A Multi-Coil Inductive Powering System for an Endoscopic Capsule with Vibratory Actuation

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Abstract:

The evolution of endoscopic capsules from passive tools to robotic devices is increasingly attracting the interest of the research community. In the past few years, significant progress has been achieved in the miniaturization of electronics and electromechanical systems. However, their use in commercial endoscopic capsules is hindered by their increased power demands, which, to present, cannot be adequately met by embedded power sources. A 3D inductive powering module, providing over 300 mW to the capsule, overcomes these limitations, thus enabling the integration of active locomotion systems, as well as advanced diagnostic and therapeutic features. This is demonstrated in the present study by a capsule prototype employing the wireless powering unit to drive an on-board vibratory motor for capsule propulsion. Simplified models are employed to illustrate the main principle of this vibratory locomotion scheme. Experimental results, involving movement of the prototype in various environments, confirm both the effectiveness of the wireless powering system, and the efficacy of the vibratory locomotion scheme.

Keywords: capsular endoscopy; wireless power supply; inductive coupling; active locomotion; vibratory motor

1. Introduction

Since their introduction in the market, endoscopic capsules have become increasingly popular in the medical world [1-4]. Specialists are looking forward to their evolution from passive tools, only capable of examining the small bowel, into robotic devices, endowed with active locomotion and therapeutic capabilities, for examining the entire gastrointestinal (GI) tract [1-4]. A major obstacle to this goal is the limited amount of power in the capsule [1-5], since a pair of watch batteries (what is currently used in commercial endoscopic capsules) can merely provide 25 mW for the required 6-8 hours of operation. This lack of power hinders the integration of the different modules currently under development for evolving robotic capsules [1,3,5,6].

Some of these modules aim at increasing the diagnosis capabilities, improving the image quality [7,8] and their transmission rate [9], providing sensing features [2,10,11] or the option of taking micro samples of suspicious tissues [12,13]. Other modules aim at improving the capsule control, providing the operator with active locomotion systems [14-17] which even allow the capsule to be anchored in a specific region [18]. Modules for therapeutic interventions are also being developed, based on local drug delivery [2] or on the release of mechanical clips to stop GI bleeding [19].

Pioneering work in the field of integrated capsules was reported by Astaras at al. within the frame of the IDEAS project [10]. The developed prototype has the size of a commercial endoscopic capsule and is characterized by real time acquisition and processing capabilities, while it also features a duplex RF link to communicate with a base station. An onboard microcontroller handles the interfacing with four sensors: pH, conductivity, temperature and dissolved oxygen [10,11]. Due to the low power design of the electronics and the limited consumption of the sensors, two watch batteries were enough for meeting the capsule's overall power requirements, which amounted to less than 14 mW [10,11].

Within the frame of the EU funded Vector project, an advanced robotic capsule is under development [20], aimed both at increasing the diagnostic capabilities of commercial capsules and at providing therapeutic functionalities. Fig. 1 depicts a block schematic of the target robotic capsule and its main modules, while Table 1 summarizes their power consumption combined with their estimated *in use* time. As it is evident from this table, electromechanical tools have the greatest impact on the power budget of the capsule, mainly due to the high power peak required when the motors engage [21]. This clearly underlines the need for an alternative solution to batteries.

A multi-coil inductive link was shown to effectively address this problem [5,6,21,22]. 3D inductive powering efficiently overcomes the intrinsic limitations of embedded cells, providing over 300 mW within the same volume as the batteries [6]. This amount of power is expected to suffice for the smooth operation of a complete robotic capsule, on the premise of judicious activation of its more power-demanding modules.

The development of active locomotion schemes for endoscopic capsules (as opposed to passive traversal of the GI tract by natural peristalsis) is expected to significantly enhance the diagnostic (and, foreseeably, the therapeutic) scope of these devices [1,2]. Vibratory actuation is currently under investigation for micro-robotic devices (such as capsule endoscopes), whereby the centripetal forces generated by a rotating eccentric mass are exploited either as a means of frictional reduction, or for obtaining self-propulsion [23,24]. To this end, a series of prototypes have been developed which integrate on-board small vibratory motors, similar to those found in mobile phones or pagers. Potential benefits of such a locomotion method include the relatively simple implementation and control, the ability for bidirectional movement, as well as the fact that the actuator can be completely encased inside the robot, without any external protrusions. However, the power consumption of these vibratory motors is quite high, a fact that highlights the potential benefits of combining vibratory actuation with an inductive powering scheme.

2. Wireless powering

Although inductive powering has been widely used to bias biomedical implants [22,25,26], applying this approach to capsular endoscopy presents unique challenges. Whereas typical subcutaneous systems are calibrated for small coil separation (1-2cm), the capsule has limited dimensions and is freely moving through the GI tract. Besides constraining the size of the power receiver, this implies the secondary coil operates over 20 cm away from the primary coil, facing angular misalignments up to 90º.

Since, for the present application, the relative position between the power transmitter and the receiver is expected to vary significantly during the examination, a multi-coil system is mandatory. The use of 3D receiving coils was proven an efficient solution by Lenaerts [5] and this approach was further optimized by the authors in [6,21]. Fig. 2 depicts a schematic of the multi coil powering system. The external base station includes a primary coil and its driver. This unit produces an alternating magnetic field that is partially detected by the secondary coils and converted into voltage. A Helmholtz coil topology was selected for the primary coil, since it produces a magnetic field more uniform and stronger than a solenoid for the same excitation current. Moreover, it allows a good confinement of the field to the patient's trunk [6], which minimizes unnecessary exposure of other body parts, hence allowing full compliancy with the strict norms that regulate this matter [27].

The receiver consists of three resonant LC tanks tuned to the carrier frequency, which has been set at 1 MHz, in order to reduce the absorption of the human body. The three contributions are separately rectified and added for further voltage regulation. Two DC lines, the levels of which are application dependant, are provided to the embedded system; in the current implementation, these were selected at 3 and 1.8 V.

In addition to the power transfer, the system supports the transmission of low data-rate control signals from the external station to the capsule, by modulating the amplitude of the power carrier. Data are recovered at the capsule side by envelope detection of the rectified voltage (Fig. 2). With a 1 MHz carrier, a transmission speed up to 9.8 kbps can be achieved.

Experiments, detailed in [6], demonstrated that 300 mW can be transferred, without time limitations and regardless of capsule orientation, from an alternating magnetic field source to a set of orthogonal coils wound on a Ø9 mm ferrite core. A comparative characterization of various 3D coils in [6] indicated that use of a ferrite core allows an increment of the received power up to 150% within the same external field (270 μT), a 33% size reduction and compliance with the associated safety norms [27].

Fig. 3 depicts the implemented external unit consisting of the Helmholtz coil and its driving board (top), as well as samples of 9 mm 3D-coil power receivers (bottom). The power received by one of these modules under different coupling conditions is shown in Fig. 4. The graph indicates that the received power is always well above 300 mW, which are considered to be sufficient for allowing the integration of modules with increased power demands, such as the electro-mechanical ones.

3. Vibratory locomotion

Vibratory actuation is currently under investigation for use in endoscopic capsules, as well as for the propulsion of other miniature robotic devices over challenging environments [23,24]. This Section presents simplified models and corresponding simulation results, which illustrate the basic related concepts. As indicated in Fig. 5, the mechanical model of the system involves the main mobile platform representing the capsule (with its mass denoted by *M*), inside which a smaller mass m describes a circular trajectory of radius *r*, as it rotates on the xy-plane, about the platform's center of mass, at a constant angular velocity *ω*. For the present study, the system is thus constrained, so that translations along the *x*-axis are the only movements permitted. The rotation of the eccentric mass generates a centripetal force, whose components in the horizontal and the vertical direction are

$$C_x(t) = mr\omega^2 \sin\left(\omega t\right) \text{ and } C_y(t) = mr\omega^2 \cos\left(\omega t\right), \tag{1}$$

respectively. Denoting by $F_R(t)$ the frictional force opposing movement of the platform over the substrate, the system's equation of motion along the x-axis is derived as:

$$M\ddot{x}(t) = C_x(t) + F_R(t) \tag{2}$$

Assuming a Coulomb friction model for the interaction with the locomotion environment (such an assumption may be an adequate first approximation for movement over sand [28], as well as for certain parts of the gastrointestinal tract [29]), and denoting with μ the associated coefficient of friction, the frictional force is obtained as:

$$F_R(t) = \begin{cases} -\mu F_N(t)\,\mathrm{sgn}(\dot{x}) & \text{for } F_N(t) > 0 \\ 0 & \text{for } F_N(t) \le 0 \end{cases} \tag{3}$$

where

$$F_N(t) = (m + M)g + C_y(t) \tag{4}$$

is the normal force exerted on the capsule (g denotes the gravitational constant). Note that, depending on the parameters involved (primarily on the rotational velocity ω), $F_N(t)$ may also assume negative values, and such a case is associated with zero frictional force in Eq. (2). Matlab has been used to numerically solve the above equations, where the values of the parameters (specified as $M = 12$ g, $m = 0.32$ g, $r = 0.85$ mm, and $\mu = 0.3$) were selected to correspond to those of the prototype presented in Section 4. Note that, in order to obtain estimates for the rotating mass m and its eccentricity r, the vibratory motor employed in the prototype has been disassembled and its parts have been analyzed with a CAD software package. Simulation results, demonstrating the generation of forward locomotion for the system, are summarized in Fig. 5b. As indicated by these results, in order for a net displacement of the platform to be obtained, the rotational velocity ω of the eccentric mass must be above a certain threshold value. It is also noted that, since no assumption of asymmetric friction is made, the overall direction of locomotion can be reversed, without any performance loss, simply by reversing the rotational direction of the eccentric mass.

4. Integrated prototype and experimental results

A capsule prototype has been developed, which integrates the inductive power receiver with a vibratory motor (Fig. 6a,b) [30]. The capsule dimensions are similar to those of a large vitamin pill (length 28 mm, diameter 16 mm, weight 12.5 g), while the outer shell has been fabricated in ABSplus material, using a 3D printer. The vibration motor is a commercially available coin-shaped Ø10 mm shaftless pager motor (Precision MicroDrives 310-101), which weights about 1.2 g and requires about 290 mW to rotate at its nominal velocity of approximately 12000 rpm at 3 V. Two permanent magnets have also been integrated on-board the prototype (Fig. 6b), for future investigation of locomotion schemes involving the combined use of vibratory actuation with an external magnetic field [31].

The capsule locomotion has been tested in several environments, Table 2 summarize the average speed achieved in the different substrates. Preliminary tests were performed using two different challenging unstructured substrates, namely sand (Fig. 7a,b) and liquid soap. These successfully demonstrated both the effectiveness of the wireless powering system and the efficacy of the vibratory locomotion scheme, with the capsule prototype attaining speeds of 9 cm/sec and 3 cm/sec, respectively. It is noted that, due to the inductive powering scheme, these experiments had to be performed within the confined space surrounded by the external coils. Therefore, due to the interaction of the device with the walls of the container, the eventual trajectories prescribed by the capsule were of a circular shape (Fig. 7a,b). In additional tests, involving similar vibratory-actuated prototypes, powered by on-board batteries, movement in unconfined spaces was along a straight line, in accordance with the simulation results presented in Section 3. Straight line trajectories were also achieved with inductive powering (for short distances in this case) over both sand and a block of solid foam without any rigid edges (Fig. 7c, video 1).

Another test was performed using a rubber hose (length 2 m, diameter 30 mm), which was twisted and positioned between the two plates of the Helmholtz coil structure (Fig. 7d). The capsule was successful in traversing the whole length of the hose, which involved overtaking a path with variable slopes, and achieving an average speed of 10 cm/sec (video 2). This test confirmed that the induced field is strong enough to sustain the motor within a large region between the two plates of the Helmholtz coil, despite significant variations of the capsule's orientation. Moreover, the effectiveness of motor action is demonstrated, as the capsule managed to overcome slopes as steep as 60%: in this case the centripetal force should overtake the combination of the frictional and the gravity forces (capsule weight is over 12 g).

Future experiments will consider movement over segments of GI tissue, in order to provide a more relevant assessment of the scheme’s potential for integration in endoscopic capsules. Work is also underway to integrate a microcontroller on-board the capsule, which will make use of the low data rate signals, transferred over the power link, to enable a more active control of the capsule's motion by regulating the speed and rotational direction of the vibratory motor.

5. Conclusions

A wireless power supply represents a significant breakthrough in the development of robotic endoscopic capsules. Guaranteeing over 300 mW in the worst coupling conditions and without time limitation, 3D inductive powering overcomes the energy shortage, which is drastically limiting the capabilities of commercial capsules. This opens the road to the integration of advanced diagnostic tools and active locomotion system. The integration of vibratory actuation in a wirelessly powered capsule was described and the locomotion capabilities of a capsule prototype evaluated. It was demonstrated that the transferred power is sufficient to allow effective capsule propulsion over various substrates.

ACKNOWLEDGEMENTS:

This work was supported in part by the European Community, through the FP6 IST R&D project VECTOR (contract number 0339970). The authors would like to acknowledge the contributions of the project partners in various aspects of the present work.

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Video 2: Vibratory capsule traversing a twisted rubber hose, positioned between the two plates of a Helmholtz coil structure used to generate the alternating magnetic field.

Tables:

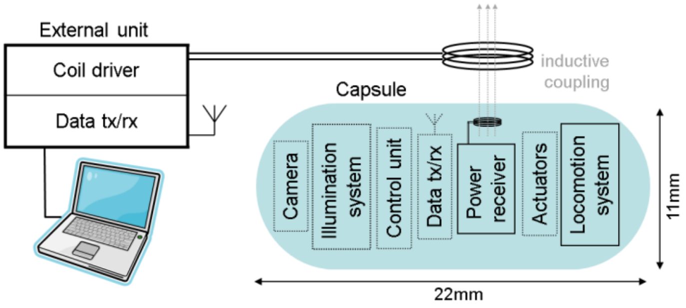
Table 1: Power consumption of the main modules embedded in a robotic capsule

Module	**Required Power**	***In use* time**
CMOS imager	40 mW	70-100%
LED illumination	4 x 10/20 mW	70-100%
µController	6 mW	100%
Image compression	7.5 mW	70-100%
Auto focus	12 mW (@ 50V)	50-80%
Sensors	2mW	1-5%
Locomotion	> 300 mW	3-10%
Actuators	>200 mW	1-2%
Transmitter	5 mW	90-100%
Receiver	< 5 mW	1-80%

Table 2: Measured speed in different environments

Testing Environment	**Speed**
Solid Foam Material	12 cm/sec
Twisted Rubber Hose	10 cm/sec
Sand	9 cm/sec
Liquid Soap	3 cm/sec

Figures:



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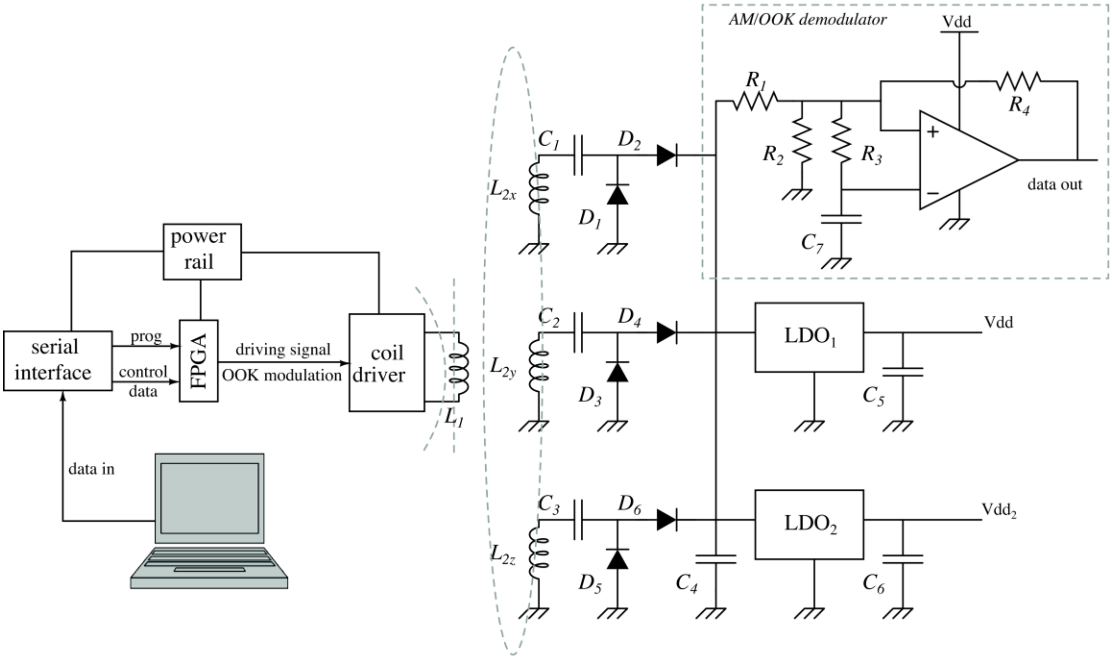
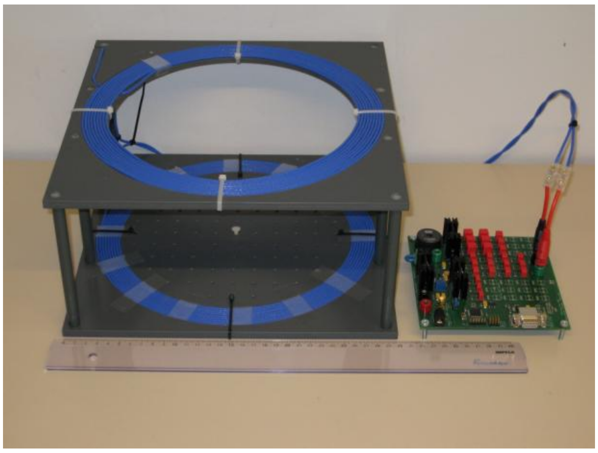
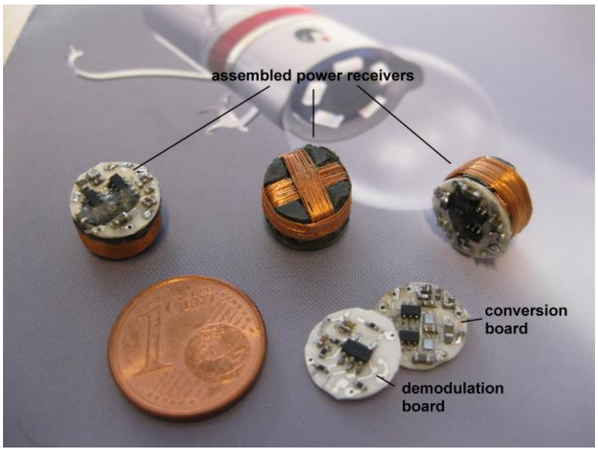


Figure 2: Schematic of the wireless powering system.





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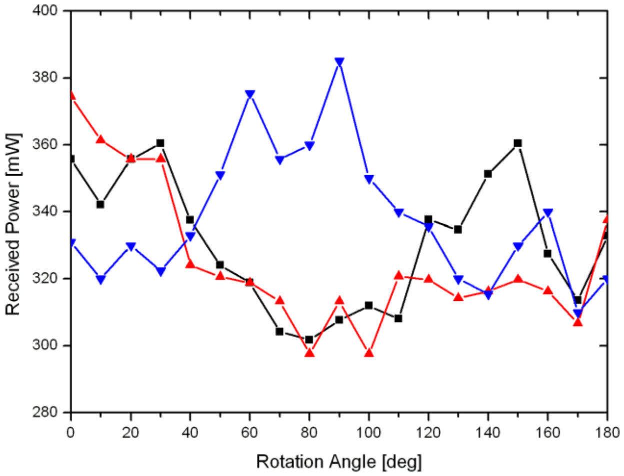

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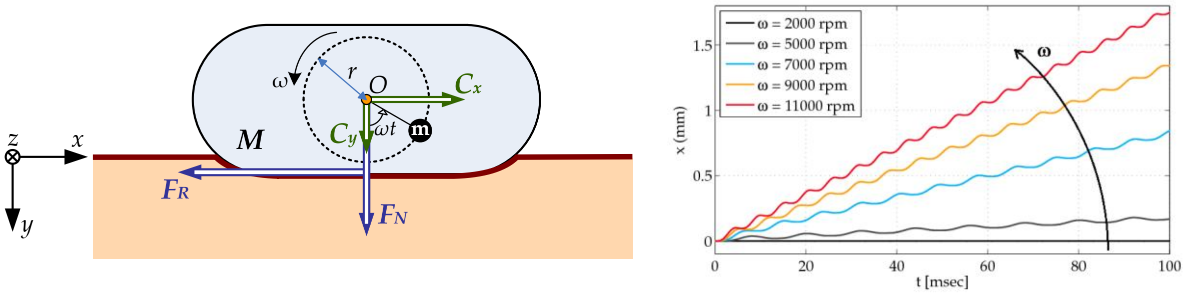

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(b)

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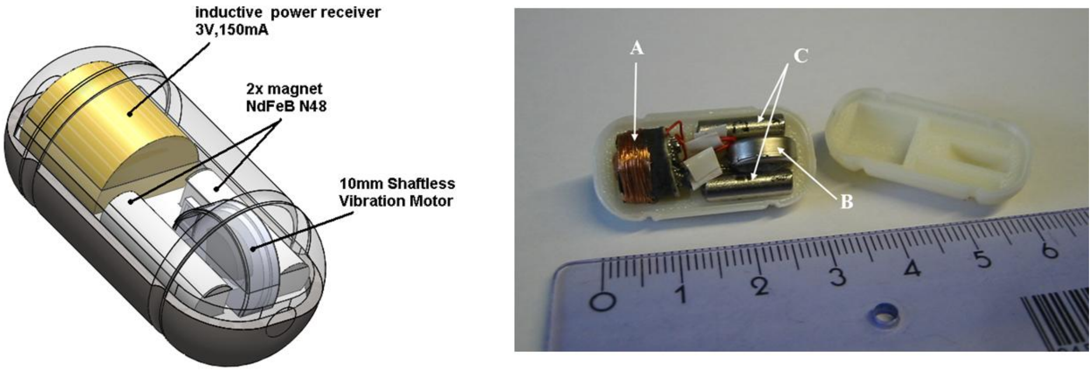

(a) (b)

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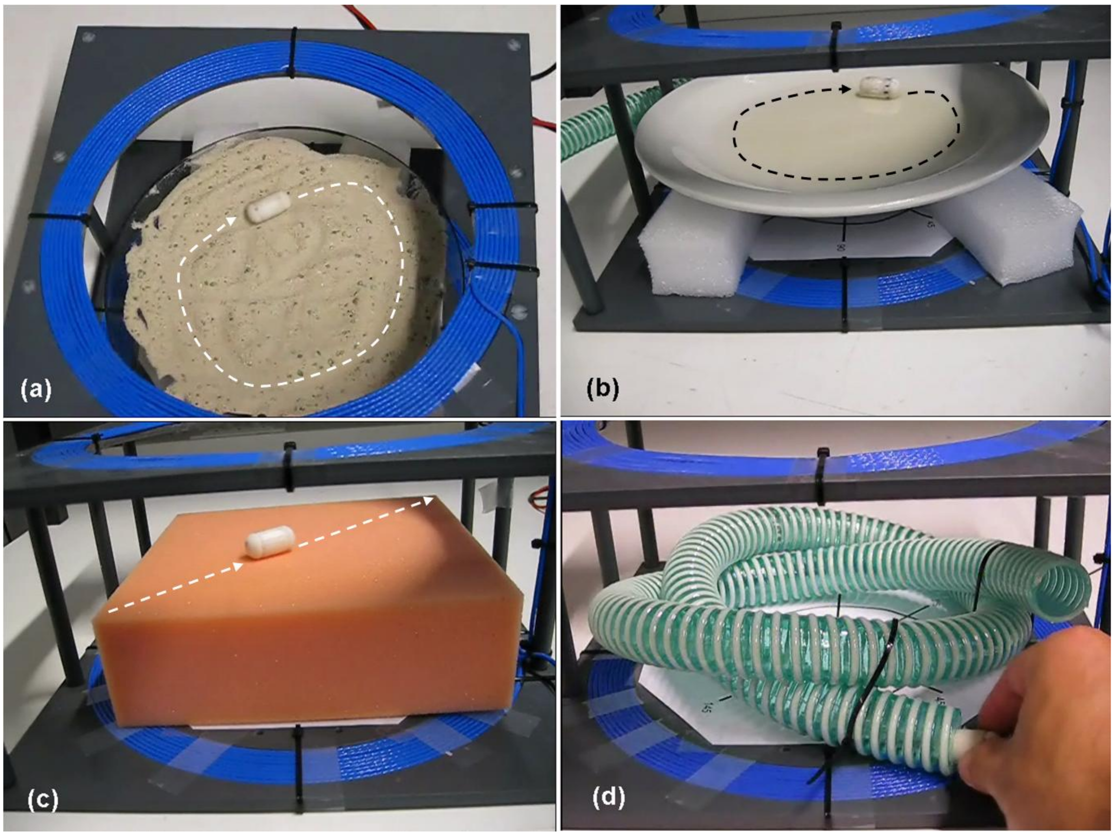

Figure 7: Snapshots of the testing activities: locomotion over (a) sand, (b) liquid soap, (c) solid foam material, and (d) traversing a rubber hose.

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At present, he is a full professor at the K.U.Leuven, teaching courses in 'Microsystems and Sensors’, in 'Biomedical Instrumentation and Stimulation', in ‘Biomedical Equipment and Regulations’, in ‘Production techniques for microelectronics’, and basic courses in ‘Electronics, System Control and Information Technology’. He is the author or co-author of more than 500 papers on biotelemetry, sensors, MEMS and packaging in reviewed journals or international conferences.

He is a Fellow of the Institute of Physics (UK), senior member of the IEEE Electron Device Society (EDS) and many others.

He is Regional Editor of the IOP Journal of Micromechanics and Microengineering, was general chair of the Eurosensors conferences between 2004 and 2010, and acted as the General Program Chairman for Transducers’07 in Lyon.