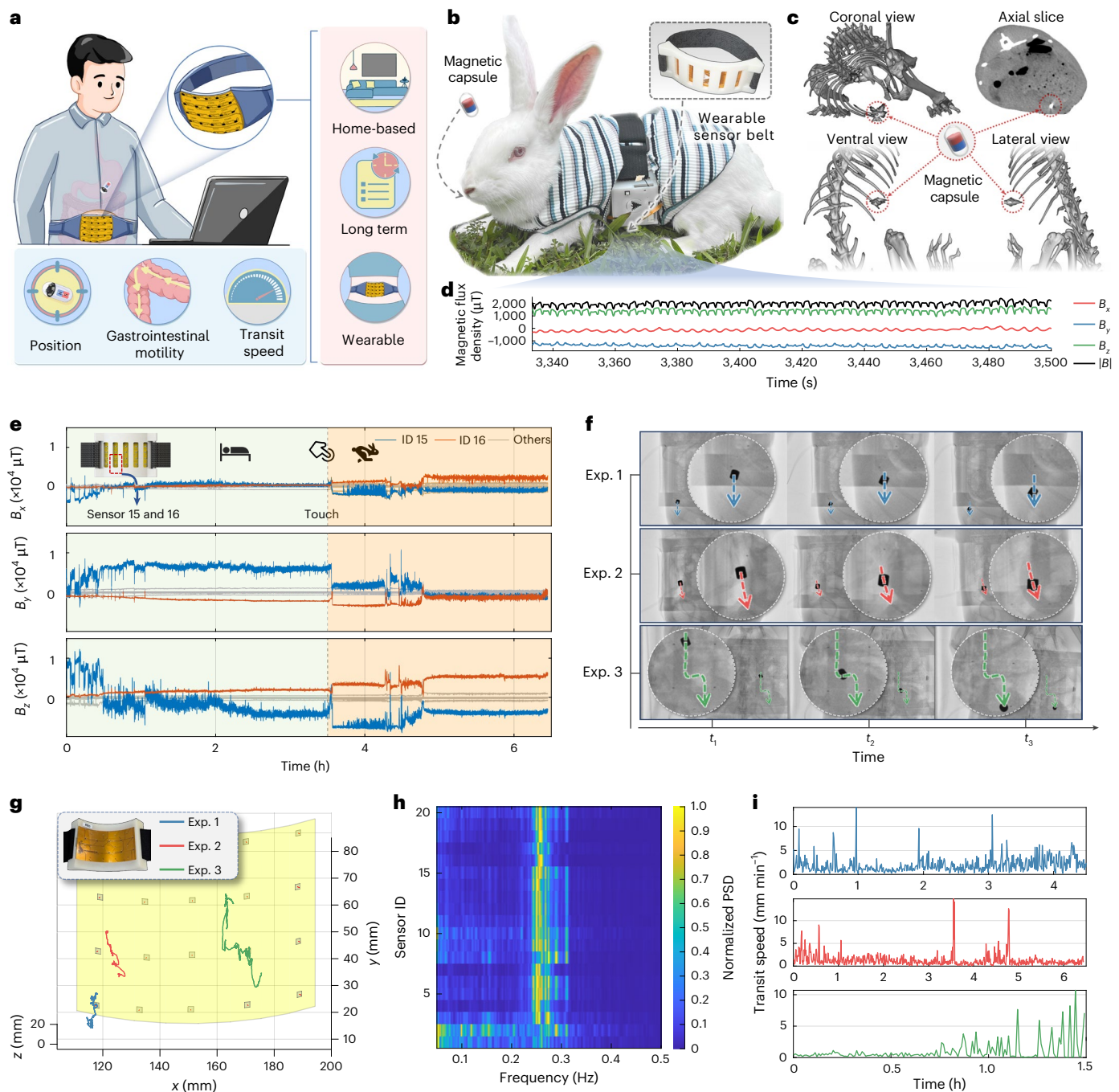


Fig. 6 | Wearable monitoring of gastrointestinal capsule. **a**, The advantages of using the flexible patch as a home-based, long-term and wearable monitoring device and its functionality in detecting gastrointestinal dynamic parameters. **b**, The experimental rabbit wearing a belt with a flexible sensor patch, designed to fit its body shape for comfortable wear. **c**, CT reconstruction image of the rabbit after swallowing the magnetic capsule. **d**, Magnetic-field reading from a single sensor in the patch during a resting period, showing periodic fluctuations. **e**, A long-duration recording of two sensors near the magnetic capsule during a single experiment. At 3.5 h, manual interference changed the gastrointestinal peristalsis, causing a sudden shift in the capsule's position and orientation,

leading to a jump in the magnetic-field measurements. **f**, The trajectory of the magnetic capsule under DSA during three experiments. The movement of the capsule is primarily driven by the gastrointestinal motility. t_1 – t_3 indicate successive time points within each experiment (absolute times differ across experiments). **g**, The localization results of the magnetic capsule and its relative position to the sensor patch. **h**, The normalized PSD was calculated for all sensors during a period of rest. The bands below the respiratory frequency could contain signal components associated with gastrointestinal peristalsis. **i**, The transit speed of intestinal contents in the rabbit across three experiments.

phases, corresponding to the pig's respiration. Figure 5i illustrates the relative position between the guidewire trajectory and the sensor patch during the vascular intervention process. The darker nodes along the curve are a result of data accumulation during X-ray imaging. When compared with the DSA images, it is evident that the results correspond very well with the X-ray fluoroscopic images.

We conducted multiple repeated experiments, analysing sample counts and localization success rates for each guidewire insertion (Fig. 5j,k). The results show consistently high accuracy, aligning with DSA images (Supplementary Figs. 26 and 27). Frequency-domain analysis in Fig. 5l reveals that power spectral density (PSD) peaks at the respiratory frequency, not the heart rate, due to greater gastrointestinal and

skin fluctuations caused by breathing. However, these rhythmic signals do not compromise the stability of the system. By using the external magnet to identify the vibration patterns of the sensors induced by respiration, most of their influence can be effectively eliminated (Supplementary Note 13).

Wearable monitoring of gastrointestinal capsule

To explore potential daily monitoring applications outside the operating room, we designed a 65 × 95 mm intermediary patch embedded in a custom 3D-printed belt for 3 New Zealand rabbits (weight, 2.5 kg; age, 5 months). Unlike current systems that require bulky external equipment, our system is lightweight, ergonomic and adaptable. These features give it advantages over traditional magnetic tracking devices, including the ability for home-based monitoring, long-term data recording and wearability. As a result, it not only enables precise localization of the in-body capsule endoscope but also allows for the inference of gastrointestinal dynamic parameters through extended monitoring data, such as gastric motility, intestinal content movement speed and other key physiological indicators (Fig. 6a). This provides valuable insights for clinical assessment of gastrointestinal function.

In the rabbit experiment, the flexible patch belt was worn to track a magnetic capsule (containing a $\varnothing$ 4 mm × 4 mm magnet) as it moved through the digestive tract via peristalsis (Fig. 6b). After the rabbit ingested the capsule, we performed a CT scan to confirm its initial position. Figure 6c shows CT reconstruction images of the magnetic capsule from various angles within the rabbit's stomach. Figure 6d shows the periodic fluctuations in the sensor's measurement during the resting state, corresponding to the regular movements of the capsule caused by intestinal peristalsis and respiration. We conducted 3 separate experiments on rabbits, with durations of 4.5 h, 6.5 h and 1.5 h, respectively. After the experiments, the capsule was naturally excreted with the rabbit's faeces.

We show a long-duration recording of two sensors near the magnetic capsule in Fig. 6e, reflecting the rabbit's physiological activity. Between 2.0 h and 3.5 h, the rabbit was calm, with slow gastrointestinal peristalsis and gradual capsule movement. Sensor 16's measurements steadily increased. At 3.5 h, manual interference made the rabbit more active, accelerating peristalsis and causing more rapid changes in the capsule's position and orientation, reflected in the increased magnetic-field fluctuations from 3.5 h to 4.5 h. Afterwards, the rabbit became calmer and the magnetic-field changes stabilized. During the three experiments, we captured multiple DSA images of the capsule's position inside the rabbit. Its trajectory is outlined in Fig. 6f with arrows in different colours. The three identified trajectories in Fig. 6g align well with the DSA images, demonstrating that this wearable localization system enables long-term monitoring and effectively indicates the capsule's location. This capability suggests extraordinary potential for applications beyond the surgical setting.

To fully explore the potential of the wearable magnetic localization system, we performed a detailed analysis of the long-term data collected. First, for the sensor measurements taken during a period of rest in the rabbit, we computed the normalized PSD across all sensors. Analysis revealed that the signal power was highest at frequencies between 0.24 Hz and 0.26 Hz and there are also additional peaks at lower frequencies below 0.2 Hz, occupying the brightest strip in Fig. 6h. This suggests that during periods of rest, the respiratory rate is approximately 15 cycles per minute, and the lower-frequency gastrointestinal peristalsis contributes to the PSD in the low-frequency range. In addition, we analysed the capsule's peristaltic velocity in all three experiments by calculating the relative displacement of the capsule per minute, as shown in Fig. 6i. These data also reflect the movement speed of intestinal contents, providing a useful metric for assessing overall gastrointestinal health. Moreover, this method could assist in the diagnosis and analysis of diseases affecting intestinal motility, such as intestinal obstruction, enteritis and Crohn's disease.

Discussion

The primary challenges of magnetic localization technology stem from its compatibility with the medical ecosystem, hindering its widespread adoption as a practical alternative for localization in the operating room. In this work, inspired by A-GPS, we developed a flexible intermediary patch and a dual-stage localization paradigm to achieve targeted, high-precision tracking of the working end across various medical scenarios. This approach unlocks the substantial potential of magnetic localization technology for a range of surgical applications.

In practical applications, the flexible intermediary patch is designed to be customized in size and shape and manufactured with an FPCB, enabling it to seamlessly attach to the area of interest or integrate with existing medical devices. This encompasses adhering to the skin, integrating into a probe or even wrapping around the tip of an endoscope. The flexible arrangement of the patch establishes an innovative magnetic localization paradigm with a source–sensors–source configuration. Compared with existing magnetic localization systems, this approach offers a more compact design, a more portable set-up and a more flexible working space.

In characterization experiments, we demonstrated a localization error of less than 300 μ m in position and 0.3° in orientation across internal magnets of different sizes (1–10 mm) within respective effective workspaces. To further validate the compatibility of the intermediary patch, we successfully used this system to locate the working ends in various in vitro and in vivo scenarios, including vascular interventions and gastrointestinal treatments. Notably, this magnetic localization technology shows promise as a more optimal alternative in procedures such as intracranial vascular interventions and ERCP, where patients are often exposed to prolonged X-ray radiation.

However, magnetic localization systems still face several pressing challenges that must be addressed in future research. These include interference from external magnetic-field noise and the limited strength of current magnetic materials. It should also be noted that the proposed patch provides mounting flexibility, but is not intended to operate under active bending during measurements, which would introduce additional degrees of freedom and compromise localization accuracy. Respiratory vibrations and deformations may also affect the patch's clinical performance, and while mitigation is possible through pattern recognition or structural reinforcement, more efficient initialization and advanced compensation algorithms are still needed.

The deformation of the flexible patch may also encounter a common challenge in flexible electronics: the degradation of onboard chip performance with increasing strain. Although the impact observed in this work is not notable, it remains an important factor to be carefully considered for broader and more reliable applications. Where necessary, its influence can be mitigated through strategies such as integrating stress sensors for online compensation^{45,46}, adopting strain-decoupled island structures^{47,48} or applying offline compensation based on estimated strain.

We envision that future work should not only address deformation- and motion-related limitations but also pursue further integration and standardization of magnetic sensor patches to enhance their versatility. With greater adaptability and scalability, these systems could become valuable tools across a broad range of disciplines beyond medical applications.

Methods

Experimental design

This study was designed to evaluate the effectiveness of the proposed flexible intermediary patch and the dual-stage magnetic localization method in clinical surgical applications. The procedures used in the study were reviewed and approved by the Institutional Animal Care and Use Committee at the Shenzhen Advanced Medical Services (AMS E2407008P). New Zealand rabbits (weight 2.5 kg; age, 5 months) and female Bama pigs (weight 30 kg; age, 8 months) were used in the

experiments to evaluate the feasibility of the flexible patch and magnetic localization method in a clinical setting. All animals were housed in individually controlled environments to minimize external influences on the experimental results. Before the experiment, animals underwent a 7-day quarantine and acclimation period, during which their health was monitored to ensure no pathological changes. The detailed surgical procedures are described in Supplementary Note 12. Post-operative monitoring was conducted to assess the animals' recovery and ensure that the procedures did not adversely affect their health. Further analysis of collected data was performed to evaluate the precision and clinical applicability of the intermediary patch.

Overview of the proposed magnetic system

Our magnetic localization system comprises three main components: the sensor patch, the data transceiver module and the external magnetic source. The sensor patch is made from a two-layer FPCB with a polyimide or polyester film substrate and a thickness of 0.16 mm (Supplementary Fig. 1d–f). Depending on the usage scenario, it can be designed in various sizes and shapes, such as spiral, planar rectangular or spiral, and can withstand considerable bending and twisting deformation (Supplementary Video 1). This adaptability makes it well suited for integration into existing medical environments. The FPCB is uniformly arranged with multiple three-axis devices (AK09973D, Asahi Kasei Microdevices). These sensors feature a compact package (1.18 mm × 0.78 mm × 0.55 mm), a wide measurement range (±36 mT) and satisfactory sensitivity (1.1 μT per least significant bit). In the typical operating range of our magnetic localization system (0.1–20 mT), the signal-to-noise ratio spans from approximately 25 dB at 0.1 mT to 71 dB at 20 mT, and can be further improved by temporal averaging or filtering improvements, thereby enabling the patch to detect weak magnetic-field signals in constrained spaces, facilitating high-precision localization. A thin FPCB interface is designed at the edge of the sensor patch for signal transmission between the sensors and the data transceiver module using the standard I2C protocol. In addition, polyimide reinforcement is used to provide stable support in areas that require repeated bending or to withstand higher stress.

The data transceiver module connects to the flexible patch via a standard 13-pin flexible printed circuit cable. It features a microcontroller (STM32F407) and a low-power Bluetooth chip (E104-BT5005A), which wirelessly transmits the filtered magnetic-field data to the host computer (Supplementary Fig. 1a,b). For the external magnetic source, we use a 60 mm × 60 mm axially magnetized N52 cylindrical permanent magnet, which is securely mounted on the end of a UR5 robotic arm using a specially designed 3D-printed fixture. Similarly, for the internal targets attached to the working end of surgical instruments, we use axially magnetized N52 cylindrical magnets, with sizes ranging from 1 mm to 10 mm in diameter.

Workflow of the dual-stage magnetic localization

Inspired by the working principle of A-GPS, where users are positioned through the collaboration of base stations and satellites²¹, we propose a dual-stage localization process based on a source–sensors–source configuration. In clinical surgeries, the external magnetic source first determines the pose of the sensor patch, which serves as an intermediary and is placed by the physician as needed. Then, the intermediary patch with a known pose identifies the pose of the target magnetic source attached to the working end of the medical instrument (Supplementary Notes 4 and 5).

The fundamental principle of this dual-stage localization process is to solve for the optimal pose of N magnetic-field sensors $[\mathbf{u}_s^i, \boldsymbol{\theta}_s^i] \in \mathbb{R}^6, i \in \{1, \dots, N\}$, and the pose of internal magnetic source $[\mathbf{u}_m, \boldsymbol{\theta}_m] \in \mathbb{R}^6$ such that the magnetic field calculated by the model⁴⁹ matches the field measured by the sensors, where i indexes the sensors

and the pose is parameterized by position $\mathbf{u}$ and orientation $\boldsymbol{\theta}$. This involves unifying multiple magnetic-field vectors under the sensor's coordinate system, ensuring that the following condition is satisfied:

$$s_i \mathbf{B}_s^{ij} = s_i \mathbf{B}_M^{ij} + s_i \mathbf{B}_m^{ij} + s_i \mathbf{G}$$

where $\mathbf{B}_M$ represents the contribution of the external magnetic source to the total measurement of the magnetic-field sensor, $\mathbf{B}_m$ represents the contribution of the internal magnetic source to the total measurement of the magnetic-field sensor, $\mathbf{G}$ is the environmental magnetic field and other biases, s_i refers to the i -th sensor and j denotes the index of the sampling point. The superscript indicates that the vector is expressed in the corresponding coordinate system.

The optimization objective for the first stage of localization with k sampling points is defined as follows:

$$F_1(\mathbf{u}_s^i, \boldsymbol{\theta}_s^i) = \sum_{j=1}^k \left\| \left(s_i \mathbf{B}_s^{ij} - s_i \mathbf{B}_s^{il} \right) - \left(s_i \mathbf{B}_M^{ij} - s_i \mathbf{B}_M^{il} \right) \right\|$$

where differentiation is applied to eliminate magnetic-field components not originating from the external magnetic source.

It is worth noting that in the first stage of localization, due to the highly nonlinear nature of the objective function, we need to specify appropriate initial values to avoid the optimization process getting trapped in local minima (Supplementary Fig. 7a). To address this, we developed a method to acquire the initial values while collecting data for optimization. The magnetic gradient tensor can be used to calculate the normalized source strength⁵⁰, which is inversely proportional to the fourth power of the distance and can be obtained by the eigenvalues of the magnetic gradient matrix $\mathcal{H}$, as expressed below:

$$\mu = \sqrt{-\lambda_2^2 - \lambda_1 \lambda_3} = \frac{3\mu_0 m}{4\pi r^4}, \lambda_1 \geq \lambda_2 \geq \lambda_3$$

where λ denotes the eigenvalue of $\mathcal{H}$, μ_0 is the vacuum permeability, m is the magnetic dipole moment and r is the distance between the measurement point and the magnetic source. The magnetic gradient matrix $\mathcal{H}$ can be determined by minor movement of the external magnetic source. Repeating this process at least three times enables triangulation to estimate an approximate position, which can be used as the initial value for optimization. In this process, each Hall sensor on the patch is calculated individually. Therefore, theoretically, regardless of how the patch is bent into different shapes, the complexity of sensor localization and calibration will not be affected.

The optimization objective for the second stage of localization, after subtracting the environmental magnetic field as a bias, can be expressed as follows:

$$F_2(\mathbf{u}_m, \boldsymbol{\theta}_m) = \sum_{i=1}^N \left\| \overline{s_i \mathbf{B}} - s_i \mathbf{B} \right\|$$

where $\overline{s_i \mathbf{B}} = s_i \mathbf{B}_M + s_i \mathbf{B}_m$ is the ideal magnetic flux density calculated by the magnetic-field distribution model. The overline signifies that the value is an estimation.

Optimization algorithm

In the optimization process, we used the Levenberg–Marquardt algorithm⁵¹, which dynamically adjusts the damping factor, combining the advantages of both the Gauss–Newton method and gradient descent. We implemented this using MATLAB R2023a's `lsqnonlin` function, explicitly specifying the Levenberg–Marquardt algorithm as the solver. It is worth noting that by default, `lsqnonlin` estimates gradients using finite differences, which is less efficient than being provided by exact gradients. To ensure real-time optimization, we derived the analytical Jacobian matrices for each optimization step. Experimental results

show that providing the analytical Jacobian improved the optimization speed by 41.1%. The main aspect of deriving the analytical Jacobian matrix involves calculating the partial derivatives of the rotated magnetic-field model to the optimization variables. The detailed derivation process is provided in Supplementary Note 6.

Operating range of the robotic arm

The operational planning of robotic arms in the operating room is crucial for ensuring safety and efficiency. In this study, we analysed both single-arm and dual-arm configurations, considering collision constraints, the effective detection range of sensors, and the workspace division for simultaneous magnetic actuation and localization (Supplementary Fig. 24). These factors are essential for optimizing robotic assistance in surgical procedures.

Calibration of the experimental set-up

To avoid introducing additional errors from the experimental set-up, we first calibrated all hardware components, including the robotic arms and magnetic sources. Up to three robotic arms were used in the characterization experiments, and their positional relationship is critical. However, owing to installation tolerances, directly measuring the relative poses based on the mounting distances on the optical platform could result in millimetre/degree-level errors. To address this, we simultaneously moved the end-effectors of the three robotic arms to several predefined positions, capturing the representation of the same point cloud in the coordinate systems of the three arms. Therefore, using this data, we applied iterative closest point⁵² to obtain a more accurate transformation between the arms' coordinate systems.

On this basis, we further optimized the calibration of the magnetic source's actual position and tilt angle within the fixture, along with other correction parameters (Supplementary Note 7). This was achieved through a grid-based sampling method using the robotic arms. We also discovered that the calculated magnet's magnetization varied across different regions, suggesting that calibration should be repeated for different operating distances in various scenarios.

Fabrication and design of the instruments equipped with the flexible patch

In addition to using the sensor patch independently for localization, we have designed several supporting hardware components to expand its application scenarios (Supplementary Fig. 17): a handheld probe, an endoscope and a belt. The handheld probe's casing is made of 3D-printed material, featuring a circular hole for easy grip. We integrated a wireless data transceiver into the probe, powered by a battery, and placed the sensor patch on the flat end of the probe, covered by a transparent plastic film to protect the circuit board.

In addition, we designed a simple tendon-driven endoscope to enhance the functionality of the flexible intermediary patch. A mini OVM6946 camera with a light-emitting-diode light is mounted on the side of the endoscope tip, along with a working channel equipped with a guiding track. The side-facing camera and working channel opening facilitate observation and guidewire navigation for targeted treatment of side-facing lesions in the gastrointestinal tract. The flexible strip-shaped patch is evenly wrapped around the endoscope's tip.

The design of the sensor belt includes a custom mould for attaching the flexible sensor patch and an elastic strap. The mould is designed to be as thin and lightweight as possible to ensure maximum comfort for the wearer.

Host and embedded software implementation

In the project, the embedded software runs on the data transceiver module, responsible for reading data from the sensor chip and transmitting it to the host computer. We used STM32CubeMX to configure the chip and generate the initialization code. The main functionality

involves selecting sensors based on the hardware's multiplexer and reading the magnetic-field measurements from each sensor via the I2C protocol.

In addition, in the lower-level system, we applied a moving average filter with a window size of 10 to the sensor readings before packaging the data into frames and transmitting it to the host computer. The host software, developed using MATLAB R2023a's App Designer, handles communication with the lower-level system via serial or Bluetooth, processes, and saves the data.

Reporting summary

Further information on research design is available in the Nature Portfolio Reporting Summary linked to this article.

Data availability

The main data supporting the results in this study are available within the paper and its Supplementary Information. Source data are provided with this paper.

Code availability

The code for data acquisition and the detailed implementation of the dual-stage magnetic localization algorithm are available at <https://github.com/xpy-zju/Dual-stage-magnetic-localization>.

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Conceptualization: H.L. and T.X. Methodology: P.X. and D.S. Investigation: P.X., D.S. and G.M. Visualization: P.X., G.M. and H.Z. Software: P.X., D.S. and H.Z. Funding acquisition: Y.W., H.L. and T.X. Project administration: H.L. and T.X. Supervision: H.L., T.X., R.X. and Y.W. Writing—original draft: P.X., D.S. and G.M. Writing—review and editing: H.L. and T.X.

Competing interests

The authors declare no competing interests.

Additional information

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Data collection	Data collection in this study was performed using custom scripts written in MATLAB R2023a. STM32CubeMX was used to configure the chip and generate initialization code in the data transceiver module.
Data analysis	Data analysis and plotting were done with MATLAB R2023a. The code for data acquisition and the detailed implementation of the dual-stage magnetic localization algorithm are available at: https://github.com/xpy-zju/Dual-stage-magnetic-localization .

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Sample size	For all experiments, sample sizes were determined based on the duration of data collection and the need to cover the relevant workspace and operating conditions. Sensors sampled at 6–12 Hz, yielding hundreds to thousands of measurements per condition, with repeated trajectories until localization errors stabilized. For in vivo studies (2 Bama pigs and 3 New Zealand rabbits), multiple trajectories were performed in each animal. Given the engineering focus and consistent performance across trials, we consider these sample sizes sufficient to demonstrate the effectiveness of the method.
Data exclusions	No data were excluded from the analyses.
Replication	All in vivo and in vitro experiments were performed at least three times, all of which were successfully completed.
Randomization	In vitro, the soft model was placed arbitrarily. In vivo, the animal's position was randomly determined by medical staff, and the magnet's trajectory was freely chosen. Each sample is individual and there is no need for group allocations.
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Wild animals	The study did not involve wild animals.
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Field-collected samples	The study did not involve samples collected from the field.
Ethics oversight	The procedures used in the study were reviewed and approved by the Institutional Animal Care and Use Committee at the Shenzhen Advanced Medical Services Co., Ltd (AMS E2407008P).

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