Biosensors and Bioelectronics 25 (2009) 845–851

Contents lists available at ScienceDirect

Biosensors and Bioelectronics

journal homepage: www.elsevier.com/locate/bios

Wireless powering for a self-propelled and steerable endoscopic capsule for stomach inspection

R. Carta[a,*], G. Tortora[b], J. Thoné[a], B. Lenaerts[a], P. Valdastri[b], A. Menciassi[b,c], P. Dario[b,c], R. Puers[a]

[a] K.U.Leuven, ESAT-MICAS, Kasteelpark Arenberg, B-3001 Leuven, Belgium
[b] CRIM Lab, Scuola Superiore Sant'Anna, Viale Rinaldo Piaggio 34, 56025 Pontedera (PI), Italy
[c] Italian Institute of Technology, Network, Genova, Italy

ARTICLE INFO

Article history:
Received 14 July 2009
Received in revised form 25 August 2009
Accepted 28 August 2009
Available online 4 September 2009

Keywords:
Capsular endoscopy
Wireless power supply
Active locomotion

ABSTRACT

This paper describes the integration of an active locomotion module in a wirelessly powered endoscopic capsule. The device is a submersible capsule optimized to operate in a fluid environment in a liquid-distended stomach. A 3D inductive link is used to supply up to 400 mW to the embedded electronics and a set of 4 radio-controlled motor propellers. The design takes advantage of a ferrite-core in the receiving coil-set. This approach significantly improves the coupling with the external field source with respect to earlier work by the group. It doubles the power that can be received with a coreless coil-set under identical external conditions. The upper limit of the received power was achieved complying with the strict regulations for safe exposure of biological tissue to variable magnetic fields. The wireless transferred power was proven to be sufficient to achieve the speed of 7 cm/s in any directions. An optimized locomotion strategy was defined which limits the power consumption by running only 2 motors at a time. A user interface and a joystick controller allow to fully drive the capsule in an intuitive manner. The device functionalities were successfully tested in a dry and a wet environment in a laboratory set-up.

1. Introduction

Wireless capsular endoscopic devices get more and more accepted as a useful tool for early diagnosis of cancer and other diseases affecting the gastrointestinal (GI) tract (Moglia et al., 2007). This practice is intended to serve as an alternative to traditional endoscopy. It is typically well tolerated by patients and is reported to be painless (Eliakim, 2004). However, commercially available pills are limited to screening and are purely passive devices. Most of them are battery powered. This involves a limitation in terms of available power and analysis duration, since the typically used button cells can only provide an average power of 25 mW for 6–8 h (Eliakim, 2004; Arena et al., 2005; Mc Caffrey et al., 2008). This energy is barely sufficient to transmit low quality images (256×256 pixels) at a low data rate (2 frames per second) to an external unit (Eliakim, 2004; Mc Caffrey et al., 2008; Iddan et al., 2000).

Energy harvesting is not an option since such systems are still far from providing adequate power to guarantee the complete function of the capsule. An exhaustive summary of the most recent achievements in this field is reported by Fiorini et al. (2008) that refers about the maximum amount of power that can be harvested. Among the available possibilities, only systems based on vibrations can practically fit in implanted devices. Up to now these systems can only deliver a few μW, in the best case scenario. Moreover, they would require a complex power management circuit and to be coupled with a rechargeable battery as they are intrinsically inefficient as a primary power source. A promising approach, generating voltage out of water-filled single-walled carbon nanotubes (SWCNT), is described by Yuan and Zhao (2009). However, the limited amount of power provided at this early stage makes this solution not suitable for capsular endoscopy.

As reported in an extensive review about capsular endoscopy and its future directions, a wireless magnetic link may offer a solution to cope with the lack of electrical power (Moglia et al., 2009). Such inductive couplings have been used over the past several years to power biomedical implants. Platforms based on a pair of aligned coils are used to supply implanted micro-pumps (Miura et al., 2006), vision (Schnakenberg et al., 2000) and orthopedic systems (Catrysse et al., 2004; Van Ham et al., 2007). These powering solutions are often referred to as Transcutaneous Energy Transmission (TET) systems. The relative position of external (transmitting) and implanted (receiving) coils is typically fixed within a given tolerance. Under such conditions and with dedicated designs, it is reported that up to 20 W can be transferred with this technique (Vandevoorde and Puers, 2001). Unfortunately such power transfers are totally out of reach in the case of endoscopic capsules. The capsule is freely moving through the GI tract and the use of

* Corresponding author. Tel.: +32 16321108; fax: +32 16321975.
E-mail address: riccardo.carta@esat.kuleuven.be (R. Carta).

0956-5663/$ – see front matter © 2009 Elsevier B.V. All rights reserved.
doi:10.1016/j.bios.2009.08.049

multiple transmitting or receiving coil-sets is mandatory in order to guarantee a good coupling for any given position of the capsule (Lenaerts, 2008). Moreover, the distance between transmitting and receiving-coils, in most TET systems, is in the order of a few centimeters, whereas this distance is much larger in the case of capsular endoscopy. Another constraint is the presence of living tissue posing severe restrictions on the field strengths. This limitation arises from exposure regulations protecting the health and safety of the patient (ICNIRP, 1998). Finally, the size restrictions do not allow such high power levels to be transferred to the tiny coils in the capsule. In the case of capsular endoscopy, it has been shown that an orthogonal coil-set at the receiving side yields the best results (Lenaerts and Puers, 2005). The receiving-coils contribute simultaneously to the powering of the embedded electronics and actuators and no active selection of the best coupled pair is required. The use of such a wireless power supply overcomes the shortage in on-board available energy and enables the integration of high power demanding modules (Swain, 2008): a more powerful and faster transmitter (Thoné et al., 2008), a better image sensor with focus adjustment (Cavallotti et al., 2008) and actuation/locomotion modules are now possible.

Most commercial pills are not equipped with active locomotion systems. While this may be sufficient for the analysis of the small bowel, it is crucial to have the possibility to steer the capsule to enable accurate investigation of some critical parts of the GI tract, especially in the stomach. Active locomotion would allow to speed-up, slow down and eventually stop the device (Swain, 2008; Quirini et al., 2007a,b).

As suggested by Quirini et al. (2007a)-EMBC, a novel approach to full GI investigation is aiming at addressing each single district with different technical solutions. In particular, it is possible to address separately esophagus, colon and stomach. The esophagus capsule must be provided with a stopping mechanism in order to slow down the fall in the stomach, allowing the inspection of the cardia sphincter, seat of important pathologies (Blot et al., 1991; Wang et al., 1986). The capsule for the colon must be equipped with an active locomotion system, designed to cope with the large lumen diameter, typically 2–3 cm. A legged locomotion module was developed by some of the authors and tested in *ex vivo* trials (Quirini et al., 2007b-ICRA). The capsule for the investigation of the stomach certainly presents unique technical challenges. It must be provided with a fully 3D steerable locomotion system for the inspection of a relatively spacious environment (about 30 cm long and 15 cm wide). The design of such a wireless capsule was preliminary described by the authors in Carta et al. (2008).

Investigation of the stomach is best conducted when the organ's cavity is distended. In traditional gastroscopy, the distension is obtained by blowing air into the stomach through the endoscope. As air distension is practically unfeasible for an endoscopic capsule, an optimal distension can be obtained by filling the cavity with water or another transparent liquid. The resulting image quality is enhanced by the presence of the liquid medium, which prevents the accumulation of organic debris in the field of view (Quirini et al., 2007a-EMBC; Rieber et al., 2008). The liquid environment allows locomotion with a lower coefficient of friction, thus at a lower power consumption.

The work towards the integration of an inductive power module and a locomotion unit, consisting of a radio-controlled set of propellers, is described in this paper. The two modules have been successfully combined and tested into a single miniaturized capsule.

2. Design

Medical considerations establish clear constraints on the design of endoscopic capsules that relate to both their size and shape. Ideally, an endoscopic capsule should be small enough to swallow and should be wireless-controlled without inducing negative effects on the patient compared to traditional endoscopy. Device design must guarantee that all the single components are embedded within a solid structure, avoiding any protruding part. The main elements of the proposed system consist of the inductive power receiver and the wireless-controlled locomotion unit. Both modules are designed to be scalable to meet the dimensions of commercial endoscopic devices (Ø 11 mm × 31 mm). The main purpose of capsular endoscopes is to acquire images from the GI tract; consequently, a compact vision module must also be integrated, that is foreseen in a future version. In the present paper the focus is on the feasibility of the combination of a wireless 3D motor control with a power link into a capsular shaped device. A free volume for the future integration of a vision module and additional functions is allocated. This resulted in a preliminary prototype of Ø 15 mm × 40 mm.

Fig. 1 depicts the block diagram of the complete system. The gastroenterologist can wireless-control the capsule through a personal computer (PC) base station, whereas the external power source is placed around the patient chest. The capsule contains the inductive power receiver, the control/telemetry unit and the locomotion system. Supplementary Fig. 1 depicts the submersible capsule in its target environment together with a zoomed view of the prototype. This section describes the different modules and the system integration process.

2.1. Wireless power supply module

A schematic of the wireless power module is depicted in Fig. 2. At the primary side, a dedicated power amplifier drives a current through the transmitting-coil, generating a magnetic field. A fraction of this field is picked up by the secondary coil, which converts it into a voltage. The received power is then regulated and distributed to the active electronic circuits. At the bottom-right of Fig. 2, a detail of the power-receiving circuitry is depicted. Every coil of the receiving-set is tuned with a series capacitance at the carrier frequency of 1 MHz and the three contributions are composed

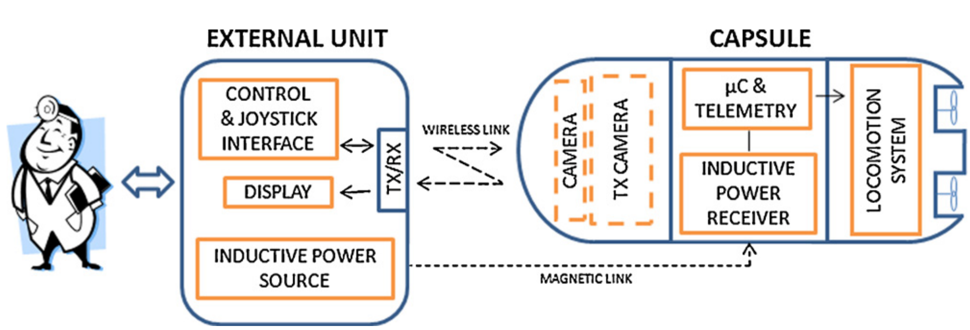

Fig. 1. Block diagram of the wireless power, self propelled and steerable capsule.

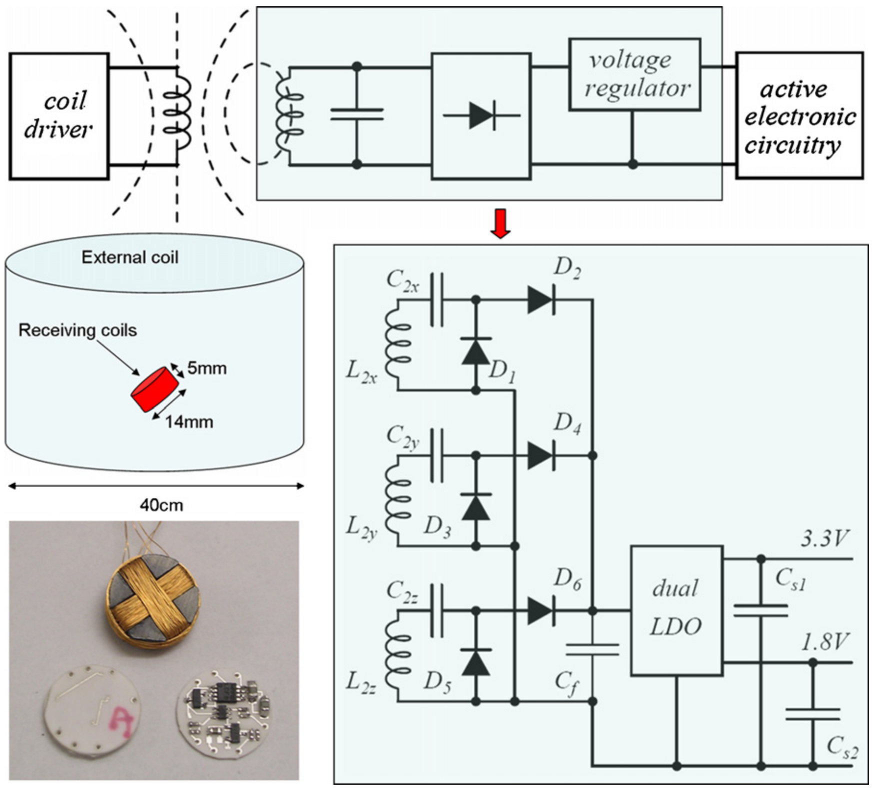

Fig. 2. General principle underlying inductive powering systems (top) and schematic of the multiple receiving coil-set (right). At the left, spatial arrangement of the coil composing the power module (top) and a detail of the 14 mm ferrite-coil receiver before assembling (bottom).

at direct current (DC) level. Further voltage regulation is obtained with Low Drop Out regulators (LDO). Again Fig. 2 illustrates the mutual position of the transmitting and receiving-coils. The external coil is placed around the patient's chest and three orthogonal coils are embedded in the capsule (Lenaerts and Puers, 2005). For the receiving-set, two possibilities are available: air-coils fitting one inside the other, with embedded power-conversion electronics, or a solution incorporating a ferromagnetic core. A prototype implementation of the first solution is presented by Lenaerts and Puers (2007). This is capable of providing up to 200 mW to the implant, for any orientation and position of the capsule within the operating volume. The second solution is promising since it substantionally improves the coupling with the external field (Miura et al., 2006). This approach either increases the power for the same coil geometry and it allows to shrinks the volume when the same amount of power is envisaged, thus leaving more space for the electronics and the actuators. The beneficial effect of a ferrite disk in close proximity of the receiver was described by Leuerer and Mokwa (2004) for planar micro-coils and it was concluded that an improvement up to 59% was possible with respect to the air-coil case.

Given a field distribution and fixed the receiver size, finite element (FE) simulations show the magnetic flux is denser inside a ferrite-coil than in the air-coil dual, hence boosting the coupling with the external source (Supplementary Fig. 2). The increment of flux density is confined to the receiver and its proximity, whereas no sensible changes are measured in the space between primary and secondary coils (occupied by the patient's chest).

Local field lines deflection inside the ferrite is not problematic due to the presence of three orthogonal coils wound around the same core. Chances are high that lines missed by one coil will be caught by the others, even if they are not perfectly aligned with the external coil. This is why maxima of received power are achieved in alignment conditions resulting in minima for air-coil orthogonal sets.

In the present work, a solution based on three orthogonal coils, wound on a 5 mm thick ferrite frame, is used. A 14 mm diameter frame was laser shaped from a sheet of 3F4 (Ferroxcube) ferrite, which guarantees low losses at the operating frequency of 1 MHz. The choice of the frame-section follows the principle that, for a given input power, the larger the cross-section the better the coupling and the higher the transferred power becomes. For this reason, when possible, the capsule cross-section should be fully exploited by the receiving-set.

The ferrite-coils are assembled with a power conversion board. This design was realized on a 0.6 mm thick ceramic substrate with the same diameter as the coil frame. Components were selected among the smallest commercially available. Two DC lines are provided, one at 3.3 V and one at 1.8 V using a dual output LDO (LP2966IMM-1833, National Semiconductor). The conversion to a dedicated ASIC (Application Specific Integrated Circuit) will be considered in the future in order to save some extra board area. Fig. 2 (bottom-left) depicts the ferrite-coil receiver and the conversion board prior to the assembly operation.

2.2. Locomotion module

The locomotion module is a modified version of a propeller based solution meant to be embedded in a battery supplied capsule (Supplementary Fig. 3). This unit, its integration and the relative testing activities were described in Tortora et al. (2009). Due to the satisfactory results in terms of controllability but its limited autonomy, this solution was chosen for integration in a wireless powered capsule. The locomotion module was designed to allow capsule navigation inside a liquid-distended stomach. The oral ingestion of 500–1000 ml of Polyethylene Glycol (Macrogol, PEG) solution achieves the gastric distension without significant discomfort for the subject (Rieber et al., 2008). PEG is considered beneficial for our purpose, since it is completely transparent to visible light, biocom-

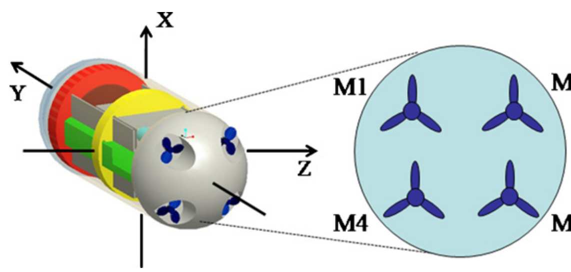

Fig. 3. Schematic of the locomotion system and actuation diagram of the propellers.

patible, non-toxic and well established in endoscopic procedures. The final density of the PEG solution is 1105 kg/m^3, which is slightly higher than water. This is helpful to reach a neutral buoyancy of the capsule, whereas its viscosity is low enough to keep the propellers from blocking. The main collateral effect of PEG solution is diarrhea. Although this might be an acceptable side effect, the use of alternative liquids can be considered in order to improve the patient's tolerability, without affecting the technical solution.

The backbone of the module takes advantage of four 3 mm in diameter propellers, actuated by electromagnetic DC motors. The actuators (Didel MK04S-24, Ø 4 mm × 8 mm) are located in the rear part of the capsule and embedded inside a protective structure in order to obtain a smooth pill-like shape. This solution improves the safety of the patient during swallowing and reduces capsule retention, which is one of the major risks in capsular endoscopy (Buchman et al., 2004). The propellers, visible in Supplementary Fig. 4, and the external shell are fabricated by urethane acrylate polymer, using the stereo-lithographic technique.

The choice of an even number of motors guarantees a good stabilization of the induced roll torque, which may be produced by inertial effects when just one propeller is active at a time. Four propellers are active if the capsule has to move forward at high speed. Alternatively, in order to reduce the power consumption, just two opposite motors (M1 and M3 or M2 and M4 of Fig. 3) can be actuated, with different spin orientation, to achieve forward motion as well, but at a lower speed. To make the capsule steer in one direction, the two propellers on the opposite side must be activated while the others are turned off. When the motors stop, the capsule maintains both its position and orientation thanks to its neutral buoyancy.

A CC2430 wireless ZigBee compatible microcontroller (μC) from Texas Instruments allows bidirectional data communication and real time control of the DC motors (Valdastri et al., 2008). The μC is characterized by low power consumption and 2.4 GHz transmission frequency which is significantly different from the frequency used for the power transfer (1 MHz), hence avoiding interference between the two signals. A graphical interface, on a PC user station, allows setting parameters like the maximum speed, retrieving information from the remote device about battery status, signal strength and propeller activation. In the final system, this interface will also include the images transmitted by the capsule, acquired with an integrated camera. This will allow the endoscopist to have visual feedback to control the capsule. For a more intuitive steering, a joystick control is implemented in order to move the capsule in the desired direction and vary its speed. The overall system architecture for the active control of the capsule is represented in Fig. 1. Supplementary Fig. 4 depicts the locomotion module to be assembled in the wireless powered capsule, the external USB transmitter and a screenshot of the graphical interface.

2.3. System integration

Both air- and ferrite-coil solutions fit inside this capsule. However, the ferrite-coil module was preferred for the final integration because of its better performances within a smaller volume.

The overall volume of the developed power receiver is 1.09 cm^3 (Ø 14 mm × 7 mm) and its weight is 3.9 g. The locomotion module was partially redesigned with respect to the battery supplied version (Tortora et al., 2009) in order to simplify the integration in the wireless powered capsule. The motors and the electronics were rearranged inside the capsule in order to fit an internal volume of 1.46 cm^3 (Ø 13 mm × 11 mm). This results in a more compact assembly, which is not dependent from the battery shape anymore. The endoscopic capsule prototype has a diameter of 15 mm and a length of 40 mm. At present, less than 60% of the overall capsule volume is required for the integration of power and locomotion modules. The rest of the available space will be used in future versions to include additional electronics/actuators and a camera module. This volume is more than sufficient for this purpose. A vision module (imager and illumination), equivalent to the one mounted on the PillCam capsule, only requires about 300 mm^3 [www.givenimaging.com].

With 400 mW, the wireless power module delivers the power demanded by the locomotion system at regime. This is sufficient to perform 3D locomotion during a complete diagnosis of the stomach. Up to 75% of the available power is consumed by DC motors, while the μC only needs about 3.0 μW in power-down (with active pins for driving motors) and 60 mW in receive mode (to get joystick direction and speed settings). The motors are controlled using a pulse width modulation (PWM) signal with a variable duty-cycle set through the user interface. Adjusting the duty-cycle will allow the gastroenterologist to modulate the speed of the capsule during the examination.

Weight adjustment, in order to obtain neutral buoyancy, is of paramount importance in order to achieve 3D steering capabilities. This is required to keep the capsule in its position and orientation when the actuators are turned off. This can be achieved by adding or reducing ballasting material into the capsule prior sealing the external shell. Moreover, it simplifies the control system since the same thrust forces are required for any direction. Fig. 4 depicts the two integrated modules before sealing the capsule. The two details at the bottom illustrate a magnified view of the locomotion module and the fully assembled capsule. Moisture resistance was obtained by sealing all joints with cyano-acrylate. This was sufficient for lab testing, performed in a liquid environment, and lasting a few hours. A complete and reliable sealing procedure was already described for the battery supplied device by Tortora et al. (2009). It consists on sealing all the junctions of the assembly with epoxy glue, using rubber rings to seal the cavities where the motor shafts are located. This allowed waterproofing the capsule for a full set of *ex vivo* and *in vivo* experiments with the battery operated propeller based capsule (Supplementary Fig. 3). The same procedure is envisaged for the further developments of the current prototype.

3. Results and discussion

The two modules were first tested separately in order to assess their performances and prove their compatibility in terms of provided/required power. The integrated capsule was then tested both in a dry and wet environment in a laboratory set-up. This section describes the characterization of the single modules and the functional testing carried on the integrated device.

3.1. Characterization of the power module

The design of the ferrite-coil receiver aimed at receiving more power within a smaller volume than the air-coils described by

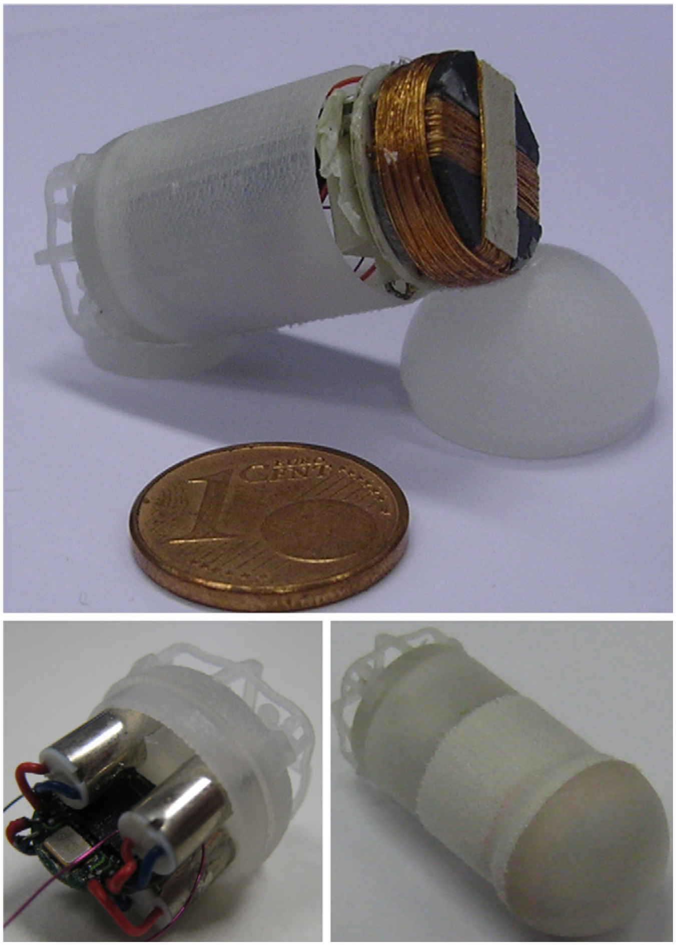

Fig. 4. Integration of the power module and the locomotion module in a single capsule. The details depict a magnified view of the ZigBee μC and motors (bottom-left) and the assembled capsule (bottom-right).

Lenaerts and Puers (2005), it is therefore useful to compare these two modules.

Power transfer measurements were performed on bench under different coupling conditions. Specifically, to transfer 200 mW to the secondary load, in the worst coupling case, a primary magnetomotive force (mmf) of 96 ampere-turns is required with the air-coil-set, whereas only 75 ampere-turns are required with the ferrite-core coils. Moreover, the overall size of the ferrite-coil module is about 30% smaller than the air-coil solution. The magnetic field is generated with a dedicated class E coil driver operating at a frequency of 1 MHz (Lenaerts and Puers, 2007). A polyvinylchloride (PVC) support allows positioning of the secondary coil-set in the center of the primary coil, both axially and longitudinally. The coil-set is held on a pivoted disc that can be rotated to test the different orientations. The rotation angle is read from a protractor (Carta et al., 2008).

Fig. 5 depicts a comparative polar plot of the received power for an externally applied mmf of 65 ampere-turns and with the orthogonal coil-set placed in the center of the primary coil. The characterization was performed by rotating the receiver 360° around the axis defined in Fig. 5, and reading the voltage output in steps of 10°. In order to de-embed the measurements from the effect of capacitive coupling to ground, readings of complementary angles are averaged. The received power by the ferrite-coils is well above 350 mW for all orientations and it almost doubles the performances of the air-coils within a smaller volume and the same external conditions.

A crucial point of this wireless powering system is the compliancy to safety standards, since the whole patient's body will be exposed to an external alternating magnetic field. An extensive description of the effects of such a field on the human tissues was done by Lenaerts and Puers (2007), who monitored the equivalent series resistance (ESR) of the primary coil in different experimental conditions in order to obtain an indirect measure of the specific absorption rate (SAR). According to the guidelines defined by the International Commission on Non-Ionizing Radiation Protection (ICNIRP), a whole-body average SAR of 0.4 W/kg is considered a safe limit to provide adequate protection for occupational exposure (ICNIRP, 1998). It was proven that reducing the impact of the conservative electric field, by shielding the inner side of the external coil, significantly decreases the absorption of the human body, still allowing transferring up to 180 mW to the air-coils within a SAR below 0.35 W/kg at 1 MHz (Lenaerts and Puers, 2007).

Although the same external magnetic field source is used in the present work, the ferrite coil-set significantly improves the coupling and hence the transferred power with the primary coil. The main outcome is that more power can be transferred to the capsule within the same external conditions and hence complying with the safety requirements.

3.2. Characterization of the locomotion module

The power consumption of the radio-controlled locomotion module was measured in a set-up supplying the unit with an external 3.3 V power source and monitoring the current drawn in several working conditions. The power demand was evaluated both in case of simultaneous and sequential actuation of the motors. As it is clearly reported in Fig. 6, the power peak required at the engagement is much higher than the consumption at regime.

In those experimental conditions, noise affects the signal in a wide frequency range. This is mainly due to the characteristics of DC brushed motors, with high current spikes. The actual amount of current drawn in the transient depends on the position of the motor-brush with respect to the magnet in the stator. This has a direct effect on the power peak which can be problematic when a limited amount of power is available, especially when more motors are engaged simultaneously. Although the inductive power module was proven to be sufficient to sustain 4 motors at regime and its output stage can provide up to 150 mA at 3.3 V, the high transient peaks may lead to a temporary voltage drop of the power line output. This might cause μC reset which would jeopardize the capsule operation. Since increasing the external magnetic field is not allowed for safety reasons (i.e. human body exposure to varying electromagnetic fields), a different solution must be sought.

In order to cope with the limited amount of power, a locomotion strategy counting on just two motors working simultaneously was devised. Fig. 6 (bottom) illustrates that approximately 200 mW (at regime) are sufficient to sustain two motors in all possible configurations. As two motors engage, the power peak is typically below 320 mW and can be totally supported by the inductive link. To move the capsule forward, two centrally opposed motors are actuated. To move left, right, up and down, two propellers on the same side are activated. A joystick control and an optimized driver enable a smooth transition from one configuration to another. All the plots reported in Fig. 6 refer to motor operation at 100% duty-cycle. Although the power consumption at regime can be dramatically reduced by decreasing the duty-cycle down to 15–25%, this is not the case for the power peaks when the motors engage. The embedded motors, which do not possess a linear behavior in terms of current supply, have to overcome higher frictions when engaging at a lower duty-cycle. This makes the use of the air coil-set not

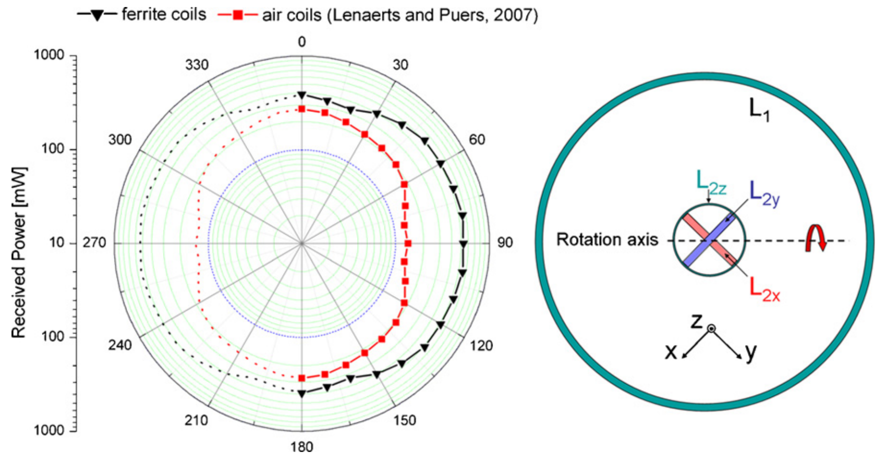

Fig. 5. Compared measurements of the power received by ferrite- and air-coils (Lenaerts and Puers, 2007) for an external mmf of 65 ampere-turns. The dotted lines represent an extrapolation of the measurements to the full 360°. *Note*: The power scale is logarithmic. The relative position of primary, secondary coils and the rotation axis are resumed in the scheme at the right.

a viable option for the propelled capsule: although the received power is sufficient to sustain two motors at regime, the module cannot cope with the transient current peaks.

An additional improvement may be achieved by using a supercapacitor as an energy buffer to absorb part of the transient current peaks, maintaining a more uniform power demand. Preliminary testing with a 220 mF supercapacitor (Ø 10 mm × 3 mm; LM035224A, Maxwell Technologies Inc.) has proven a transient peak reduction up to 30%, requiring a mere 5% of the overall capsule volume.

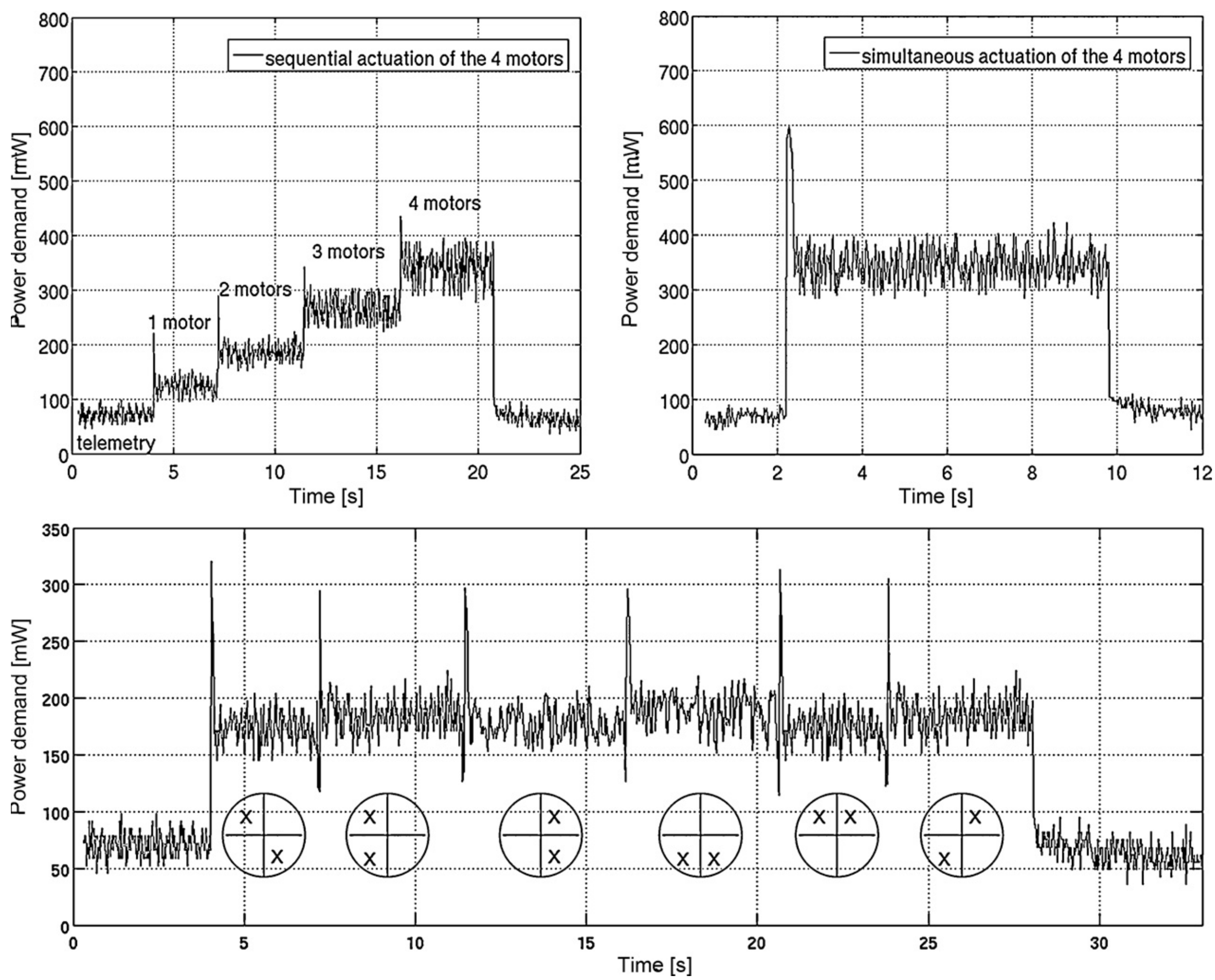

Fig. 6. Measurement of power absorbed by the control/locomotion module related to sequential (top-left) and simultaneous (top-right) actuation of the 4 motors. At the bottom, a 3D locomotion strategy based on the simultaneous use of only two motors.

3.3. Functional testing

The set-up used to test the performance of the integrated device was composed by the external magnetic field source and a control base station consisting of a laptop, a 3D joystick and the transmitter/receiver unit (Supplementary Fig. 5).

Dry and wet environments were created for the capsule operation. The dry set-up was the same used to characterize the 3D inductive link described in Section 3.1. This test shows the controllability of the propellers through the wireless transmission system, actuating them either sequentially or simultaneously. For liquid environment tests, a jar containing 5 l of water was placed inside the primary coil. The main focus of these experiments was to demonstrate the feasibility of integrating wireless power supply together with the propeller based locomotion module. A thorough characterization of the locomotion module, in a battery supplied capsule, was described in Tortora et al. (2009).

This work demonstrated that a propelled capsule is able to move in all directions in the gastric environment under wireless-control by the endoscopist. Freestyle swimming was shown to be feasible inside a stomach mock-up. The steering precision was demonstrated by controlling the capsule through small ring-shaped targets. *Ex vivo* and *in vivo* experiments confirmed the feasibility of the propelled solution, enabling locomotion inside an explanted porcine stomach and in the gastric cavity of a living pig.

The wirelessly transferred power was proven to be sufficient to supply up to 4 propellers at low speed for diagnostic purposes as well as at high speed (7 cm/s) for moving the capsule to target areas. As expected, the main problem was the high current peak required as the motor engages. To overcome this problem, a strategy based on the simultaneous actuation of two motors was successfully applied. Both the steering capability on the surface (2D locomotion) and the swimming from the bottom of the jar to the liquid surface were tested. In both cases, two motors active at a time allowed to achieve quasi-straight trajectories, thus enabling a sufficient control to overcome obstacles and invert the advancing direction. At a duty-cycle of 100%, the capsule speed is about 20 cm/s, which is too high to allow a good control in the stomach. A more usable duty-cycle lies between 15 and 25% providing a speed of 2–4 cm/s. In this case the power consumption of the locomotion module is about 130 mW using 2 motors. Supplementary Fig. 6 depicts the testing set-up and snapshots of dry and wet environment experiments. The video-clip (cfr. electronic version) resumes the three phases of the testing activities performed with the integrated device.

4. Conclusions

Inductive powering boosts the amount of available power in miniaturized mobile devices by one order of magnitude, and overcomes the problem of finite battery lifetime. This enables the integration of active locomotion modules required to actively steer endoscopic capsules during the exploration of the GI tract. The use of a ferrite frame in the receiving coil-set improves the electromagnetic coupling and consequently the power transfer. This opens the possibility of reducing the external magnetic field or shrinking the volume of the coil-set for a given amount of required power. A wireless power module, based on three orthogonal coils wound on a ferrite-core, was integrated in a self-propelled capsule designed for operation in a liquid-distended stomach. Integration and functionality tests were successfully performed: the power module was able to simultaneously feed the telemetry control system and up to 4 motors connected to the propellers. The capsule was tested both in a dry and a wet set-up with satisfactory results. *Ex vivo* trials are scheduled in the near future.

In addition to the present application, an integrated power module with infinite lifetime in miniaturized devices can open unexpected possibilities in many applications, going well beyond capsular endoscopy.

Acknowledgements

The work is supported by the European Community, within the 6th Framework Programme, through the VECTOR project (contract number 0339970). The authors wish to thank the project partners. Special thanks to Mr. Michel De Cooman, Mr. Nicodemo Funaro and Mrs. Noella Gaethofs for their invaluable help on parts fabrication and assembly.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version at doi:10.1016/j.bios.2009.08.049.

References

Arena, A., Boulougoura, M., Chowdrey, H., Dario, P., Harendt, C., Irion, K.M., Kodogiannis, V., Lenaerts, B., Menciassi, A., Puers, R., Scherjon, C., Turgis, D., 2005. In: Bos, L., et al. (Eds.), Medical and Care Compunetics, vol. 2. IOS Press, pp. 167–174.
Blot, W.J., Devesa, S.S., Kneller, R.W., Fraumeni, J.F., 1991. JAMA 265, 1287–1289.
Buchman, L., Miller, F.H., Wallin, A., Chowdhry, A.A., Ahn, C., 2004. Am. J. Gastroenterol. 99, 2171.
Carta, R., Lenaerts, B., Thoné, J., Tortora, G., Valdastri, P., Menciassi, A., Puers, R., Dario, P., 2008. Proceedings of Eurosensors XXII, Dresden, Germany, pp. 1369–1372.
Catrysse, M., Hermans, B., Puers, R., 2004. Sens. Actuators A: Phys. 115, 221–229.
Cavallotti, C., Focacci, F., Piccigallo, M., Susilo, E., Valdastri, P., Menciassi, A., Dario, P., 2008. Proceedings of Eurosensors XXII, Dresden, Germany, pp. 1366–1368.
Eliakim, R., 2004. Israel Med. Assoc. J. 6, 560–561.
Fiorini, P., Doms, I., Van Hoof, C., Vullers, R., 2008. Proceedings of ESSCIRC 2008, Athens, Greece, pp. 4–9.
International Commission on Non-Ionizing Radiation Protection (ICNIRP), 1998. Health Phys. 74 (4), 494–522.
Iddan, G., Meron, G., Glukhovsky, A., Swain, P., 2000. Nature 405, 417.
Lenaerts, B., 2008. Inductive links for omnidirectional wireless powering. Ph.D. Thesis. K.U. Leuven, Belgium.
Lenaerts, B., Puers, R., 2005. Sens. Actuators A: Phys. 123–124, 522–530.
Lenaerts, B., Puers, R., 2007. Biosens. Bioelectron. 22, 1390–1395.
Leuerer, T., Mokwa, W., 2004. Sens. Actuators A: Phys. 116, 410–416.
Mc Caffrey, C., Chevalerias, O., O'Mathuna, C., Two, K., 2008. Pervasive Comput. 7, 23–29.
Miura, H., Arai, S., Kakubari, Y., Sato, F., Matsuki, H., Sato, T., 2006. IEEE Trans. Magn. 42, 3578–3580.
Moglia, A., Menciassi, A., Schurr, M., Dario, P., 2007. Biomed. Microdevices 9, 235–243.
Moglia, A., Menciassi, A., Dario, P., Cuschieri, A., 2009. Nat. Rev. Gastroenterol. Hepatol. 6, 353–362.
Quirini, M., Scapellato, S., Valdastri, P., Menciassi, A., Dario, P., 2007a. Proceedings of IEEE EMBC 2007, Lyon, France, pp. 2827–2830.
Quirini, M., Webster III, R.J., Menciassi, A., Dario, P., 2007b. Proceedings of Robotics and Automation 2007 IEEE International Conference, Rome, Italy, pp. 1856–1862.
Rieber, F., Tognoni, V., Cenci, L., di Lorenzo, N., Schurr, M., 2008. Capsule endoscopy of the entire GI tract. SMIT Annual Meeting, Session 4.1.
Schnakenberg, U., Walter, P., Vom Bögel, G., Krüger, C., Lüdtke-Handjery, H.C., Richter, H.A., Specht, W., Ruokonen, P., Mokwa, W., 2000. Sens. Actuators A: Phys. 85, 287–291.
Swain, P., 2008. World J. Gastroenterol. 14, 4142–4145.
Thoné, J., Turgis, D., Carta, R., Puers, R., 2008. Proceedings of Eurosensors XXII, Dresden, Germany, pp. 347–377.
Tortora, G., Valdastri, P., Susilo, E., Menciassi, A., Dario, P., Rieber, F., Schurr, M.O., 2009. Minimally Invasive Therapies and Allied Technologies (MITAT) 18 (5), 280–290.
Valdastri, P., Menciassi, A., Dario, P., 2008. IEEE Trans. Biomed. Eng. 55, 1705–1710.
Van Ham, J., Reynders-Frederix, P., Puers, R., 2007. Proceedings of Transducers07, Lyon, France, pp. 427–430.
Vandevoorde, G., Puers, R., 2001. Sens. Actuators A: Phys. 92, 305–311.
Wang, H., Antonioli, D., Goldman, H., 1986. Hum. Pathol. 17, 482–487.
Yuan, Q., Zhao, Y.-P., 2009. J. Am. Chem. Soc. 131 (18), 6374–6376.