Preliminary Mechanical Characterization of the Small Bowel for In Vivo Robotic Mobility

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In this work we present test methods, devices, and preliminary results for the mechanical characterization of the small bowel for intra luminal robotic mobility. Both active and passive forces that affect mobility are investigated. Four investigative devices and testing methods to characterize the active and passive forces are presented in this work: (1) a novel manometer and a force sensor array that measure force per cm of axial length generated by the migrating motor complex, (2) a biaxial test apparatus and method for characterizing the biomechanical properties of the duodenum, jejunum, and ileum, (3) a novel in vitro device and protocol designed to measure the energy required to overcome the self-adhesivity of the mucosa, and (4) a novel tribometer that measures the in vivo coefficient of friction between the mucus membrane and the robot surface. The four devices are tested on a single porcine model to validate the approach and protocols. Mean force readings per cm of axial length of intestine that occurred over a 15 min interval in vivo were 1.34 ± 0.14 and 1.18 ± 0.22 N cm⁻¹ in the middle and distal regions, respectively. Based on the biaxial stress/stretch tests, the tissue behaves anisotropically with the circumferential direction being more compliant than the axial direction. The mean work per unit area for mucosadhesion over the entire length of the porcine small bowel is approximately 0.55 J. The mean in vivo coefficient of friction (COF) of a curved 6.97 cm² polycarbonate sled on live mucosa traveling at 1 mm s⁻¹ is 0.016 ± 0.002 . This is slightly lower than the COF on excised tissue, given the same input parameters. We have initiated a comprehensive program and suite of test devices and protocols for mechanically characterizing the small bowel for in vivo mobility. Results show that each of the four protocols and associated test devices has successfully gathered preliminary data to confirm the validity of our test approach. [DOI: 10.1115/1.4005168]

Keywords: small bowel, mechanical characterization, peristalsis, tribometry, mucus adhesivity, biaxial

Introduction

The small bowel is a challenging environment for in vivo micro robots. A robot must navigate a tortuous path often steering around 180° bends, ascending and descending mucous membrane covered surfaces, and traveling against peristaltic forces generated by the migrating motor complex (MMC). In addition, the intestine wall is soft, delicate, and narrow, which constrains the size of a robot and provides minimal mechanical support [1]. If mobile within the lumen, robots could perform tasks that are currently not possible, such as gallstone removal, distal bowel biopsy acquisition, distal identification and repair of ulcers, and expansion of the capabilities of natural orifice transluminal endoscopic surgery (NOTES) [2–4]. Current research in this area is growing, with various teams developing robotic endoscopes using varying methods for locomotion [2,5–10]. Despite the progress indicated by these projects, limited research has been done to investigate mechanical properties of the GI tract. Some numerical modeling has been developed [11]; however, with the exception of a few groups [12–17], little work has been done to physically measure and understand the amount, duration, and local nature of these forces. In general, the forces experi-

enced by an intraluminal robot can be explained by four phenomena: active forces from the myenteron, biomechanical response of tissue, mucus adhesivity, and tribology.

Active Forces. The motor activity of the small bowel and the distal stomach creates the forces that act on an intraluminal robot. The myenteron of the small intestine is bilayered with mutually orthogonal circular and longitudinal muscle layers. The circular muscles are primarily responsible for segmentation waves, which increase peripheral resistance and slow transit of intraluminal contents. Longitudinal muscle contractions, on the other hand, facilitate transit [18]. Together, the two layers create the complicated rhythmic contractions within the bowel known as the MMC [19]. The intraluminal pressures and traction forces associated with the MMC are well understood [20–24]; however, the radial forces directly exerted by the bowel wall on a bolus have not been experimentally determined.

Biomechanical Response. The small bowel is an incompressible, viscoelastic, and multilayered organ. The morphology of the small bowel, and presumably the biomechanics, vary throughout its length. For example, the duodenum and proximal end of the jejunum are generally wider, more vascular, and have greater wall thickness than the ileum [25]. The myenteron provides most of the mechanical support while the mucous and serous layers

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provide little [3]. The cells in the two muscle layers are oriented in the circular and longitudinal directions and are mutually orthogonal. The cross-section of the lumen is somewhat asymmetrical due to its natural curvature and the attachment of the mesentery. The attachment point is highly vascular, whereas the outside radius of the lumen contains fewer blood vessels; therefore, the biomechanical response of the tissue could vary along the circumferential as well as longitudinal directions. Though important, the biomechanical response investigated in this work does not include the effects of the mesentery, whose contribution to the total force acting on the robot will vary depending on the locomotion technique of the robot. For example, the mesentery may contribute more to the biomechanical response of inchworm-type robots that generate significant deformations in the longitudinal direction [26], than to wheeled or legged devices that may cause more circumferential distension.

Mucoadhesion. The lumen of the small bowel, in its unanimated state, is a soft organ and will collapse if unsupported or empty. In its collapsed condition, the mucosa from opposite sides adheres to itself. For in vivo mobility within the small bowel, a robot must expend energy to overcome the adhesion of the opposing sides of the lumen in order to expand it sufficiently for passage.

Tribology. Characterizing friction between the robot's surface and the mucosa is important for mobility in terms of energy losses as well as mechanical advantage for movement. Current intestinal tribological research focuses on a particular robot of interest, and thus, the corresponding literature reports only data for the specific device being tested [27,28]. Other experiments move devices through an intestinal sample submerged in water in order to simulate the in vivo intra-abdominal pressure [9]. Some studies have conducted similar experiments by sliding rectangular and capsule shaped sleds along a flattened specimen [28] and sliding capsules of varying geometries and weights through a closed intestinal specimen [29]. Lacking in this subject area is in vivo experimentation and a generalized model.

The aim of this work is to create a suite of devices and test methods to mechanically characterize these phenomena: (1) an in vivo device, called the Migrating motor complex Force Sensor (MFS), consisting of a monometer and linear array of radial force sensors to measure myenteric force per cm axial length, (2) an in vitro biaxial stress/strain test apparatus to measure the mechanical response of the intestinal wall tissue to stress, (3) an in vitro mucus

adhesivity protocol consisting of a tensile tester to measure work required to overcoming mucoadhesion, and (4) an in vivo tribometer consisting of a linear actuator, sliding platen, and force sensor to measure friction forces between the mucosa of the intestine and the robot's casing. The results presented herein are preliminary and will be followed by multiple porcine studies in future work.

Methods

A test protocol was approved by the Institutional Animal Care and Use Committee (IACUC protocol 87,909(05)1D), and the following devices were used in a single live porcine model. To ensure the entire length of small bowel was clear of solids during surgery, the test animal was fed nothing but water and Jell-O® 48 h prior to surgery and then only water the last 24 h. After in vivo testing, the animal was euthanized, and sections of the duodenum, jejunum, and ileum were excised for further in vitro testing.

Active Force Characterization. The MFS is a novel device designed to measure the in vivo radial forces exerted by the myenteron (Fig. 1). The device consists of a temperature sensor, monometer, and an array of custom natural latex rubber force sensor balloons that expand to the nominal inner diameter of the intestine. The manometer measures intraluminal pressure, whereas the balloons measure the combined intraluminal pressure and the pressure generated by contact with the wall. The balloon array is segmented to independently measure the spatial and temporal changes in radial force along its 46 mm length. PTFE Tubing (0.81 mm ID x 1.5 m length) connects each balloon and the monometer to the ex vivo data acquisition and inflation system. Each balloon is independently monitored by a pressure transducer (Measurement Specialties 4426-05G). The transducer differential outputs are recorded to a personal computer via a data acquisition system (National Instruments, USB-6218).

To calibrate the MFS sensor, the balloon segments are fixed and a flat plate attached to a load cell is brought into contact with the sensor's surface. Sensor pressure and contact force are recorded and their relationship is accurately described with a quadratic polynomial. The load distribution in vivo, however, is radial as opposed to contact, so further testing was performed to verify that force versus sensor pressure change is independent of the location and distribution of the load on the sensor. The pressure outputs of each sensor segment are mapped to external load per unit length by solving the quadratic polynomial for force and dividing by sensor width.

$$F_{lm}(t) = \left(-c_{f2m} + \sqrt{c_{f2m}^2 + 4c_{f1m}(\Delta P_{s,dm}(t) - c_{Tm}\Delta T_{sm}(t) - c_{am}P_e(t))} \right) (2w \times c_{f1m})^{-1} \quad (1)$$

Fig. 1 Schematic of the MFS in vivo (left). Cross section of a single MFS balloon segment (right).

Fig. 2 MFS insertion (left) and fully inserted (right) into the jejunal region of the small bowel

where F_l is the external load on the sensor per unit length, $\Delta P_{s,d}$ and ΔT_s are the changes in sensor pressure and temperature due to the in vivo environment, P_e is the in vivo ambient gauge pressure, c_T and c_a are temperature and ambient pressure compensation coefficients, w is the width of the sensor segment, c_{f1} and c_{f2} are the contact load quadratic polynomial's coefficients, and m is an index indicating the particular sensor segment. Each compensation coefficient is determined through additional calibration procedures and contributes error to in vivo measurements.

Two MFSs are placed in the middle and distal regions of the small intestine to simultaneously gather data. To insert the sensors, an incision is made in the abdominal wall and a section of the middle bowel is pulled out and placed on the abdomen. An incision is made in the bowel and the sensor inserted 20 cm into the bowel past the incision (Fig. 2). The incision is sutured closed, and the process is repeated in the distal region. Once both sensors are in place, the small bowel is placed back inside the peritoneal cavity, the abdomen is sutured closed, data acquisition initiated, and the animal is left untouched in the dorsal recumbent position for approximately 15 min. During this time, blood pressure, core body temperature, respiratory rate, and heart rate are recorded along with the MFS data.

Biomechanical Response Characterization. The purpose of this characterization is to provide preliminary insight into the biomechanical behavior of the small bowel tissue. Specifically, we measure the biaxial stress-stretch response and the in-plane shear that occurs during the test. We use a biaxial test apparatus (Fig. 3) based on plans and software developed by Grashow [30] and Sacks [31], who present a detailed description of the equipment and its functionality. Briefly, square tissue samples are mounted on the device and pulled along two orthogonal axes. The deformation of the tissue is tracked via video of four black marks placed

on the sample surface. Four 1000 g load cells accurate to 0.20% of full scale (Honeywell Model 34) measure the force on each side of the sample. The camera (Sony XCD-X710) and telephoto lens provide an undistorted image and have a resolution of 1024×768 pixels with a field of view about $1 \text{ cm} \times 0.8 \text{ cm}$, which yields a displacement resolution of about $10 \mu\text{m}$. The linear actuators (404XR, Parker Daedal, Irwin, PA) have repeatable positioning within $3 \mu\text{m}$. Location error due to the video system resolution and machine vision software algorithm affects accuracy; however, it was not investigated for this preliminary work [32].

Square segments of porcine small bowel are excised within minutes of euthanization. Samples for testing are cut from the proximal, middle, and distal regions of the small intestine opposite the mesentery and stored in phosphate buffered saline at 3°C . Sixteen samples are tested (five proximal, five middle, and six distal), with mean side length and thickness of $26.6 \pm 1.9 \text{ mm}$ and $2.0 \pm 0.4 \text{ mm}$, respectively. Testing is performed at room temperature within 72 h of excision. Four black marks are placed in the center of the serous surface of the sample, which is then secured to the biaxial test device via nylon string and four equally-spaced hooks per sample side (Fig. 3). The black markers form the corners of a $\sim 2 \text{ mm}$ square, whose positions are tracked in real-time and are treated as nodes of a four-sided finite element, which allows direct measurement of deformation from the material, rather than assuming deformation from the linear actuator displacements. The area encompassed by the markers is about 1% of the sample size, which makes the marker area essentially free from edge effects due to specimen tethering [33]. The longitudinal and circumferential directions of each sample are aligned with the device axes one and two, respectively (indicated by the indices in Eq. (2) through Eq. (6) below). Prior to each test, the sample is preconditioned with 13 contiguous cycles of equibiaxial tension. The maximum tension for the samples is determined by stressing samples equibiaxially until tearing is visually observed in the submucosa at the hook-tissue interface (tearing of the serosa occurs at higher tensions). The maximum tension level for the tests is set to 140 N m^{-1} , which is 90% of the failure level. Tension is defined as

$$\mathbf{N} = \begin{bmatrix} \frac{P_{11}}{L_{22}} & 0 & 0 \\ 0 & \frac{P_{22}}{L_{11}} & 0 \\ 0 & 0 & 0 \end{bmatrix} \quad (2)$$

Table 1 The nine biaxial stress test protocols

Test number	1	2	3	4	5	6	7	8	9
Longitudinal tension (N_{11}), N m^{-1}	140	140	140	140	140	105	70	35	14
Circumferential tension (N_{22}), N m^{-1}	14	35	70	105	140	140	140	140	140

Fig. 3 Biaxial test apparatus and intestine sample ready for processing (inset)

Fig. 4 Steps in Mucus Adhesivity Characterization: Tubular intestine sample adhered to lower test mount (A). Upper grip brought into contact with sample (B). Cuts are made through the intestinal wall prior to separation (C). Two halves of the sample are pulled apart while recording force (F) and displacement (s).

where P are the loads, and L are the length and width of the sample in its reference configuration acting along the device axes. Note that N_{13} and N_{31} are assumed to be zero due to the assumption of plane stress, and N_{12} and N_{21} are assumed to be zero because the loading regimen is normal to the edges. Piola stress is calculated from tension by

$$\mathbf{T} = h^{-1} \mathbf{N} \quad (3)$$

where h is the thickness of the sample.

The test protocol consists of load-controlled biaxial testing where the ratio $N_{11}:N_{22}$ varies according to Table 1. The protocols are designed to be exhaustive, i.e., to probe the biomechanical response of the tissue from near uniaxial loading in the longitudinal direction (test one) to the near uniaxial loading in the circumferential direction (test nine). After 13 preconditioning cycles of equibiaxial tension, the sample undergoes the nine stretch tests in random order. The tension along each axis is ramped from near-zero pre-stress to the peak tension, back down to near-zero pre-stress using a triangular waveform [34]. Tension is applied at $0.14 \text{ N m}^{-1} \text{ s}^{-1}$, which is physiological considering interaction between a typical robotic endoscope and the tissue. Based on existing studies of the viscoelastic response of soft tissue, we assume that the stress-stretch response is independent of this, and similar, tension rates [35]. Each test is cycled 13 times, and all data reported in this study are taken from the loading curve of the 13th cycle. Data acquisition occurs at approximately 10 Hz, and at each data point, the locations of the four markers are recorded and Piola stress and tension are calculated from Eqs. (2) and (3). The biaxial deformation of the tissue sample is calculated as follows, similarly to work by Sacks and Sun [36].

$$x_i = \lambda_1 X_1 + \kappa_1 X_2, \quad x_2 = \lambda_2 X_2 + \kappa_2 X_1, \quad x_3 = \lambda_3 X_3 \quad (4)$$

where x_i and X_i are the vectors of a particle on the sample's surface in the reference and deformed states, respectively, λ_i and κ_i are the stretch ratios and measures of in-plane shear, and indices one and two correspond to the longitudinal and circular directions of the smooth muscle layers. Solving for λ_i and κ_i at each time point yields, in real-time, the deformation gradient tensor

$$\mathbf{F} = \begin{bmatrix} \frac{\partial x_1}{\partial X_1} & \frac{\partial x_1}{\partial X_2} & \frac{\partial x_1}{\partial X_3} \\ \frac{\partial x_2}{\partial X_1} & \frac{\partial x_2}{\partial X_2} & \frac{\partial x_2}{\partial X_3} \\ \frac{\partial x_3}{\partial X_1} & \frac{\partial x_3}{\partial X_2} & \frac{\partial x_3}{\partial X_3} \end{bmatrix} = \begin{bmatrix} \lambda_1 & \kappa_1 & 0 \\ \kappa_2 & \lambda_2 & 0 \\ 0 & 0 & \lambda_3 \end{bmatrix} \quad (5)$$

The Piola stress from Eq. (3) and stretch ratios from Eq. (5) are plotted against each other to obtain the stress-stretch curves. The shear angle is computed by

$$\alpha = \sin^{-1} \frac{2E_{12}}{\sqrt{1 + 2E_{11}} \sqrt{1 + 2E_{22}}} \quad (6)$$

where $\mathbf{E} = 1/2(\mathbf{F}^T \mathbf{F} - \mathbf{I})$, and $\mathbf{I}$ is the identity tensor [37]. The maximum shear angle that occurs during the uniaxial tests is recorded for each sample, then averaged based on intestinal region (proximal, middle, and distal).

Mucus Adhesivity Characterization. In this method, the energy required to separate two adhered mucosal surfaces from each other is determined. Adhesivity values from this test are useful for a preliminary evaluation of energy needed for a robotic endoscope to overcome mucoadhesion. Future work will determine adhesivity between the tissue and the robot's surface, as well as the effect of peel angle on the work of adhesion. An Insight II tensile testing machine (MTS Systems) measures the adhesivity using a 2 N load cell (MTS Systems, PN 100-090-965). The load cell's nonlinearity and hysteresis are no greater than 0.05% and 0.03% of full scale, respectively. Thirty tubular samples of intestine (each $25 \pm 1 \text{ mm}$ in length) are excised along the longitudinal axis of the proximal, middle, and distal small bowel (ten samples from each region). Samples are adhered to the test grips using cyanoacrylate (Fig. 4) and the intestinal wall is severed on either side of the sample without disturbing the mucus bond. The sample is then pulled apart at $2.5 \text{ mm per second}$ until the mucus is totally separated. The separation speed is chosen based on the anticipated speed and diameter of an in vivo robot. Future studies will investigate the rate dependence of the separation energy. Thirty tissue samples are tested. Force and position

Fig. 5 Tribometer with curved specimen tray (left). In vivo tribometer test (right).

Fig. 6 Radial contact force exerted by small bowel on bolus

during separation are recorded at 100 samples s^{-1} . Total work required for separation per unit area of intestine is calculated by

$$W_L = A^{-1} \int F(s) \cdot ds \quad (7)$$

where A is the mucosal surface area of the test sample, $F(s)$ is the force registered by the load cell during separation, and s is the separation displacement. The total energy required to overcome adhesivity of the collapsed bowel is

$$E_T = \frac{A_T}{2} W_L \quad (8)$$

where A_T is the total area of the collapsed mucosa of the small bowel.

Tribological Characterization. The purpose of this experimental method is to create a hardware platform that is capable of measuring in vivo tribological characteristics of the small bowel, which is something that has not been performed. A single sled with one geometry, material, speed, and pressure is tested both in vivo and in vitro to compare how the coefficient of friction (COF) varies on living versus excised tissue. In this preliminary test, a lumped frictional coefficient is experimentally derived. Future work will investigate the separate contributions of dry friction, tissue viscoelasticity, and mucus rheology to the overall friction coefficient. The tribometer is comprised of a linear actuator (Haydon-Kerk, 25,844-05-001ENG) that propels a load cell (Loadcell Central, ESP4-1KG) along a linear slide, with the load cell pulling a curved polycarbonate sled along the intestinal specimen (Fig. 5). A motor driver (Sparkfun, A3967), data acquisition system, and bridge strain measurement module (National Instruments, USB-

6218, USB-9237) control the motion of the sled and record position, time, and force.

The radius of curvature of the tray is 1.5 cm, which corresponds to the average diameter of the porcine small bowel and allows the position of the villi to face radially inward to maintain their natural in vivo orientation. The radius of the leading edge of the sled is 1.2 mm. To test, a live section of the porcine small bowel is pulled through a small incision in the abdominal wall. With the mesentery intact, the bowel is cut open along the longitudinal axis and placed on the tribometer specimen tray. The weighted polycarbonate sled is placed on the exposed mucosa and pulled at a rate of 1 mm s^{-1} . The sled weighs 0.53 N and has a contact surface area of 6.97 cm^2 , which creates contact pressure on the mucosa by the sled of 0.76 kPa. Total sample length and travel distance of the sled is approximately 90 mm. Two samples each from the proximal, middle, and distal small bowel are tested. The experiments are repeated in vitro with porcine specimens excised from the same animal and preserved in phosphate buffered saline (PBS) at 3 °C. The COF is calculated by

$$COF = \frac{F_t}{F_n} \quad (9)$$

where F_t is the force required to pull the sled at 1 mm s^{-1} , and F_n is the weight of the sled (0.53 N). Throughout the tests, the mucosa is kept hydrated with PBS at room temperature.

Results

In the following results, error bars and confidence intervals are one standard deviation of the mean.

Active Force Characterization. The mean radial contact forces per cm of axial length experienced by each sensor balloon segment are shown in Fig. 6. The overall means for each sensor are 1.34 ± 0.14 and 1.18 ± 0.22 N cm^{-1} in the medial and distal regions of the small bowel, respectively. Note the alternating force (high, low) pattern over the length of the sensor that is manifest both in the middle and distal bowel.

Biomechanical Response Characterization. A subset of the results from the stretch-tension tests are shown in Fig. 7 (left). The results are for three tissue samples, one each from the proximal, middle, and distal regions of the small bowel. The three pairs of curves in each graph correspond to the equibiaxial (thick line), near uniaxial (thin line), and 100% circumferential versus 50% longitudinal (dotted line) stretch ratios. Note that in the near uniaxial cases, the control algorithm achieves close to zero stress (but

Fig. 7 Left: stretch-stress results for proximal, middle, and distal tissue samples. Curves that are indicated with an L are stretched along the longitudinal direction, and C indicates the circumferential direction. **Right:** mean stretch values of all equibiaxial tissue samples at a post-collagen engagement stress level of 53 kPa.

Table 2 Maximum shear angle versus bowel region

Bowel Region	Shear angle (degrees)
Proximal	3.17 ± 0.82
Middle	5.34 ± 3.11
Distal	3.08 ± 2.42
Overall	3.81 ± 2.42

still positive) in the no-load direction. The tissue exhibits behavior typical of collagenous material, i.e., the biomechanical response of the tissue is stable and repeatable after preconditioning, and there are low and high modulus regions depending on the amount of stretch. Shown in Fig. 7 (right), are the mean stretch values of all equibiaxial tissue samples at a post-collagen engagement stress level of 53 kPa. Note that the tissue behaves anisotropically, with the circumferential direction exhibiting more distensibility than the axial direction. The maximum shear angle experienced by each sample during uniaxial testing is shown in Table 2. Also noteworthy is that preconditioning increased the nominal size of the sample by $119 \pm 18\%$.

Mucoadhesion Characterization. The mean work per unit area for mucoseparation of the small bowel is 0.09 ± 0.03 , 0.07 ± 0.03 , and 0.07 ± 0.02 mJ cm⁻² for the proximal, middle, and distal small bowel, respectively. The total energy to overcome adhesivity of the mucosa to itself over the entire 15 m length of the porcine small bowel is approximately 0.55 J.

Tribological Characterization. The mean COFs are 0.016 ± 0.002 and 0.019 ± 0.004 for the in vivo and in vitro tests, respectively. There is no significant difference in the COF for proximal, middle, and distal regions of the small bowel. Figure 8 shows the COF versus sled position for three runs on a single sample taken from the distal small bowel. The COF varies over the length of the test segment, which is a similar result to Baek et al., who suggest that the heterogeneity in the COF is “related to the local change in the viscoelastic property of the intestine” [29]. Another explanation is that the local changes in topography due to the many wrinkles and folds in the muscularis mucosae affect the COF. The inset in Fig. 8 shows the relative size of these wrinkles and folds, which are the same scale as the local variations in the graph of the COF. Also note that the first time the sled is pulled over the mucosa, it produces a higher COF than subsequent runs, which all yield similarly smaller COFs.

Fig. 8 COF versus sled position for three runs on a single sample (in vitro test, distal small bowel). Inset: portion of an intestine sample illustrating relative size of mucosa topographical features.

Discussion and Conclusions

We have initiated a comprehensive program for mechanically characterizing the small bowel for in vivo mobility. The results in this study are preliminary but have laid a foundation for the development of a comprehensive model of the intraluminal environment. The creation of such a model will assist in the design and optimization of an in vivo micro robot capable of intraluminal motion.

With the MFS, we have shown the feasibility of directly measuring the radial forces exerted by the myenteron on an intraluminal bolus. The measurements quantified in this work mark the first time these forces have been measured. The preliminary results are significant because they confirm theoretical force values found in the literature, e.g., findings by Miftahof et al. that the total force exerted by the bowel on a solid bolus is in the range of 0.15 to 1.9 N cm⁻¹ [11,38,39]. Knowing the radial force exerted by the myenteron provides designers of in vivo micro robots with additional biomechanical data that can be used for design optimization.

Although the tissue exhibits anisotropic behavior, with the circumferential direction appearing less stiff than the longitudinal direction, it is not certain these are axes of symmetry. Small in-plane shearing was measured (on average, less than $\sim 5^\circ$), which may indicate that axes of symmetry do not lie along the circular and longitudinal directions, or may simply be a result of misalignment of tissue samples during excision or mounting.

Several factors should be considered regarding the estimated energy consumption to overcome mucoadhesion. For one, the results in this work assume full separation of the entire mucosa, yet the cross-sectional profile of the robot will affect the degree of mucoseparation needed for passage through the lumen. A robot with a smaller profile could cause less separation in which case the energy expenditure would be smaller. In addition, the mucoadhesion test pulled apart rigidly supported parallel tissue samples. As a robot travels within the lumen, the peel angle of the mucosa will be less than 90° and will depend on the profile of the leading edge of the robot.

With the tribometer, we have shown feasibility of measuring the in vivo COF of a polycarbonate sled on live mucosa. The in vivo tribometry tests yield a COF that is slightly lower than that of the in vitro tests; therefore, previous work that is based solely on in vitro measurements is possibly slightly overestimating the COF.

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