

RESEARCH ARTICLE

Machine Learning-Enhanced Soft Robotic System Inspired by Rectal Functions to Investigate Fecal Incontinence

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Abstract

Fecal incontinence (FI), which can arise from various pathogenic mechanisms, has attracted considerable attention worldwide. Despite its importance, the reproduction of the defecatory system to study the mechanisms of FI remains limited, largely because of social stigma and being considered inappropriate. Inspired by the rectum's functionalities, we developed a soft robotic system that includes a power supply, pressure sensor, data acquisition system, flushing mechanism, stage, and rectal module. The innovative soft rectal module includes actuators inspired by sphincter muscles, both soft and rigid covers, and a soft rectum mold. The rectal mold, which was fabricated from materials that closely mimic human rectal tissue, was produced using a mold replication fabrication method. Both the soft and rigid components of the mold were created using 3D printing technology. The sphincter muscle-inspired actuators featured double-layer pouch structures that were modeled and optimized based on multilayer perceptron methods to obtain high contraction ratios (100%), generate high pressure (9.8 kPa), and have a small recovery time (3 s). Upon assembly, this defecation robot could smoothly expel liquid feces, perform controlled solid fecal cutting, and defecate extremely solid long feces, thus closely replicating the functions of the human rectum and anal canal. This defecation robot has the potential to facilitate human understanding of the complex defecation system and contribute to the development of improved quality-of-life devices related to defecation.

1. Introduction

Fecal incontinence (FI) is characterized by the involuntary discharge of bowel contents, including gas, mucus, and both liquid and solid feces, thus indicating an inability to control defecation [1]. The etiology of FI includes various pathogenic mechanisms, which can commonly be attributed to rectal anomalies, sphincter muscle dysfunction, nerve impairment, and other medical or physiological conditions [2]. In response to these challenges, the establishment of physiological organ simulators and computational models offers alternative approaches for the evaluation of potential treatment methodologies during the preliminary development phases and enhances the comprehension of the biomechanical processes underlying the diverse disease states [3][4]. Computational models, while invaluable in the study of human physiology and the simulation of organ functions, often do not capture the complex interplay among multiple organs. This limitation primarily stems from the inherent challenge of accurately replicating the intricate and dynamic interactions that occur within the human body. These interactions involve the mechanical and physical forces as well as the biochemical processes that govern organ function and interorgan communication. Consequently, the lack of authentic reproducibility of these models hinders our ability to fully understand the holistic functioning of the body's systems and to accurately predict the outcomes of medical interventions [5].

Soft robotics have recently emerged as a cutting-edge approach for simulating or enhancing human organ functions [6][7][8][9][10]. For example, implantable soft robotic devices have demonstrated promising outcomes in strengthening cardiac function in cases of isolated heart failure, whether the left or right side of the heart is affected [11]. Similarly, stomach simulation robots have been developed to provide a controlled testing environment for innovative food products and pharmaceuticals [12][13], while the biomimetic robotic soft esophagus serves as an *in vitro* model to evaluate endoprosthetic stents designed to manage dysphagia [14]. These developments highlight the potential of soft robotics as both a complementary and supportive technique in conjunction with mathematical models, particularly to explore human physiology and validate medical interventions [15]. Notably, anal sphincter (AAS) devices utilizing a circular cuff design have been used for the treatment of FI in humans [16][17]. Researchers have utilized different actuation techniques, such as rigid clamping mechanisms [18], fluidic actuators [19][20][21], cable-driven actuators [22], magnetic actuators [23], and shape memory alloy actuators [24], to investigate their ability to reproduce sphincter pressures observed in the clinic. In addition, other actuators have exhibited potential for occluding the canal [25][26][27][28]. Moreover, a cost-effective reusable phantom model has been developed that combines realistic anatomy with realistic tissue properties [29]. Although current endeavors have primarily involved the AAS device, research focused on reproducing the defecatory system and investigating the mechanisms associated with FI is limited. The scarcity of these studies can primarily be attributed to the stigma and social avoidance related to the subject, which considerably impedes the progress of understanding and addressing FI through innovative soft robotic simulations.

In this study, we propose a soft and lightweight pneumatically driven defecation robot, which includes several soft ring-shaped actuators within soft cases, to replicate the rhythmic peristaltic motion and opening–occlusion motion of the anal canal at the symmetrical level.

Fig. 1 presents a detailed visual analysis of the defecation process, alongside a conceptualized soft robotic system designed to replicate these physiological motions. The muscular anatomy includes the puborectalis muscle, which plays a vital role in continence, as well as the internal and external sphincter muscles, which are crucial for the regulation of stool passage (

Fig. 1A). In our study, we set the anorectal angle to 165° , which is beneficial for fecal defecation. The cross-sectional view of the rectal architecture indicates its composite structure, which consists of the mucosal surface, underlying intestinal gland, and muscular layers, specifically the circular and longitudinal muscles (

Fig. 1B). The rectum is a muscular tube consisting of a continuous longitudinal muscle layer that connects with the underlying circular muscles. The longitudinal and circular muscles propel the fecal matter along the gastrointestinal tract and contract radially to crush and grind the feces [30]. Three distinct defecation scenarios generally occur in humans: (i) diarrhea, illustrating the process of expelling liquid feces; (ii) cut feces, indicating the expulsion of solid feces that are abruptly truncated; and (iii) defecating long feces, demonstrating the continuous expulsion of a solid stool (

Fig. 1C). In this study, we simulated these scenarios to understand the rectum's role in defecation. The peristaltic movement of fecal matter from the sigmoid colon into the rectum is portrayed via a dynamic cross-sectional visualization of the fecal bolus, where a decrease in the cross-sectional area indicates the peristaltic contraction rate (α) (

Fig. 1D). The successive cross-sectional illustrations (1–5) visually quantify this reduction from 100% to approximately 30%. In addition, a segmented soft pouch actuator system was designed to mimic the natural peristaltic contractions of human bowel movements, with the actuator segments demonstrating varying degrees of contraction to replicate the rectum's complex muscular movement during defecation.

Fig. 1 Conceptualization of a Soft Robotic System to Simulate Defecation. **A:** Schematic overview of key physiological components of the defecation system. **B:** Anatomical structure of the muscle layers of the rectum. **C:** Three unique defecation scenarios generally occur in humans. **D:** Defecation process within the rectum, including the contraction rate of each rectal section and strategic placement of actuators within the robotic model (A_0 represents the initial area when the actuator is fully extended and A_1 denotes the contracted area during actuation).

2. Results and discussion

2.1 Rectum model

The tensile properties of human rectal tissue exhibit viscoelasticity, as evidenced by an increase in material stiffness at larger displacements [31][32]. To develop materials with elastic properties comparable to the human rectum, we investigated various soft and hard materials, including AR-M2, Elastic 50A, AR-G1L, Agilus 30-1, and Ecoflex 00-30 (Fig. 2A & B). AR-M2 was classified as a hard material, while the remaining materials were considered soft. We evaluated the soft materials by analyzing their stress–strain relationships and prepared the samples according to the dumbbell-type specimen standards of JIS No. 7. For the construction of the rectum model, we used various methods,

including 3D printing and mold replication (Fig. S1). The results indicated that Ecoflex 00-30 most closely replicates the human rectum's stress-strain characteristics. The fabrication process required multiple molds, including inner, bottom, upper, and cross molds (Fig. 2C). The inner mold was produced in two halves and was subsequently joined with adhesive because of its length. These molds were assembled into a single unit, which the prepared Ecoflex 00-30 mixture (A:B = 1:1) was poured into. This assembly was then cured at 60°C for 1 h before being de-molded (Fig. 2D). To enhance the installation and testing process of the subsequent actuators, we partitioned the model into five segments (denoted as S-1 to S-5). The radius at the entrance of both ends of S-2 was set at 17 and 21 mm. Similarly, for S-3, the entrance radius at both ends was designed to be 21 and 28 mm. In the case of S-4, the entrance radius at both ends was 28 and 32 mm. The radius of S-5, which is cylindrical in shape, was 32.5 mm. The constructed rectum model was segmented into five parts, which represent the short anal canal and the upper and lower rectum, which exhibit complex curvatures and twisted angles along the axial direction of the rectum (Fig. 2E). These five segments were reproduced using the described methods (Fig. 2F).

Fig. 2 Material Selection and Fabrication Method of the Rectum Model. **A:** Stress-strain characteristics of various materials compared with those of the human rectum. **B:** Optical images of the rectum models printed from four types of soft elastic materials. **C:** Fabrication procedure for the rectum model. **D:** Optical image of the rectal molds. **E:** Photos of the rectum model alongside a 3D model. **F:** Fabrication method and optical images of each segment.

2.2 AAS muscle-inspired actuators

Inspired by the motion of the anus stimulated by the AAS muscles, we designed ring-shaped actuators to simulate the external sphincter's function in opening and occluding the anus (Fig. 3A). The external anal sphincter, a voluntary muscle under conscious control, enables individuals to contract and relax it at will, thereby facilitating control over bowel movement timing. To mimic this functionality, we introduced a soft pneumatic pouch actuator based on the principle that introducing air into a flat pouch causes it to contract longitudinally, as the air induces lateral expansion of the actuator [33][34]. Aiming for durability, we selected three types of thermoplastic materials: polypropylene (PP), biaxially oriented polypropylene film (BOPP), and high-density polyethylene (HDPE), and each measured 130 mm in length and 25 mm in width (Fig. 3B). To ensure a reliable connection to the pneumatic pumps, we used a commercial adapter (VRF206, MonotaRO, Japan), coupled with a relatively rigid polyurethane tube (inner diameter = 2 mm, outer diameter = 4 mm), and reinforced with a soft silicone tube (inner diameter = 3 mm). Subsequently, we applied a cyanoacrylate adhesive (Loctite 401) to seal the areas around the connecting parts. In the experimental setup, we applied a control signal characterized by a square wave with a period of 4 s and a duty cycle of 50%. To prevent potential damage from high pressure, a directional valve was employed to vent the system, thereby safeguarding the soft actuator from overexpansion.

Fig. 3C illustrates the durability of the three developed pouch actuators under applied pressures ranging from 20 to 100 kPa. The results indicated that the pouch actuators made from PP exhibited superior durability, outperforming those made from BOPP and HDPE. Specifically, the PP-made actuators maintained functionality for over 5400 cycles at pressures below 40 kPa, while their lifetime decreased significantly at pressures exceeding 60 kPa. In contrast, actuators made from BOPP and HDPE fail at pressures exceeding 50 and 30 kPa, respectively. To convert the linear displacement of pouch actuators into circumferential contractions, we construct ring-shaped actuators by rolling the flat pouch actuators and sealing the overlapped areas with double-coated adhesive tape (No. 5000NS; Nitto) (Fig. 3D & E).

First, we constructed three types of actuators to assess the influence of a soft shell on their performance: Type I (a pure pouch actuator), Type II (encased in polyethylene terephthalate [PET] paper), and Type III (encased in an A80 ring). The production of the designed PET paper ring and A80 ring was performed using a laser cutting machine (TROTEC Speedy 100) and a 3D printer (Form 3+, Formlabs), respectively. To evaluate the pressure output of these actuators, we devised a simple pneumatic circuit comprising a power module, a pressure sensor, and a rectal balloon catheter (46Fr 75 mL, Create Medic Co., Ltd.) (

Fig. 3F). The testing procedure involved inflating the catheter within the testing apparatus to a pressure of 10 kPa. Upon closing the valve and resetting the pressure sensor to 0 kPa, the actuators were mounted on the catheter, which allowed for the evaluation of their performance by inflating them to specified pressures.

Fig. 3G illustrates the three types of ring-shaped actuators, operating in both static and inflated states (input pressure 30 kPa). We determined the α values for each actuator type under variable applied pressures (

Fig. 3H). The results indicated that α increases with the application of higher pressures, reaching peak values above 20 kPa. Specifically, the maximum contraction ratio for Type I actuators approached 100%. Conversely, the maximum contraction ratios for Types II and III were significantly lower, which is attributable to the stiffness differences between the PET paper and the A80 ring. In addition, we assessed the pressure generated within the anal canal, thereby revealing a direct correlation between the increased actuator pressure and sensor readings (

Fig. 3I). It is noted that the Type III actuator ranks first when comparing the three actuator types because the relatively rigid cover restricts the spherical actuation and thus applies more pressure to the sensor. The peak pressure recorded for Type III actuators reached 6.2 kPa at an applied pressure of 30 kPa. Regarding the recovery time, Type III actuators demonstrated an enhanced ability to revert to their original shape, which is attributed to the rigidity of their materials (

Fig. 3J).

Fig. 3 Design, Fabrication, And Performance Characteristics of Anal Sphincter (AAS) Muscles-Inspired Actuators. **A:** Mimicking the process of AAS muscle movement using ring-shaped actuators. **B:** Pouch actuators made of three different materials and their connectors to the outer sections. **C:** Durability assessment of the three pouch actuators in response to varying levels of inflation pressure. **D:** Fabrication

procedure of the ring-shaped actuators with various encasements. **E:** Photos of the pouch actuator encapsulated in different materials. **F:** Photos of the measuring system for the generated pressure of the actuators. **G:** Comparative photos of the AAS muscle-inspired actuators in both static and inflated states for Types I, II, and III. **H:** Performance comparison of the three types of actuators in terms of area contraction ratio. **I:** Performance comparison of the three types of actuators in terms of generated pressure. **J:** Performance comparison of the three types of actuators in terms of recovery time.

2.3 Machine learning-enhanced modeling and optimizing process

To identify actuators that exhibited optimal performance characterized by significant contraction ratios, high generated pressures, and minimal recovery times, we engineered a variety of actuators with diverse structures, cover types, lengths, widths, and turns under static pressure (0–30 kPa). Subsequently, we performed experiments to assess their performance based on the three criteria: recovery time (t), contraction ratio (α), and generated pressure (P_g). Fig. 4A shows the workflow for the modeling and optimization process. The experimental setup comprised an air compressor, pressure gauge, pneumatic actuator, timer, and a catheter. Our data analysis protocol included data gathering, computation, model training, prediction, and parameter refinement to derive the most effective input parameters for the AAS-inspired pouch actuators. Fig. 4B shows various actuator types with distinct structural properties, ranging from Type 1 to Type 4, which were fabricated using the thermosetting process (Video S1 & Fig. S2). We further explored the impact of length on the contraction ratios of the pouch actuators under various pressure settings (Fig. 4C). In Fig. 4D, the actuators with varying widths (15–30 mm) were studied. In addition, the effect of changing the number of turns (1–3) on the performance of the Type 1 pouch actuators was systematically assessed (Fig. 4E & Video S2).

Considering the modeling process, preliminary steps involved data preprocessing to accommodate both categorical and numerical data types, followed by the segmentation of the dataset into training and testing subsets. In this study, we adopted a multilayer perceptron (MLP) model to estimate the output of the actuators, as it has been used in several fields [35][36]. The model, incorporating two hidden layers with 150 and 100 neurons, was implemented. We used the mean absolute error, mean squared error, and coefficient of determination (R^2) to evaluate the accuracy of the estimated models, which were 0.41, 0.59, and 0.99, respectively. Furthermore, we examined the error distribution for the contraction ratio, generated pressure, and recovery time of the pouch actuators (Fig. 4F–H). The error distribution histograms revealed that the majority of errors cluster near zero, which suggests that our model achieved a high level of prediction accuracy. The investigation of feature importance (Fig. 4I) highlighted that the cover material, coil turns, and structural design significantly affected the output parameters of the pouch actuators. Subsequently, this highly accurate model was used to identify the optimal inputs within the bounds of the physical constraints. To convey this objective, we utilized a straightforward function, which is described by the following equation:

$$f = \text{Min}[t / (P_g \times \alpha)] \quad (1)$$

By applying the modeling and optimization process, we determined the optimal values for the six input features: a length of 130 mm, a width of 25 mm, a single coil turn, an input pressure of 30 kPa, a paper material for the cover, and a Type 4 structural design. The comparison with the other actuators (Table

S1) revealed that our actuator achieved complete occlusion in large-diameter hollow tubes, typically those with diameters greater than 30 mm but less than 50 mm.

Fig. 4 Machine Learning-Enhanced Optimization Process Characterized by a Series of Systematic Steps and Outcomes. **A:** Workflow of multilayer perceptron (MLP)-based learning, encompassing an experimental system setup, data analysis, model training, predictive assessment, and performance evaluation. **B:** Feature 1: pouch actuators with different cross-section shapes; **C:** Feature 2: length (L); **D:** Feature 3: width (W); **E:** Feature 5: turns (T); **F:** error distribution for contraction ratio; **G:** error distribution for generated pressure; **H:** error distribution for recovery time; and **I:** analysis of feature importance.

2.4 Rectal module

The rectal module comprises a rectum model, five double-layer actuators, and five casings, both rigid and soft (Fig. 5). Fig. 5A illustrates the components and assembly process of the rectal module. For segment S-1, we selected the ring-shaped actuator coupled with a paper ring because of its superior occlusion capabilities, which were characterized by a higher area contraction ratio. For segments S-2 to S-5, the A80 ring was selected because 3D printing technology enables the production of soft shells in various shapes. The actuators, denoted as A1–A5, are highly adaptive, double-layer, pouch-based designs. They were engineered to reproduce a simplified system of anal opening–occlusion motion (A1), mimic the amplitude and width variations of the smooth muscles in the rectum to facilitate peristaltic motion (A2–A4), and prevent fecal backflow (A5), respectively. To control the radial expansion, four soft cases (SC1–SC4) and one rigid case (RC5) were created. Each segment of the rectum model incorporates curled pouch actuators and a 3D-printed soft shell, thereby enabling radial contraction. The assembly of the soft actuators and covers was accomplished by sewing using a needle and thread.

Fig. 5B shows the assembled rectal module from top and elevation views. The prepared actuators and cases are shown in Fig. 5C. Fig. 5D shows the actuators in their initial and inflated states, thus demonstrating that actuators A1, A2, and A3 could tightly occlude segments S-1, S-2, and S-3, respectively, while A4 and A5 do not achieve full occlusion of segments S-4 and S-5, respectively (Video S3). These results are in agreement with the expected range illustrated in

Fig. 1. To assess the effectiveness of the actuator systems in expelling the simulated fecal matter, we employed ordinary clay to mimic feces, shaping it into elongated forms and affixing it to the device. Upon application of pressure, it was observed that the actuator system could segment the simulated feces, thereby facilitating the automatic detachment of the cut portion (Fig. 5E & Video S4). Further analysis of the pressure generated by the system revealed that it was comparable to the minimum residual pressure and maximum squeeze pressure observed in the female anal region, as indicated in Fig. 5F [37]. By adjusting the pressure within 30–120 kPa, we matched the pressure levels found in the female human rectum. Regarding the contraction ratios, it was noted that segments S-2 and S-3 could effectively close, as viewed from perspectives A and B. The contraction ratios for S-4 were measured at 0.98 and 0.96 in views A and B, respectively (Fig. 5G). Furthermore, segment S-5, which was designed to prevent the

backward movement of the simulated stool and secure the entire model, maintained a certain degree of contraction, with ratios around 0.64 and 0.66. While S-2 and S-3 achieved a maximum contraction ratio of 1, S-4 also exhibited satisfactory contraction. Because its radius was 32.5 mm, S-5 displayed the lowest contraction ratio, which is in alignment with its primary function of simulating fecal retention rather than complete closure and mainly served as a barrier. Lastly, the recovery time of each segment following decompression (at an applied pressure of 30 kPa) was examined (Fig. 5H).

Fig. 5 Rectal Module: Rectum Mold and its Segments, Soft Cases, Rigid Case, and Actuators. **A:** Components and assembly process of the rectal module. **B:** Assembled prototype from top and elevation views. **C:** Optical images of the five pouch actuators (A1–A5), including four soft cases (SC1–SC4) and one rigid case (RC5). **D:** Photos of the pouch actuator deformation between their initial and inflated states. **E:** Demonstration of the small system (S-1, SC1, and A) severing the simulated fecal matter. **F:** Comparison of the pressures generated by the system (S-1, SC1, and A1) against the maximum squeeze pressure and minimum residual pressure typically observed in the human female anal region. **G:** Performance comparison of each segment inflated by actuators in terms of contraction ratios. **H:** Recovery time required for the actuators to return to their original shape.

2.5 Defecation robotic system

We designed the defecatory system, power supply system, pressure sensing system, and data acquisition system (Fig. 6). The overall system includes mechatronics, flush, stage, and rectal modules (Fig. 6A). Fig. 6B shows the schematics of the power supply, data acquisition system, and control system. The pneumatic supply system comprises an air compressor (SR-L30MPT-01), a combined pressure controller and filter unit (FRF300-03-MD), electromagnetic valves (CKD Company, MEVT series), a Koganei solenoid valve (030E1-L DC24V), an RS PRO switching power supply (PDF-600-24), a controller (Arduino Mega), pressure sensors (PSE564-A2-28), and a data acquisition system (LABVIEW, NI instruments). Throughout the experimental phase, the input pressure was adjusted to 500 kPa. We designed five pressure-regulating valves and solenoid valves to ensure precise gas supply and exhaust for each soft actuator. For accurate input pressure measurement, five pressure sensors were employed. Data acquisition is conducted via the LABVIEW DAQ system for pressure metrics, while the simulated weight is quantified using a precision balance (Fig. S3).

In this study, actuators A1 and A5 were primarily utilized for opening and closing the anus, as well as for preventing the backward movement of simulated feces. Actuators A2, A3, and A4 were utilized to generate the peristaltic motion for feces transportation. We performed tests on the movement capabilities of each simulator component (Fig. 6C). It was observed that upon activation of the various actuators, the interior of the rectum model underwent the corresponding deformations. A1 was able to effectively seal the anus with an input pressure of 30 kPa. During actual operation, control modes for A1 and A5 were configured to operate independently, whereas those of A2, A3, and A4 were managed via a pattern of inflation and deflation at predetermined intervals. The trajectory of defecating the cylindrical feces can be observed from the top and front views (Fig. 6D).

We formulated simulated feces with a specific viscosity by combining clay and olive oil to evaluate defecation performance. Fig. 6E shows that the robot achieves significant defecation capabilities when the mass ratio reaches 9%, with the utilized feces being cylindrical, having a diameter of 20 mm and a length of 70 mm (Video S5). Therefore, we selected a 9% ratio for further investigation. In addition, we examined the influence of the applied pressure (p) and one-quarter of the period (t) on the defecation efficacy of the robotic system (Fig. 6F & G). It can be concluded that the defecation speed consistently increases with the increase of applied pressure when t is maintained at 1 s. Moreover, we investigated the impact of t on defecation performance. As the actuator's squeezing speed increased, the defecation speed notably declined, thus indicating that feces were more effectively expelled along the inner wall of the rectum. Consequently, the optimal values for pressure and t were determined to be 30 kPa and 1 s, respectively, and were selected to maximize defecation velocity, thus closely mirroring the frequency of human defecation, as evidenced by magnetic resonance imaging findings. We further prepared feces samples with various diameters, lengths, and shapes (Fig.S4). Fig. 6H shows that both spherical and cylindrical shapes (diameter < 30 mm) can be expelled by our robot, whereas larger specimens (diameter > 30 mm) could not be expelled because the diameter of the simulated anus was 30 mm.

Fig. 6 Defecation Robotic System and Associated Components. **A:** Optical images of the defecation robot along with its pneumatic systems. **B:** A detailed schematic highlights the power supply system, data acquisition system, and control system. **C:** Top view revealing the movements of each actuator in relation to the input pressure sensors designated for each actuator. **D:** Optical photos of the defecation process from both top and front perspectives. **E:** Relationship between the defecation speed and the oil-to-clay weight ratio. **F:** Correlation between the defecation speed and the applied pressure. **G:** Relationship between the defecation speed and control time (t : quarter of the period). **H:** Defecation performance of the different types of feces with various shapes.

When humans experience FI, various situations may arise, such as diarrhea, constipation resulting in the anal to cut stool, and the occasional passage of excessively long stools. To address these issues, we have used our robot to simulate and replicate these scenarios. For the diarrhea (Fig. 7A & Video S6), we synthesized liquid feces using sodium alginate and water. In this scenario, actuators A2, A3, A4, and A5 were rendered non-operational, while A1 was closed for a period before being opened to simulate this condition. Fig. 7B illustrates the effective sealing capability of actuator A1, thereby preventing any leakage. The liquid feces can be rapidly expelled into the flush upon the activation of the A1 actuator. Experiments with pure water were also performed (Fig. S5). Moreover, our robot could simulate the human anal canal's ability to cut feces (Fig. 7C). Here, we adjusted the pressure on A5 to 1 kPa and set A4, A3, and A2 to simulate normal peristaltic motion at 30 kPa, with a control time (t) of 1 s each. After several peristalsis cycles, A1 initiated the feces-cutting process. The length and type of feces were regulated by the pulse-width modulation wave duration applied to A1. Short feces were produced when t_0 , t_1 , and t_2 were set to 12 s, 2 s (on), and 2 s (off), respectively (Fig. 7D). Conversely, extremely long feces were produced by setting t_0 , t_1 , and t_2 to 0 s, 1 s (on), and 1 s (off), respectively (Fig. 7E & F). In

this manner, A1, A2, A3, and A4 work in a coordinated wave for the peristaltic motion. Moreover, our robot achieved multiple feces-cutting actions by mimicking the human anus (Fig. S5), and it has the advantages of low input pressure, a long lifespan, and fast defecation speed. Finally, we compared our defecation robot or simulator with that of others and found that our robot is flexible, suitable for both liquid and solid feces, capable of discharging extremely long feces, and incorporates the rectal peristaltic characteristics (Table S2).

Fig. 7 Simulating Three Unique Defecation Scenarios Encountered in Humans Using our Developed Defecation Robot. **A:** Simulation of human diarrhea and the operational control strategies for each actuator. **B:** Optical images showing our defecation robot replicating the condition of human diarrhea. **C:** Motion control in the simulator's anal mechanism for cutting solid feces. **D:** Optical images showing our defecation robot replicating the cutting of solid feces. **E:** Replication of scenarios involving the expulsion of extremely long solid feces, accompanied by detailed control strategies. **F:** Optical images showing our defecation robot enables the production of extremely long, solid feces.

3. Conclusions

In summary, this study introduces a rectum-inspired, machine-learning-enhanced soft robotic system that aims to replicate three distinct defecation scenarios to investigate FI. The system encompasses a power supply system, pressure sensors, and a data acquisition system, along with a flushing mechanism, a stage, and a rectal module designed to mimic the peristaltic movement of fecal matter through the anal canal. The rectal module integrates a rectum model with actuators inspired by AAS muscles and is encased in both rigid and soft materials. Our evaluation of the various soft materials, through stress-strain analysis, identified Ecoflex 00-30 as the material that most closely mirrors the stress-strain properties of the human rectum. The replication mold method facilitated the fabrication of both the rectum model and its segments. In addition, we developed, produced, and assessed the AAS muscle-inspired actuators and optimized their functionality based on six criteria: pressure, structure, covering material, length, width, and turns. The initial modeling was achieved using an MLP model, which provided highly accurate predictions (R^2 near 1), thus enabling the identification of the optimal configurations within the physical limitations. Subsequently, the optimal parameters for the double-layer pouch actuator (Type 4) sheathed in PET paper were integrated into the rectal module, which significantly enhanced the contraction ratios and pressure generation, and reduced the recovery time. The completed robotic system effectively simulates peristaltic movements within the rectum and the occlusion of the anal canal, thereby reproducing common human conditions such as diarrhea, normal feces, and scenarios of prolonged fecal discharge under various control strategies. This innovative robot provides a foundation for new research into constipation-related factors through physical simulation and may facilitate the development of defecation-assistive devices for the elderly.

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Author Contributions

Z.M., S.S., and H.N. were responsible for the conceptualization, design, fabrication, and performance evaluation of the devices, in addition to writing and managing the project. S.M., K.S., and S.M. supervised the projects and facilitated the acquisition of funding. All authors have read and approved the final manuscript.

Conflicts of Interest

The authors declare that there are no competing interests.

Data Availability

Data supporting the findings of this study are available in the main text or the Supplementary Materials

Supplementary Materials

Fig. S1: Molds, fabrication methods, and equipment used in this study; stress–strain characteristics of the printed material; and photos of the AR-M2 samples.

Fig. S2: Manufacturing processes of the four actuator types.

Fig. S3: LabVIEW interface.

Fig. S4: Design and fabrication of simulated feces.

Fig. S5: Water defecation and cutting of feces twice.

Table S1: Performance comparison of actuators

Table S2: Performance comparison of defecation robots and simulator

Video S0: Code execution

Video S1: Four pouch actuator types

Video S2: Fabrication process

Video S3: Rectum module

Video S4: Demo of the cutting of simulated feces

Video S5: Defecation of the simulated feces with or without lubricant

Video S6: Simulation of three unique defecation scenarios encountered in humans

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