Odocapsule: Next Generation Wireless Capsule Endoscopy with accurate Lesion Localization and Video Stabilization Capabilities

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Science & Technology, Chongqing, China), and e) CapsoCam®SV1 (CapsoVision, Inc., Saratoga, USA).

Abstract — In this paper, a new capsule endoscopy is proposed to achieve accurate localization of small-bowel lesions and endoscopic video stabilization in wireless capsule endoscopy. Up to now, research modules rely on the use of external magnetic fields and triangulation methods to calculate the position vector of the capsule, leading to considerable error margins. Our platform, entitled OdoCapsule (a synthesis of the words Odometer and Capsule), provides real-time distance information from the point of duodenal entry to the point of exit from the small-bowel. To achieve this, OdoCapsule is equipped with 3 miniature legs. Each leg carries a soft rubber wheel, which is made with human-compliant material. These legs are extendable and retractable thanks to a micro-motor and 3 custom-made torsion springs. The wheels are specifically designed to function as micro-odometers: each rotation they perform is registered. Hence the covered distance is measured accurately in real-time. Furthermore, with its legs fully extended, OdoCapsule can stabilize itself inside the small-bowel lumen thus offering smoother video capture and better image processing. Recent ex-situ testing of this concept, using porcine small-bowel and a commercially available (custom-modified) capsule endoscope, has proved its viability.

I. INTRODUCTION

An early conceptual abstract on wireless capsule endoscopy (CE), entitled “An endorobot for flexible endoscopy: a feasibility study”, was published in 1994 [38]. In 1997 two groups of pioneers, initially working independently in Israel and London, joined forces to achieve wireless digestive endoscopy [40]. By that time, miniaturization of electronic components allowed radical changes to the design of complex diagnostic structures. Consequently, at the Digestive Disease Week meeting of the millennium and almost concurrently in Nature [41], [42] Professor P Swain presented the world’s first minimally invasive capsule endoscopy. Over the last decade, CE has been established as a prime-imaging tool, over other modalities, in the investigation of small-bowel diseases [43]. A wealth of scientific data on indications, diagnostic yield, safety profile and technological evolution has proven the superiority of this ‘disruptive’ technology over other imaging modalities [43]. Consequently, CE has now been established as the main form of wireless digestive endoscopy, re igniting clinicians’ interest in the study of the small-bowel [44]. At present, there are 5 commercially available wireless CE platforms: a) PillCam®SB (Given®Imaging Ltd, Yokneam, Israel), b) MiroCam® (IntroMedic Co, Seoul, South Korea), c) EndoCapsule® (Olympus®Medical Systems Co, Tokyo, Japan), d) OMOM® (Jinshan...
B. Why is Video Stabilization important for CE?

The shape of current commercial capsule endoscopes is sufficient to allow them to pass through anatomical sphincters (pylorus and ileocecal valve) of the small-bowel without an obstruction risk ([23], [24]). However, this size predisposes the capsule to rotate (or tumble) within the small-bowel lumen resulting – frequently – in deficient luminal coverage. Furthermore, through segmentation and peristaltic contractions of the gut, the viewpoint changes constantly. This is better illustrated in

Fig. 2 sequential CE videos frames from two different procedures show the effect of tumbling i.e. oblique forward, perpendicular and rotational movement.

![Fig. 2: CE frame sequences (from two different patients) showing the constant vector change effect. Top row: a PillCam®SB recording. Bottom row: a MiroCam® recording](image)

Apart from erratic/incomplete mucosal coverage, tumbling of the capsule results in non-smooth (‘staccato’) video recording. This, in conjunction with limitations imposed by the capsule’s low frame rate and resolution, can reduce the performance of computer-aided diagnosis (CAD) systems ([28], [29], [30], [31], [32], [33], [34]). Software-based approaches ([25], [26], [27]) that mostly rely on image registration methods have been proposed in order to compensate for de-stabilization in CE videos. Their performance varies greatly because they depend on identifying good matching key points between consecutive images.

Therefore, further attempts for CE video stabilization, using hardware improvements, have been undertaken [2], [3]. More specifically, in [2] the authors modified a commercial capsule (MiroCam® from IntroMedic Co Ltd) into a self-stabilizing device by adding a mesh of expandable polymer granules at the non-imaging dome of the capsule. With ex-vivo experiments, a considerable average improvement in the automated tracking of locales in recorded videos, leading overall to better imaging, was confirmed. In [3], a conceptual capsule design was presented offering stabilizing capabilities using springs and wheels.

Furthermore, having achieved video stabilization various motion models [4] can be applied effectively to create maps of the GI tract. These maps can be further enhanced using Shape-from-Shading (SFS) [35] and Shape-from-Motion (SM) ([36], [37], [38]) techniques, leading to a better discovery of pathologies and GI landmarks through computer-aided algorithms or visual inspection.

II. RELATED WORK

From commercially available CE devices to research projects there has been a consistent effort to achieve capsule localization. A recent comprehensive review on capsule localization systems was published by Than et al [6]. In this section, we briefly present these systems, discuss their advantages and disadvantages, and finally we introduce the reader to our own proposed solution, the Odocapsule.

In commercial approaches a standard array of receivers is placed on the exterior of the patient abdomen, while a transmitter inside the capsule sends data (i.e. images, other sensor data, etc.) to the receivers. Positional info is calculated based on the principle that the closer the capsule is to a receiver the stronger the received signal should be. There has been an attempt for localization by Given Imaging Ltd which has been largely abandoned because of low precision (3.77 cm) [49], [6], [46], [49]. While this approach is very attractive for its simplicity - it does not require extra equipment-, it suffers from external electromagnetic noise and complex radio wave absorption properties of human tissue [7]. On the other hand, another innovative capsule-based platform, motility monitoring system (MTS2) by Motilis Medica SA, Lausanne, Switzerland enables monitoring of regional transit time and a more accurate recording of capsule position. Furthermore, a more practical capsule platform, called SmartPill® (Given Imaging Ltd, Yokneam, Israel) measures pressure, pH and temperature as it travels through the GI tract to assess GI motility, thus allowing recognition of entry of the capsule device in the duodenum and caecum.

In research domain two main approaches have been explored to retrieve position information: a) Magnetic Field Strength-based (MFS) methods and Electromagnetic Wave-based (EW) methods [6].

In MFS methods, a permanent magnet is incorporated in the capsule and an external array of magnetic sensors is placed outside the patient’s body. As the capsule (and its magnet) moves, its magnetic flux changes - in magnitude and direction- and the external sensors can measure these magnetic signals. Using mathematical modeling, the location and orientation can be calculated. Various systems using the MFS methods have been proposed by Weitschiles et al [8], Schlageter et al [9], Chao et al [10], Azziz et al [11] and Wu et al [12]. These systems managed to achieve location information accuracy in the range of millimeters. However, the aforementioned systems suffer from a number of drawbacks such as sensitivity to interference and motion, complex mathematical models, required calibrations, complex sensor arrangement, physical size, cost and bulkiness (see Fig. 3).

![Fig. 3 Proposed localization system based on MFS method from Wu et al. [12]](image)

In EW methods, different electromagnetic waves have been utilized with lower position information accuracy. To date, only radio waves (RF), visible waves, x-ray and gamma ray have been explored in literature because of their high penetrability through human tissue [6]. Arshak & Adepoju [13] used an empirical signal propagation model to measure received signal strength indicator (RSSI) related with distance between transmitter and receiver. Shah et al [14] used a lookup table to estimate position. Furthermore, Lujia et al [15] improved RSSI calculations taking into consideration tissue absorption and antenna orientation. More interestingly, Kuth et al [16] used x-ray and image processing to detect the location of the capsule. Wilding et al [17] proposed a method based on gamma scintigraphy technique to locate a drug delivery capsule. While the EW methods are characterized by improved ease of use compared to
MFS, they suffer with large errors in position accuracy and many of them have to ignore important factors to simplify the modeling and calculations.

To overcome the aforementioned disadvantages and implications of the EW approaches, we propose a new wireless capsule with localization capabilities that takes into consideration size limitations, robustness, efficiency and ease of use. In the following paragraphs, the components and principles of this novel design are presented in detail.

III. PROPOSED CAPSULE

OdoCapsule is the name we selected for this proposal and that’s for a good reason; it stands for odometer and capsule. An odometer is a familiar instrument that measures covered distance in vehicles. And it is the same principle that functions in the core of OdoCapsule when measuring distance. In contrast to other research proposals (paragraph II), OdoCapsule does not address the lesion localization problem indirectly, that is calculating the capsule position through triangulation of distance from external receivers. Instead, it offers the distance of a lesion from unchanged anatomical landmark i.e. the duodenum.

It is well known that the GI tract of the human constantly curls, twists and changes its shape [22], [48]. Therefore, the distance (vector) between the capsule and external points of reference (e.g. receivers) is not a reliable measurement for accurate lesion localization. On the contrary, measuring the distance from the start of the small bowel (i.e.pyloric opening) is independent of any shape achieving robust localization. Especially when device-assisted enteroscopy is contemplated, it is crucial for the operator to know not just the approximation of the location of lesion in a 3D space, but the accurate distance of a lesion from a stable anatomical landmark, such as the pyloric opening to the small-bowel.

OdoCapsule is an ongoing research project that started in 2010. In [18], we introduced the concept; more recently, in [19] we presented a more advanced version of the capsule in ex-vivo experiments using porcine intestine, which has attracted considerable interest in the medical field [51]. In contrast to other proposed capsules we tried to maintain the dimensions of the OdoCapsule within that of existing commercially available capsules. OdoCapsule’s proposed diameter is 13mm and length 30mm i.e., slightly bigger than Given’s® Imaging’s PillCamSB (11mm x 26mm). Fig 4. details the internal components of the OdoCapsule.

In Table I the technical specifications of the components are presented.

<table>
<thead>
<tr>
<th>Table 1</th>
<th>TECHNICAL SPECIFICATIONS FOR ODOCAPSULE PARTS</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Electronic Parts</strong></td>
<td><strong>Manufacturer</strong></td>
</tr>
<tr>
<td>CMOS camera</td>
<td>Toshiba CMOS camera (TCM8240MD)</td>
</tr>
<tr>
<td>Illumination</td>
<td>4 LED lights Agilent (HSMW-C191)</td>
</tr>
<tr>
<td>Printed circuit board</td>
<td>Custom printed</td>
</tr>
<tr>
<td>Microcontroller</td>
<td>Texas Instruments Microcontroller MSP430</td>
</tr>
<tr>
<td>RF Transceiver</td>
<td>Zarlink ZL70102</td>
</tr>
<tr>
<td>Motor</td>
<td>Fraulhaber Brushless Flat DC-Micromotor Series 1307</td>
</tr>
<tr>
<td>pH sensor</td>
<td>ISFET sensor</td>
</tr>
<tr>
<td>Batteries</td>
<td>2 Lithium coin batteries (CR1025 Energizer)</td>
</tr>
</tbody>
</table>

OdoCapsule incorporates a pH sensor and 3 legs—placed at 120° intervals—equipped with wheels. The proposed length of each leg is 15mm. The rest of the CE components have been re-positioned to achieve a compact, yet functional, design. The camera is CMOS VGA technology with an array of 1300x1040 pixels and a built-in JPEG encoder. It can capture up to 15 fps. In order to accommodate the data rate specifications of the low-energy RF transceiver (800kbps) we programmed the camera to capture at 3 frames/sec in VGA mode (640x480 pixels). Based on these requirements the data rate of the camera is 3x640x480x8bit = 7,372,800 bits. This is still beyond the capacity of the RF transceiver specifications. However using JPEG compression on the raw data we can achieve 10-fold reduction so that data can fit into the data rate (737.28 kbps / 800 kbps) while preserving good image quality. Data are sent from the camera via an 8-bit parallel port to the micro-controller, which performs buffering, packetization and finally transmission via the SPI interface to Zarlink’s wireless transceiver.

In the following two paragraphs, IV and V, we describe in detail the leg mechanism and the odometer concept, respectively.

IV. MECHANISM FOR RETRACTING AND EXTENDING LEGS

The brushless micromotor is in charge of retracting or extending the three (3) legs. Before the procedure begins the capsule is deactivated, the motor is off and each leg is extended outwards of the capsule body because the torsion spring is deflected to its free angle of 35° degrees. Right before the patient swallows the capsule, the motor is activated and retracts the legs by pulling in the cables (see cables in Fig. 4) to make the capsule swallowable. At the same time, the latches (see latches in Fig. 4) are turning to lock the legs in their new position while the micromotor is turned off for power efficiency. From physiology [20] we know that pH in the stomach has a level of 1.5 while its level in the intestine (small and large) jumps to around 7.5. This difference is detected by the capsule’s pH sensor and the micromotor releases the legs and it turns off once again. In the small-bowel, the springs deflect to their free angle (35° degrees) again and the legs are fully released. The diameter of the extended capsule is 30.2 mm, which is sufficient for full contact with the mucosa of an
average human intestine (20.5-30.0 mm in diameter [50]). Peristaltic forces are exerted to the capsule and they push the capsule while friction forces between the wheels and mucosa rotate the wheels. These rotations are translated to covered distance using the odometer system discussed in detail in paragraph V. This information along with the video feedback is transmitted through the RF transmitter to a receiver worn by the patient, which is the standard procedure used by most commercial capsules. Finally, when the capsule exits the small-bowel and enters the large-bowel, there is a significant momentary reduction in the pH, which is again detected by the pH sensor [20]. The micromotor is then momentarily activated to retract the legs and resume the compact diameter (15mm) of the capsule.

A. Analysis of forces – Measuring the friction coefficient

An analysis of the forces acting upon the capsule is shown in Fig. 6. Miftahof [21] estimated the amplitude of the peristaltic force $P$ to be 17.2 g/cm in the axial direction and 26.9 g/cm in the radial direction. For the proposed capsule, the peristaltic force $P$ results into an axial force $P_{ax} = 510$ mN and a radial force $P_{rad} = 807$ mN. To be as accurate as possible, we followed the same force analysis described by Glass et al [22] and Woods et al [23] in similar CE projects.

The total force $F_{TOTAL}$ is equally distributed among the 3 points of contact (wheels). The force analysis follows below:

$$F_{TOTAL} = P + mg = P_{ax} + P_{rad} + mg = 3 \cdot F_{ax} + 3 \cdot F_{rad} + mg$$ (1)

We also know that in order for each wheel to be able to rotate the following requirement must always apply:

$$F_{FRICTION} < \frac{P_{ax} + mg \cos \theta}{3}$$ (2)

with the following additional conditions:

$$F_{NORMAL} = \frac{P_{rad} + mg \sin \theta}{3}$$ (3)

$$F_{FRICTION} = \mu \cdot F_{NORMAL}$$ (4)

where $\mu$ is the friction coefficient of the wheels with the mucosa and $\theta$ is the angle of weight in relation to the main axis of the capsule.

Substituting (3) and (4) into (2) we end up with the following requirement for the friction coefficient:

$$\mu < \frac{P_{ax} + mg \cos \theta}{P_{rad} + mg \sin \theta}$$ (5)

Plotting this function (Fig. 5) gives the maximum allowed friction coefficient, $\mu = 0.4963$, for which (2) is always true and therefore the wheels keep turning.

![Fig. 5 Plot for equation (5): friction coefficient vs. angle of capsule's weight. The maximum coefficient friction to satisfy (2) is the minimum value (0.4963) on this plot.](image)

The required value for such a coefficient friction can be achieved by fabricating the wheels with a biocompatible polymer polydimethylsiloxane, (e.g. PDMS, Sylgard 184, Dow Corning) [22]. Although in the above calculations the static friction has been neglected, for the sake of simplicity, we know from [22] that it can reach up to a level of 10 mN for the specific polymer material. Additionally, to limit the static friction the wheels have been designed with notches (see Fig. 4) to let the mucosal surface liquid go through while sustaining adhesion. As mentioned before, the validity of this design has been tested in a previous experiment using a porcine intestine in a Tyrode’s solution [19].

![Fig. 6 Exerted peristaltic forces in the axial and radial direction (shown in blue) and the reacted forces (shown in red).](image)

B. Spring specifications

One of the key parts is the torsion spring for each leg. The purpose of the spring is dual: a) to provide continuous contact with mucosa so that the wheels can rotate, and b) to dynamically adjust its deflection angle based on the radial force $F_{rad}$ (see Fig. 6). When no radial force is exerted to the spring ($F_{rad} = 0$), the spring is deflected to its free angle (35° degrees). As $F_{rad}$ increases, the spring compresses and therefore the leg is retracted. Finally, when $F_{rad}$ reaches its maximum value ($F_{rad} = 807$mN), the spring is fully compressed to 0° degrees. To accommodate these requirements the spring was custom designed by Lee Spring Company, Brooklyn, NY. Its specifications are shown in Table II.

<table>
<thead>
<tr>
<th>Spring Specifications</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Material</td>
<td>Polydimethylsiloxane (PDMS)</td>
</tr>
<tr>
<td>Thickness</td>
<td>0.1 mm</td>
</tr>
<tr>
<td>Diameter</td>
<td>0.5 mm</td>
</tr>
<tr>
<td>Torsion Angle</td>
<td>35°</td>
</tr>
<tr>
<td>Maximum Load</td>
<td>807 mN</td>
</tr>
<tr>
<td>Minimum Load</td>
<td>0 mN</td>
</tr>
</tbody>
</table>

Table II: Spring Specifications
TABLE II
TECHNICAL SPECIFICATIONS FOR TORSION SPRING

<table>
<thead>
<tr>
<th>Parts</th>
<th>Specifications</th>
</tr>
</thead>
<tbody>
<tr>
<td>Material</td>
<td>Music Wire (MW)</td>
</tr>
<tr>
<td>Outer diameter</td>
<td>3.73mm</td>
</tr>
<tr>
<td>Wire diameter</td>
<td>0.457mm</td>
</tr>
<tr>
<td>Body length</td>
<td>3.835mm</td>
</tr>
<tr>
<td>Total coils</td>
<td>7.403</td>
</tr>
<tr>
<td>Free angle</td>
<td>35° degrees</td>
</tr>
<tr>
<td>Leg 1</td>
<td>10mm</td>
</tr>
<tr>
<td>Leg 2</td>
<td>4mm</td>
</tr>
<tr>
<td>Rate/turn</td>
<td>0.0034 m x kg/turn</td>
</tr>
</tbody>
</table>

V. Odometer

Based on the analysis in paragraph IV, we showed that a constant contact between the wheels and the mucosa has been established. To identify the lesion location whilst in the small-bowel, the wheel rotations need to be detected and converted to distance. To achieve this, we designed a miniature odometer system attached into the core of each wheel (Fig. 7).

![Resistor - R](resistor.png)

![Conductive Slider](slider.png)

![Fig. 7](https://dl.dropboxusercontent.com/u/7591304/1035301R.AVI)

Fig. 7 a) 3D model of the odometer parts, b) corresponding 2D projection and c) its circuit schematic.

The system consists of a potentiometer (Fig. 7-a). It is a ring-shaped resistor made with thin film process. As the wheel rotates at angle \( \omega \) and the slider moves, potentiometer’s resistance \( R \) changes. The capsule’s microcontroller samples the electric potential \( V_R \) to measure rotations. Solving for \( V_R \), we have:

\[
V_R = \frac{V_o}{r + R} R \quad (6)
\]

with

\[
R = \rho \frac{l}{A} \quad (7)
\]

\[
I = \omega \ mod(2\pi) \quad (8)
\]

where \( V_o \): battery source, \( r \): chip resistor, \( \rho \): material resistivity, \( l \): length of resistor, \( A \): cross-sectional area of resistor, \( a \): radius of ring and \( \omega \): rotation angle. Using a resistor material of Germanium we have \( \rho = 4.6 \times 10^{-3} \). Its dimensions are 2mm wide 1mm high and 2\( \pi \) \times 2mm = 12.56nm long. Substituting (8) in (7) we have:

\[
R(\omega) = \frac{4.6 \times 10^{-3} \times 2 \times 10^{-3}}{2 \times 10^{-3} \times 1 \times 10^{-3}} \omega \ mod(2\pi) = 460 \cdot \omega \ mod(2\pi)
\]

And (6) can finally be rewritten:

\[
V_R(\omega) = \frac{3}{10^4 + R} R = \frac{1380 \omega \ mod(2\pi)}{10^4 + 460 \omega \ mod(2\pi)}
\]

With the aforementioned parameters, the odometer circuit has a mean power consumption of 8.6 mW, small enough to be accommodated by coin batteries for the duration of an average procedure.

In Fig. 8 the electric potential, \( V_R \), is given for three (3) cases of wheel rotation.

![Fig. 8](https://dl.dropboxusercontent.com/u/7591304/1035301R.AVI)

Fig. 8 Simulations for various cases of movements depicting \( V_R \) versus angle. a) Forward movement (positive rotation angle), b) Backward movement (negative rotation angle). c) Random movement (positive and negative rotation angle)

The distance covered by the wheel can finally be calculated using the following formula:

\[
\text{Distance} = (\text{Total Accumulated Angle}) \times (\text{Wheel Radius}) = 2.5 \times (\text{Total Accumulated Angle}) \ (\text{mm})
\]

In the simulations above (Fig. 8) we assumed that all 3 wheels rotate at the same pace (equal angles). However to validate this assumption we performed ex-vivo experiments using a porcine intestine. A glass tank (50 cm \( \times \) 20 cm \( \times \) 20 cm) with fixed points for the intestine (metal tubes) and two entry points for the prototype capsule was constructed (see Fig. 9). A freshly harvested porcine (Large White x Landrace 15-mo-old female sow) small intestine was attached to both ends of the tank. The tank was filled with Tyrode’s solution to mimic the friction specifications of gastrointestinal fluids. A string was attached to the prototype capsule and it was pulled from one entry point to the other one using weights hung from string’s end, thus creating an axial force between the intestine and the wheels, mimicking the axial peristaltic load present in the GI tract [22]. After running the capsule through the intestine four (4) times (video sample: https://dl.dropboxusercontent.com/u/7591304/1035301R.AVI) we were able to identify three (3) different cases of rotation: a) all three wheels rotated at the same pace, b) one wheel was dragged and it did not rotate, while the other two rotated and c) the wheels rotated at
different pace. We confirmed that for 85% of the time the capsule’s traversal followed case a, while we never noticed an obstruction of the capsule’s movement confirming our belief that OdoCapsule can be used in humans. Fig. 10 and Fig. 11 present cases b and c in regards to the voltage output of the three (3) wheels.

![Fig. 9 Top: Ex-vivo experiment setup using a porcine intestine, Bottom: First generation prototype apparatus attached to a commercial capsule used in the ex-vivo experiments](image)

![Fig. 10. Wheel A is not rotating (Vr steady) while wheel B is rotating slower than wheel C.](image)

![Fig. 11. Wheel A rotates much faster than wheel B but quite close to the rotation speed of wheel C.](image)

More specifically, Fig. 10 presents the case in which one wheel is dragged without rotation. This obstructs the rotation of the other wheels and even shifts the center of gravity of the capsule forcing wheel B to rotate slower than wheel C. On the other hand Fig. 11 shows a more common scenario: due to friction forces wheel B rotates slower than the rest two (2) wheels (39% slower to wheel A, 25% slower to wheel C) while wheel A runs slower at 22%. In this case the averaging the measurements is taken into account to reduce the error.

We further examined the effect of error to the measured distance by performing extensive simulations. For this purpose we had the distance error of each wheel be randomly selected between 5% and 30%, which is reasonable. For one wheel the accumulated error had a μ: mean value = 0.1752, σ: standard deviation = 0.0011, for two wheels μ = 0.1749, σ = 0.0011 and for three wheels μ = 0.1746 and σ = 0.0012. It is no surprise that there are no significant deviations in the error between the 3 setups (one wheel, two wheels, three wheels). However by having multiple wheels instead of one not only do we achieve better capsule stabilization as mentioned in previous paragraphs but we are able to correct the error in realtime too. By calculating the standard deviation between the measured values of the three wheels we can retract the wheels and release them instantly to reset calculations. Figure 12 depicts the projected error (red line) for three wheels, the reduced error (blue line) using a threshold (0.05) on the standard deviation of the wheels (green line) to reset the capsule. By tuning the threshold we are able to reduce the projected error.

To identify the relationship between the reduction of the projected error (ratio of corrected error to projected error) and the threshold we performed an extensive simulation (100 experiments) by modifying the threshold from 0 to 0.15 and measuring the ratio. Figure 13 shows these results along with the fitted line. There is a tradeoff between the number of resets and the error, which is experimental. In our simulations we used the extreme case when the errors for all three wheels are randomly generated.
V. CONCLUSION AND FUTURE DIRECTIONS

In this paper we proposed a new CE technology. It manages to address two important limitations of current CE technology: lesion localization and video stabilization. Accurate localization of small-bowel pathology is essential for successful endoscopic therapy with device-assisted endoscopes and (potentially) targeted drug delivery in the future. Furthermore, being able to measure the exact distance of a lesion from an anatomical landmark i.e. pylorus can help guide subsequent follow-ups (diagnostic and/or therapeutic) more accurately.

We showed that the new design overcomes current technological drawbacks at a minimal cost in size, complexity and power efficiency. The design is based on the concept of conventional odometer, which has been modified to meet smaller scaling constraints. With this simple approach wheel rotations are registered and overall travelled distance is measured inside the lumen. Moreover, using 3 wheels we can acquire multiple simultaneous measurements to reduce overall distance error. Finally the star formation of the extended legs minimizes the tumbling effect and it helps with capsule stabilization.

Certainly, there are weak points to this conceptual design. One of them is the possibility one or more of the legs getting stuck in bowel tissue (Fig. 9). In this case, since wheel rotations are constantly monitored, the capsule will activate the micro-motor to retract the legs momentarily and resume its locomotion. The possibility of perforation, although remote due to the absence of acute points, cannot be entirely excluded and use of soft manufacturing materials should be considered. Finally, there is the strong possibility that the capsule enters the pylorous backwards. In this case, the radial peristaltic forces will push the legs inwards first and then the axial peristaltic force will push the capsule forward allowing it to continue its journey while measuring distance.

We are planning to develop a functional prototype to carry ex-vivo experiments. As mentioned in paragraph V we want to test and evaluate the measurements acquired from each odometer circuit for various cases.

Finally, we believe that the design proposed herein could serve as an alternative model for a colon CE too.

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REFERENCES
