A mm-Sized Wirelessly Powered and Remotely Controlled Locomotive Implant

Daniel Pivonka, Anatoly Yakovlev, Ada Poon, Senior Member, IEEE, and Teresa Meng, Fellow, IEEE

Abstract—A wirelessly powered and controlled implantable device capable of locomotion in a fluid medium is presented. Two scalable low-power propulsion methods are described that achieve roughly an order of magnitude better performance than existing methods in terms of thrust conversion efficiency. The wireless prototype occupies 0.6mm × 1mm in 65 nm CMOS with an external 2mm × 2mm receive antenna. The IC consists of a matching network, a rectifier, a bandgap reference, a regulator, a demodulator, a digital controller, and high-current drivers that interface directly with the propulsion system. It receives 500 µW from a 2 W 1.86 GHz power signal at a distance of 5 cm. Asynchronous pulse-width modulation on the carrier allows for data rates from 2.5–25 Mbps with energy efficiency of 0.5 pJ/b at 10 Mbps. The received data configures the propulsion system drivers, which are capable of driving up to 2 mA at 0.2 V and can achieve speed of 0.53 cm/sec in a 0.06 T magnetic field.

Index Terms—Biomedical telemetry, drug delivery, implantable biomedical devices, low power, micro-scale fluid propulsion, noninvasive, wireless health monitoring, wireless powering.

I. INTRODUCTION

Implantable devices capable of in vivo controlled motion can serve a variety of existing applications, and they also open up new possibilities for emerging noninvasive medical technologies. Drug delivery is an especially attractive application, as drugs can be precisely targeted to problematic regions with minimal disturbance to the rest of the body. Additionally, precision guidance through fluid cavities could enhance both endoscopic and cardiac procedures that currently rely on catheter systems, such as angioplasty and coronary stent treatments, cardiac arrhythmia ablation surgeries, and diagnostic techniques like endomyocardial biopsies. Heart disease is the leading cause of death, and this technology could improve the effectiveness of these procedures as well as reducing costs. With these enhancements and cost reductions, it is possible to develop new noninvasive procedures for cardiac care or endoscopy that can aid in prevention, early detection, and treatment of a variety of conditions.

The key challenge in designing an effective locomotive implant is the limited power budget, as existing fluid propulsion methods have significant power requirements and batteries hinder the potential for miniaturization. Existing wireless devices rely on batteries and have very limited or no motion control. These devices are not small enough to travel through the circulatory system, and are most widely used in the gastrointestinal tract for endoscopy [1]. There are many proposed systems being researched that attempt to enhance motion either with mechanical actuation or by manipulating passive magnetic structures with external fields [2], [3]. Mechanical techniques tend to require significant power, have complex designs and moving parts, and have very low thrust efficiency as they are scaled down. Devices relying on passive magnetic fields require complex external equipment have limited controllability in small regions (typically a few cm²), and move very slowly as they are scaled down [4]. Because of the challenges in developing a highly efficient and scalable propulsion method, wireless locomotive devices have not been possible.

An alternative to both the mechanical and passive magnetic propulsion techniques can be accomplished with the manipulation of Lorentz forces on current-carrying wires. These forces require a simple static magnetic field, and are power efficient and controllable. In our prior work, we developed two mechanisms for generating propulsion with these forces [5], [6]. The first is based on magnetohydrodynamics (MHD), in which current is driven directly through a conductive fluid. This current flowing through the fluid experiences a force in the magnetic field, and the device experiences an equal and opposite force that propels it forward. The second method relies on asymmetries in fluid drag forces experienced by an oscillating structure. The oscillations are achieved with alternating currents flowing through a loop of wire, which experiences torques in the magnetic field. With loops in different orientations, the structure can be oscillated to create “swimming” motions similar to fins on a fish. Both of these methods have significant potential for operating efficiently at very small sizes, and can enable fully wireless locomotive implants.

In this paper, we will present a fully wireless device capable of controlled motion in water. An external transmitter powers the device wirelessly with data modulated on the power carrier. Propulsion is achieved via the described methods based on the manipulation of Lorentz forces, and the prototype operates with either method. Fig. 1 shows the conceptual operation of the prototype travelling through the bloodstream with MHD propulsion. The constructed device is comprised of a 2mm × 2mm receive antenna and an integrated circuit that includes a matching network, a rectifier, a regulator, a demodulator, a digital controller, and high-current drivers that interface with the propulsion system. This prototype can travel through any fluid and can potentially be navigated through.
the circulatory system, enabling a variety of new medical procedures.

The organization of the paper is as follows. Section II presents the analysis and simulation of the fluid propulsion methods based on Lorentz forces. Section III describes the design of the wireless power transmission system as well as the data receiving architecture. The circuit implementation is presented in section IV, and section V discusses the experimental results and summarizes performance. Section VI then concludes the paper.

II. ELECTROMAGNETIC PROPULSION

Propulsion for implantable devices has not been possible because of the high power requirement for mechanical designs, and the high complexity of passive magnetic designs. Our prior work based on Lorentz forces demonstrates two methods with significant advantages over existing techniques in terms of power efficiency, scalability, and controllability [5], [6]. The first method drives current directly through the fluid using magnetohydrodynamics (MHD), and the second switches current in a loop of wire to oscillate the device, which experiences asymmetric drag fluid forces. In both methods, the force is proportional to current, and therefore maximizing current will maximize the speed.

The thrust forces work against fluid drag forces, which are velocity dependent. This dependence varies with the Reynolds number of the fluid flow. The Reynolds number is a dimensionless representation of the ratio of the inertial forces to the viscous forces, and is given by

\[ Re = \frac{\rho v^2 D}{\mu} \]  

where \( \rho \) is the density of the fluid, \( v \) is the velocity, \( D \) is a characteristic dimension, and \( \mu \) is the fluid viscosity. For high Reynolds numbers (>1000), the drag force is given as

\[ D = \frac{1}{2} \rho v^2 C_D A \]  

where \( A_f \) is the frontal area of the device, and \( C_D \) is the shape factor. These forces scale with area, and as will be shown, the thrust forces for both propulsion methods scale linearly with length. This means that in the high Reynolds regime, less current is needed to maintain a constant speed as the device is scaled. As the Reynolds number decreases, viscous forces become dominant. For extremely low Reynolds numbers (<1), the drag force scales linearly with the size of the device as predicted by Stokes Law. In the low Reynolds regime, the current must be kept constant as the device is scaled to maintain a constant speed. For mm-sized devices moving at cm/sec speeds in water, the Reynolds number ranges from roughly 10–100, so numerical fluid simulations are necessary for an accurate analysis of the fluid drag forces.

A. Magnetohydrodynamic (MHD) Propulsion

MHD propulsion drives electric currents through fluids, so the efficiency of this method depends on the fluid conductivity. The basic principle of motion is described in Fig. 2. The conductivity of human blood varies approximately from 0.2 S/m to 1.5 S/m depending on the concentration of blood cells [7]. This translates to a load of less than 300 \( \Omega \) at the device, which varies with the size, shape, and distance between the electrodes as well as the temperature and applied voltage. Stomach acids tend to have higher conductivities but also vary significantly with normal biological processes. In the following analysis, the required current for a given speed will be estimated as a function of the size of the device and the background magnetic field. This will give insight into the scalability of the propulsion method and also provide a design target for the circuitry.

The thrust force for MHD propulsion is the Lorentz force on the current flowing through the fluid. These forces are given in the equation below, where \( I \) is the current in the wire, \( \mathbf{L} \) is a vector denoting the length and direction of the wire, and \( \mathbf{B} \) is the background magnetic field:

\[ \mathbf{F} = I \mathbf{L} \times \mathbf{B} \]  

These forces scale linearly with the length of the wire \( \mathbf{L} \), which allows for the operation of very small devices. It scales more slowly than high Reynolds drag forces, which means that for smaller devices constant current scaling results in higher speeds; and it scales evenly with low Reynolds drag forces, which means that constant current scaling results in a constant speed. Additionally, the amount of force is linearly proportional to the background magnetic field, so the performance of this method improves with stronger magnetic fields. To accurately estimate the speed, numerical simulations
of the fluid mechanics are performed. Fluid simulations based on incompressible Navier-Stokes flows predict the fluid drag forces, and from these forces the steady-state velocity can be extracted. In Fig. 3, the required current is estimated for a given speed as a function of the size of the device with a background magnetic field of 0.1 T, which can be generated with permanent magnets. This analysis shows that mm-sized devices should be able to achieve speeds on the order of cm/sec with approximately 1 mA of current.

The amount of current that can be driven is a strong function of the fluid conductivity, and has significant nonlinear variations with electrode area, electrode materials, applied voltage, and the types of ions in the fluid. To drive 1 mA through blood (which has the lowest conductivity of the targeted fluids), roughly 300 mV is required, resulting in a power consumption of around 300 $\mu$W. As the fluid conductivity increases, the required power decreases. These power requirements are within the bounds of optimized wireless powering techniques through tissue, so miniaturized locomotive implantable devices are possible with this method.

**B. Asymmetric Fluid Drag Propulsion**

The second fluid propulsion method relies on asymmetries in fluid drag created by an oscillating asymmetric structure. The structure is oscillated by alternating currents in a loop of wire that is placed in a background magnetic field. The basic principle of operation is described in Fig. 4. The forces generated with this method are a function of the fluid viscosity, which for most bodily fluids is similar to water. The performance of this method is enhanced as the number of loops is increased, and the amount of current that can be driven is limited by the internal resistance of the circuitry and the amount of power delivered through the antenna. The following analysis estimates the required current as a function of the size of the device and the desired speed. This analysis predicts the device scalability and also specifies the requirements on the circuitry.

The thrust forces result from asymmetric fluid drag on a structure that oscillates with electromagnetic torque of

$$\tau_{em} = IL^2B$$

(4)

where $I$ is the current on the loop, $L$ is the length of the wire, and $B$ is the background magnetic field. The asymmetry in the fluid drag is represented by the shape factor, $C_D$. By integrating the fluid drag along one side of the device, the net force can be represented as

$$F_p \propto (C_{D,H} - C_{D,L})L^4\omega^2$$

(5)

where $C_{D,H}$ and $C_{D,L}$ represent the different shape factors due to the asymmetry, $L$ is a side length of device, and $\omega$ is the rotation frequency. Assuming small angle rotations and constant angular acceleration, which is true when the electromagnetic torque dominates the fluid drag torque, the average angular velocity over a half-cycle is

$$\omega_{avg} = \sqrt{\theta \tau_{em}/4I_{int}}$$

(6)

where $\theta$ is the angle of rotation and $I_{int}$ is the moment of inertia. Realizing that $\tau_{em} \propto L^2$ and $I_{int} \propto L^5$, constant current scaling results in the average angular velocity scaling as $\omega \propto L^{-3/2}$. Using this result in the equation for the net force, we again find that these thrust forces scale linearly with $L$. This method scales in the same way as MHD propulsion and allows for the operation of very small devices. As the Reynolds number decreases, the fluid drag becomes much more shape dependent, which complicates analytical analysis. For accurate estimations of the forces on these devices, we again rely on numerical simulations of the fluid mechanics.
For this propulsion method, the simulations predict both the average fluid drag torque and the average net force over a cycle as a function of the rotation frequency and the size of the device. The fluid drag torque and the average force are shown in Fig. 5. These simulations agree with the predicted scaling behavior in terms of size and rotation frequency. From the fluid drag torque simulation, the current required to achieve a given rotation frequency can be estimated. The simulated net forces can then predict the speed, which relates to the current shown in Fig. 6. From these simulated results, mm-sized devices with a single loop of wire require currents of approximately 1 mA to achieve cm/sec speeds in water with a 0.1 T magnetic field. Additional loops of wire enhance the performance, essentially multiplying the current experiencing a force.

III. WIRELESS CHIP ARCHITECTURE

The purpose of the chip was to create a wireless prototype that demonstrates the effectiveness of the propulsion system at the mm-scale. The specifications were derived from the requirements of the propulsion methods, which need approximately 1 mA of current for cm/sec speeds. The integrated circuit (IC) must receive both power and data from the external receiver to propel and navigate the device, and must operate with a limited power budget. The chip architecture is shown in Fig. 7, and the IC consists of a matching network, a charge-pump connected rectifier, a regulator, a bandgap reference circuit, a demodulator, a digital controller, and configurable electrode drivers. There are no external components except for the receiving antenna. The key challenge in this design is driving the high-current propulsion system efficiently and controllably while continuously harvesting RF energy. Power is the primary limitation, and minimizing power consumption was critical for the design.

The non-linear electrode-fluid resistance limits the minimum voltage required to drive the current, and is estimated at approximately 200–300 mV. The propulsion system dominates the power budget consuming over 90% of the total delivered power to the chip. The required 1 mA of current for propulsion...
needs to be sourced from no more than 300 mV while the active circuitry requires a regulated voltage of 700 mV and draws approximately 15 μA. Using a linear regulator for the propulsion system is inefficient, and a switching regulator requires large passive components, accurate on-chip clock, and complex controllers [8]. Therefore, the chip was designed to drive the propulsion system from the first rectifier stage, which provides an unregulated 200–300 mV supply depending on the received power and can source the required current. Because the loading from the propulsion system varies with navigation, an adaptive loading network is also necessary to maintain effective matching at the antenna. The first rectifier stage is followed by three additional stages to boost the voltage, which is then regulated for the analog and digital circuits.

The size requirements prohibit the use of external energy storage components, so power must be continuously transmitted to the device. Power transmission must adhere to FDA safety regulations for tissue heating. From prior work, mm-sized antennas can receive approximately 200–300 μW at low-GHz frequencies safely [9]. A 2mm×2mm antenna provides sufficient power for this design, and performing a frequency sweep with the antennas yields an optimal frequency of 1.86GHz. It is important that the modulation scheme minimally affects the power transfer to the device because of the limited power budget. Frequency-shift keying (FSK) and phase-shift keying (PSK) operate with a constant envelope, but the demodulator requires either a frequency or phase-locked loop for carrier synchronization, which consumes significant power at high frequency [10]. Amplitude modulation does not require carrier synchronization, and the modulation depth and duty cycle can be designed to minimize the impact on power delivery. For this reason, we implemented amplitude shift keying (ASK) with low modulation depth (minimum of 9%), and the pulse width (PW) encodes the data allowing for asynchronous clock and data recovery with simple circuitry.

A high-level description of the data receiver is shown in Fig. 8. The demodulator provides both the clock signal for the digital controller and decodes incoming data. The demodulator interface with the matching network uses two rectifiers: the first has a small time constant and tracks the envelope, and the second has a large time constant and approximates the average of the envelope. These two signals are input to a comparator to generate the digital signal V_{out1}. This signal is buffered to produce a digital clock. V_{out1} is also integrated and compared to a threshold to decode the data. With this implementation, long pulses produce high output and short pulses produce low output. The demodulated data is captured on the falling edge of the clock by a low-power digital controller, which configures the high-current electrodes for driving the propulsion system.

IV. CIRCUIT IMPLEMENTATION

A. Antenna and Matching Network

The antenna dominates the size of the prototype, and is implemented with a 2mm×2mm loop on a PCB using Rogers 4350 substrate. External components are not possible due to size constraints, so a balanced L-match consisting of only capacitors was implemented because on-chip inductors have significant loss and occupy large area [11]. The total quality factor of the antenna and the matching network in air is estimated at 39. The chip input impedance is dominated by the propulsion system, and loading varies significantly during normal operation. Therefore, an adaptive loading network was implemented to maintain an effective match. When the chip is powered on and before the controller is reset, the gate of transistor M_{n4} in Fig. 9 is weakly pulled up by V_{unreg}, which shunts the first rectifier stage with an internal 200 Ω resistor. After the digital supply is enabled, the weak pull-down transistor M_{n3} slowly turns off the shunt resistor. Once the power-on reset (POR) signal has been issued, the digital controller is reset and takes control of the network, adjusting the resistance based on incoming data.

B. Start-up and Power-on Reset Circuits

Start-up circuitry for the initial power-on is necessary to ensure that the antenna impedance maintains a match and that the chip enters a known state. A start-up network that turns on a pass transistor for the digital supply voltage is shown in Fig. 9. The cross-coupled inverters are skewed in opposite directions to prevent metastability, and the delay is controlled.
by a capacitor at the supply of the cross-coupled inverters that slowly charges through a weak current source. This delay ensures that the analog supply voltage has reached a stable 700 mV before powering on the digital circuits. Once the digital supply is enabled, a POR pulse is issued after an additional delay. This pulse generation is shown in Fig. 10, and is similar to the design used in [12]. The pulse width is set by the delay of a capacitively loaded inverter chain that provides a sufficient duration pulse to reset the controller.

C. Power Management

When the antenna receives 500 µW, the RF input voltage to the rectifier is 350 mV. Conventional diode-capacitor ladder rectifiers suffer from low efficiency at low input voltage. Therefore, charge-pump connected self-driven synchronous rectifiers (SDSR) based on [13] are used with low-Vt devices. The first stage of the rectifier is sized 10 times larger than the consecutive stages because the propulsion system is driven directly from this first stage. It outputs an unregulated 200–300 mV and drives roughly 1 mA of current. The remaining three stages are all sized the same and output 0.9–1.2 V while driving 15 µA. The pump capacitance between these three stages is 5 pF. The simulated efficiency of the rectifier is approximately 55%.

The unregulated supply voltage fluctuates significantly with variations in available power due to varying link gain as the device moves, propulsion driver strength, and switching noise from the digital circuits. The device must also be insensitive to temperature variations. To create a stable 700 mV supply for the active circuitry, we implemented a low drop-out voltage regulator that relies on a bandgap reference circuit similar to [14]. A total of 86 pF of smoothing capacitance was used to maintain stable voltage at the supply. The schematic of the regulator is shown in Fig. 11. The regulated voltage is sampled via a resistive voltage divider and is compared to the bandgap reference output voltage of 525 mV. The resistive divider also outputs a voltage of $V_{dd}/2$, providing a reference for the demodulator. Capacitor C1 is added to help stabilize the feedback loop. The regulator has an overall efficiency of 58%. However, the dissipated power due to the rectifier inefficiency is only

$$\eta_{\text{degradation}} = \frac{P_{\text{lost}}}{P_{\text{total}}} = \frac{(V_{\text{unreg}} - V_{\text{reg}})I_{\text{reg}}}{P_{\text{propulsion}} + P_{\text{circuits}}} = 3\% \quad (7)$$

of the total power consumption because the unregulated propulsion system dominates power usage.

D. Clock and Data Recovery

The low modulation depth and fluctuating input power make it impossible to use a fixed reference voltage for the ASK threshold detector. Instead, a dynamic reference voltage is generated concurrently with envelope detection. The schematic of the envelope detector and dynamic reference generator is shown in Fig. 12. Both circuits use cross-coupled PMOS transistors to achieve full-wave rectification. The envelope detector RC time constant filters out the carrier and passes the data. In the dynamic reference generator, the RF input voltage is resistively divided to weakly turn on the cross-coupled transistors. The higher on-resistance and larger load capacitance form a large RC time constant, which effectively averages the envelope. The resistor at the output of envelope detector aligns the average of the envelope with the dynamically generated reference voltage.

Clock and data signals are recovered from the envelope and the dynamic reference, which are first input to a comparator to generate the full-swing digital signal $V_{\text{out1}}$. This comparator consists of two differential amplifier stages followed by a Schmitt-trigger inverter as shown in Fig. 13. Two low-power differential amplifiers ensure that the gain remains high for a
Fig. 12. Envelope detection and dynamic reference voltage generation circuits.

Fig. 13. First comparator that converts the envelope into digital signal.

Fig. 14. Integrator and second comparator for data decoding.

E. Controller

The digital controller receives data and clock signals from the demodulator, and configures the propulsion system drivers and the adaptive loading network. Data transmission begins with a 5-bit prefix that, when received, enables a shift register to begin accepting the 55-bit data packet. While data is being shifted into the register, the prefix detection circuitry is disabled. Once the entire packet is received, the shift register pushes all the data to a memory register that stores it until the next valid transmission. By only enabling the necessary circuitry in each stage of data reception, power consumption is minimized. Because the clock is derived from the data signal, when no data is being received the only current drawn is due to leakage. The estimated average power consumption of the digital controller while receiving data is 2 µW, and it occupies 0.009 m².

F. Configurable High-Current Drivers

The chip has 6 high-current electrode drivers with configurable strength to accommodate both propulsion mechanisms. Each of the drivers can be independently set to V_{propulsion} from the first rectifier, ground, or left floating. Additionally, the driver strength can be controlled with 4 parallel transistors, and ranges from 20–1000 Ω. This configurability is necessary to adapt to uncertainty in electrode-fluid resistance and to enable speed and steering control. Data in the memory register directly controls the electrode driver state and strength.

V. EXPERIMENTAL VERIFICATION

Experimental tests verified all the elements of the design including wireless power transmission, the ASK-PWM data transfer, the analog and digital circuitry, and the two propulsion schemes. Independent tests evaluated the wireless link and the circuit performance, and testing of the complete system demonstrated navigation and propulsion through fluids. Each experiment will be described in detail in this section. The overall circuit performance is summarized in Table I, and the annotated chip micrograph is shown in Fig. 20.

A. Wireless Power Transmission

The transmitter consists of a signal generator, a high-frequency amplitude modulator, a power amplifier, and a 4cm×4cm loop antenna fabricated on PCB. The IC was wire bonded to a 2mm×2mm antenna fabricated on a Rogers 4350 substrate to minimize RF losses. A frequency sweep of the link gain was tested at a separation distance of 5 cm both in air and with the device placed on the surface of water. The measurements are shown in Fig. 15. From this plot, the quality factor in air is 39 for the antenna including the matching network. The rectified output voltage and the regulated voltage are plotted as a function of input power in Fig. 16, showing that the device first powers on with roughly –7 dBm. With a rectifier efficiency of approximately 55%, roughly 2 W must be transmitted to receive 500 µW, resulting in approximately 250 µW of usable power after rectification.
B. ASK-PWM Data Transfer

Data modulation was designed to minimize impact on power delivery with a low power circuit implementation. To accomplish this, an asynchronous design was implemented that operates with minimal modulation depth and without carrier synchronization circuitry. This method allows for variable data rates and modulation depths. In order to test the range of operation, a versatile high-frequency modulator was constructed. The data signal was generated from an FPGA and input to the modulator, which modulates the output from the signal generator at an adjustable depth from 0–100%. The FPGA was able to stream data at up to 25 Mbps, and the chip properly received data from 2.5–25 Mbps. Additionally, the chip functioned with as low as 9% modulation depth. The spectrum of the carrier modulated at 9% with an 8.3 MHz clock is shown in Fig. 17, and the received clock and data signals on chip are shown in Fig. 18. The power consumption of the demodulating circuitry is approximately 5 µW at 10 Mbps, resulting in energy efficiency of 0.5 pJ/bit.

C. Fluid Propulsion

The IC was designed to function with either of the described fluid propulsion mechanisms. The chip and receive antenna are encapsulated in RF-transparent epoxy to protect them from the fluid. The leads from the electrodes are exposed to adapt the device for use with either of the fluid propulsion methods. For MHD propulsion, these leads are positioned to directly connect to a conductive fluid, and salt water was used for testing. For the method relying on asymmetric fluid drag forces, the electrodes are connected to loops of wire that oscillate the device. In both test cases, the device floats on the surface of the water with a neodymium magnet placed next to the fluid to provide a magnetic field. Even though testing was performed on floating devices, both propulsion methods can function when fully submerged.

The experimental setup for MHD propulsion is shown in Fig. 19. During propulsion testing, the external antenna tracked the device at a distance ranging from 2 to 5 cm. Data is continuously transmitted with commands to control the motion. The device achieves speeds of up to 0.53 cm/sec in a 0.06 T field with approximately 1 mA, and can be navigated successfully along the surface of the water. Performance improves as the magnetic field is increased, so MRI systems will generate approximately 100 times as much propulsion force.

The setup for asymmetric fluid drag propulsion is very similar to MHD propulsion. The device is connected to 40 loops of wire, which are oriented to oscillate it. The prototype has an attached fin that experiences asymmetric fluid drag when oscillating. By changing the orientation of the magnetic field, the device can oscillate along the surface of the water, or into and out of the water. The external antenna is again placed
above the device and continuously transmits data. The forces on the device are much stronger for this method because the additional loops and smaller load; however propulsion is much more difficult to control. This method is also more sensitive to non-uniformities in the magnetic field. Additionally, the antenna link degrades as the device rotates, causing frequent errors in data reception. For this method to operate effectively, a new antenna link and a feedback controller are necessary.

VI. CONCLUSION

In this work, we have demonstrated a fully wireless 3mm×4mm prototype capable of controlled motion in a fluid environment, requiring only a static background magnetic field generated from permanent magnets. The device is wirelessly powered and operates with approximately 250 µW, and travels controllably at 0.53 cm/sec in a 0.06 T field. Additionally, data transfer is fast and efficient, achieving rates of 25 Mbps and consuming only 0.5 pJ/bit at 10 Mbps. These devices can serve as a versatile tool for a variety of medical treatments that require precise guidance including drug delivery, diagnostics, and cardiac catheter treatments.

Two propulsion methods have been demonstrated, and both can operate at sub-mm sizes. MHD propulsion is simple to control with high efficiency, though it requires a conductive fluid. The second method relies on asymmetric drag forces experienced by an oscillating structure. This method has higher forces and can function in any fluid, but is more sensitive to field non-uniformities and is more difficult to control. Future work can improve the robustness and controllability of these methods, and can further optimize wireless power transfer to desensitize it to the motion of the device.

REFERENCES


Daniel Pivonka (S’09) received his B.S. in engineering from Harvey Mudd College in 2007, his M.S. in electrical engineering from Stanford University in 2009, is currently working at Stanford University toward his Ph.D. degree in electrical engineering. He interned at the Beckman Laser Institute in 2005, at the Southwest Research Institute in 2006, and at ViaSat in 2007. His current research interests is in wireless implantable devices capable of locomotion through fluid cavities.

Daniel Pivonka is a recipient of the Stanford Graduate Fellowship and the Clay/Wolkin Fellowship.

Anatoly Yakovlev (S’04) received his B.S. degree in electrical engineering from Cal Poly Pomona in 2007, where he graduated valedictorian of the School of Engineering. In 2009, he received his M.S. degree in electrical engineering with emphasis in analog/RF circuits from Stanford University and is now continuing to pursue his Ph.D. degree. His Ph.D. work focuses on wireless powering and data communication for biomedical devices.

Between 2007 and 2008 he was with the Advanced Channel Architecture team at Western Digital. In summer 2011, he was with VLSI Group at Oracle Labs working on clock and data recovery circuits for optical interconnects.

Ada Poon Ada S. Y. Poon (S’98–M’04–SM’10) was born in Hong Kong. She received the B.Eng and M.Phil. degrees in Electrical and Electronic Engineering from the University of Hong Kong, and received the M.S. and Ph.D. degrees in Electrical Engineering and Computer Sciences from the University of California at Berkeley. In 2004, she was a senior research scientist at Intel Corporation, Santa Clara, CA. In 2005, she was a senior technical fellow at SiBeam, Inc., Fremont, CA. In 2006–2007, she was an assistant professor at the Department of Electrical and Computer Engineering in the University of Illinois at Urbana-Champaign. Since 2008, she has been at the Department of Electrical Engineering in Stanford University, where she is currently an assistant professor. Her research focuses on applications of wireless communication and integrated circuit technologies to biomedical and health care.

Teresa Meng (S’82-M’83-SM’93-F’99) received the Ph.D. degree in electrical engineering and computer science from the University of California, Berkeley, in 1988.

She is the Reid Weaver Dennis Professor of Electrical Engineering at Stanford University, Stanford, CA. Her current research interests focus on neural signal processing and bioimplant technologies. In 1999, she left Stanford University and founded Atheros Communications (NASDAQ: ATHR), which is a leading developer of semiconductor system solutions for wireless communications products. She returned to Stanford University in 2000 to continue her research and teaching.

Dr. Meng received the 2009 IEEE Donald O. Pederson Award, the DEMO Lifetime Achievement Award, the McKnight Technological Innovations in Neurosciences Award in 2007, the Distinguished Lecturer Award from the IEEE Signal Processing Society in 2004, the Bosch Faculty Scholar Award in 2003, the Innovator of the Year Award by MIT Sloan School of Management in 2002, and the CIO 20/20 Vision Award, a Best Paper Award from the IEEE Signal Processing Society, a National Science Foundation Presidential Young Investigator Award, an ONR Young Investigator Award, and an IBM Faculty Development Award, all in 1989. In 2002, she was named one of the Top 10 Entrepreneurs by Red Herring for 2001. Dr. Meng is a member of the National Academy of Engineering.