

MagnetoStalsis: Generating Peristalsis in an Artificial Bowel for Treatment of Short Bowel Syndrome

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Efforts to develop an artificial intestine for treatment of short bowel syndrome have thus far failed to achieve coordinated peristalsis. To address this, we introduce an implantable, biomimetic, magnetically-actuated peristaltic pumping apparatus. The system consists of a magnetic pump surrounding an intestinal graft and a rotating external magnetic field generator that produces peristaltic motion in the graft as the field direction alternates. We performed computational finite element modeling to simulate deformation and stress concentration on the pump. We then tested the ability of the system to produce fluid flow through porcine subintestinal submucosa at actuation frequencies from 0 to 4 Hz. The system achieved a maximal flow rate of 6.03 mL/min at 3 Hz actuation frequency, which is in the physiological range of human intestinal flow rates. This device represents a novel proof-of-concept for achieving untethered peristalsis in an implantable intestinal graft that is primed for preclinical testing in animal models of short bowel syndrome.

Keywords: magnetic actuation; biomimetic peristalsis; medical robotics; soft robots.

1. Introduction

Short bowel syndrome (SBS) is the primary cause of intestinal failure in children, presenting both life-altering and life-threatening challenges due to the residual intestine's inability to uphold nutritional homeostasis [1]. The causes of short bowel syndrome include necrotizing enterocolitis (a form of neonatal intestinal inflammation and ischemia affecting preterm infants requiring bowel resections in up to half of cases), congenital anomalies of the gastrointestinal tract (including intestinal atresia, gastroschisis, malrotation with midgut volvulus), and various rarer diagnoses such as Hirschsprung disease, trauma, and vascular etiologies [2, 3]. Medical therapies for SBS include long-term parenteral nutrition, which requires a long-term vascular catheter and is associated with bloodstream infections, vascular thromboses, and liver disease. Intestinal lengthening and tapering operations may be offered to children with severe bowel dilation, bacterial overgrowth, or failure to progress toward enteral autonomy with medical treatment alone [4]. Ultimately, children with persistent severe hyper-

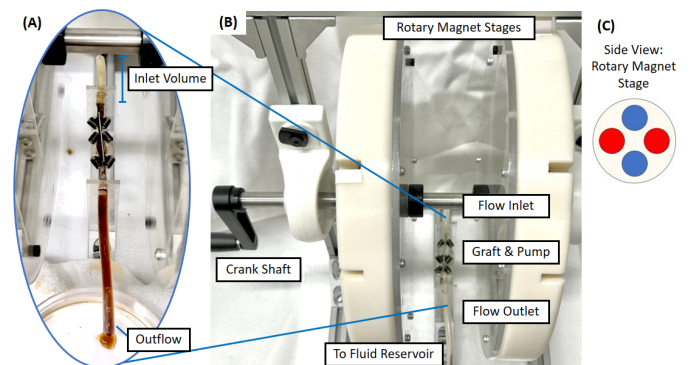


Fig. 1: The bowel graft within the pump is secured to an inlet and outlet for benchtop testing (A) within the full actuation system (B). Each rotary stage contains 4 disc magnets, arranged with alternating polarity as in (C).

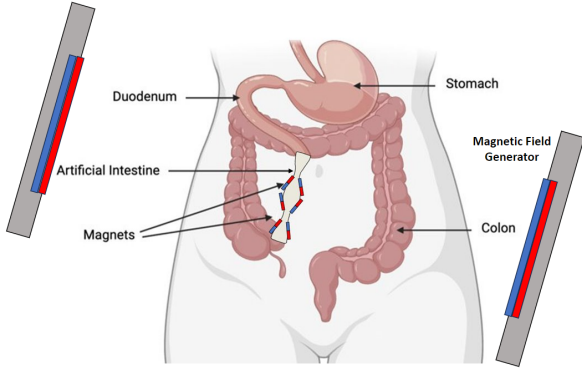


Fig. 2: Clinical configuration of the artificial intestine and pump, amidst the surrounding abdominal anatomy.

-bilirubinemia, requiring two or more intensive care unit admissions, or loss of most central venous access sites, may be referred for small intestinal transplantation [5], though allograft availability is certainly limited and complications of long-term immunosuppression can be severe.

A remaining novel option for treatment is the implantation of an artificial intestine. Hackam et al. have shown that a child's stem cells can be layered onto a scaffold that, when implanted in animals, acquires a blood supply and allows for nutrient absorption [6–10]. However, this artificial intestine has thus far not achieved coordinated peristalsis and thus is essentially an inert tube with limited ability to undergo digestion. The implementation of mechanical peristalsis to an artificial intestine, such as in Fig. 2, is necessary for this solution to become a viable treatment option for children suffering from short bowel syndrome.

Existing biomimetic peristaltic pumps employ a variety of actuation approaches (e.g., roller pumps, magnets, pneumatics [11,12], shape memory alloys [13,14]) to achieve peristaltic flow for diverse use cases [15]. However, none are specifically designed for providing long-term peristalsis in an implantable intestinal graft. Fuhrer et al. embedded magnetic microparticles in a silicone tube and actuated the tube with external solenoids [16]. However, their device was designed for stand-alone peristalsis, rather than within a biological graft, and was not tested for its ability to replicate the flow rates and contraction frequencies of native small bowel, which are critical considerations for functional implantation of an artificial intestine. Zhang et al. created a series of soft peristaltic pumping modules using a permanent magnetic elastomer [17]. The soft pumping modules were attached to a soft tube, and as current was applied to a coil within each module, the module deformed and squeezed the tube, facilitating peristaltic flow. Since a wired current source is required for actuation, this pump could not be implanted tetherlessly into the abdomen. Esser et al. introduced a pneumatic actuator to create peristalsis in a silicone device [18], and a similar pneumatic device by Dirven et al. was designed to mimic esophageal swallowing [19]. These pneumatic devices in their current form are too large for an implantable system, and are limited by the

need to include the actuation apparatus on the pump itself and the need for an external compressed air source. Lastly, there have been recent advances in achieving peristalsis using dielectric elastomers, flexible materials that change in size in response to an electric field [20]. However, dielectrics require an onboard power source and electronic components for actuation, which pose risks in long-term implantation.

Considering the limitations of other actuation approaches, we use magnets to achieve mechanical peristalsis. Magnetic fields are safe, can be used at large amplitudes with no adverse effects, and can allow untethered manipulators to be wirelessly controlled, making them potentially useful in a variety of medical applications. Shape-programmable swimmers [21], spiral microrobots [22, 23], and robotic capsules [24] have navigated to anatomical targets when actuated with varying fields. Magnetic nanorods [25,26], microgrippers [27], needles [28], and catheters [29] have been designed for specialized tasks such as grasping and suturing. Magnetic manipulators can tetherlessly generate controlled torque within the body, an ability that could be advantageous in controlling the contraction of a graft within the body. Our design is motivated by the need for an untethered, low-profile pump that can interface with pediatric small bowel and be actuated from a distance. In this paper, we present MagnetoStalsis (Fig. 1), the first biomimetic peristaltic system specifically designed to produce peristalsis in small bowel for therapeutic use. Unlike peristaltic pumps that require electronics or pneumatic airflow, our magnetic design is fully untethered, taking advantage of magnetism's actuation from a distance. Our design is low-profile in comparison to previous pumps, which require an onboard battery or a large device volume to achieve their functionality. We also test our pump's ability to produce peristalsis in a graft mimicking native small bowel tissue and employ strategies for attaching the pump to the graft, which have not been explored in other studies. Meeting these criteria, our system presents an innovative technical advance towards an implantable intestinal graft suitable for testing in small animals. The MagnetoStalsis pump (Fig. 3) is comprised of small rigid permanent mag-

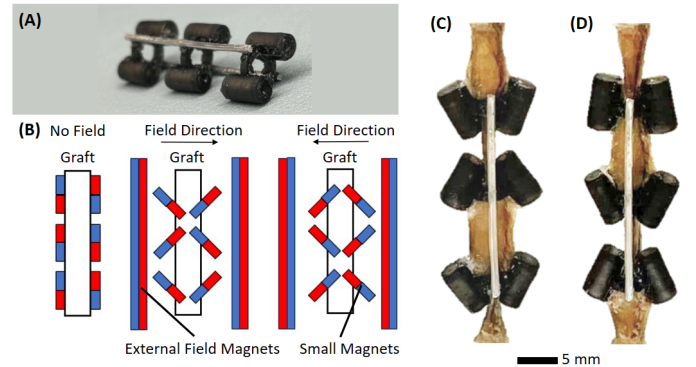


Fig. 3: MagnetoStalsis pump (A) and the corresponding magnetic polarity diagram for its actuation by an alternating external magnetic field (B). The pump deforms a bowel graft when acted on by a field in one direction (C), and the other (D).

nets encased in silicone and affixed to a biological graft. An alternating external field is generated by rotating stages of permanent magnets (Fig. 1) with controllable strength (depending on stage distance) and frequency. As the external field alternates in direction, the small rigid magnets in the pump respond with alternating torques that produce wavelike contractions in the graft. In this paper, we discuss our technical specifications and fabrication processes, and we use finite element analysis to simulate our pump’s deformation and stress concentration. We then characterize our system’s ability to induce volumetric contractions and pump simulated intestinal chyme at a range of actuation frequencies.

2. System Design

2.1. Technical specifications

The design of MagnetoStalsis was informed by the clinical constraints of small intestinal implantation. Our long term goal is to use this system in pediatric patients with short bowel syndrome, so the necessary dimensions, actuation frequencies, and flow rates were benchmarked against human pediatric values to inform our pump design. In the short term, we aim to demonstrate feasibility in a small animal, which is why our actuation system was built at that scale. Thus, our rotary stages are 4 inches apart, housing magnets that are strong enough to apply sufficient actuation torque from that prescribed distance. However, human-scale stages with the same torque capacity could be constructed with a larger distance and stronger rotary magnets. The graft material used was constructed from porcine small intestinal submucosa, which was chosen to approximate the material properties of Hackam et al.’s scaffolded artificial intestine for human implantation. Since transplanted pediatric human small intestine segments have measured approximately 5.5 cm in length and 4 mm in diameter, we used those dimensions for our graft. This limited our pump to three sets of magnets.

It was also necessary for the system to pump at contraction frequencies consistent with clinical values. Healthy adult humans undergo intestinal contractions at frequencies ranging from 0.12 to 11 Hz [30]. Our rotary stages are equipped with a crankshaft that can be paced to induce that full range of pumping frequencies. Lastly, we benchmarked our system’s flow rates against standard human clinical values (the intestinal flow rate in healthy adults ranges from 2.5 to 20 mL/min [31]). We aimed to achieve volumetric peristaltic contraction measurements consistent with human values; however, precise human measurements of bowel contraction could not be found in our literature review. Thus, we used the contraction of the native rat bowel as a benchmark, which contracts from 5 mm to 4 mm diameter [32]. Since the graft material contributed negligible resistance to deformation, it was sufficient to balance the external magnet strength, small magnet strength, and geometry to yield the desired deformations and flow rates.

Computational modeling and iterative fabrication were performed to design a working solution.

2.2. Fabrication

Our pump’s discrete magnet holders were printed on an Elegoo MARS3 printer using RESIONE F80 Black resin, cleaned, and cured for 15 seconds. Each cylindrical magnet chamber holds five N42 NdFeB disc magnets (K&J Magnetics) of 1/8 inch diameter. The pump’s backbones are 18 gauge aluminum, which is nonmagnetic, workable, readily available, and sufficient to prevent the cylindrical magnets from causing buckling and torsion in the graft. The pump is secured to the graft with conventional superglue.

The rotary magnet stages (Fig. 1) were printed on a Stratasys F170 printer. Each stage holds four N52 Ni-Cu-Ni disc magnets of 3 inch diameter (K&J Magnetics), arranged such that the magnets in one side of the stage attract the magnets in the other side, and that the four magnets in each stage alternate in polarity with respect to their angular position. This configuration generates an alternating field gradient at a position equidistant between the two stages (e.g., the pump’s location in Fig. 1), where the motion of the alternating field is approximately linear along the length of the pump due to the large radius of the stages compared to the small scale of the pump length.

An inlet and outlet were constructed from silicone tubing and adhered to the graft with superglue. Rigid holders constrained the inlet and outlet in air, and a fluid reservoir collected outflow. The graft was constructed from a sheet of porcine small intestinal submucosa extracellular matrix (CorMatrix Cardiovascular, Inc., Roswell, GA) that was cut to size and tubularized to a diameter of 4 mm with 8-0 nylon suture in running-locking fashion. The graft is suspended in air with a negligible pressure (0.2 cm H₂O) within its lumen due to the flow. This configuration is reasonable to replicate the anatomical pressure differential between the intraluminal and intra-abdominal pressures (which range between 6-13 cm H₂O [33–35] and 6-10 cm H₂O [36, 37], respectively). Due to the overlap of these ranges, we take the anatomical pressure differential to be negligible; however, we do not believe that a differential of ± 6 cm H₂O would impede the function of our system.

3. Device Modeling

3.1. Model setup

The MagnetoStalsis pump was modeled using COMSOL Multiphysics for analysis of deformation and stress concentration. The simulation used an imported geometry of the CAD file of a magnet holder pair. COMSOL’s built-in silicone material was employed to simulate the material characteristics of RESIONE F80 resin, as the RESIONE datasheet indicated that F80 exhibits properties closely resembling those of silicon rubber. The final material was defined with a density of 2329 kg/m³, a Young’s modulus

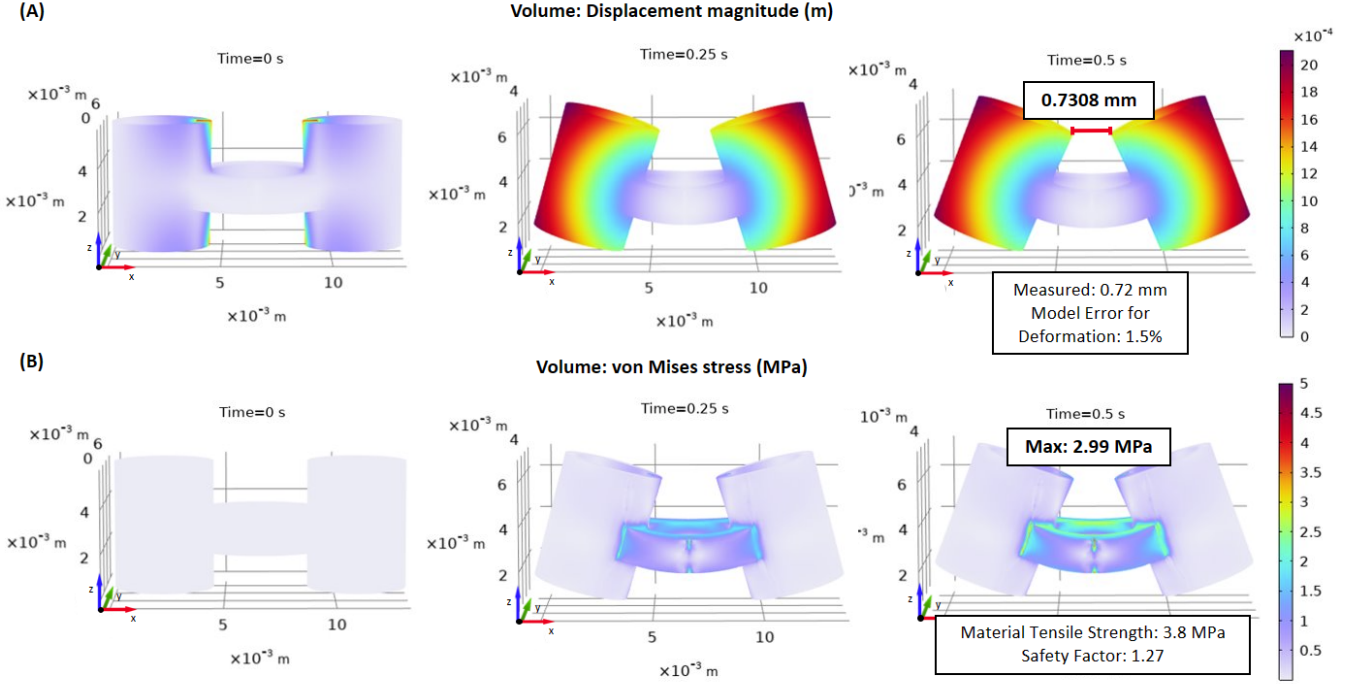


Fig. 4: Simulated 3D deformation of the device and resulting simulated gap width (A). Simulated 3D stress concentration of the device and resulting simulated maximum stress (B).

of 27 MPa, and a Poisson's ratio of 0.28. Within the Solid Mechanics interface, the entire domain was specified as hyperelastic and all initial values for displacement and structural velocity were set to zero. An x-y-z coordinate system was used. Fixed boundary constraints were added where the magnet holder would be affixed to the rigid backbone.

To simulate the forces generated by the magnets' motion, edge loads were prescribed along both magnet chambers. These edge loads are described by Eq. (1), where the load on one magnet chamber is $F(z)$ and the load on the opposite magnet chamber is $-F(z)$, facilitating opposing sinusoidal force profiles. In this equation, M is the maximum force in Newtons generated by a cylinder of small magnets (where the force profile of the cylinder ranges from $+M$ to $-M$, at its peak pivot angle), L is the length of the cylinder in meters, f is the frequency of external field alternation in Hertz, and t is time in seconds.

$$F(z) = M(z - (L/2))(2/L) \sin(2\pi ft) \quad (1)$$

The total torque τ (N · m) produced by a pivoting cylinder of magnets is calculated using Eq. (2), where m is the magnetic moment of the cylinder and B is the external magnetic field. This equation can be rewritten in terms of B_T , B_r , r , L , and μ_0 , where B_T is the strength of the external magnetic field in Tesla, B_r is the residual magnetism of the small magnets in Tesla, r is the cylindrical radius of the small magnets, L is the cylinder length, and μ_0 is the magnetic permeability constant $\mu_0 = 4\pi \times 10^{-7}$ henries per meter. The maximum edge force during pinching, M in

Newtons, can then be approximated as the torque τ over the moment arm L .

$$\begin{aligned} \tau &= m \cdot B \\ \tau &= \frac{2B_TB_r\pi r^2L}{\mu_0} \\ M &\approx \frac{\tau}{L} \end{aligned} \quad (2)$$

After prescribing sinusoidal edge loads in this way, we created a user-controlled extra fine mesh and set up the time-dependent solver to run from 0 to 2 seconds with a time step of 0.01 seconds. The simulation solved for 303141 degrees of freedom and converged in 1526 seconds.

3.2. Deformation

Since printed rubber materials are prone to variation and inconsistency, they can be quite complicated to model accurately in computation. Our approach, after building the model, was to incrementally apply changes to the built-in silicone's Young's Modulus until the deformation of the model matched the deformation of the pump (to a model error for deformation of 1.5%). Essentially, the modified Young's Modulus captures the other relevant physics not explicitly included in the model to account for the observed displacement. Then, with this sufficient model of deformation, we could go on to obtain reasonably accurate stress concentrations. The simulated 3D deformation of the pump

is shown in Fig. 4. The plot shows the pump at 0, 0.25, and 0.5 seconds, at which time the pump reaches one of its maximum deformation state. After fine-tuning the model, the pinched gap width of the pump in simulation was 0.7308 mm, compared to a measured pinched gap width of 0.72 mm $\pm$ 0.0147 mm ($n=3$) in our physical prototype, taken with a digital caliper. These values yield a model error for deformation of 1.5%, supporting that this model is sufficient to use for stress analysis of the design. However, the model still has many limitations, and future work will be required to accurately model the physics of the entire system (e.g., the dipole-dipole interactions of the cylindrical magnets and the resistance from the graft and chyme).

3.3. Stress concentration

A 3D plot of the stress concentration on the pump (Fig. 4) was generated from the same model. The maximum stress induced on the pump was 2.99 MPa, which yields a safety factor of 1.27 when compared to the tensile strength of the RESIONE F80 material. This safety factor is sufficient for initial proof-of-concept testing but should be increased for future animal tests in which the implanted pump should be able to withstand thousands of cycles of actuation.

4. Experiment and Results

4.1. Experimental setup

For experimentation, the full actuation system was configured as shown in Fig. 1. The graft was inserted into the pump's lumen using a dowel for guidance, and then superglued along each pump cylinder. The graft's ends were then glued flush with the silicone inlet and outlet. The inlet and outlet were press-fit within the tube holders, which constrained them in air. A mixture of water and coffee grounds was used to mimic the consistency of intestinal chyme. A volume of this mixture was syringed into the inlet such that the inlet volume was completely full and some of the mixture entered the graft. As the crank shaft was turned manually, the mixture was pumped through the graft and outlet. By recording the volume of mixture in the inlet, and the time for the inlet volume to fully clear, flow rate could be calculated by taking inlet volume over flow time.

4.2. Characterizing volumetric contraction

To characterize the contraction of the graft induced by our pump, measurements were taken with a digital caliper ($n=3$) of the graft's closed and open widths, constrained between stationary rotary stages, as shown in Fig. 5. The cross-sectional areas were approximated as ellipses, and the measured widths were taken to be elliptical axes (i.e., the closed width measurement of 0.72 mm being twice the semi-minor axis of the closed ellipse a_c and the open width measurement of 5.16 mm being twice the semi-major axis of the open ellipse b_o). Since the graft's circumference was

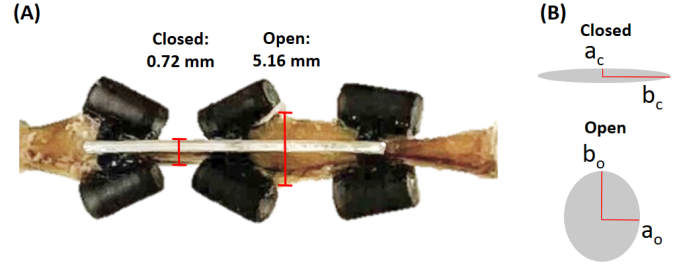


Fig. 5: Experimental measurements of the graft in open and closed states (A) and visual representation of the elliptical approximation (B).

known to be 13.35 mm, the Ramanujan ellipse perimeter approximation [38] was then used to calculate the remaining unknown axis lengths. With both elliptical axes known, it was possible to calculate the area of both ellipses. The closed ("pinched") elliptical area was calculated to be 3.71 mm², and the open elliptical area was calculated to be 13.05 mm². This yielded a percent area reduction of 71.6% from the open to closed state. However, since only a portion of the length of the graft is closed at a given time (45.5% of the graft's length), this percent area reduction is multiplied by that length to yield a volumetric contraction of 32.6%. In comparison, the full length of the rat bowel changes in diameter from approximately 4 mm to 5 mm during contractions [32], equating to a 35.97% volumetric contraction. Our pump's ability to induce contractions is quite similar to that of the natural rat bowel (9.83% difference), supporting that our pump is sufficient to induce peristaltic motion by generating volumetric contractions in the graft.

4.3. Flow rates at varying frequencies

Next, we tested the ability of our system to pump fluid. The frequency was controlled by manually rotating the crankshaft and using a metronome to pace the rotations (since the rotary magnet stages consist of 4 alternately polarized magnets, one full rotation of the magnetic stage corresponds to two full peristaltic cycles of contraction and expansion). It was initially hypothesized that a slight pressure differential between the inlet and outlet may cause fluid to flow through the graft in the absence of peristaltic motion. However, no fluid movement (0 mL/min) occurred in the absence of peristalsis (0 Hz), verifying that all fluid motion through the graft is attributable to the peristaltic motion of our magnetic system.

The measured flow rates at a range of tested frequencies, for five identically fabricated prototypes, are shown in Fig. 6. We are able to obtain the full human peristaltic contraction frequency range (0.12 to 11 Hz [30]) with our rotating actuator stages. Within this range, we ran consecutive tests starting at 0 Hz, increasing the frequency by 0.5 Hz for the first four measurements and then by 1 Hz, until the pump experienced fatigue breakage. Two prototypes

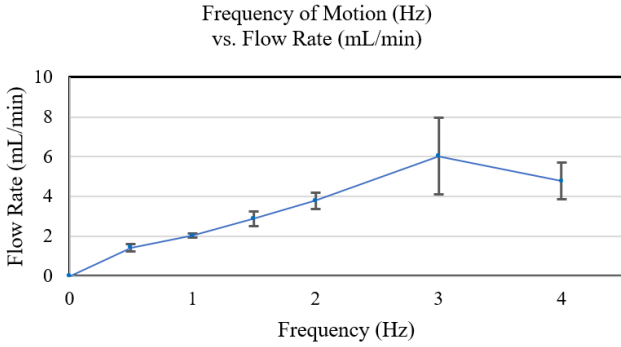


Fig. 6: Frequency of peristaltic motion (Hz) vs. the flow rate induced by the magnetic device (mL/min).

experienced breakage after 2 Hz, and three prototypes experienced breakage after 4 Hz (limiting the 3 Hz and 4 Hz data to only three prototypes). Flow rates were observed to increase with peristaltic frequency, which is consistent with other mechanical peristaltic pumps [39] and is reasonable given that biomimetic peristalsis is driven by an abrupt volume change (where more abrupt volumetric contractions occur at higher frequencies). However, flow rates were observed to decrease after 3 Hz, perhaps indicating that this system has an optimal actuation frequency above which marginal returns of flow rate can be achieved. We observed an average maximal flow rate of $6.03 \text{ mL/min} \pm 1.92 \text{ mL/min}$ at 3 Hz. While this is in the lower end of the range of intestinal flow rates in healthy adults (2.5 to 20 mL/min [31]), this flow rate is satisfactory for our purposes of facilitating nutrient absorption in a human bowel. As shown in [31], a jejunal flow rate of 5 mL/min, compared to 10 and 20 mL/min, resulted in a higher permeability ratio and higher absorption rates in humans. Our data show that our system is capable of producing flows within the physiologic rates of healthy humans to allow for nutrient absorption through an artificial intestine. Moreover, our results show that our system induces peristaltic pumping in comparison to a 0 Hz test case with no flow.

5. Discussion

Here, we have presented MagnetoStalsis, a magnetically-actuated mechanical system to achieve untethered peristalsis in an implantable intestinal graft. We designed a pumping device and magnetic actuation apparatus which are capable of producing controlled peristaltic motions from a distance. Using this setup with a subintestinal submucosa intestinal graft, we have successfully achieved fluid flows within physiologic rates at peristaltic frequencies within the range of native intestinal bowel. We further characterized the deformations and stresses on our pump through computational modeling to inform future iterations of this device. Taken together, our data demonstrate that our system is a viable approach to produce peristalsis in artificial intestinal

grafts, which thus far have resisted efforts to achieve spontaneous peristalsis through nervous system integration.

Several characteristics of our system are advantageous in terms of its intended clinical use for intestinal replacement therapy. The use of magnets harnesses their ability for action at a distance, enabling us to offboard the actuation mechanism from the implantable pump. This allows us to reduce the number of implantable components and the form factor of the pump, both of which may reduce patient morbidity and risk of device failure. However, as the strength of the magnets incorporated into the pump decreases, the external magnetic field strength must increase to achieve the same force and flow in the graft. Balancing this trade off will be critical in devising a useful and feasible system for clinical use. The modular design of the pump also presents a straightforward approach to producing peristalsis in intestinal grafts of different lengths by simply varying the number of magnetic pumping units. Pumps of various sizes could be manufactured to accommodate implants of different lengths or even personalized based on patient-specific intestinal geometries. Lastly, our setup would have the flexibility to be deployed either as definitive long-term therapy for patients requiring intestinal replacement, or as a temporary peristalsis aid to allow time for infiltration of the native enteric nervous system into the graft. Previous efforts to develop an artificial intestine have struggled to achieve enteric nervous system infiltration into an aganglionic graft [40]. However, enteric neural crest-derived neurospheres and human induced pluripotent stem cell (iPSC)-derived neural crest progenitors have shown promise in inducing peristalsis-like motion in intestinal organoid models [41, 42]. *In vivo* testing will be required to determine whether enteric nerves can successfully integrate in an implanted intestinal graft and whether these nerves can produce coordinated peristaltic waves without the aid of an artificial pump.

Besides being the only system to date designed to produce peristalsis in an intestinal graft, our device compares favorably to existing biomimetic peristaltic pumps in the literature. Fuhrer et al.'s silicone pump with embedded magnetic microparticles achieved a maximal flow rate of 78.5 mL/min at a 4 Hz actuation frequency [16]. Zhang et al.'s device produced a flow rate of 3.74 mL/sec in a cycle of four modules [17]. While Fuhrer's and Zhang's devices were not presented for any specific biomedical application, their concept of introducing permanent magnetic elastomer (PME) into a soft tubular pumping device is an appealing alternative to our use of permanent magnets that we intend to test in the future. While their devices were actuated with close-range solenoids and built-in coils, it is conceivable that a PME device could be actuated from a distance using our rotating magnetic field generator or more powerful electromagnets.

Our current system has several key limitations that we aim to address in future designs. First, the pump struggles to stay adhered to the intestinal graft with repeated peristaltic cycles using conventional superglue. Preliminary testing with other commercially-available adhesives and sutures has yet to identify a suitable solution to the adhesion

problem that will be required for long-term peristalsis. Second, the pump currently fractures after approximately 50 peristaltic cycles. These fractures occur near points of maximal stress identified in our simulations and are likely attributable to the relatively low safety factor of our device. To overcome this challenge, future prototypes could incorporate modified device geometry to minimize focal stress and stronger material composition. Additionally, while the cylindrical actuating magnets were sufficient for rapid prototyping a proof-of-concept demonstration, flat bar magnets could give the pumping device a smaller form factor while maintaining its functionality. Rubbing and pinching of surrounding tissue is also a potential concern; however, infants with small bowel syndrome have a more vacant abdominal cavity than healthy infants, and thus it is possible that the pumping device would not be in close contact with neighboring anatomy. Still, if inflammation or pinching is detected after small animal implantation, future devices could be modified to eliminate sharp edges in the pump or encased in a bio-compatible coating to reduce pinch points.

Another limitation of future clinical importance is the need for the external field to be near orthogonal to the magnetic dipole moment of the pump. If the pump significantly deviates from its orthogonal alignment, there may be undesired torques or an inability to induce the correct deformations for pumping. To prevent this, the implanted pump could be anchored, such as to the peritoneum as in Muramatsu et al.'s work [43], to constrain it orthogonal to the rotating actuators. An actuation system controlled instead by electromagnetic gradients [28, 44], and coupled to an appropriate sensor may be able to dynamically adjust the magnetic field in response to changes in pump orientation, allowing the pump to be actuated successfully in any orientation.

The current pump is also limited to an approximately straight path, and could not be used for complex S- or loop-shaped trajectories, since orthogonal control for one segment could not control the rest of the curved length. A longer straight pump could be controlled by increasing the radius of our rotary stages. A gradient-based electromagnetic system could perhaps control a pump placed along a curved or loop-shaped bowel segment by varying its field strength and direction across space, as in [28, 44].

Translating this technology for use in patients could take a variety of forms. In one approach, the rotating external magnetic field generator could be positioned on the sides of a crib or chair while the pediatric patient is stationary. These sessions could be conducted at regular intervals after meals, depending on the kinetics of food passage through the GI tract. Once the patient has passed their intestinal contents, they would be free to resume normal activity until the next session, being careful to avoid sources of strong magnetic fields. An electromagnetic actuator could potentially be incorporated into an even smaller form factor, such as a portable vest that the patient or a caregiver could turn on or off as needed.

6. Conclusion

In summary, we have designed, built, and characterized a system for producing peristalsis in an intestinal graft using magnets. Our setup represents a novel approach to address persistent shortcomings in the development of functional artificial intestine, which thus far have been incapable of achieving coordinated peristalsis. Combining a tissue-engineered intestinal graft capable of nutrient absorption with a biocompatible magnetically-actuated peristaltic pump like the one described has the potential to treat patients with short bowel syndrome and other disorders requiring intestinal replacement. Future work will focus on optimizing the pump's structure and mechanical properties to increase its durability, adding a motor to the magnetic actuation apparatus to improve control over actuation frequency, and modifying the pump to include only biocompatible components suitable for *in vivo* testing in a porcine model of short bowel syndrome.

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