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Novel technology for the provision of power to implantable physiological devices

David M. Budgett,¹ Aiguo Patrick Hu,² Ping Si,² Wayne T. Pallas,¹ Mark G. Donnelly,¹ Jared W. T. Broad,¹ Carolyn J. Barrett,³ Sarah-Jane Guild,³ and Simon C. Malpas^{1,3}

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Budgett DM, Hu AP, Si P, Pallas WT, Donnelly MG, Broad JW, Barrett CJ, Guild S-J, Malpas SC. Novel technology for the provision of power to implantable physiological devices. *J Appl Physiol* 102: 1658–1663, 2007. First published January 11, 2007; doi:10.1152/jappphysiol.00105.2006.—We report the development of a novel technology that enables the wireless transmission of sufficient amounts of power to implantable physiological devices. The system involves a primary unit generating the magnetic field and a secondary pickup unit deriving power from the magnetic field and a power conditioner. The inductively coupled system was able to supply a minimum of 20 mW at all locations and pickup orientations across a rat cage, although much higher power of up to 10 W could be achieved. We hypothesized that it would be possible to use this technology to record a high-fidelity ECG signal in a conscious rat. A device was constructed in which power was utilized to recharge a battery contained within a telemetry device recording ECG signal sampled at 2,000 Hz in conscious rats (200–350 g) living in their home cage. Attributes of the ECG signal (QT, QRS, and PR interval) could be obtained with a high degree of accuracy (<1 ms). ECG and heart rate changes in response to treatment with the beta blocker propranolol and the proarrhythmic alkaloid aconitine were measured. Transmitters were implanted for up to 4 mo, and the characteristic circadian variation in heart rate was recorded. Such technology allows potentially lifetime monitoring without the need for implant refurbishment. The ability to provide suitable power levels to implanted devices without concern to the orientation of the device and without causing heating provides the basis for the development of new devices to record or influence physiological signals in animals or humans over significantly longer time periods than can currently be accommodated.

telemetry; inductive power; battery; biopotential; electrocardiogram

THE DEVELOPMENT of telemetry-based implantable circuits has revolutionized the ability to record physiological parameters under more natural conditions and for longer periods (7, 15). Technology developments continue to reduce the size of electronic circuits. However, despite some improvements in the energy density of batteries, supplying power to implantable devices has become the major limiting factor. While implantable physiological devices for use in animals have been developed that inductively supply power to the device and do not use batteries (2) (e.g., those produced by MiniMitter, Bend, OR), the amount of power that can be provided to the implant has been extremely small. This has meant that this approach has been suitable for measuring only very-low-bandwidth signals such as temperature and locomotor activity. While larger amounts of power can be transmitted (up to 25 W) and have

been implemented for powering artificial heart systems (12, 13) (referred to as transcutaneous energy transfer), the major impediment for the use of this approach as a power supply in animals is that the distance between the external supply and the internal power receiver must generally be very small (1–3 cm) (11, 12), and the alignment between the supply and receiver must be tightly controlled. A further limitation of transmitting large amounts of power is that poor efficiency leads to heat generation, a major concern for any implantable device. With regard to monitoring in animals, the mobility of a small animal, such as a rat, living within a cage means that inductive power supply systems have not proved suitable. The final issue has been that the size of the implanted power receiver is considerably larger than that which could reasonably be used in a rat (13). The above restrictions mean that for high-bandwidth signals requiring significant power, such as ECG, all currently available telemetry-based biopotential amplifiers utilize a battery that eventually requires replacement (e.g., those produced by Data Sciences, St Paul, MN). While these devices provide power for a considerable length of time, the ability to make recordings is limited by the lifetime of the battery.

The ability to provide power is also a limiting factor in the development of new implantable devices in animals and humans. This is particularly significant when the physiological signal and associated monitoring device have such a high current drain that the useable battery life prohibits its application. Examples include monitoring very-high-bandwidth signals such as sympathetic nerve activity (1), blood flow measurement by ultrasound (16), and the operation of implanted motors (10). The successful implementation of new technology to power such devices has the potential for opening new areas for research and for allowing the development of new clinical diagnostic or treatment instrumentation.

We hypothesized that it would be possible to adapt inductive power-transfer technology used currently for industrial applications to record via telemetry a high-fidelity ECG signal in a conscious freely moving rat and demonstrate the ability to power implantable physiological devices by induction.

METHODS

Principle. The principle underlying the transmission of power across an air gap is referred to as inductively coupled power transfer (ICPT). The fundamental feature of ICPT is that it allows power to be transmitted to a mobile unit across an air gap, thus removing the need for wires. An ICPT power supply system can be regarded as some-

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where between an electrical transformer and a transmitter and receiver configuration as in a radio system. The key issue involved in this technique is the power conversion from a low-frequency system, such as direct current or 50- to 60-Hz main power supply, to a much higher-frequency system with a frequency between 10 and 200 kHz. The rapidly changing rate of the magnetic field induces a strong induction effect. This power conversion makes power flow across an air gap feasible in practice. An ICPT system comprises two electrically isolated parts (Fig. 1). The primary system generates the magnetic field and consists of a power converter and a primary conductive path that can be an elongated track or, in the case of the present application, a coil under a plastic rat cage. The main function of the power converter is to supply a constant high-frequency alternating current along the path. The secondary system derives power from the magnetic field and consists of a secondary pickup unit and a power conditioner. Power from the secondary system is used to operate the other electronic circuits (which may include a battery charger) of the implantable device. Because of the mutual magnetic coupling between the primary track loop and the secondary pickup unit, an inductive electromotive force is induced in the pickup unit that forms a voltage source. Since the magnetic coupling is loose, i.e. low, compared with normal transformers, the induced voltage is usually very weak and is unsuitable to be used directly, but a power conditioner can boost and regulate the power into the form required by the implantable device. With the development of power semiconductor switching devices and power conversion techniques, the transfer of larger amounts of power (up to 60 W) across an air gap is practical.

Problems to overcome. With regard to the development of an ICPT system suitable for providing power to an implantable device, there are several unique problems to overcome. In industrial applications and some human applications (4, 12, 13), the distance and orientation between the primary field and the secondary pickup unit can be precisely defined and regulated. This is not the case for an implant in a conscious animal where the orientation between primary track and secondary pickup can be highly variable. Second, the distance between the primary field and the implant will vary as the animal moves around its cage. Third, in order for the application of the technology to be as wide as possible, the size of the secondary power pickup unit must be suitable to be implanted in small animals such as a rat. We have developed a solution to address these issues.

Electronic design. A key part of the primary system is a resonant tank made up of the conductive path or coil and a capacitive element. Energy is injected into the resonant tank (using soft switching techniques) to maintain resonance in the presence of losses and energy drawn from the magnetic field by the secondary system. The primary system produces a magnetic field with a sinusoidal amplitude distribution with very low distortion. This minimizes the losses and electromagnetic interference (EMI) that would be introduced if higher-order harmonics were present. The amplitude and the resonant frequency of the magnetic field are varied in response to the feedback received from the implant device using an inherent radio link.

The secondary system has stringent constraints, including the need to provide adequate power to other implant circuits, occupying minimal volume, and having good efficiency to minimize heating. The

frequency of the magnetic field is ~ 200 kHz. Although more complex to produce from a primary power electronics switching perspective, the high frequency permits physically smaller reactive elements in the secondary device. The secondary power converter also has the ability to adjust the resonance frequency of the pickup circuit. This is an effective way of detuning the circuit to regulate power flow to match the immediate need of the implant device (and hence minimize heat generation).

The voltage induced in the secondary pickup system is detected by the telemetry system, and this information is embedded into the digital data packets containing physiological data samples. The primary system receives these data and will adjust field strength or frequency to match the power delivery of the inductive link to the power needs of the implant.

For the application of monitoring rat ECG, a small button cell lithium-ion rechargeable battery was incorporated into the implant. The battery is recharged periodically using the inductive link as required. This arrangement allows many rats to be monitored without the need to provide primary systems for each cage. When an implant requires charging, the cage is placed near the primary unit for the duration of the charging process.

Animal testing. To test the hypothesis that it is possible to provide power to an implanted physiological device, we recorded ECG signals in conscious rats. A telemetry-based biopotential amplifier and transmitter was provided by Telemetry Research (www.telemetryresearch.com, Auckland, New Zealand). This unit incorporates an amplifier, 12-bit analog-to-digital converter sampling at 2 kHz, and transmitter (2.4-GHz band, range 5 m). The power consumption of the unit is ~ 15 mW. A 35-mA/h lithium-ion rechargeable button cell (diameter 20 mm, thickness 3.2 mm) provided continuous operation of the amplifier and transmitter. The implant included a secondary pickup, power electronics to regulate resonant frequency and power flow, power rectification and smoothing circuits, and battery-charging elements. The different subsystems were coordinated by an eight-bit 8051 microcontroller. The complete implant was encapsulated in medical-grade silicone and measured $35 \times 23 \times 11$ mm, with a weight of 13 g. The various subsystem components of the implanted device are shown in Fig. 2.

A receiver was responsible for reconstructing the transmitted data signal. The typical range was 5 m. A wireless power-charging unit was responsible for controlling the magnetic field generated by the primary unit. The charging unit also incorporated a receiver unit to decode the feedback information on field and battery status sent by the implant.

When the primary field was activated, the energy derived from the secondary pickup was used to recharge the lithium-ion battery using a constant current followed by a float voltage-charging method. Feedback of field strength and charge status was sent from the implant to the field control unit to regulate power flow and terminate charging when appropriate.

All animal procedures were approved by the University of Auckland Animal Ethics Committee. Wistar rats weighing between 200 and 350 g were anesthetized either using pentobarbital sodium (50 mg/kg ip) or with a combination of medetomidine (Dormitor), 0.5 mg/kg, and ketamine 75 mg/kg ip (Southern Veterinary Supplies, Hamilton, New Zealand). Rats also received an antibiotic (gentamycin sulfate 2.5 mg im) and an analgesic (butorphanol tartrate 0.5 mg sc) at the same time. Using sterile procedures, a midline incision was made through the skin and underlying abdominal muscle layer. The body of the implantable device was positioned within the abdominal cavity, and the muscle layer was sutured closed. The transmitter was not sutured to the abdominal muscle but allowed to move freely with the peritoneal cavity. ECG leads (constructed from 1-mm-OD stainless steel spring and inserted through silicon tubing with a 5-mm tip exposed) were tunneled subcutaneously. Two placement profiles were used to attempt to achieve the best-quality ECG during body movement. In five rats, one lead was positioned to lie 1 cm to the left of the

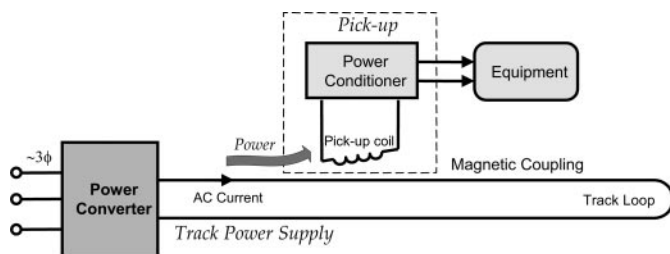


Fig. 1. Basic structure of a typical inductively coupled power transfer system. AC, alternating current.

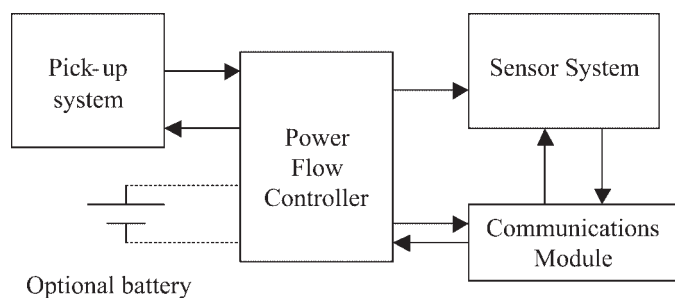


Fig. 2. The various subsystem components of the implanted device. The power flow controller is responsible for matching the power available from the pickup system to the power needs of the other elements (including recharging the battery if present). The communications module will typically multiplex data from the sensor system with data from the power flow controller. The communications module is bidirectional and can receive data for the purpose of altering the actions of the power flow controller.

xiphoid process, and the other lead was positioned on the right pectoral muscle. This placement equates most closely to lead II in the human. In another five rats, one lead was maneuvered on the xiphoid process and the other lead was maneuvered subcutaneously toward the head and then pushed along the trachea into the mediastinum region. This approach is adapted from Sgoifo et al. (14). In all cases, the end of the electrode was firmly sutured to the underlying tissue. Animals were left to recover for 24 h before recordings were commenced. The ECG signal from the implantable unit was received via a dedicated receiver (Telemetry Research). This signal was band-pass filtered between 1 and 2,000 Hz, and the reconstructed analog signal was displayed using a PowerLab data-acquisition system (sampling at 4,000 Hz) with associated Chart software (model ML870, ADInstruments, Castle Hill, NSW, Australia). The ECG signal was further analyzed using the ECG analysis module software (MLS360) from ADInstruments.

RESULTS

The average water intake in a group of five rats before implantation surgery was 25–43 ml/day (average 36 ± 6 ml). In the 2 days postsurgery, this intake fell between 5 and 10 ml. However, within 3 days all rats resumed drinking at preimplantation levels. Similarly, the average body weight before implantation was 320 ± 23 g and fell between 5 and 10 g over the first 2 days after surgery before animals recommenced normal eating and body weight continued to increase (average weight gain across the next 4 wk was between 15 and 30 g). Although a detailed analysis of behavior was not undertaken, the animals appeared in good health with grooming and normal body movements observed.

The ECG signal was recorded in each animal intermittently under a variety of conditions over an extended period (Figs. 3 and 4). A recording was obtained over a 4-mo period in one rat. The ECG signal showed no sign of deterioration over this period. In all cases, placement of the leads on the xiphoid process and along the trachea in mediastinum region gave the best-quality ECG signal. With the leads placed across the chest (lead II), movement of the animal caused small but observable artifacts. Some variation in the R-wave amplitude with respiration was also observed with a lead II orientation. When necessary, the encapsulated button cell battery within the transmitter was recharged using ICPT. The ICPT circuit was able to supply power to the transmitter for recharging while the rat was freely moving, and no restraint was required. Many charges and monitoring cycles were performed in each rat. During the period of charging (~100 min), no ECG signal could be recorded because of the magnetic field interference with the electronic circuitry. No behavioral changes in the rats were observed during the charging period. A representative averaged ECG signal obtained from a sample of 20 heartbeats using ECG analysis software (Chart from ADInstruments) is shown in Fig. 5. The width of the QRS complex was 12.5 ms, the QT interval was 56 ms, and the PR interval was 45 ms in this example recording. The implantable amplifier sampled the ECG signal at 2,000 Hz and had a 4-mV input range and with a 12-bit resolution of the analog-to-digital conversion offered sufficient accuracy with regard to measurement of the amplitudes of the P, R, and T waves.

To determine the usefulness of the telemetry system in four rats, the beta blocker propranolol was administered (10 mg/kg ip). A clear reduction in heart rate between 50 and 80 beats/min was observed in the rats (Fig. 6).

In several rats anesthetized with medetomidine and ketamine (as described in METHODS), the proarrhythmic alkaloid aconitine was administered (50 μ g/kg iv). This neurotoxin opens tetrodotoxin-sensitive sodium channels in the heart and produced distinct ventricular arrhythmias in each rat (Fig. 7).

To record the diurnal variation in heart rate, the R wave of the ECG was detected using Chart software. Data were collected over a 7-day period in one rat with the transmitter turned on for 10 min of each hour. This scheduling utilized a software extension in Chart. A diurnal variation in heart rate was observed (Fig. 8). In the configuration of the transmitter used in the present study, the coin-sized battery used allowed for 7 h of continuous monitoring between recharges. Switching the transmitter off via the receiver allowed the monitoring duration

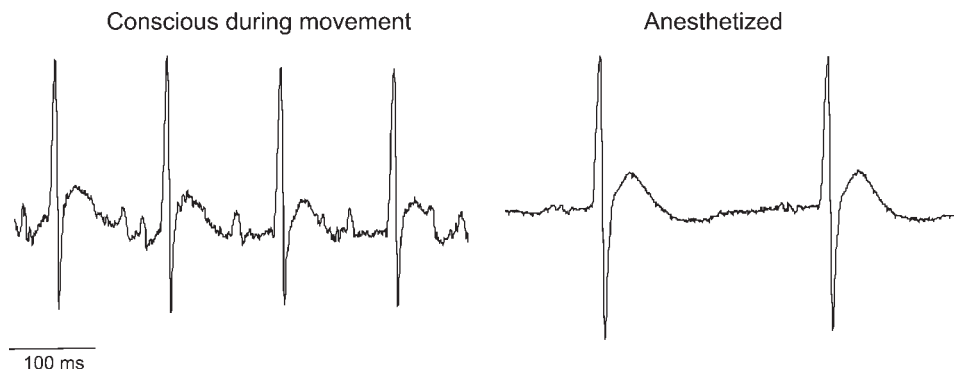


Fig. 3. Sample ECG waveforms obtained from a single rat 1 wk after transmitter implantation during movement around a cage (heart rate 450 beats/min; left) and subsequently under medetomidine and ketamine anesthesia (heart rate 238 beats/min; right).

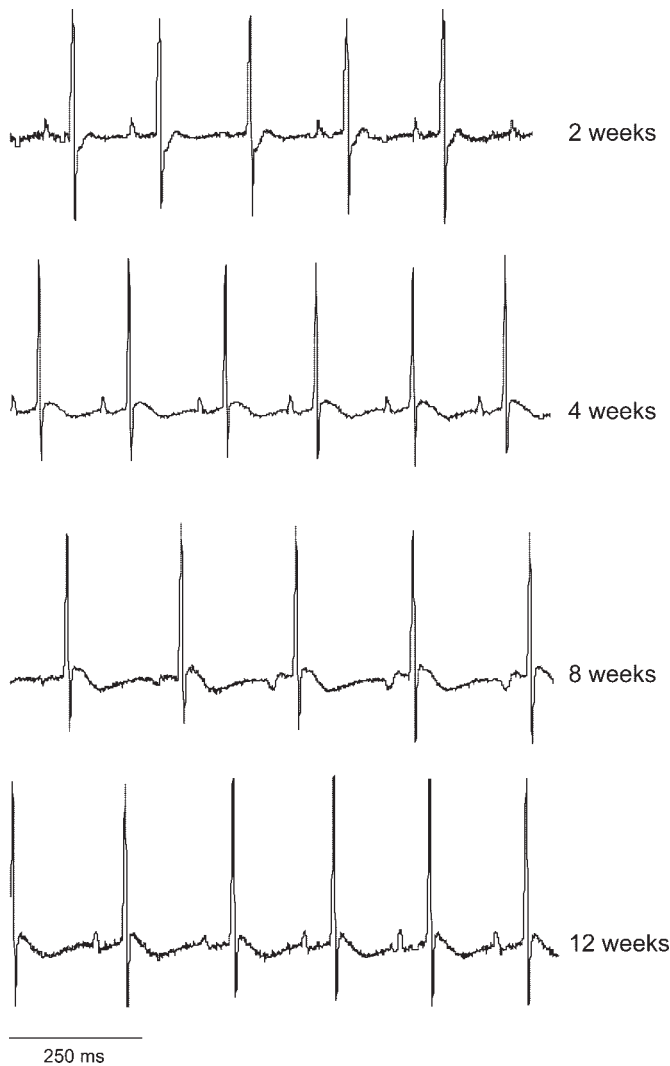


Fig. 4. ECG records from a single conscious rat over a 12-wk period.

between recharges to be greatly extended (and minimized the volume of data that needed analysis). Utilizing larger battery size would clearly extend the monitoring period as well, and in larger animals with a greater ability to carry larger transmitters, this arrangement could be used. With regard to the extended

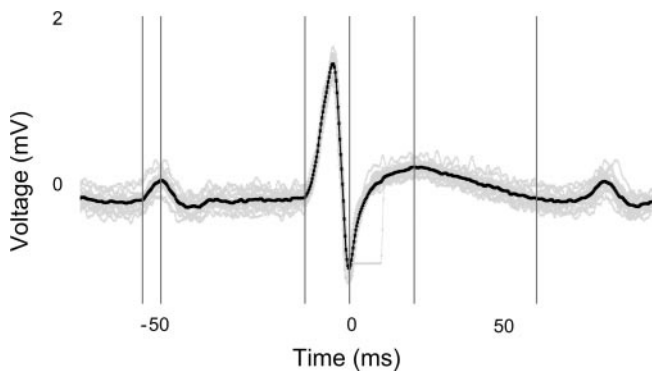


Fig. 5. Averaged ECG waveform obtained from 20 heartbeats in a conscious rat (thick line) and the individual ECG waveforms used to construct the average waveform (faint lines).

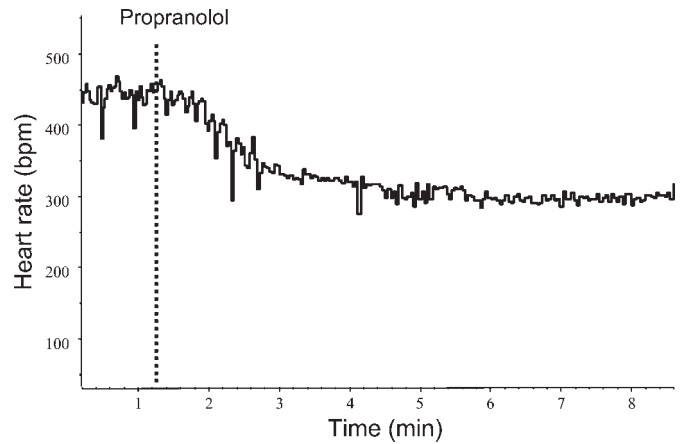


Fig. 6. Change in heart rate detected from the ECG waveform after administration of 10 mg/kg ip propranolol in a conscious rat. bpm, Beats/min.

monitoring, because we found that the recharging of the transmitter could take place when the transmitter was still within the animal, we found that it was possible to “top up” the battery by using a brief period of recharging, e.g., 30 min.

Figure 8 was made by scheduling the transmitter to turn off and on, e.g., 10 min on and 50 off. However, in the off periods because the inductive recharge field could be easily applied, it was a simple matter of switching the recharge field “on” for 30 min every now and then to “top up” the battery. We have considered that a more useful approach would be to automatically schedule the recharging, and this method remains to be studied.

DISCUSSION

In the present study, we tested the hypothesis that inductive power technology could provide power to implantable physiological devices and allow the telemetric recording of a high-fidelity ECG in freely moving rats. This system contains a rechargeable battery that can be recharged while the implant is still in the animal. Thus this technology provides continuous, potentially lifetime recording and transmission of high-bandwidth biopotential signals. The novel features of our system are the ability to transmit greater power levels over a larger distance than previously attainable in animals and that the flexible orientation of the implanted device relative to the primary power field generator meant that power was provided to the unit with the rat in any position within the cage, e.g., standing, lying, or walking.

A critical feature of our technology is that power is provided to the implant with minimized production of heat (<5°C). This is achieved using precise control in both the primary and secondary systems. On the primary side, the output current from the power converter, which generates the magnetic field, is adjusted automatically using feedback from the load requirement of the secondary. In most cases the sensor does not require the maximum power that could be delivered. Matching the power delivery to the load requirement results in a lower primary current and reduced losses. On the secondary side, the pickup circuit is tuned adaptively to supply only the power needed; therefore no excessive heat is generated. This is a significant improvement over the hard-switching techniques used in traditional ICPT systems.

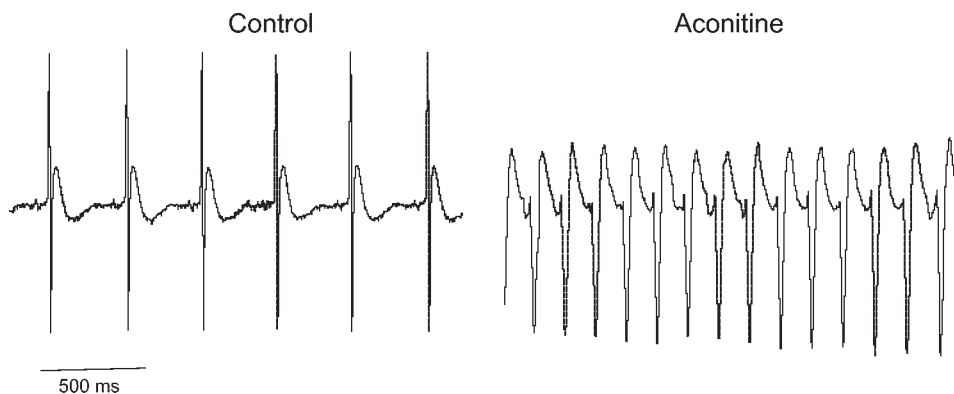


Fig. 7. Representative recording of the ECG responses to the proarrhythmic alkaloid aconitine (50 $\mu\text{g}/\text{kg}$ iv) administered to an anesthetized rat.

The ability to recharge the implantable transmitter while the implant was still present within the rat meant that more power was available for high sampling frequencies and large range (5 m) without compromising the longevity of the recording. Unfortunately, ECG recording could not be undertaken during the recharging phase because of interference between the magnetic field and the telemetry transmission. With appropriate filtering at the amplification stage, it is likely that this interference could be removed, and thus monitoring could take place during recharging. This could potentially allow use of a much smaller battery or even no battery and thus considerably reduce the size of the transmitter.

We chose to implement a rechargeable battery within the implant that was recharged when necessary by placing the cage holding the rat on a charging pad. This provided for greater flexibility in that the primary power field did not need to be on continuously and allowed for normal housing conditions. Such an approach is also applicable when monitoring in larger animals such as dogs or cattle where it would not be possible to provide a primary power field to cover the entire cage. In this setting it could be possible to bring the charging pad to the animal for the period required to recharge the battery and to employ a larger battery to provide for a greater duration of monitoring before recharging was required. Because the orientation between the implant and the primary field generator

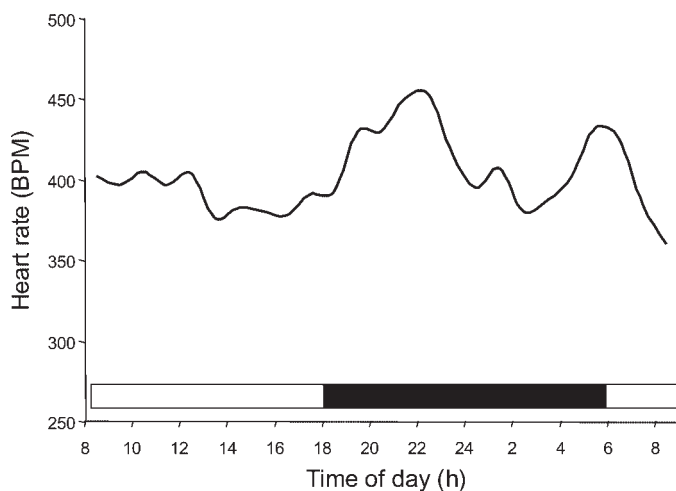


Fig. 8. Average diurnal rhythm in heart rate obtained from a single rat over a 7-day period. The ECG was recorded 10 min each hour, and the average heart rate for that period was calculated. Bars indicate the periods of light and dark.

can be highly flexible and still allow power to be transferred, it is not necessary to restrain the animal in any way.

Currently available telemetry-based biopotential amplifiers suitable for the measurement of signals such as ECG in rats provide power via a battery for a considerable length of time (warranted up to 6 mo; Data Sciences); however, these units eventually require costly refurbishment to replace batteries. To reduce the current drain on the battery unit, the range of these implants is often limited to ~ 1 m, and the sampling frequency of the signal often restrained at a lower than optimal level (in the case of the Data Sciences biopotential transmitters, the transmitter frequency is up to 500 Hz, giving a maximum timing resolution of 2 ms; Wade DePas, Data Sciences, personal communication). For accurate determination of particularly fast events, for example during high heart rates, higher sampling frequencies may offer considerable advantage in allowing accurate determination of, for example, QT intervals. A high sampling frequency is critical for some signals such as sympathetic nerve activity where the signal is normally band-pass filtered between 50 and 3,000 Hz (9).

A further important aspect of our technology is that the large amount of power that can be provided allows for the development of implantable devices for the long-term monitoring or regulation of physiological processes using devices such as motors, sensors, and pumps. Currently, such devices generally require either exteriorized wires for power supply or a prohibitively large battery. Another significant attribute is that with greater power levels available, the range of the transmitter can also be greater (5 m in the present case). We chose to implement this new technology to power an implantable device for recording the ECG in a rat as this signal has well-described characteristics and is routinely measured via telemetry. The rat is an animal widely used in drug development, genomic research, and basic physiology, and implementation of a fully implantable amplifier/transmitter and power module capable of implantation in animals >200 g illustrates the applicability of this technology. We observed normal drinking and eating patterns in the rats after transmitter implantation and a diurnal variation in heart rate (6, 8). The observations made during the propranolol and aconitine interventions indicate the applicability of this approach to recording the ECG.

While we have chosen to develop a power-transfer system for implantable devices for use in small animals, the technology also has great potential for human-implantable applications requiring large amounts (2–10 W) of power to drive motors, such as in ventricular assist devices.

It has been known for more than a century that power can be transferred from one place to another in electromagnetic forms. Magnetic couplings are widely used in traditional transformers and electrical machines. However, until recently it was considered impractical, if not impossible, to transfer high levels of power across large air gaps. It is the development of power electronics, particularly the rapid advances in semiconductor power switches, that make inductive power transfer viable. This has resulted in several industrial applications, including an overhead monorail system (5), road stud lighting, and recharging of electric buses. In these applications, no biological effects of the electromagnetic field have been observed (3).

In conclusion, we have developed and implemented a novel power supply system that enables power to be supplied to an implantable device. Power was utilized to recharge a battery contained within a telemetry device recording a high-fidelity ECG signal in a conscious rat living in its home cage. The ability to provide suitable power levels to implanted devices without concern to the orientation of the device and without causing heating provides the basis for the development of new devices to record physiological signals in animals or humans over significantly longer time periods than can currently be accommodated. The technology underpinning this study has been licensed to Telemetry Research and will be commercially available from the beginning of 2007.

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GRANTS

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DISCLOSURES

D. M. Budgett, A. P. Hu, and S. C. Malpas have a shareholding in the company Telemetry Research Ltd. The company is a new company launched by the University of Auckland to commercialize inductive power transfer for biomedical applications. Wireless monitoring equipment is available through the company. D. M. Budgett, A. P. Hu, and S. C. Malpas have patents pending.

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