Measurement of Peristaltic Forces Exerted by Living Intestine on Robotic Capsule

Muhammad Rehan, Ibrahim Al-Bahadly, Senior member IEEE, David G. Thomas, Ebubekir Avci, Member IEEE

***Abstract*—Using robotic capsules for assessing gut health has been an emerging field since the early 2000s with researchers attempting to perform diagnosis, monitoring and therapeutic functions inside the gut. The knowledge of peristaltic forces inside the intestine are crucial for designing the actuation mechanism of robotic capsules, however the impact of peristalsis on a capsule has not yet been quantified. In this work, an analytical model is presented to study the peristaltic movement of the small intestine. For the first time, finite element simulations were conducted in COMSOL Multiphysics to generate intestinal peristaltic forces, and analyse their impact on a robotic capsule. A capsule prototype (30 mm x ϕ12 mm) was developed to measure the peristaltic forces from living intestinal tissue, while an embedded system was used simultaneously to record the live data from the capsule-intestine interaction. In in-vitro experiments, the intestine applied an average axial force of 226 mN and contraction cycles of 9 times/min, while the capsule prototype experienced maximum radial force of 180 mN. A specialized in-vitro setup is developed to keep fresh ex-vivo intestine samples alive for up to 6 hours, while the capsule prototype measured the intestinal forces from the living tissue. This in-vitro experimental setup provided an excellent model for the in-vivo environment in terms of generating peristaltic movements, hence this force analysis will help in developing efficient prototypes for locomotion, anchoring, localization, biopsy, drug delivery and sampling mechanisms for robotic capsules.**

I. INTRODUCTION

The beginning of 21^{st} century was marked by a revolution in medical imaging when Iddan et al developed a minimally invasive capsule endoscope as an alternative to the tethered endoscopy [1]. A capsule endoscope moves passively inside the gut, in a similar way to food, with the help of peristaltic movement and captures images of gut lesions. One of the limitations of a capsule endoscope is its uncontrollable movement, which restricts its navigation inside the gut and it is not possible to spend significant time at the region of interest. Another limitation is its incapability to determine its precise location within the 9 meter long gut. Therefore, locomotion mechanisms have been developed to anchor at, or navigate towards, the target site [2]–[5]. Similarly, localization mechanisms were proposed to estimate the position of the capsule endoscope [2], [3], [6]. These advancements in the field of capsule endoscope laid the foundation for devices, with similar size, to perform diagnostic, monitoring and therapeutic functions like sampling [7], [8], tissue biopsy [9], sensing [2], [10] and drug delivery [11].

Advance techniques in robotic capsules such as locomotion, localization, biopsy, drug delivery and sampling, require knowledge of peristaltic forces to develop suitable actuation mechanisms. In locomotion based designs, the stopping mechanism needs to overcome the peristaltic forces to anchor at the site of interest [5]. Similarly, the contraction rate, peristaltic pressure and gas sensing can be used for localization, as this information can assist in estimating the position of the capsule [10], [12]. The biopsy tool counters the peristaltic forces in order to penetrate the gut wall, to capture the tissue [9]. Likewise, the sampling device also acts against the peristaltic forces to open its mechanism, to collect the specimen [7]. Once the peristaltic forces, applied by the intestine on the capsule endoscope or robotic capsule, are known, an accurate actuation mechanism can be designed for incorporating the additional features. Therefore, quantification of intestinal peristaltic forces can produce significant contribution for all these devices and add-on mechanisms.

Methods have been devised to measure the intestinal pressure, which can later be converted to peristaltic contraction forces based on the size of the measuring device, but they have certain limitations. Mostly, endoscopic manometry is used to measure the pressure inside the gut; however, this tethered method limits its reach into the small intestine. Secondly, this method involves a high risk of gut perforation and bleeding, and the procedure is invasive and unpleasant for a patient [2], [13]. A commercial capsule, SmartPill motility capsule (Medtronic, Minneapolis, US), and several other laboratory prototypes have measured gut pressure using minimally invasive robotic capsules with MEMS based sensors [12], [14]–[16]. These capsules have mainly recorded the intraluminal pressure within the gut, which was the cumulative pressure of each region, and it is not possible to extract the peristaltic pressure from the overall pressure signal [13], [14]. Furthermore, these sensors often capture breathing and heartbeat signals which need to be separated from the primary signal [12], [15]. Although the intraluminal pressure is helpful in treating gut related diseases, it provides less information on gut motility (peristaltic behaviour) and hence it is not possible to quantify the peristaltic forces.

Thus, this article describes how a tactile sensor based

*This research was supported by the KiwiNet grant 2020. Support was also provided by the Higher Education Commission (HEC), Pakistan under grant no. 5-1/HRD/UESTPI(Batch-V)/2421/2018/HEC.
Corresponding author: Ebubekir Avci (email: e.avci@massey.ac.nz).
The authors are with Massey University, Palmerston North, New Zealand.
M. Rehan is also with Department of Electronic Engineering, Sir Syed University of Engineering & Technology, Karachi, Pakistan.
E. Avci is also with MacDiarmid Institute for Advanced Materials and Nanotechnology, Wellington, New Zealand.

robotic capsule was developed to measure the peristaltic forces, directly from the living intestine. A biomechanical analytical model of the small intestine was developed for analysing the peristaltic forces. Finite element simulations are conducted in COMSOL Multiphysics, modelling the peristaltic behaviour of the small intestine, as elaborated in Fig. 1. In addition, an investigation of the impact of the peristaltic contraction forces on a robotic capsule prototype was conducted, which has not been done before. Simulation provided an in-depth analysis of peristaltic behaviour by providing the freedom to change the design parameters, such as intestinal diameter, and observe its impact on to the robotic capsule. A capsule prototype was then fabricated with a force sensor and an embedded system was devised to measure the peristaltic forces from living intestinal tissue. A specialized in-vitro setup was utilized to keep the intestine alive, while the robotic capsule recorded the peristaltic forces of the intestine. The three main contributions of this work are as follows.

1) The development of an analytical model and then simulation in COMSOL Multiphysics to measure the impact of peristaltic forces on the robotic capsule,
2) The development of a capsule prototype with a force sensor and an embedded system to measure the peristaltic forces from living intestine,
3) Rigorous discussion about quantification of peristaltic forces and its impact on robotic capsule designs.

The peristaltic motion developed in COMSOL in this study can be replicated by researchers to test advance techniques in robotic capsules like locomotion or anchoring mechanisms, by incorporating their design into robotic capsules. Similarly, the in-vitro setup in this study, which closely replicates in-vivo studies as it maintains the intestine alive for 6 hours, can be used to examine the robotic capsule under living intestinal tissue conditions. Furthermore, the impact of peristaltic forces on a capsule inside the intestine is quantified in this study and this knowledge will help in developing efficient tools for robotic capsules for biopsy, drug delivery, sampling, locomotion and localization mechanisms.

II. Materials and Methods

A. Biomechanical modelling of the small intestine

The movement of food through the small intestine is relatively poorly studied, as this region of the gut cannot be assessed through tethered tools. A pulpy fluid containing digested food and gastric juices known as chyme enters the small intestine from the stomach. The chyme is simultaneously mixed and moved along the length of the intestine using two main phenomena, namely peristalsis and segmentation. Peristalsis creates a wave-like motion by contracting and relaxing the muscles in circular direction, hence moves the chyme along the small intestine. Whereas, segmentation is mainly responsible for mixing of the chyme by creating a forward and backward motion in a longitudinal direction. The flow inside the intestine is described by the conservation of momentum and conservation of mass, which can be expressed by Navier-Stokes equations and continuity equation respectively. The

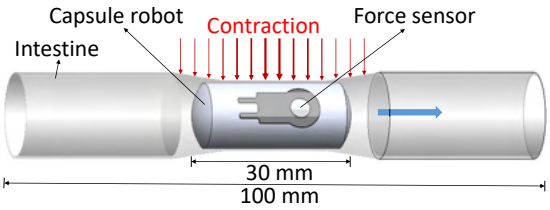


Fig. 1: Biomechanical model of the small intestine.

Navier-Stokes momentum equation in Cauchy momentum form can be defined as [17]

$$\frac{Du}{Dt} = \frac{1}{\rho}\nabla \cdot \sigma + g \tag{1}$$

Where, u is the fluid velocity, ρ represents the density, ∇ is the divergence, σ is the Cauchy stress tensor, g represents the gravity, and $\frac{D}{Dt}$ is the material derivative (Lagrangian derivative) that is the time rate of change of fluid and can be defined as

$$\frac{Du}{Dt} = \frac{\partial}{\partial t} + u \cdot \nabla \tag{2}$$

Where, $\frac{\partial}{\partial t}$ represents the change with respect to time at a given point. The Cauchy stress tensor (σ) expresses the changes in the material due to the deformation. This can be defined as a sum of deviatoric stress (changes in shape) and volumetric stress (changes in volume). Deviatoric stress can be expressed as a function of viscosity while volumetric stress can be considered as pressure, which are defined below

$$\sigma = -p + \mu(\nabla u + (\nabla u)^T) - \frac{2}{3}\mu(\nabla \cdot u)I \tag{3}$$

Where, p represents the pressure, μ is the dynamic viscosity, ∇u is the velocity gradient, $(\nabla u)^T$ is the transpose of ∇u and I is the identity matrix. The overall flow through the intestine can be described by the conservation of mass in the form of continuity equation as follows [17],

$$\frac{D\rho}{Dt} + \rho\nabla \cdot u = 0 \tag{4}$$

For incompressible fluids, the density of the fluid remains constant so (4) can be reduced to [18]

$$\rho\nabla \cdot u = 0 \tag{5}$$

Considering the assumption that the flow inside the intestine is incompressible (refer to (5)), (3) can be re-written as,

$$\sigma = -p + \mu(\nabla u + (\nabla u)^T) \tag{6}$$

Navier-Stokes momentum equation (1) can be modified by using (2) and (6), and expressed as follows [18]

$$\rho\frac{\partial u}{\partial t} + \rho(u \cdot \nabla)u = -\nabla p + \nabla \cdot \mu(\nabla u + (\nabla u)^T) + \rho g \tag{7}$$

Therefore, (7) can be described as Newton's second law of motion in which the left side defines the inertial forces wihin the fluid while right side is a sum of pressure forces, viscous forces and external forces. Equation (5) and (7) are used in COMSOL multiphysics to define the fluid flow. The effect

of gravity on fluid flow was negligible so it was ignored in simulations and the flow was considered as newtonian fluid with dynamic viscosity (μ) of $0.0014Pa \cdot s$ and density (ρ) of $1040Kg/m^3$ [18].

The peristaltic movement is generated when the intestinal muscles apply the forces inward, which results in deformation of the intestine. The original location of a point on the material is denoted by X while the new location of the point after deformation is indicated by x and the displacement vector between the two points is expressed by $w(X,t)$. The momentum conservation for an arbitrary (undeformed) volume V_0 is

$$\frac{d}{dt}\int_{V_0}\rho_0 v dV = \int_{V_0} f_V dV + \int_{\partial V_0} T dA \tag{8}$$

Where, ρ_0 is the undeformed density, v is the velocity of deformation, f_V represents the volumetric forces in the undeformed region and T is the traction of forces acting in the undeformed area. The traction using its spatial components (T_i) and normal vector using its material components (N_J) can be further defined as

$$\int_{\partial V_0} T_i dA = \int_{\partial V_0} P_{iJ} N_J dA = \int_{V_0} \frac{\partial P_{iJ}}{\partial X_J} dV \tag{9}$$

Where, small indices (e.g. T_i) are used to define spatial components, and capital indices (e.g. N_J) are used for the material components. P_{iJ} is the tensor component and X_J expresses the original location of the material particle. The velocity of deformation (v) can be represented as a function of displacement (w)

$$v = \frac{\partial w(X,t)}{\partial t} \tag{10}$$

The volume is arbitrary, so the differential form of momentum balance can be achieved by substituting (9) and (10) in (8), as shown below

$$\rho_0 \frac{\partial^2 w_i}{\partial t^2} = f_{V_i} + \frac{\partial P_{iJ}}{\partial X_J} \tag{11}$$

Where, w_i is the spatial displacement, f_{V_i} is the spatial component of volumetric forces and P_{iJ} is the spatial tensor component. The tensor form of (11) is shown below

$$\rho_0 \frac{\partial^2 w}{\partial t^2} = f_V + \nabla_X \cdot P^T \tag{12}$$

Where, P is the first Piola-Kirchhoff stress tensor that signifies the relationship between forces acting in the spatial direction to the original location (undeformed configuration). This can be related to Cauchy stress tensor (σ) as follows

$$PNdA = \sigma n da \tag{13}$$

Where, N is the normal vector before deformation, dA is the area before deformation, n is the normal vector after deformation and da is the area after deformation. The deformation will change the area, which can be computed by Nanson's formula

$$nda = JF^{-T}NdA \tag{14}$$

Where, F is the deformation gradient tensor and J is the volume factor which can be computed by the determinant of F. Hence, the (13) can be re-arranged as

$$P = J\sigma F^{-T} \tag{15}$$

Similarly, the Kirchhoff stress tensor τ can be expressed as,

$$\tau = J\sigma \tag{16}$$

Therefore,

$$P^T = \tau^T F \tag{17}$$

The density of undeformed and deformed materials can be related, based on mass conservation as,

$$\rho = J^{-1}\rho_0 \tag{18}$$

The momentum balance equation in terms of Cauchy stress tensor can be written as

$$\rho \frac{\partial^2 w}{\partial t^2} = f_v + \nabla_x \cdot \sigma \tag{19}$$

Where, ρ denotes the density of the deformed material and f_v represents the forces per deformed material. These analytical equations are used to develop a peristaltic wave model for the small intestine. The details for the modelling and simulations are shown in the next sections.

B. Design considerations for simulations

A small intestine sample 100 mm in length and 15 mm in diameter, was used in this study, as shown in Fig. 1. A robotic capsule (30 mm x ϕ12 mm) was placed inside the section of intestine. The robotic capsule was moved inside the intestine, mainly, by peristaltic movements. While interacting with a solid object, like a robotic capsule, the dominant function of intestine is peristaltic movement as compared to segmentation which was not considered in these simulations. The forces were repeatedly applied at the centre of intestine section, after an interval of 7 seconds [19]. The deformation of intestine was defined by the radial contraction ratio and was selected to be 78% based on a related study [19]. The robotic capsule received the pressure from intestinal muscles based on deformation and is recorded over the time to measure the impact of radial contraction forces on the capsule. The biomechanical model of the small intestine was simulated by considering the following assumptions.

1) The geometry of the small intestine is considered as a cylindrical shaped tube with a mean diameter of 15 mm in relaxed state.
2) A small section of 100 mm from the small intestine is modelled to reduce the computational complexity.
3) The force is applied once in each cycle, at the centre of the 100 mm section of the small intestine.
4) An incompressible Newtonian fluid flow is considered inside the small intestine.

A Finite Element Analysis (FEA) based software, COMSOL Multiphysics (version 5.5, COMSOL AB, Stockholm, Sweden), was used to develop the biomechanical model as shown in Fig. 1. The modelling parameters were selected based on related studies [18], [19].

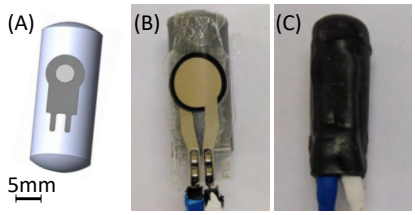


Fig. 2: Development of robotic capsule prototype. (A) CAD model (B) Capsule prototype with force sensor glued on the capsule. (C) Capsule prototype with water resistant lamination.

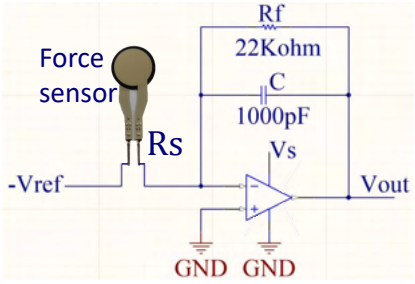


Fig. 3: Inverting Op-amp circuit for reading force measurements.

C. Design requirements and fabrication of the capsule

The dimensions of a robotic capsule should be small enough to allow transit through the entire GI tract. A capsule with size of 30mm x ϕ 12mm is considered to be safe [7], [12], [20]. The force measurement system should be sensitive to small forces, such as 10mN, so it can account for all activities related to the forces inside the intestine [21]. In addition, the measurement should be fast enough to process the force readings in less than 1s [21]. Furthermore, the robotic capsule needs to measure the contraction force in live animals or humans, hence the force sensor should operate between 35 and 40°C and in high moisture conditions. The computer-aided design (CAD) of robotic capsule with force sensor, developed in Solidworks (Dassault Systèmes SolidWorks Corporation, USA) is shown in Fig. 2 (A). The force sensor (FlexiForce ESS301 Sensor, Tekscan Inc., USA) is glued on to the outside of the robotic capsule as shown in Fig. 2 (B). The force sensor was laminated with water resistant tape to eliminate direct contact with the fluid as shown in Fig. 2 (C) and it was calibrated before each trial so its measurements were not affected. The sensor was exposed to the gut tissue; hence, the readings were obtained directly from the peristaltic forces of the intestine.

The capsule prototype was fabricated with Digital Light Processing (DLP) technique using a 3D printer (Hunter, Flashforge, China) with the resolution of 25 μm. The overall length of the capsule prototype, as shown in Fig. 2, was 30mm and its diameter was 12mm. The force sensor was calibrated to measure forces from 0 to 4N and its resolution was 5mN. The baud rate for collecting the data was set to 9600 bps, which processed the data in 5 μs. The force sensor had an operating

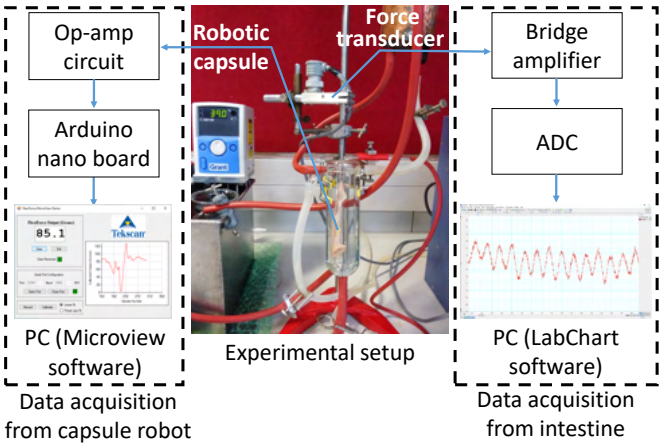


Fig. 4: Data acquisition system for peristaltic and axial force measurements. (Left side) data acquisition system of robotic capsule measures the peristaltic forces. (Right side) data acquisition system of in-vitro system measures the axial forces.

range from -40 °C to 85 °C. The force measurements were recorded by an embedded system, which is explained in the next section.

D. Data acquisition system for robotic capsule

The calibration is important in tactile sensors and it becomes more evident in this study as the gut environment is dynamic in nature. An electronic kit (FlexiForce Prototyping Kit, Tekscan Inc., USA) equipped with an Arduino nano board and an operational amplifier (op-amp) circuit was used to collect the force readings from the robotic capsule sensor. The op-amp circuit provided the flexibility required for calibrating the sensor inside the dynamic intestinal environment, by adjusting the feedback resistance and drive voltage, as shown in (20). An inverting op-amp configuration was used with a 22 KΩ resistor and 1000 pF capacitor in the feedback loop, as shown in Fig. 3. The V_{ref} signal was set to an square wave with 20 Hz frequency and 20% duty cycle. The drive voltage was made variable in the circuit, having multiple selections between 0V and 5V.

$$V_{out} = -V_{ref}\frac{R_f}{R_s} \tag{20}$$

Where, V_{out} is the output voltage, V_{ref} is the reference voltage, R_f is the feedback resistance and R_s is the resistance offered by the force sensor.

A specialized software (FlexiForce MicroView, Tekscan Inc., USA) developed for reading the force measurements, was used to record the force data. The op-amp circuit converted the force measurements to analog data, which were converted to digital form using built-in analog to digital converter (ADC) of Arduino nano board. The Arduino nano board was connected to the computer through a usb port. The MicroView software displayed the data on-screen for real-time utilization and also recorded the data for future analysis. A systematic diagram with force sensor, electronic kit and GUI view of the

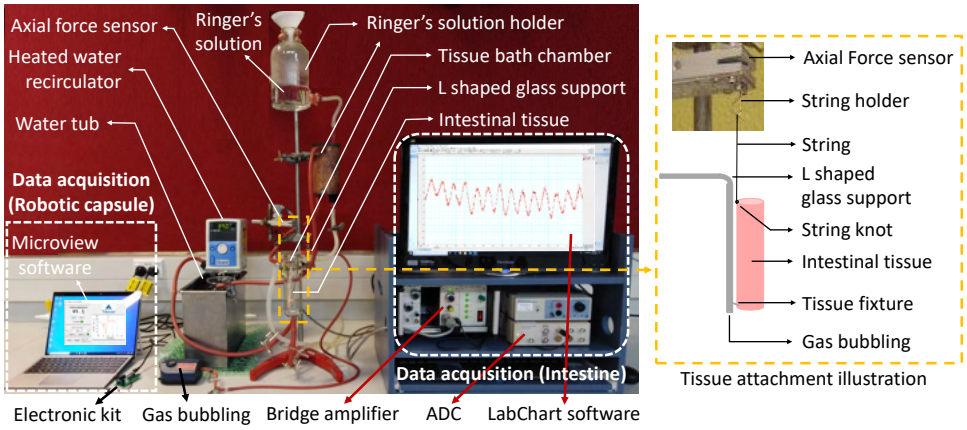


Fig. 5: In-vitro experimental setup for postmortem tissue with data acquisition systems for robotic capsule sensor and in-vitro axial force transducer. The intestinal tissue attachment inside the tissue bath chamber is elaborated seperately on right side.

MicroView software is shown in Fig. 4. Figure 4 shows two separate data acquisition systems, one for measuring the radial peristaltic forces by force sensor on the robotic capsule (left side) and a second for measuring the axial forces by force transducer in an in-vitro experimental setup (right side).

E. In-vitro experimental setup

An in-vitro experimental setup, which kept fresh intestinal tissue alive for up to 6 hours inside a tissue bath chamber, was used to test the capsule prototype. The in-vitro experimental setup is shown in Fig. 5. A test tube shaped tissue bath chamber was filled with ringer's solution, which provided the nutrients to keep the tissue alive [22]. The ringer's solution was kept oxygenated from an L shaped glass tube as shown in Fig. 5. A heated water recirculator was used to continuously circulate the heated water through the tissue bath chamber and the ringer's solution holder, which maintained the temperature of the entire system at the body temperature of lamb i.e. 39 °C. The ringer's solution holder was used to store the ringer's solution at body temperature, which was added to the tissue bath chamber as needed during the experiment.

The in-vitro setup also included a force transducer (MLT0420, ADInstruments Ltd., Dunedin, NZ) which measured the axial forces from the intestinal tissue, while the tissue was placed inside the tissue bath chamber. The force transducer was connected to the PC through a bridge amplifier (Bridge Amps, ADInstruments Ltd., Dunedin, NZ) and ADC (PowerLab 4/16, PowerLab, ADInstruments Ltd., Dunedin, NZ) as shown in Fig. 4 (right side). A dedicated software LabChart (LabChart, ADInstruments Ltd., Dunedin, NZ) was used to record and plot the data of axial forces from the living intestinal tissue.

III. Results

A. Simulations of interaction between peristaltic forces and robotic capsule

The analytical model, developed in this work, is implemented in COMSOL Multiphysics. A 2D-Axisymmetric model was developed and its mesh consisted of 3112 domain elements and 504 boundary elements. The internal domain (inside the intestine) was defined with laminar flow while the robotic capsule was also moving inside this domain. Whereas, the outside domain (intestine wall) was defined with the structural mechanics module and it goes under the deformation in order to exert the peristaltic force on the internal domain. Equation (5) and (7) were used to define the internal domain, while (12) and (19) were used for outside domain. A repetitive peristaltic wave of 7 seconds per cycle was generated across the cross section of an intestine as shown in Fig. 6 (A)-(C).

The progressive wave applied the force on the capsule inside the intestine based on its deformation. The peristaltic force was applied through the boundary load function through a built-in Heaviside function in COMSOL as shown below.

$$F_{A_r} = L_{max} \cdot load(zs, ts) \tag{21}$$

Where, F_{A_r} is the force in radial direction, L_{max} is the maximum load applied from the intestinal wall, $load$ is the Heaviside function, zs and ts are the dimensionless arguments for the Heaviside function.

The literature on intestinal motility suggested that the peristaltic force applied by the intestine is 1.72 g/mm in a longitudinal direction and 2.69 g/mm in a radial direction [23]. This study was considered by various researchers to design their locomotion [5], localization [24], drug delivery [25] and sampling [7] mechanisms. Based on these studies, the overall load for the 100 mm section of small intestine was calculated as 1.7 N in an axial direction and applied through the load function in (21). The peristaltic wave driven by the $load$ function, deforms the intestine and pressurizes the capsule inside the intestine. The interaction between the intestine and the robotic capsule can be seen in Fig. 6 (D)-(F).

The small intestine exerted force in an axial direction, which resulted in the deformation of the intestine in a radial direction. The robotic capsule records the pressure and Fig. 6 (G) shows an example of pressure values for an instant time. The radial pressure applied to the capsule is 2.69 g/mm [23] which will

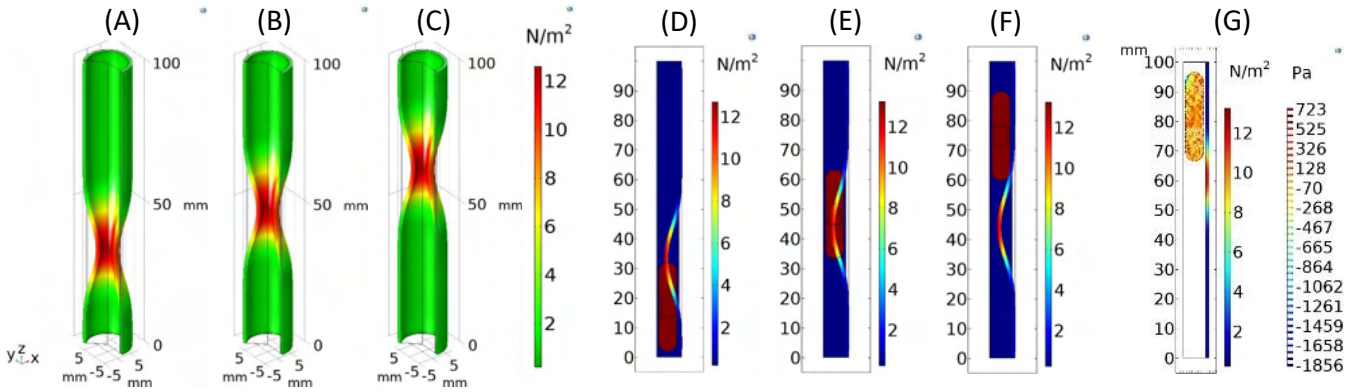


Fig. 6: Simulations performed in COMSOL Multiphysics. (A)-(C) 3D view of repetitive peristaltic wave of 7 seconds/cycle generated across the 100 mm intestine (A) t = 0.5 sec, (B) t = 3.5 sec, (C) t = 6.5 sec. (D)-(F) 2D-axisymmetric view of robotic capsule interacting with the intestine due to the repetitive peristaltic movement (D) t = 0.5 sec, (E) t = 22 sec, (F) t = 32 sec. (G) Pressure experienced by the robotic capsule due to radial contraction of the intestine.

result in 160 mN force which is equivalent to 890 Pa. In these simulations, the robotic capsule received an average pressure of 500 to 1065 Pa which is equivalent to a force of 90 mN to 192 mN.

B. Experimental results

1) Sample preparation for contraction force measurements: Fresh intestines of 5 lambs, dissected 1 hour before the experiments, were obtained on different days. The duodenum region of the small intestine was selected for experiments, as this region produces the highest frequency of contractions as compared to the other regions of the small intestine. The intestine was cut in to 100 mm long tissue samples, and immediately stored in chilled ringer's solution, to maintain its physiological properties until the start of the experiments. During the experiment, one end of the tissue sample was affixed to L shaped glass tube's support and the other end was tied with suture material. This material was stretched and fixed at the string holder of the axial force sensor as shown in Fig. 5. The intestinal tissue samples were kept under tension so the axial movements along the vertical axis were detected by the force transducer of the in-vitro system. The environment inside the tissue bath chamber maintained the postmortem tissue which continued to behave like living intestine, and started to produce peristaltic forces. The radial forces compressed (deformed) the tissue, which resulted in shrinking of the tissue in axial direction, which were recorded by the LabChart software through axial force transducer as shown in Fig. 4 (right side). Simultaneously, the robotic capsule measured the radial peristaltic forces, which were recorded in MicroView software through peristaltic force sensor as shown in Fig. 4 (left side).

2) Calibration of force sensor for capsule prototype: The force sensor inside the capsule prototype was calibrated each time before recording the peristaltic forces from the intestine. The tethered capsule prototype was inserted inside the tissue bath chamber and linearly increasing loads were applied to the capsule and the data was measured through the electronic kit as shown in Fig. 4 (left side). The data was simultaneously recorded and displayed through the MicroView software. Similarly, dynamic loads were also applied and radial force sensor (robotic capsule) readings were recorded. The radial force sensor (robotic capsule) showed less than 2% error for values between 0 and 200 mN, while 6% for readings greater than 250 mN.

3) Axial and peristaltic force results: In-vitro experiments were conducted on 13 intestinal samples from 5 lambs. The range of axial force measurements from each sample, measured by the in-vitro force transducer, is shown in Fig. 7. The Fig. 7 shows the minimum force (bar at bottom) and maximum force (bar at top) applied by each intestinal tissue sample. This allows to visualize the variation among different range of forces, applied within each sample. The (rectangular) box for each intestine sample between the minimum and maximum bars represents the 25^{th} percentile (first quartile) and 75^{th} percentile (third quartile), which is used to determine the effective range of forces applied by each intestinal tissue sample. The Figure shows the axial force behaviour of each sample under the in-vitro system in more detail. The mean forces recorded by the in-vitro axial force transducer are also shown in Table I along with the maximum readings observed by the robotic capsule under each intestine sample. The axial force transducer recorded the accumulated axial force measurements from the intestine, as any radial contraction throughout the 100 mm section of the intestine leads to shrinking of the longitudinal muscles. However, the robotic capsule only detected the radial (peristaltic) forces exerted exactly on the body of the capsule. The peak values of the peristaltic forces detected by the robotic capsule are shown in Table I. Peristaltic forces recorded by the robotic capsule are less than those recorded by the axial force transducer, because the capsule only recorded radial contractions which occurred directly on the face of the sensor. See supplementary video for further details about simulations and experiments.

Mean axial forces measured by the axial force transducer

TABLE I: Peristaltic forces recorded by the force transducer and robotic capsule

Intestine sample	1	2	3	4	5	6	7	8	9	10	11	12	13
Mean axial force measured by in-vitro force transducer (mN)	216	248	229	232	239	310	290	292	346	113	92	183	152
Peak radial force detected by robotic capsule (mN)	140	60	45	180	0	120	30	155	135	90	0	55	65

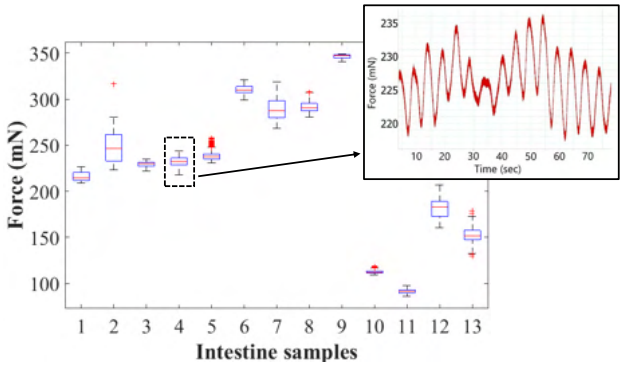


Fig. 7: Variations in axial forces generated by 100 mm intestinal tissue samples measured by the axial force transducer. In each sample, top and bottom bars show the range of axial forces, rectangular box indicates first and third quartile, red line shows the median and red plus sign shows the outliers. Window at top right shows a sample of force measurement data, plotted by LabChart software.

(in-vitro system) and peak peristaltic forces detected by the radial force sensor (robotic capsule), for each trial, are shown in Table I. The average axial force generated by the 13 selected samples was 226 mN while the average of peak peristaltic forces detected by the robotic capsule was 83 mN. Data was recorded for at least 30 minutes from each sample and the average is computed from 10 minutes of stable force output. The mean contraction cycle rate was 9 cycles per minute. The contraction rate used in simulation was 8.57 cycles per minutes and both contractions are in accordance with the relevant studies [5], [23].

IV. Discussion

The analytical modelling of the small intestine provides an understanding of gut motility. The model presented in this work is used as a base to simulate the contractile motion of the small intestine in COMSOL Multiphysics. The assumptions made in modelling and simulation are common to similar motion related studies [18], [26]–[28]. Simulation of a robotic capsule to study the impact of peristaltic forces, is a unique contribution which would help researchers to evaluate the designs of locomotion, localization, biopsy, sampling and drug delivery mechanisms. The simulation of peristaltic waves allows to study the impact of various parameters on a robotic capsule, such as change in radius of the intestine or the robotic capsule. The force analysis on the robotic capsule in this work can be used to study the pressure locations, which would allow installation of the actuation mechanism at the least exposed areas of the capsule, or optimize the tools to increase the efficiency of the device. Although literature [23] suggested that the robotic capsule in this study will experience a pressure of 890 Pa, it didn't specify which part would be exposed less. The simulations results revealed that the robotic capsule, as shown in Fig. 6 (G), will be exposed to a range of pressure levels and this information will determine the best location for installing the tools for performing diagnosis, monitoring and therapeutic functions inside the intestine. Furthermore, this study shows that the robotic capsule would be exposed to a range of pressures from 500 Pa to 1065 Pa, which is equivalent to a load of 90 mN to 192 mN.

In order to verify the findings of our analytical modelling and simulations, a specialized in-vitro setup was utilized in this work, which keeps a fresh intestine alive for up to 6 hours. The in-vitro system recorded the intestinal contractions through an axial force transducer and plotted the axial forces through LabChart software, which ensured that the intestinal tissue was alive. The average force recorded by the axial force transducer in 13 trials of tissue from 5 different lambs was 226 mN and the range of axial force measurements were from 92 mN to 346 mN, which agrees with previous work based on an in-vivo measurement from anaesthetize pig of 215 mN to 328 mN [12]. Another study suggested that a 100 mm section of intestine will generate 198 mN contraction (peristaltic) forces in a radial direction [23], which is also within the range of forces shown in Table I. The axial force transducer was fixed at the stretched ends of the intestinal tissue, hence it recorded the accumulated contraction forces of the entire section in axial direction. However, the robotic capsule only observed the peristaltic (radial) forces applied to its body. The peristaltic forces experienced by the capsule prototype were in the range of 30 mN to 180 mN, whereas the simulations predicted the range from 90 mN to 192 mN, which shows that the efficiency of predicting the peak peristaltic force was 94%. In two trials, the capsule prototype didn't recorded any force and this was either due to wrong calibration or a lack of peristaltic wave contracting at the position of the capsule. The capsule prototype recorded comparatively low peristaltic forces in some of the trials, this could occur when the peristaltic force applied to the capsule body wasn't captured fully by the sensor. Furthermore, we tested the robotic capsule by attaching the force sensor at 2 more locations, as compared to the Fig. 2 (B), one at the front face of the capsule and second on the front edge of the capsule. The capsule with front face sensor resulted in measuring less forces while the capsule on front edge detected almost similar results as the case shown in Fig. 2 (B). The experimental results verified, similar to simulations, that the radial peristaltic forces occur more at the side walls of the robotic capsule as compared to the front side, which could allow us to determine the low pressure points to place our tools for performing diagnosis,

monitoring and therapeutic functions.

The robotic capsule effectively measured the peristaltic forces from the living tissue. Although the robotic capsule under this study was tethered, the force sensor was calibrated before each trial. Calibration ensured that the wire did not affect the results from the force sensor. The force sensor was pasted on the exterior of the capsule, to record the peristaltic forces directly from the intestinal tissue. This method was specially adopted based on the availability of the in-vitro setup in this study, but for in-vivo trials this will not be possible and a proper encapsulation will be required. To further ensure results, in future works the robotic capsule will be tested in-vivo after incorporating a telemetry system.

V. Conclusion

This paper presented a novel method of measuring the impact of peristaltic forces on a robotic capsule in gut tissue. An analytical model was developed and simulated in COMSOL Multiphysics. The model measured the impact of peristaltic forces from the intestine on the robotic capsule. Later, an in-vitro system, which maintained intestinal tissue for up to six hours, was used to directly measure the forces from the intestine. The forces generated by the intestine were recorded by the axial force transducer and robotic capsule sensor simultaneously. The axial force transducer measured all the contraction forces, which can be understood as an accumulated force of the entire tissue. While, the robotic capsule recorded the radial forces acting directly on its body. Both forces were well-aligned with the simulation results and the related literature. These results will be useful in understanding the gut motility and quantification of peristaltic forces. In addition, knowing the impact of peristaltic forces on the capsule will enable the development and optimization of novel robotic capsules for locomotion, localization, biopsy, sampling and drug delivery purposes.

References

[1] G. Iddan, G. Meron, A. Glukhovsky, and P. Swain, "Wireless capsule endoscopy," *Nature*, vol. 405, no. 6785, p. 417, 2000.

[2] K. Kalantar-Zadeh, N. Ha, J. Z. Ou, and K. J. Berean, "Ingestible sensors," *ACS sensors*, vol. 2, no. 4, pp. 468–483, 2017.

[3] J. Min, Y. Yang, Z. Wu, and W. Gao, "Robotics in the gut," *Advanced Therapeutics*, vol. 3, no. 4, p. 1900125, 2020.

[4] J. Gao, Z. Zhang, and Y. Guozheng, "Locomotion analysis of a clamper-based capsule robot in a compliant tube," *IEEE/ASME Transactions on Mechatronics*, vol. 26, no. 1, pp. 55–65, 2020.

[5] H. Zhou and G. Alici, "A novel magnetic anchoring system for wireless capsule endoscopes operating within the gastrointestinal tract," *IEEE/ASME Transactions on Mechatronics*, vol. 24, no. 3, pp. 1106–1116, 2019.

[6] D. Son, S. Yim, and M. Sitti, "A 5-d localization method for a magnetically manipulated untethered robot using a 2-d array of hall-effect sensors," *IEEE/ASME Transactions on Mechatronics*, vol. 21, no. 2, pp. 708–716, 2015.

[7] M. Rehan, I. Al-Bahadly, D. G. Thomas, and E. Avci, "Capsule robot for gut microbiota sampling using shape memory alloy spring," *The International Journal of Medical Robotics and Computer Assisted Surgery*, vol. 16, no. 5, p. e2140, 2020.

[8] P. Shokrollahi, Y. P. Lai, S. Rash-Ahmadi, V. Stewart, M. Mohammadigheisar, L.-A. Huber, N. Matsuura, A. E. Zavodni, J. Parkinson, and E. Diller, "Blindly controlled magnetically actuated capsule for noninvasive sampling of the gastrointestinal microbiome," *IEEE/ASME Transactions on Mechatronics*, 2020.

[9] D. Son, H. Gilbert, and M. Sitti, "Magnetically actuated soft capsule endoscope for fine-needle biopsy," *Soft robotics*, vol. 7, no. 1, pp. 10–21, 2020.

[10] K. Kalantar-Zadeh, K. J. Berean, N. Ha, A. F. Chrimes, K. Xu, D. Grando, J. Z. Ou, N. Pillai, J. L. Campbell, R. Brkljača *et al.*, "A human pilot trial of ingestible electronic capsules capable of sensing different gases in the gut," *Nature Electronics*, vol. 1, no. 1, pp. 79–87, 2018.

[11] F. Munoz, G. Alici, H. Zhou, W. Li, and M. Sitti, "Analysis of magnetic interaction in remotely controlled magnetic devices and its application to a capsule robot for drug delivery," *IEEE/ASME Transactions on Mechatronics*, vol. 23, no. 1, pp. 298–310, 2017.

[12] P. Li, V. Kothari, and B. S. Terry, "Design and preliminary experimental investigation of a capsule for measuring the small intestine contraction pressure," *IEEE Transactions on Biomedical Engineering*, vol. 62, no. 11, pp. 2702–2708, 2015.

[13] G. Cummins, B. F. Cox, G. Ciuti, T. Anbarasan, M. P. Desmulliez, S. Cochran, R. Steele, J. N. Plevris, and A. Koulaouzidis, "Gastrointestinal diagnosis using non-white light imaging capsule endoscopy," *Nature Reviews Gastroenterology & Hepatology*, vol. 16, no. 7, pp. 429–447, 2019.

[14] Y. Kim, G. Lee, S. Park, B. Kim, J.-O. Park, and J.-h. Cho, "Pressure monitoring system in gastro-intestinal tract," in *Proceedings of the 2005 IEEE International Conference on Robotics and Automation*. IEEE, 2005, pp. 1321–1326.

[15] P. Li, C. Kreikemeier-Bower, W. Xie, V. Kothari, and B. S. Terry, "Design of a wireless medical capsule for measuring the contact pressure between a capsule and the small intestine," *Journal of Biomechanical Engineering*, vol. 139, no. 5, 2017.

[16] S. A. Woo, Z. Mohy-Ud-Din, and J. Cho, "Telemetry capsule for measuring contractile motion in the small intestine," *Biomedical microdevices*, vol. 15, no. 1, pp. 63–72, 2013.

[17] J. Monreal, "General navier–stokes-like momentum and mass-energy equations," *Annals of Physics*, vol. 354, pp. 565–569, 2015.

[18] J. Karthikeyan, D. Salvi, and M. V. Karwe, "Modeling of fluid flow, carbohydrate digestion, and glucose absorption in human small intestine," *Journal of Food Engineering*, vol. 292, p. 110339, 2021.

[19] A. Fuyuki, H. Ohkubo, T. Higurashi, H. Iida, Y. Inoh, M. Inamori, and A. Nakajima, "Clinical importance of cine-mri assessment of small bowel motility in patients with chronic intestinal pseudo-obstruction: a retrospective study of 33 patients," *Journal of gastroenterology*, vol. 52, no. 5, pp. 577–584, 2017.

[20] D. Son, M. D. Dogan, and M. Sitti, "Magnetically actuated soft capsule endoscope for fine-needle aspiration biopsy," in *2017 IEEE International Conference on Robotics and Automation (ICRA)*. IEEE, 2017, pp. 1132–1139.

[21] P. Boscan, S. Cochran, E. Monnet, C. Webb, and D. Twedt, "Effect of prolonged general anesthesia with sevoflurane and laparoscopic surgery on gastric and small bowel propulsive motility and ph in dogs," *Veterinary anaesthesia and analgesia*, vol. 41, no. 1, pp. 73–81, 2014.

[22] K. Wolf, "Physiological salines for fresh-water teleosts," *The Progressive Fish-Culturist*, vol. 25, no. 3, pp. 135–140, 1963.

[23] R. N. Miftahof, "The wave phenomena in smooth muscle syncytia," *In silico biology*, vol. 5, no. 5, 6, pp. 479–498, 2005.

[24] L. Barducci, J. C. Norton, S. Sarker, S. Mohammed, R. Jones, P. Valdastri, and B. S. Terry, "Fundamentals of the gut for capsule engineers," *Progress in Biomedical Engineering*, vol. 2, no. 4, p. 042002, 2020.

[25] S. P. Woods and T. G. Constandinou, "Wireless capsule endoscope for targeted drug delivery: mechanics and design considerations," *IEEE Transactions on Biomedical Engineering*, vol. 60, no. 4, pp. 945–953, 2012.

[26] M. D. Sinnott, P. W. Cleary, J. W. Arkwright, and P. G. Dinning, "Investigating the relationships between peristaltic contraction and fluid transport in the human colon using smoothed particle hydrodynamics," *Computers in biology and medicine*, vol. 42, no. 4, pp. 492–503, 2012.

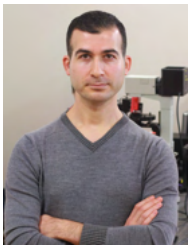
[27] M. D. Sinnott, P. W. Cleary, and S. M. Harrison, "Peristaltic transport of a particulate suspension in the small intestine," *Applied Mathematical Modelling*, vol. 44, pp. 143–159, 2017.

[28] Y. Zhang, P. Wu, R. Jeantet, D. Dupont, G. Delaplace, X. D. Chen, and J. Xiao, "How motility can enhance mass transfer and absorption in the duodenum: Taking the structure of the villi into account," *Chemical Engineering Science*, vol. 213, p. 115406, 2020.

Muhammad Rehan received his BS degree in Electronic Engineering from Sir Syed University of Engineering & Technology (SSUET), Karachi, Pakistan in 2008 and MS degree in Electrical (Control) Engineering from National University of Sciences & Technology, Karachi, Pakistan in 2013. Since 2009, he has been with SSUET, where he is currently working as an Assistant Professor. Rehan is pursuing his PhD in Micro-robotics from Massey University, Palmerston North, New Zealand. His current research interests include robotics in general and robotic capsule for gut analysis in particular.

Ibrahim Al-Bahadly received a B.Sc. degree from Baghdad University of Technology in 1987, followed by M.Sc. and Ph.D. from Nottingham University, in 1990 and 1994, respectively, all in electrical and electronic engineering. From 1994 to 1996, he was a research associate with the electric drives group at the University of Newcastle upon Tyne. Since 1996, he has been with Massey University, where he is currently Associate Professor. His research interests include power electronic, variable speed drives, renewable energy, and robotics.

Dr David Thomas obtained a Bachelor of Science Honours degree in Zoology from Dundee University in 1990, and a PhD from Institute of Zoology, University College London in 1994. Joined the teaching staff at Massey University in 2003. Specialises in commercial and applied Feline and Canine nutrition research. Currently an Associate Professor in Companion Animal Science in the School of Agriculture and Environment and has supervised, or is currently supervising, 32 post graduate students (2 honours, 14 MSc and 16 PhD students) and through his research activities has published over 150 peer-reviewed scientific articles.



Ebubekir Avci received his PhD degree in robotics from Osaka University, Japan in 2013. Between 2014 and 2015, he was a research associate with the Hamlyn Centre for Robotic Surgery, Imperial College London, UK. He is currently a senior lecturer in mechatronics at Massey University, Palmerston North. Ebu is an associate investigator at The MacDiarmid Institute for Advanced Materials and Nanotechnology. His current research interests include micro-robotic systems for bio-medical applications such as nano-scale robots for molecule analysis, micro-scale robots for cell manipulation, and millimetre-scale robots for gut analysis.