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A magnetically controlled capsule robot for dual-target constant-rate drug delivery and stable anchoring

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Abstract: Gastrointestinal (GI) drug delivery remains challenging because sustained constant-rate infusion and positional stability are difficult to achieve simultaneously. Conventional magnetic capsules often rely on burst-type or nonlinear release mechanisms. To address these limitations, a magnetically controlled dual-target constant-rate drug delivery capsule system (M-DCDC) is presented. The system enables highly linear and quantitatively adjustable drug infusion through specialized screw–piston transmission. This mechanism establishes a direct relationship between the magnetic rotation frequency and the delivery rate, achieving a maximum flow rate of 0.1 mL/s with predictable dosage control. Sequential dual-target delivery is achieved through functional decoupling. By reversing the direction of the magnetic field, two independent reservoirs, each with a volume of 0.6 mL, can be selectively actuated at separate target sites. To resist intestinal peristalsis, a self-locking mechanical anchoring module is integrated to maintain positional stability without continuous power consumption. Analytical modeling, finite element analysis (FEA), benchtop tests, and ex vivo experiments using porcine intestines demonstrate that M-DCDC can withstand a peristaltic force of 0.32 N while achieving localized staining. By combining stable anchoring with decoupled constant-rate infusion, the proposed system offers an effective engineering approach for targeted treatment of multifocal GI diseases.

Key words: Capsule robot; Magnetic actuation; Targeted drug delivery; Active anchoring

1 Introduction

Gastrointestinal (GI) diseases often exhibit multifocal involvement and chronic progression, and effective treatment frequently depends on achieving sufficiently high local drug concentrations. These characteristics pose substantial challenges for existing drug delivery strategies (Chu and Traverso, 2022). Although conventional oral preparations cause less discomfort, they often fail to maintain effective drug levels at target sites because of gastric degradation, enzymatic breakdown, and variable intestinal absorption (Hua, 2020). Systemic administration via intravenous injection can bypass some absorption barriers but is commonly associated with off-target exposure and dose-limiting adverse effects (Ezike et al., 2023). These limitations have stimulated growing interest in device-assisted approaches for localized GI drug delivery.

Among these approaches, magnetically controlled capsule systems have gradually evolved from a diagnostic modality into a multifunctional platform capable of integrating imaging and

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therapeutic functions (Chen et al., 2022; Cao et al., 2024). Recent studies have explored the incorporation of drug delivery modules into capsule systems to improve the spatial control of drug administration (Fu et al., 2022; Yang et al., 2025). From a drug delivery perspective, existing magnetically actuated capsule systems still face several practical limitations, including limited controllability of release rates, insufficient capacity for delivery to spatially distinct sites, and unstable positioning under intestinal fluid dynamics and peristalsis.

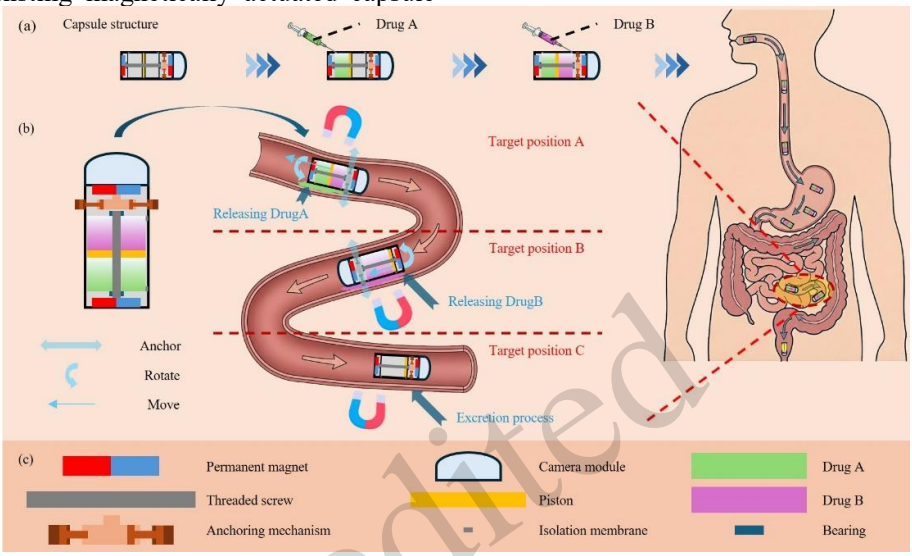

Fig. 1 Structural design, function and application of M-DCDC. (a) The structure of M-DCDC and the drug injection process. (b) Schematic illustration of M-DCDC anchoring and multitarget drug release in the intestinal tract. (c) Simplified structural composition of M-DCDC

Regarding drug delivery mechanisms, current systems remain limited in their ability to achieve precise, sustained, constant-rate infusion. Passive approaches, including chemically triggered systems and soft-body deformation–based mechanisms (Esendag et al., 2024), typically exhibit nonlinear release profiles characterized by rapid initial release followed by a gradual decline in release rate (Sun et al., 2023). Such time-dependent release behavior complicates dose prediction and limits their suitability for dose-sensitive therapies (Cai et al., 2024). Although active mechanisms based on springs, torsional elements (Cao et al., 2025), or shape-memory alloys can provide mechanical actuation, they often operate in a discrete or burst-like manner rather than enabling sustained and adjustable flow control (Zheng et al., 2023; Zhang et al., 2025). In addition, capsule systems originally designed for fluid sampling may generate insufficient or poorly regulated driving pressure, further restricting their applicability to controlled drug infusion (Wu et al., 2025).

Beyond release-rate control, spatially selective drug delivery remains a major challenge for magnetic capsule systems, as most existing designs support only single-drug, single-target operation. Although multichamber capsules can increase payload capacity (Valdastri, 2025), the lack of effective functional decoupling limits the independent control of multiple drug compartments (Li et al., 2025; Ye et al., 2025), making sequential or region-specific dual-drug delivery difficult to achieve (Zheng et al., 2022; Wang et al., 2025). Moreover, intestinal peristalsis and fluid motion can displace capsules during delivery, thereby reducing localization accuracy and reliability. Although sensor fusion and localization algorithms can improve tracking performance (Bianchi et al., 2022; Zhang et al., 2022; Zhang et al., 2024), they do not provide physical stabilization (Wang et al., 2021). Existing anchoring approaches either carry a risk of mucosal injury or are difficult to integrate and decouple within the limited capsule volume (Song et al., 2022; Yu et al., 2025). These challenges are further exacerbated by the limited and variable efficiency of wireless power transfer, which restricts the use of high-power or continuously

actuated stabilization mechanisms (Wen et al., 2024; Han et al., 2025).

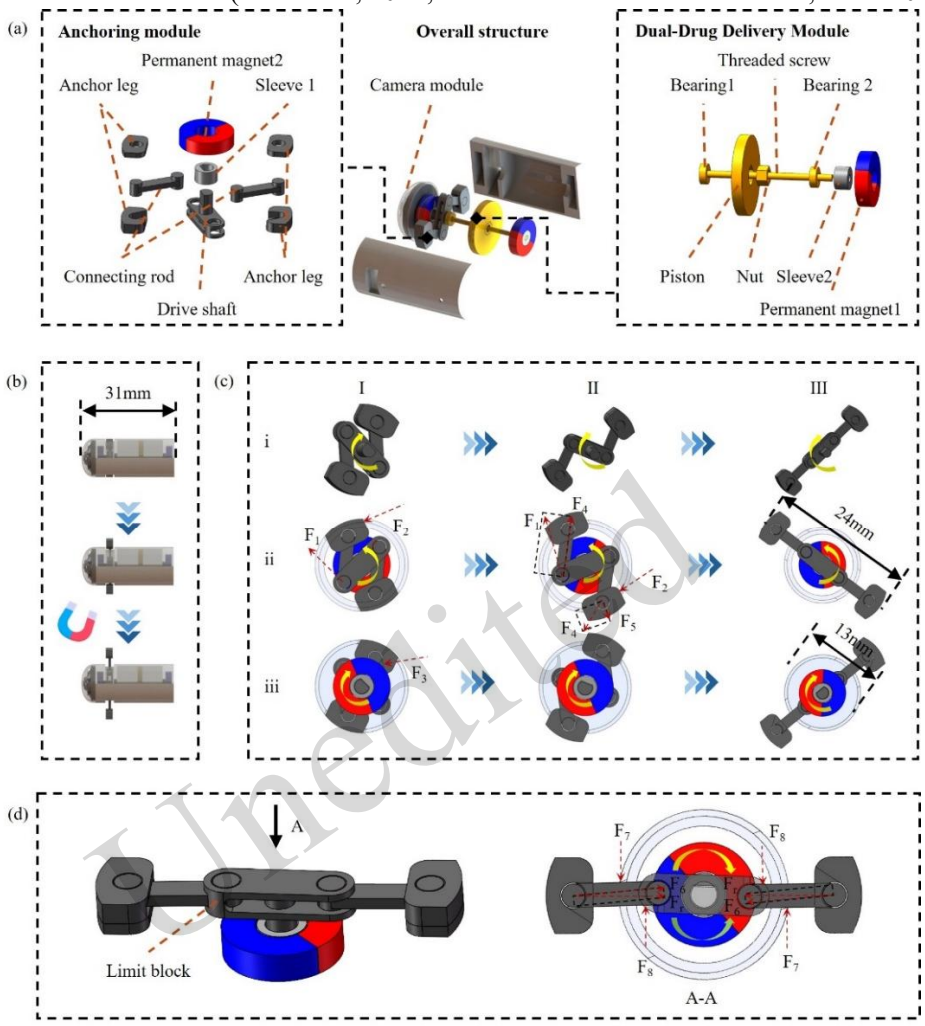

Fig. 2 Design of the M-DCDC. (a) Overall structure and detailed design of the M-DCDC. (b) Schematic diagram of the M-DCDC unfolding process from the contracted state to the anchored state; (c) Kinematics and force principles of the anchoring module: (I) Initial contracted state; (II) During the drive-deployment phase, demonstrating the transmission of the link drive force F_1 and the leg radial drive force F_4; (III) Fully deployed state. (i) Isometric projection view. (ii) Rear view. (iii) Front view. (d) Mechanical self-locking design of the anchoring module.

M-DCDC was introduced as a magnetically controlled dual-target drug delivery capsule system designed to overcome the critical limitations of conventional GI drug delivery, specifically the lack of rate control and positional instability, as illustrated in Fig. 1. The innovative contributions are as follows:

1. Unlike traditional mechanisms prone to burst-type release, M-DCDC utilizes specialized screw-piston transmission. This mechanism converts the magnetic rotational frequency into a highly linear and quantitatively adjustable drug infusion, achieving a flow rate of up to 0.1 mL/s.

2. A key advance lies in the functional decoupling of the delivery sequence. By reversing the magnetic field direction, two distinct 0.6 mL reservoirs can be independently actuated, enabling sequential delivery to spatially separated lesions within a single procedure.

3. To ensure stability against peristalsis, the system integrates a mechanical self-locking anchoring module. This module can resist peristaltic forces without requiring continuous power consumption or constant magnetic actuation, thereby improving dosing reliability and localization accuracy.

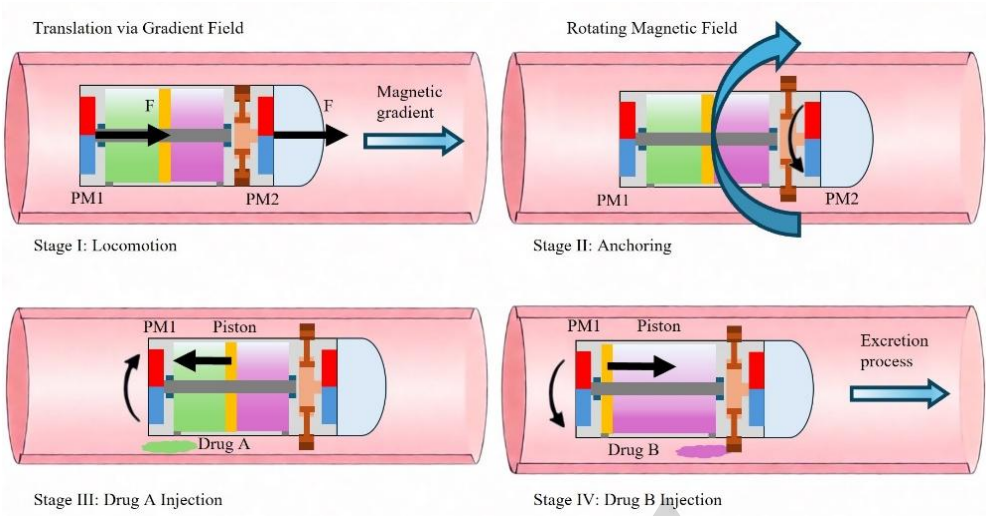

Fig. 3 Dual-drug injection workflow. PM2 is rotated by an external magnetic field to activate anchoring. Subsequently, PM1 is controlled to rotate clockwise or counterclockwise clockwise/counterclockwise through the external magnetic field, releasing Drug A or Drug B. By adjusting the rotation speed of the external magnetic field, the drug delivery rate and dosage can be precisely controlled. After reanchoring, drug delivery to the next lesion site can proceed.

2 Robot structure design

2.1 M-DCDC

Fig. 2 illustrates the overall structure of the proposed capsule robot. With a diameter of 13 mm and a length of 31 mm, the capsule is primarily designed for use in the intestinal environment and can be administered orally. Given that the adult esophagus has a diameter of approximately 20 mm and a length of approximately 250 mm, sufficient anatomical space is available for smooth capsule passage. The system integrates three core functional modules: a camera module, an anchoring module, and a drug delivery module. The camera module enables real-time observation of the intestinal environment. The anchoring module employs a linkage mechanism to deploy anchoring legs, enabling reliable fixation of the capsule while minimizing the risk of intestinal wall damage, thereby providing a stable platform for targeted drug delivery. The drug delivery module can accommodate either a single drug reservoir or two separate reservoirs, with a maximum capacity of 1.3 mL for single-drug loading or 0.6 mL per reservoir for dual-drug operation. Under external magnetic actuation, the system enables controlled and adjustable-rate release of one or two drugs. Following drug delivery, the anchoring mechanism can be retracted to restore capsule mobility for subsequent navigation.

It should be noted that the camera module integrated into the current physical prototype serves as a structural and spatial placeholder. Because the primary focus of this study is to validate the mechanical anchoring and dual-drug delivery mechanisms, reserving realistic volume and weight for a standard vision module ensures that the proposed actuation structures can be accommodated within clinically viable capsule dimensions, thereby paving the way for future fully integrated systems.

2.2 Dual-drug delivery module

To achieve multifunctional operation within the limited capsule volume, this study proposes an integrated module based on screw–piston transmission. Positioned at the rear of the robot, the module comprises a radially magnetized permanent magnet (PM1), a lead screw, a piston with an embedded nut, and an isolation membrane. This design uses a single actuation source to generate bidirectional linear motion. PM1 is rigidly attached to the screw and axially supported by bearings. Because the rotation of the piston is constrained by the capsule wall, an external rotating magnetic field drives PM1 to rotate, which in turn rotates the screw and converts rotational torque into axial reciprocating motion of the piston, as shown in Fig. 3. By exploiting the reversibility of the screw–nut mechanism, the direction of piston motion

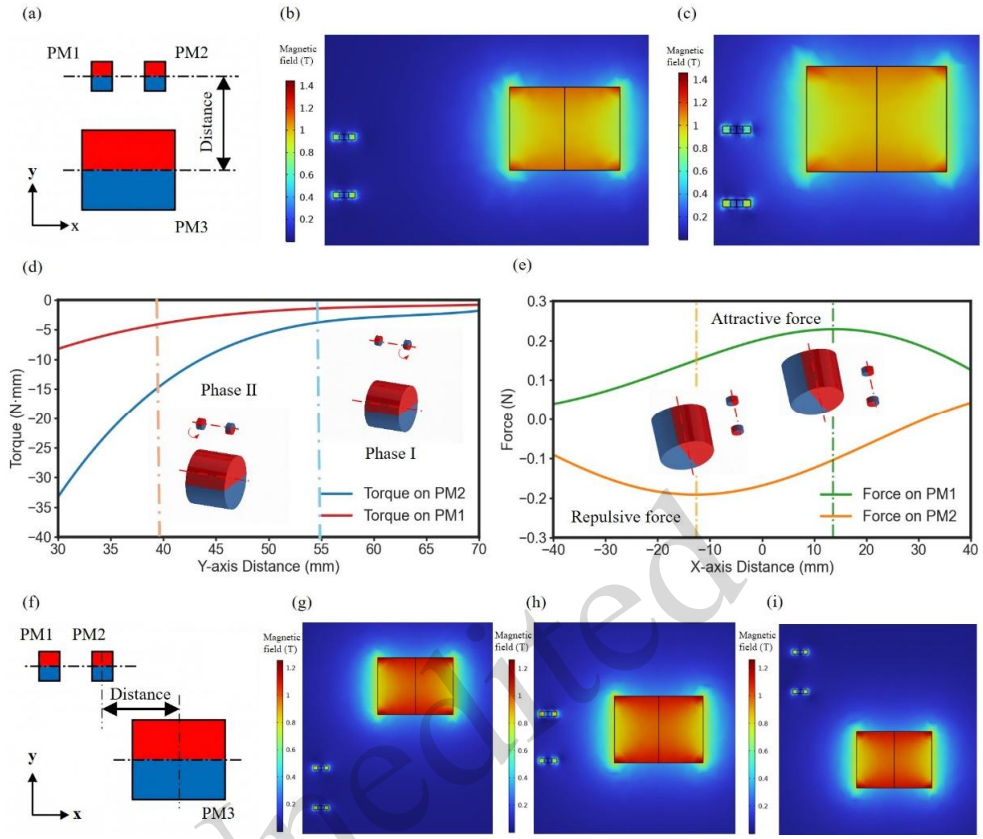

Fig. 4 Simulation Analysis and Characterization of Magnetic Drive Systems. (a) Schematic of the simulation model showing the external magnet (PM3) moving along the Y-axis; (b)–(c) finite element analysis (FEA) heatmaps at different Y-axis positions; (d) calculated characteristic curve of magnetic coupling torque versus distance, indicating torque decay trends at different stages; (e) variation curve of magnetic driving force with X-axis displacement; (f) schematic diagram of the simulation model with the external magnet moving along the X-axis; (g)-(i) magnetic field distribution maps at different positions along the motion path.

can be controlled by switching the rotation direction of the external magnetic field. This bidirectional capability enables precise, sequential, and selective release of two distinct drugs, thereby improving space utilization and enhancing therapeutic flexibility.

2.3 Anchoring module

To achieve effective anchoring within a limited space, this study proposes a linkage-based module comprising a central magnetic shaft and telescopic legs. During the cruising phase, the mechanism retracts to a diameter of 13 mm to ensure smooth passage. Upon activation, the magnetic shaft drives the linkages to expand the module to a diameter of 24 mm. The nonlinear relationship between the radial expansion L and the rotation angle θ enables adaptation to variations in intestinal diameter while converting magnetic torque into an effective radial force F_4. Furthermore, to minimize energy consumption during long-term anchoring, a mechanical self-locking mechanism is implemented. At maximum expansion, the interaction between the intestinal resistance F_6 and the shell support force F_7 creates a countertorque through stop block F_8, ensuring that the anchoring structure remains fully deployed and engaged without an external magnetic field.

3 Analytical model of magnetic fields

3.1 Magnetic control system

The driving force of the capsule robot originates from the interaction between the magnetic field $\boldsymbol{B}$ generated by the external permanent magnet (EPM) and the magnetic moment $\boldsymbol{m}$ of the internal permanent magnet. Similar principles have been applied in

navigation systems composed of dual permanent magnets for accurate posture control (Kim et al., 2024). When a rotating magnetic field is applied, the magnetic torque $\boldsymbol{T}$ exerted on the internal permanent magnets (PM1/PM2) serves as the primary actuation source for the anchoring and lead-screw mechanisms. This torque can be expressed as:

$$\boldsymbol{T} = \boldsymbol{m} \cdot \boldsymbol{B} = |\boldsymbol{m}||\boldsymbol{B}| \sin\theta \cdot \boldsymbol{n} \tag{1}$$

where θ is the angle between the magnetic moment and the external magnetic field.

When a gradient magnetic field is applied, the magnetic force $\boldsymbol{F}$ driving the capsule's cruising motion is expressed as:

$$\boldsymbol{F} = (m \cdot \nabla)\boldsymbol{B} \tag{2}$$

Based on the aforementioned model, a simulation analysis of the magnetic drive system was performed using COMSOL Multiphysics software. The grades and dimensions of the permanent magnets were selected by considering the required actuation torque, magnetic decoupling, and spatial constraints. PM1 and PM2, which are embedded inside the capsule, were selected as N35 magnets to provide sufficient torque for drug delivery and anchoring while reducing undesired magnetic coupling between the two internal magnets. PM3 was selected as an N52 magnet to compensate for the rapid decay of magnetic field strength with distance and to ensure sufficient actuation capability within the working space. To provide a comprehensive understanding of the magnetic interactions, Fig. 4 presents both quantitative characteristic curves and the corresponding finite element analysis (FEA) heatmaps.

Figs. 4b and 4c visualize the magnetic flux density distribution when the external magnet (PM3) is placed at different vertical distances along the Y-axis. These field maps are consistent with the nonlinear decay of the magnetic coupling torque experienced by PM1 and PM2, as quantified by the characteristic curves in Fig. 4d. Similarly, Figs. 4g-i illustrate the dynamic spatial variations in the magnetic field as PM3 translates along the X-axis. These field distributions explain the transitions between the attractive and repulsive forces acting on the capsule, which directly correspond to the force variations plotted in Fig. 4e. Together, the theoretical analysis and field visualization confirm the feasibility of the proposed magnetic drive scheme within the effective workspace.

3.2 Dual-drug delivery analysis

To estimate the driving force required for controlled drug infusion and to guide the design of the release mechanism, the flow behavior of liquid or weakly viscous drug formulations within the microneedle was modeled using the Poiseuille flow equation. Under the assumption of steady, laminar flow in a cylindrical needle, the relationship between the volumetric flow rate Q and driving pressure difference ΔP can be expressed as:

$$\Delta P = \frac{8\eta L Q}{\pi r^4} \tag{3}$$

where $\Delta P = p_2 - p_1$ represents the pressure difference required to overcome fluid resistance (p_2 is the driving pressure within the cavity, $p_1 \approx 0$ is the ambient pressure in the intestinal tract), η denotes the dynamic viscosity of the drug, L is the length of the injection needle, and r is the inner radius.

In the proposed screw–piston drive mechanism, the volumetric flow rate Q is geometrically related to the linear advancement velocity v_{piston} of the pitson, $Q = v_{piston} \cdot A_{chamber}$. The linear velocity of the piston is directly related to the angular velocity ω of the external magnet through the lead P_h of the screw:

$$v_{\text{piston}} = \frac{\omega}{2\pi} P_h \tag{4}$$

By combining the above equations, a linear mapping relationship can be established between the rotational frequency and the drug flow rate Q:

$$Q(\omega) = A_{\text{chamber}} \frac{P_h}{2\pi} \cdot \omega \tag{5}$$

This relationship indicates that, for a given mechanical configuration and drug viscosity, the infusion rate is proportional to the rotational speed of the external magnetic field. Provided that the magnetic torque is sufficient to generate the piston thrust $F_{\text{thrust}} = \Delta P \cdot A_{\text{chamber}}$, the release rate and delivered dose can be adjusted in a predictable manner by modulating ω, offering a continuous alternative to discrete, switch-based magnetic actuation strategies.

3.3 Anchoring analysis

To prevent the capsule from being swept away

by intestinal peristalsis during its mission, it is essential to ensure that the axial resistance F_{anchor}

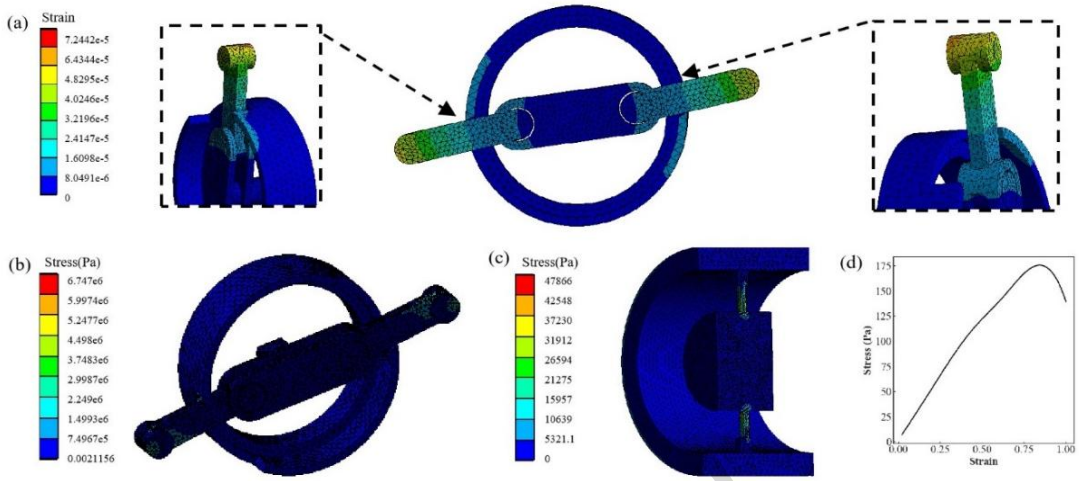

Fig. 5 Mechanical simulation performance evaluation of the anchoring mechanism. (a) Strain distribution contour at maximum deployment, highlighting critical connection points. (b) Stress distribution contour at maximum deployment. (c) The stress at the time of capsule anchoring during intestinal peristalsis. (d) The stress-strain characteristic curves of the anchoring mechanism.

generated by the anchoring mechanism exceeds the maximum peristaltic force $F_{peristalsis}$ of the intestine. According to relevant physiological studies, the axial peristaltic force density in the small intestine is approximately 17.2 g/cm (Miftahof., 2005). For a capsule robot of length L_{cap}, the maximum peristaltic thrust can be estimated as:

$$F_{peristalsis} \approx 17.2 \times 10^{-3} \times L_{cap} \times g \tag{6}$$

Based on the length and weight of this capsule, this force is approximately 0.32 N. After the anchoring legs extend, they compress the intestinal wall, with the resulting anchoring force primarily composed of friction. Let the radial support force exerted by the anchoring legs on the intestinal wall be F_r. According to Coulomb's friction law, the anchoring resistance F_{anchor} is:

$$F_{anchor} = \mu_c \cdot \sum F_r \tag{7}$$

In this study, the equivalent friction coefficient between the capsule anchoring legs and the intestinal mucosa was estimated to be approximately 0.08–0.20 based on existing biomechanical evaluations of ex vivo porcine intestinal tissues (Kim et al., 2006). For the theoretical estimation and numerical simulation, a friction coefficient of 0.2 was adopted to evaluate the anchoring reliability.

To ensure delivery stability, the system design must satisfy the following mechanical conditions:

$$F_{anchor} > F_{peristalsis} \tag{8}$$

A simulation analysis of the anchoring device was performed using Ansys software, as shown in Fig. 5. First, the stress and strain distributions under the self-locking state of the linkage mechanism were evaluated, and the corresponding stress–strain relationship was analyzed. Second, the stress condition induced by intestinal peristalsis was investigated. The contact area of each anchoring foot was approximately 10 mm², indicating that the device could withstand an intestinal peristaltic force of approximately 0.48 N. This result satisfies the stability condition defined in Eq. (8).

4 Experiments

4.1 Prototyping

To experimentally validate the proposed design, a physical prototype $(\varnothing 13mm \times 31.5mm)$ was fabricated, as shown in Fig. 6. The capsule shell and internal supports were manufactured via high-precision SLA 3D printing. The internal mechanism integrates a custom-machined brass lead screw (P = 0.35 mm) with radially magnetized permanent magnets for precise actuation, while a highly elastic TPE film serves as an isolation barrier for the drug storage module. To withstand the gastrointestinal environment, this component is

bonded using medical-grade adhesive to prevent leakage.

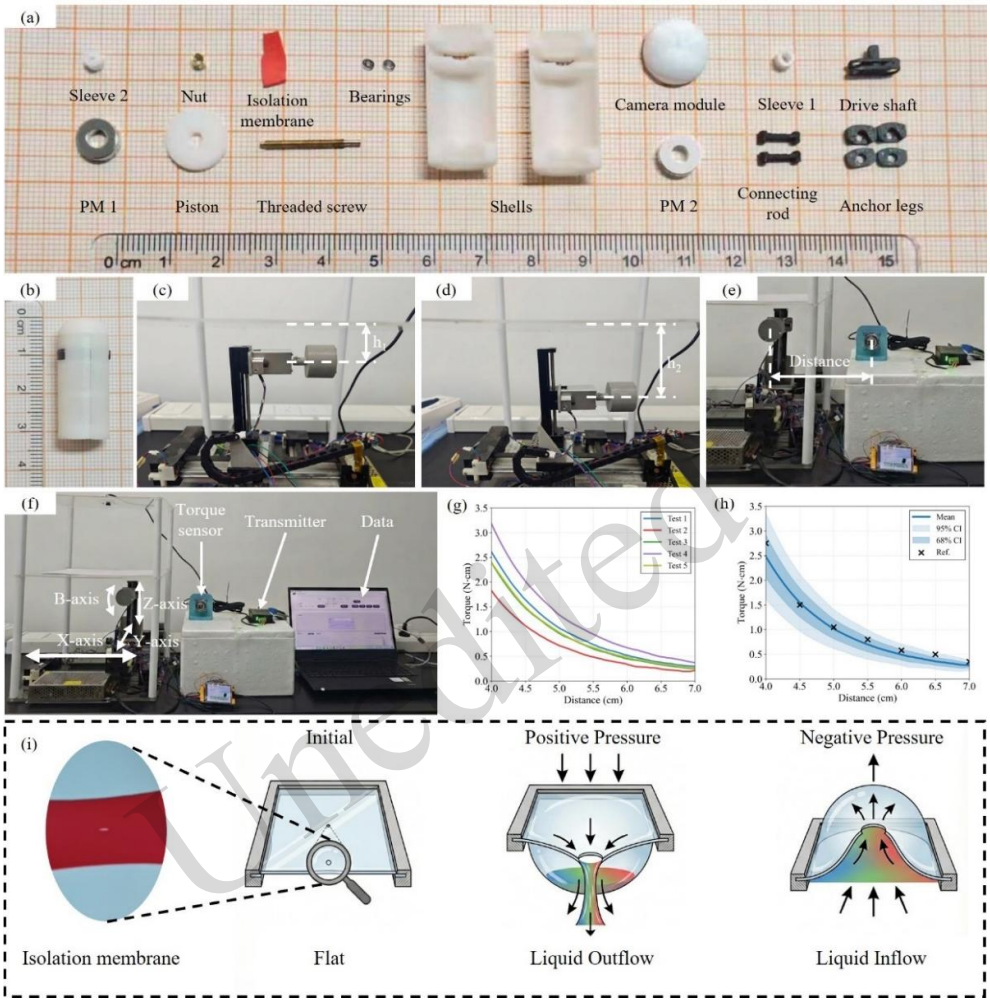

Fig. 6 M-DCDC prototype structure and magnetic drive characteristic testing. (a)-(b) Exploded view of the robotic prototype and details of its internal core components. (c)-(d) Demonstration of the effective working space, indicating the effective distance range of magnetic control. (e)-(f) Schematic of the magnetic drive torque measurement experiment and physical test platform incorporating a torque sensor. (g) Raw experimental curves showing magnetic coupling torque decay with distance across five distinct test sequences. (h) Statistical analysis of magnetic torque characteristics, presenting comparisons of the average torque curve, confidence interval (CI), and theoretical reference value (Ref.). (i) Working principle of the isolation membrane.

membrane maintains a microscopic orifice that effectively prevents drug leakage. When the drug chamber is subjected to an outward positive pressure, this orifice expands, allowing for proper drug release. Conversely, under inward negative pressure, the orifice also expands slightly. However, owing to the relatively high viscosity of the intestinal fluid, fluid ingress is minimal. Consequently, this design reduces fluid backflow into the drug reservoir, as illustrated in Fig. 6i.

4.2 Decoupling height test

In the absence of applied pressure, the isolation

To characterize the operational separation between the anchoring and drug delivery modules, decoupling tests were conducted to map the effective magnetic workspace based on magnetic field attenuation with height and the distinct torque requirements of each module. As shown in Figs. 6c and 6d, the anchoring module requires a relatively lower operating height to generate sufficient magnetic torque to overcome the mechanical resistance of the linkage mechanism. In contrast, the drug delivery module, which incorporates low-friction bearings,

can be actuated under reduced magnetic torque.

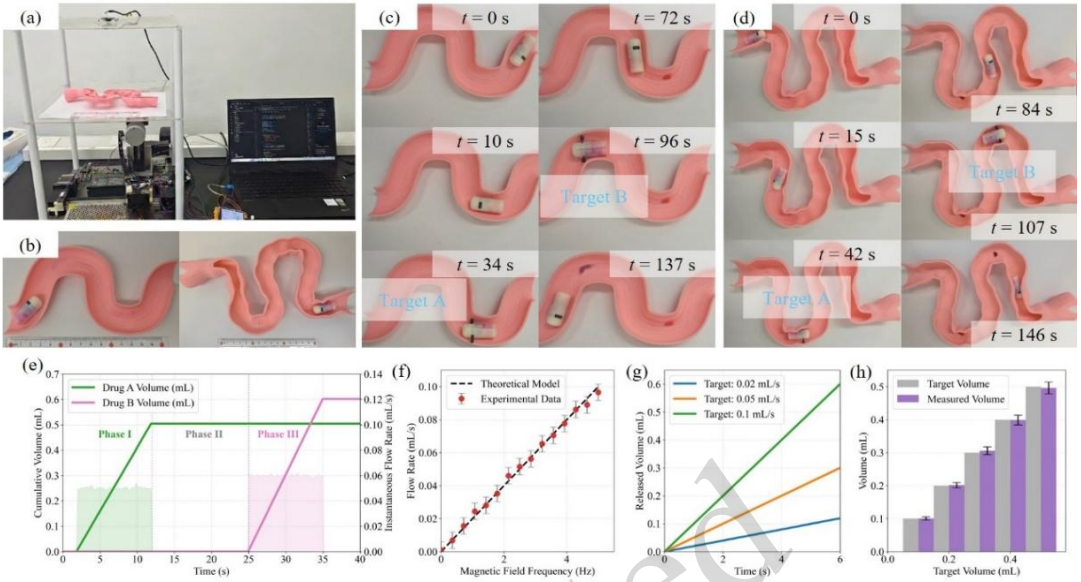

Fig. 7 In vitro evaluation of dual-drug delivery performance. (a)-(b) Experimental setup and simulated gastrointestinal model. (c)-(d) Time-lapse sequences of the dual-target release process. (e) Cumulative release volume showing decoupled control. (f) Linear fit between flow rate and magnetic frequency. (g) Release volume profiles at different set flow rates. (h) Accuracy analysis comparing target versus measured volumes.

These measurements reveal a separation in the height-dependent actuation thresholds of the two modules, which can be exploited for selective control.

Based on the experimentally identified thresholds, a sequential control strategy was implemented. In this strategy, the external magnet is first positioned at a lower height to activate the anchoring mechanism and is subsequently raised to a higher position to actuate the drug delivery piston while maintaining the anchored state. The results demonstrate that modulation of the external magnet height provides a feasible approach for achieving functional decoupling between anchoring and drug delivery under the tested conditions.

4.3 Magnetic model validation

To experimentally examine the validity of the magnetic torque model, a dedicated torque measurement platform was constructed, as shown in Figs. 6e and 6f. The capsule prototype was rigidly mounted on a high-precision dynamic torque sensor, while the external permanent magnet was positioned at different distances ranging from 4.0 to 7.0 cm. At each distance, the peak magnetic coupling torque was recorded, and five independent measurements were performed to obtain the mean value.

Fig. 6h compares the experimentally measured torque values with the theoretical predictions obtained from the magnetic simulation model. The experimental results show close agreement with the simulated curve and remain within the 95% confidence interval across the tested distance range, indicating good consistency between the model predictions and physical measurements. These results support the applicability of the proposed magnetic model for estimating the available driving torque within the intended operating workspace. In particular, at a representative working distance of 6.0 cm, the measured torque exceeds 0.5 N·cm, providing an adequate actuation margin for subsequent screw-driven drug delivery and anchoring operations under the experimental conditions considered.

4.4 Dual-drug delivery accuracy test

To rigorously evaluate the quantitative control capability of the screw–piston mechanism, a series of bench tests was conducted to validate the relationships among magnetic drive frequency, flow rate, and dosing accuracy, as shown in Fig. 7. The linearity of the drive system was first assessed. As derived from Eq. (5), the drug flow rate is theoretically proportional to the rotational speed of

the external magnetic field.

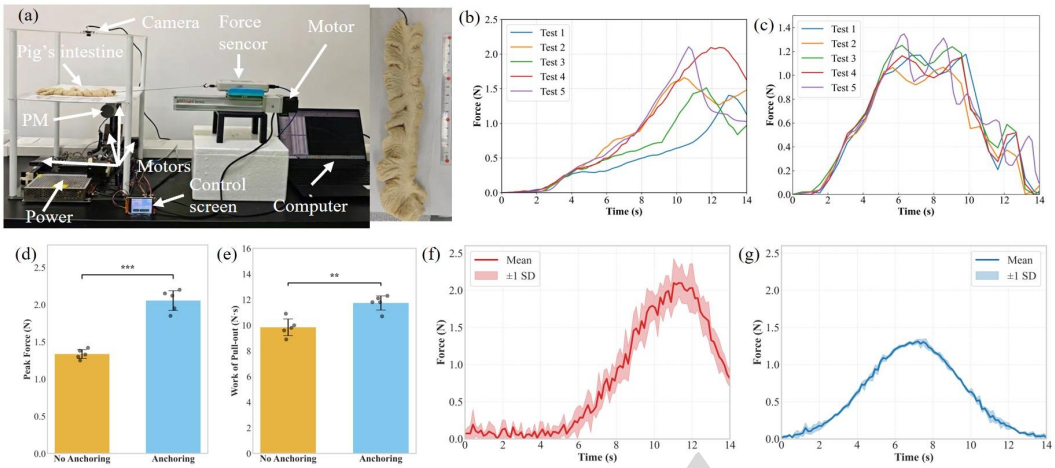

Fig. 8 Mechanical ex vivo performance evaluation of the anchoring mechanism. (a) Experimental setup for anchoring force testing on ex vivo porcine colon. (b)-(c) Raw drag force-time curves for 5 independent trials in anchored and unanchored states. (d)-(e) Statistical comparison of the peak force and work of pull-out (*p<0.001, **p<0.01). (f)-(g) Mean resistance curves (± SD) comparing anchored and unanchored states.**

The instantaneous flow rate was measured at different magnetic field frequencies. The results plotted in Fig. 7f demonstrate a highly linear correlation between the input frequency and the output flow rate, with the experimental data closely matching the theoretical model. This confirms that the drug delivery rate can be precisely modulated by adjusting the rotational speed of the external magnet.

Experiments were performed at three target flow rates: 0.02 mL/s, 0.05 mL/s, and 0.1 mL/s to verify the temporal stability of the infusion process. Fig. 7g shows that the cumulative delivered volume increased approximately linearly with time under all tested settings, indicating consistent infusion behavior throughout the delivery duration. Minor fluctuations were observed but remained within a limited range across repeated trials. In addition, dosing accuracy was evaluated by setting target release volumes between 0.1 mL and 0.5 mL and comparing them with the experimentally measured delivered volumes. The results in Fig. 7h show only small deviations between the target and measured values, with low standard deviations across repeated measurements. These results suggest that the proposed system is capable of repeatable and controllable drug dosing under the tested experimental conditions.

4.5 Anchoring test

Anchoring stability and resistance to intestinal peristalsis were evaluated through ex vivo dragging experiments using freshly harvested porcine intestines, as shown in Fig. 8. The capsule was subjected to controlled axial pulling in both unanchored and anchored configurations, and the resulting resistance forces were recorded for comparative analysis. Deployment of the linkage-based anchoring mechanism resulted in a marked increase in mechanical resistance, primarily arising from frictional interaction with the intestinal folds. Quantitative analysis showed that both the mean peak resistance force and the withdrawal work were significantly higher in the anchored state than in the unanchored state (**$p < 0.01$; ***$p < 0.001$). The measured peak resistance forces fell within the range of peristaltic forces reported in the literature for the small intestine. These findings suggest that the proposed anchoring mechanism can provide enhanced positional stability under physiologically relevant mechanical disturbances during ex vivo testing.

4.6 Ex vivo integrated experiments

Ex vivo integrated experiments were conducted using freshly harvested porcine intestines to evaluate the feasibility of the proposed M-DCDC system under physiologically relevant conditions, as shown in Fig. 9. A sequential operation protocol involving navigation, anchoring, and drug release was

implemented. Guided by a gradient magnetic field, the capsule was navigated to the first predefined target location, where the anchoring mechanism was deployed to stabilize the device against the intestinal wall, followed by the release of Drug A through the screw–piston delivery module.

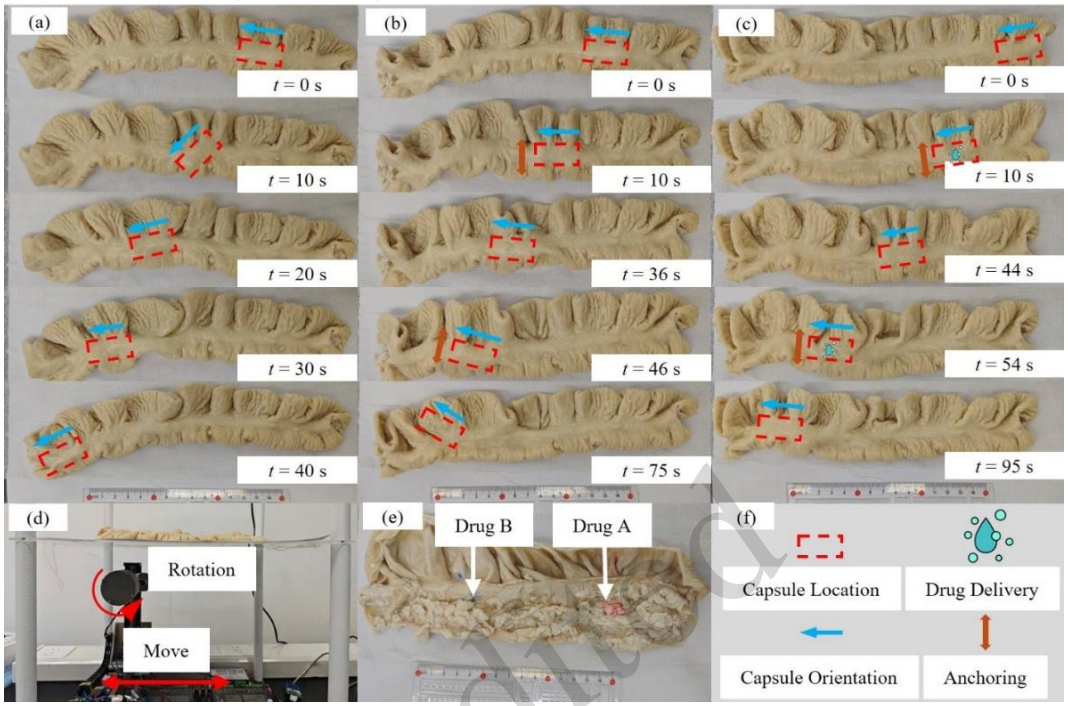

Fig. 9 In vitro intestinal integrated experimental validation of the M-DCDC robot. (a) Time-series screenshots of capsule movement within the intestine; (b) Time-series snapshots of capsule movement and stationary anchoring; (c) Time-series snapshots of dual-drug delivery execution; (d) Schematic of motion control for the magnetically controlled platform; (e) Demonstration of staining effects within the intestine after drug injection; (f) Legend explaining the labeling symbols used in the figure.

After the first release step, the anchoring mechanism was retracted to restore capsule mobility, and the device was subsequently guided to a second downstream target. The anchoring and drug release procedures were then repeated for Drug B delivery. Visual inspection after the experiments revealed two spatially separated localized staining regions corresponding to the intended release sites. These observations demonstrate that the system can perform sequential multisite drug delivery while maintaining positional separation under ex vivo experimental conditions.

5 Discussion

The proposed M-DCDC system addresses several persistent challenges in gastrointestinal drug delivery that have been widely reported in the literature, including limited control over the release rate, difficulty in achieving spatially separated delivery, and reduced positional stability during drug administration. Rather than introducing a fundamentally new actuation paradigm, this study focuses on the system-level integration of magnetic actuation, mechanical anchoring, and quantitative infusion within a single capsule platform.

As summarized in Table 1, many existing magnetic capsule robots exhibit inherent limitations in drug release controllability. Approaches based on chemical reactions (Esendag, et al., 2024) or magnetic torsion springs (Cao, et al., 2025) typically rely on threshold-triggered or discrete actuation mechanisms, which limit their ability to provide adjustable and sustained infusion rates. Soft capsule robots driven by ferrofluids or deformable structures have improved compliance and reduced contact stress with intestinal tissue (Hua et al., 2022). However, drug release in these systems is generally governed by deformation or squeezing processes, which can lead to nonlinear release profiles and transient bursts that are difficult

Table 1 Comparison of the Proposed Robot with Related Works

Work	Robot Size (mm)	Powered by	Anchoring Capability	Drug Release Mechanism	Rate Controllability	Quantitative Dosage Control	Multi-Target / Dual-Drug
(Esendag, et al., 2024)	11(D)×26(L)	Chemical Reaction	No	Pressure Expansion	No	Yes	No
(Cao, et al., 2025)	14(D)×29(L)	Magnetic Field	No	Torsion Spring	No	Yes	No
(Wu, et al., 2025)	9.8(D)×20(L)	Magnetic Field	No	Magnetic Valve	No	Yes	Yes
(Ye, et al., 2025)	13(D)×29(L)	Magnetic Field	No	Balloon Puncture	No	Yes	Yes
(Song, et al., 2022)	16(D)×50(L)	Magnetic Field	Yes	Valve Opening	No	Yes	No
(Hua, et al., 2022)	Soft Capsule	Magnetic Field	No	Squeeze	No	Yes	No
Ours	**13(D)×31(L)**	**Magnetic Field**	**Yes**	**Screw-Piston**	**Yes**	**Yes**	**Yes**

to regulate over time. In contrast, the M-DCDC employs a rigid screw–piston transmission mechanism, enabling a direct and predictable relationship between the rotational speed of the external magnetic field and the volumetric flow rate. Although this rigid configuration introduces additional mechanical complexity compared with soft designs, it provides a practical means of achieving adjustable and repeatable release behavior under controlled experimental conditions.

Spatially selective drug delivery presents an additional challenge for capsule-based systems. Although recent work has demonstrated a dual-chamber capsule capable of delivering two drugs, the absence of a dedicated physical anchoring mechanism may limit positional stability during the release window, particularly in the presence of intestinal peristalsis (Ye, et al., 2025). Other magnetic capsule systems have addressed stability or tissue interaction through orientation locking (Erin et al., 2021) or origami-inspired mechanisms for biopsy and sampling (Huang et al., 2025). However, these designs do not incorporate actively controlled continuous drug infusion in combination with mechanical stabilization.

The M-DCDC integrates a self-locking mechanical anchoring module with a height-modulated, functionally decoupled dual-drug delivery mechanism, allowing sequential delivery at multiple locations while maintaining positional stability during ex vivo testing. This integrated design demonstrates a feasible pathway for combining localized stabilization and quantitative infusion within the size constraints of a capsule robot, thereby providing a complementary approach to existing soft or purely magnetically stabilized systems.

Despite the promising ex vivo results, several limitations should be noted. The drug flow model and experiments mainly considered liquid or weakly viscous formulations under laminar flow, and higher-viscosity or non-Newtonian drugs may require further characterization. Moreover, ex vivo porcine intestines cannot fully replicate in vivo gastrointestinal dynamics, such as active peristalsis, variable luminal pressure, and long-term tissue responses. In addition, the experiments focused on short-term anchoring and delivery, without assessing prolonged deployment, repeated operating cycles, or potential tissue irritation. Future work should therefore emphasize in vivo studies, long-duration testing, and optimization of the anchoring design and actuation parameters to further evaluate safety, robustness, and clinical feasibility.

Furthermore, although the anchoring mechanism operated smoothly in the clean ex vivo environment, the presence of food residues or complex debris in the in vivo gastrointestinal tract may pose a risk of the anchoring legs becoming stuck during deployment or retraction. Addressing this mechanical reliability issue through optimized leg geometry, anti-adhesive coatings, or redundant retraction mechanisms will be a critical focus of future in vivo studies.

6 Conclusions

This study presents the M-DCDC, a magnetically actuated capsule system that integrates mechanical anchoring with quantitatively controllable and spatially selective drug delivery. By combining a screw–piston transmission mechanism with magnetic actuation, the system enables adjustable-rate infusion and provides an alternative to the burst-type release mechanisms commonly used in passive or spring-based capsule systems. Functional decoupling further enables sequential delivery of two drugs to spatially separated sites within a single capsule platform. In addition, a compact self-locking anchoring mechanism is incorporated to enhance positional stability during drug administration. Ex vivo experiments demonstrated that the anchoring module increased resistance to mechanically relevant disturbances while remaining compatible with magnetic navigation and release operations. Simulation and experimental results further confirmed predictable actuation and dosing behavior under controlled experimental conditions.

Overall, the M-DCDC provides an integrated engineering framework for localized and controlled gastrointestinal drug delivery using magnetic capsule robots. Although further in vivo validation is needed, this approach represents a promising platform for advancing targeted drug delivery in minimally invasive gastrointestinal interventions.

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Author contributions

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Conflict of interest

The authors declare that they have no conflict of interest.

Declaration on the use of generative AI tools

During the preparation of this work, the authors used ChatGPT to improve language and readability. After using this tool, the authors reviewed and edited the content as needed and take full responsibility for the content of the publication.

Data availability

Data are available on reasonable request.

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中文概要

题目： 一种用于多靶点恒速递药的磁控胶囊机器人

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目的： 针对现有肠道递药胶囊机器人易受蠕动扰动、释放速率难控、以及难以实现多靶点独立递药等问题，本文提出一种磁控多靶点恒速递药胶囊机器人，以实现复杂肠道环境下的稳定驻留、定量和顺序药物释放。

创新点： 1.提出螺杆—活塞式磁控恒速递药机制，建立磁场转速与药物流量的线性关系，实现连续可调的恒速输注；2. 构建双药腔解耦释放策略，通过磁场旋转方向切换分别驱动两个独立药腔，实现空间分离靶点的顺序递药；3. 集成机械自锁锚定机构，在无需持续供能的条件下增强胶囊抗蠕动位移能力。

方法： 1.设计并制备直径约 13 mm、长度约 31 mm 的磁控双靶恒速递药胶囊机器人，集成相机模块、机械锚定模块和双药物释放模块；2. 采用径向充磁永磁体驱动螺杆旋转，通过螺杆—螺母结构实现药液连续释放；通过改变磁场旋转方向，实现双药腔选择性驱动；3. 设计连杆展开式锚定机构与机械自锁结构，使胶囊在递药阶段稳定贴附于肠壁，降低蠕动导致的位移风险；4. 通过有限元仿真、磁力矩测试、递药精度实验及离体猪肠实验，验证了所提出系统的驱动能力、剂量可控性和锚定稳定性。

结论： 1. 胶囊可实现磁场转速与药物流量的线性调控，最大流量达 0.1 mL/s；2. 两个 0.6 mL 储药腔可独立释放，实现空间分离靶点的顺序递药；3. 锚定模块可抵抗约 0.32 N 肠道蠕动力。离体实验中，胶囊完成导航、锚定与双靶点释放，并形成清晰分离的局部染色区域。

关键词： 胶囊微型机器人；磁性驱动；靶向药物递送；主动锚定