Recording Brain Activity Wirelessly

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Inductive Powering in Miniature Implantable Neural Recording Devices

The continuous monitoring of electrical brain activity with implanted depth electrodes is essential for understanding the neural substrates of many physiological and pathological brain functions, such as emotion and epilepsy [1]. Existing instrumentation uses wires to power and record directly from the brain. The wires hooked up to the implanted electrodes restrain and interfere with the animals' natural behaviors. Consequently, it has been difficult to correlate truly normal behavior with neuronal activity. It would be ideal if brain activity could be monitored while animals live in a natural and enriched environment. Animals reared in such complex, natural environments, rather than in standard laboratory cages, have been shown to have altered brain development, a larger number of cells in certain brain areas, and considerable changes in neuronal morphology [2]. Because of the complexity of environments with natural hiding places and burrows, recordings of brain activity with wires attached to the animal are impossible. These challenges could be overcome by the use of wireless recording devices.

Conventional recording devices are powered by means of direct wire connections or onboard batteries. Although small batteries can be used to avoid interference with animal behavior, this greatly limits the operating lifetime of the device due to necessary and frequent battery changes. Thus, the use of direct wire connections is impractical and the use of batteries is inconvenient when performing long-term continuous recordings of neural signals from freely behaving animals in their natural environment.

To reduce the overall device size and eliminate the need for battery replacement, inductive powering can be used. The design, manufacture, and packaging of such a device is dependent on the ability to establish a reliable inductive power link to the implant. Magnetically coupling power from a large external coil to a smaller coil located in the implantable device would solve this problem if enough power could be reliably delivered in this manner. Previously, researchers have described devices that made use of inductive powering to drive biological implants [3]–[5], while others have described the theory involved [6].

In this article, we describe the theory involved in every aspect of the design of an inductively powered system with emphasis on the trade-offs involved, from theory through to the details involved in developing a practical working system. In addition, we provide the designer with a detailed guide to the design of such systems, shedding light on some of the details omitted in the description of previous work. A specific real-world application of inductive powering is detailed, illustrating how many of the design decisions were made to develop a successful system for recording neural signals from awake and freely behaving animals.

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Design

Magnetic Fields, Issues, and Difficulties

From Maxwell's equations [7], we know how time-varying magnetic and electric flux gives rise to electric and magnetic fields, respectively. From these, Ampere's law can be derived as

$$I = \oint_c \overline{H} \cdot d\bar{l} \tag{1}$$

and is illustrated in Figure 1. Ampere's law relates the current I flowing through a conductor to the closed-contour integral of the magnetic field H flowing around it. Intuitively, this means that a magnetic field flowing around an infinitesimal current carrying a wire element gives rise to a current in that element. Alternatively, forcing a current through a wire element will give rise to a magnetic field around that element. If a series of conductive elements is in a coil, as is done in transformers, then a time-varying current flowing around the coil will give rise to a time-varying magnetic flux through the coil. It is this flux we will attempt to couple onto an implanted device. To understand why, we look at Faraday's law

$$V = \oint_c \overline{E} \cdot d\bar{l} = -\frac{\partial}{\partial t}\int_s\int \overline{B} \cdot d\bar{S} = -N\frac{\partial \Phi}{\partial t}, \tag{2}$$

which relates voltage V to the integral of the electric field E present around a contour C, which is equal to the negative rate of change of the integral of the magnetic flux density B passing through a surface S. In a simpler form, and in the case of our coil, the instantaneous potential across that coil is equal to the rate of decrease in magnetic flux $\Phi = B \cdot S$ through the coil multiplied by the number of turns N in the coil. Faraday's law is illustrated in Figure 2.

By using Ampere's law to describe how electric current can result in the production of a magnetic field and Faraday's law to describe how a time-varying magnetic flux density can produce electrical energy, the process of inductive powering can be understood.

One common example we already have referred to is the use of a circular looping conductor, a coil, to generate the ac magnetic field and another circular coil to convert the time-varying flux through it into electrical energy. The first coil is commonly referred to as the primary or drive coil, and the second coil is commonly referred to as the secondary or pickup coil. See Figure 3 for an illustration of this arrangement.

In a typical transformer, the drive coil and the pickup coil are the same size and located next to each other. This arrangement maximizes the efficiency of power transfer from the primary coil to the secondary coil. However, in the application of interest here, the secondary coil will be much smaller than the primary coil because, in a practical experimental setup, the primary coil must enclose the entire space within which the animal implanted with the secondary coil will be, giving the animal the greatest possible freedom of movement. At the same time, the pickup coil must be small enough to be implanted in the animal. Consequently, the efficiency of power transfer is much worse than in a transformer because the amount of flux generated by the primary coil and coupled into the secondary coil is equal to the ratio of the two coil areas, going from 1:1 in a transformer to 1,000:1 in a realistic setup such as the one described in this article. A strategy for improving the power transfer efficiency is to use a ferrite core in the secondary coil with a high magnetic permeability μ. The ferrite core amplified the magnetic flux density picked up in the secondary coil by a factor of μ since, for a given magnetic field H, the flux density B is equal to μH, and it is the flux, not the field, that is being coupled.

The ac voltage induced (as predicted by Faraday's law) across the terminals of the pickup coil can be converted to positive and negative dc voltages with rectifiers and charge capacitors. A schematic circuit diagram of the inductive powering system is shown in Figure 4 along with the series parasitic

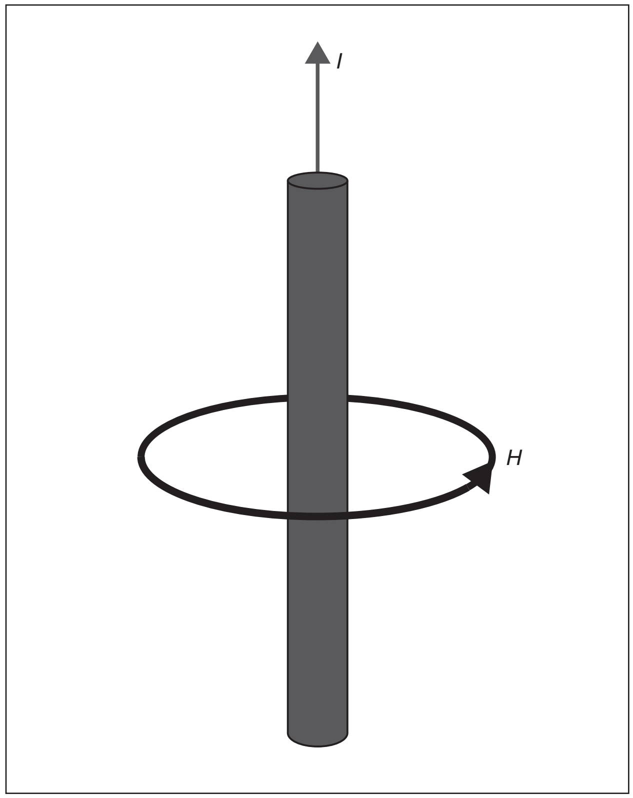


Fig. 1. Magnetic flux density surrounding a current-carrying conductor.

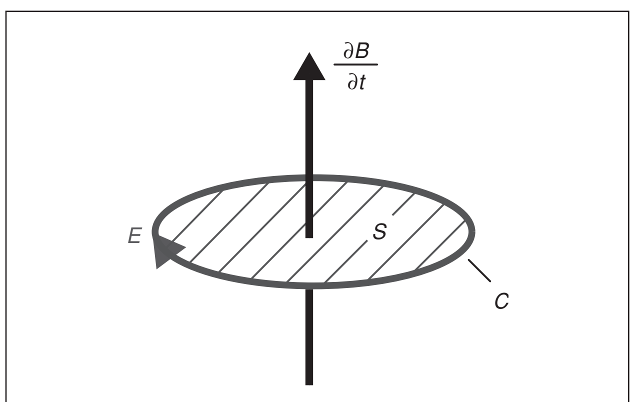


Fig. 2. Electric field induced in a conducting loop of *n* turns by a time-varying magnetic flux density.

resistance R_{s1} of the drive coil. The details of how each of these components is selected or built are covered below.

Drive Coil

The design process for the drive coil consists of a series of trade-offs aimed at obtaining the necessary amount of drive power everywhere in the desired three-dimensional enclosure space. The first step in designing an inductive power system is to choose the drive frequency. Higher frequencies have the advantage that they transfer more energy per unit of time because the received electrical power is dependent on the time derivative of the magnetic flux (2). Unfortunately, at the same time, higher frequencies exhibit greater loss in biological tissues, with the result that less of the available energy will reach the implanted device. A good compromise frequency used in practical applications [3]–[5], as well as the specific frequency detailed later in this article, is 2 MHz. In some cases, biological implants have been powered by drive coils operating at as high as 20 MHz [6].

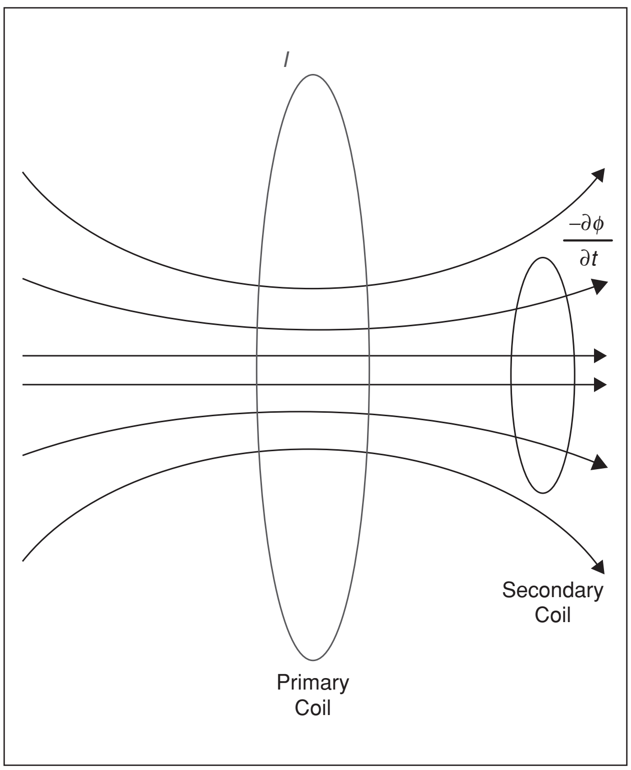


Fig. 3. A schematic illustration of the inductive powering process of a secondary coil by a primary coil.

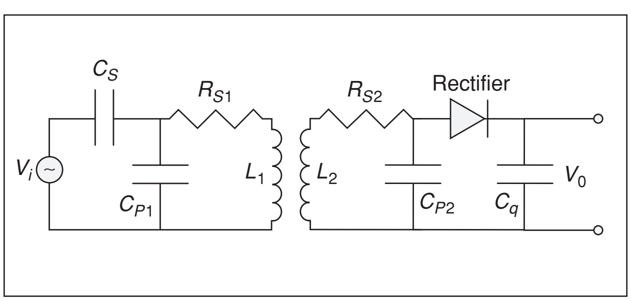


Fig. 4. The inductive power link schematic.

Once the drive frequency has been selected, the next step is to design the geometry of the drive or primary coil. From Ampere's law, we know that increasing the number of turns in the primary coil should increase the magnetic field generated by the same factor. However, at some frequency, dependent on the physical design of the coil, the impedance due to the capacitance between the turns becomes less than that due to the inductance of the coil. The frequency at which these two impedances are equal is called the self-resonant frequency (SRF). The SRF represents the point at which half the current flows between turns in the coil and thus the magnetic field strength generated by the primary coil drops despite, and in fact as a direct result of, increasing numbers of turns. Although the interturn capacitance can be reduced by increasing the spacing of the turns in the coil and by making the coil diameter smaller, there are practical limits to both of these approaches. In practice, it is sufficient to space the turns by one or two wire diameters. In addition, the diameter of the primary coil must be as large as the enclosure in which the biological implant will operate. When the primary coil drives an implant in a stationary subject, coil diameter can be quite small. However, when studying activity in subjects that are awake, the enclosure, and thus the diameter, of the drive coil must necessarily be larger. It is almost always preferable to have a circular drive coil rather than a rectangular or other-shaped coil since circular coils produce a uniform magnetic field in their interior. Having determined the coil diameter and turn spacing, the number of turns is increased until the SRF is no less than four or five times greater than the drive frequency. This is done to facilitate impedance matching of the drive coil to the power supply. For example, large primary coils on the order of 0.5 m diameter being driven at 2 MHz might have no more than five or six turns.

When driven with high-frequency ac signals, free-electrons are driven back and forth along the wire at the driving frequency. Additionally, they are driven to the surface and towards the center of the wire each full cycle by the relative potential of the wire with respect to the exterior. However, the force with which they are driven towards the center is offset by the natural tendency of electrons to repel each other. At very high frequencies, the period of the drive cycle becomes smaller than the time required for the electrons to redistribute in the wire according to their mobility. Thus, as drive frequencies increase, free-electrons tend to concentrate progressively along only the outer skin of the wire. Since this effectively decreases the cross-sectional area of the wire with free-electrons, impedance of the wire increases proportionately. This overall phenomenon is known as the *skin effect*.

The drive coil can be made of various kinds of wire, including single-strand wire, stranded wire, and braided wire. Although single-strand wire is inexpensive and widely available, it becomes susceptible to the skin effect at high frequencies. The increased impedance results in a higher inductance and a lower SRF than would be calculated for the given wire gauge using dc measurements. Because of this discrepancy, the inductance of a coil is measured using an impedance analyzer at the actual drive frequency, and both the number of turns and the value of the discrete series capacitance are optimized at this frequency. The use of multistranded wire can improve the efficiency of a coil by increasing the surface area available for current flow and reducing the impact of the skin effect and, thus, the parasitic resistance. This is at the expense of a corresponding increase in the parasitic capacitance between the individual strands of wire and a lowering of the SRF of the primary coil. A particular form of multi-stranded wire known as *Litz wire* has multiple strands carefully arranged to minimize the length of wire at which any two strands run parallel to one another. By keeping individual strands orthogonal, the parasitic capacitances within the wire can be minimized and the SRF of the coil maximized for a fixed number of turns. For applications where power conservation in the drive coil is unimportant, conventional and inexpensive single-strand wire is sufficient.

A critical aspect of drive coil design is the exact matching of the coil impedance to the output impedance of the driver to maximize the power transferred to the coil. This operation is performed with the tunable capacitors placed in series with and in parallel to the drive coil. The two capacitors are tuned at the drive frequency to match the impedance of the coil, complete with its parasitic resistance, to the output impedance of the signal generator. In our case, this means tuning the drive coil to have an impedance of 50 Ω at the drive frequency of 2 MHz. The impedance of the tank circuit, which consists of a series tuning capacitor *Cs*, a parallel tuning capacitor C_{p1}, and the coil with its parasitic resistance R_{s1}, is a function of frequency $\omega = 2\pi f$ and can be expressed in terms of its complex impedance

$$Z = \frac{1}{j\omega C_s} + \frac{1}{j\omega C_{p1} + \frac{1}{R_{s1} + j\omega L_1}}. \tag{6}$$

The value of Z at $\omega = 2\,\text{MHz}$ is 50 Ω. The value of R_{s1} and L_1 can be measured at the drive frequency using an impedance analyzer. Rearranging the equation to solve for C_s as a function of C_{pl} we obtain

$$C_s = \frac{1}{j\omega\left(Z - \frac{1}{j\omega C_{p1} + \frac{1}{R_{s1} + j\omega L_1}}\right)}, \tag{7}$$

which can be solved numerically using readily available math software packages such as MATLAB.

Once the coil has been matched to 50 Ω, it is ready to be powered. The way in which this is done depends largely on the specific application. In circumstances where the power supply to the driver is limited due to the portable nature of the system [3], the efficiency of the driver is very important and must be designed separately. Efficient nonlinear Class C–F power amplifier topologies are well suited and have been well documented elsewhere in the literature [8]. In particular, Class E power amplifiers have been designed for just this application with high efficiencies reported in portable setups [9]. For a stationary setup, such as when the drive coil surrounds an enclosure containing an implanted test subject, the driver can be plugged into an ac line and need not be particularly efficient. In such applications, inexpensive drivers can usually be purchased commercially from electronics vendors such as Ham radio stores. In either case, it is desirable to have a driver with an output frequency that can be finely tuned to compensate for small variations in coil impedance. In order to supply sufficient but not excessive power to the implanted device, it is also necessary to be able to regulate the amount of output power.

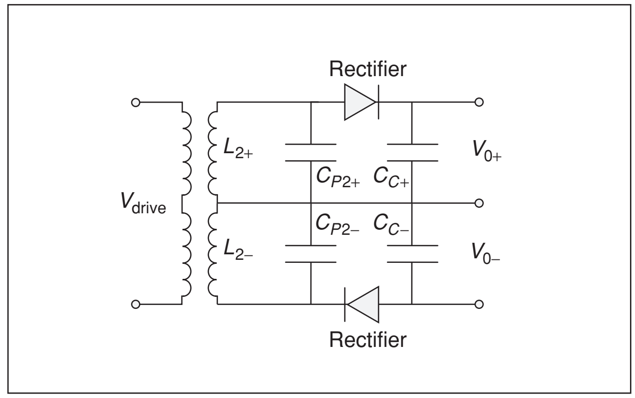


Fig. 5. The pickup coil circuit schematic.

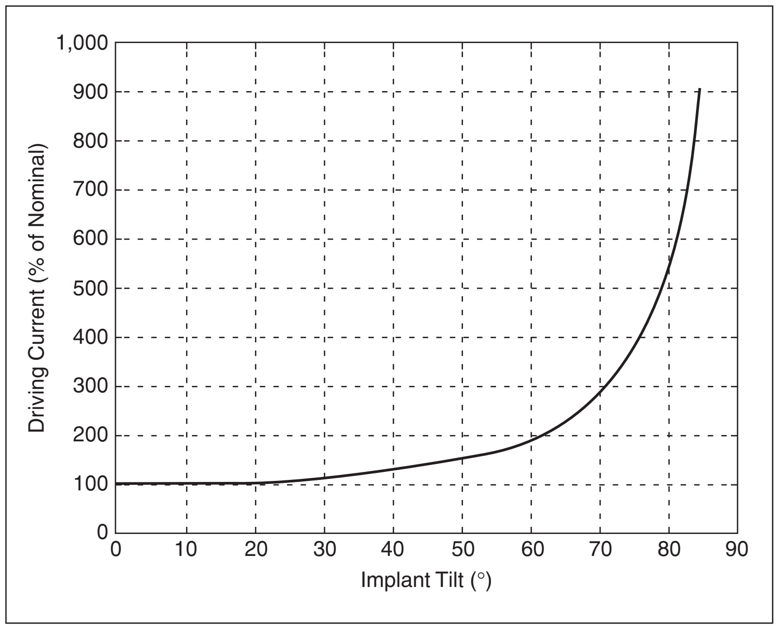


Fig. 6. The required nominal driving current versus implant tilt.

Pickup Coil

The design of the pickup (or secondary) coil is simpler; though in many ways, it is similar to that of the primary. The number of turns in the coil should be maximized to pick up as much of the primary field as possible without having the SRF drop below the drive frequency. The diameter of the coil should, unlike that of the primary, be maximized to increase the amount of magnetic flux density captured. This is offset by the need in implant applications to keep the overall size of the implanted device small. When the implanted device is small, the pickup coil will have many turns about a small radius. The amount of magnetic flux density seen by the coil for a given magnetic field can be increased by the addition of a ferrite core with a high permeability. From Faraday's law (2), we know that a time-varying magnetic flux will produce a time-varying voltage across the coil terminals. The capacitor placed in parallel with the pickup coil serves to maximize the impedance seen by this voltage and reduces the loss of power in the coil itself. The impedance seen looking at the pickup coil and capacitor in parallel is given by

$$Z = \frac{1}{j\omega C_{p2} + \frac{1}{j\omega L_2 + R_{s2}}} \tag{8}$$

and approaches infinity in a perfectly matched ideal tank circuit. In reality, and just as in the drive coil, the pickup coil has a parasitic resistance associated with it. This resistance lowers the overall impedance that can be achieved at the resonant frequency and affects the value of C_{p2} required to achieve resonance. The parasitic resistance of the pickup coil can be expressed as a quality factor Q of the tank circuit, where Q is the ratio of the energy lost to the amount of energy present in the tank circuit per unit cycle. Typically, pickup coils with $Q > 50$ and matching capacitors can be purchased off the shelf in surface-mount packages ideal for small implantable devices.

Fig. 7. The implant location.

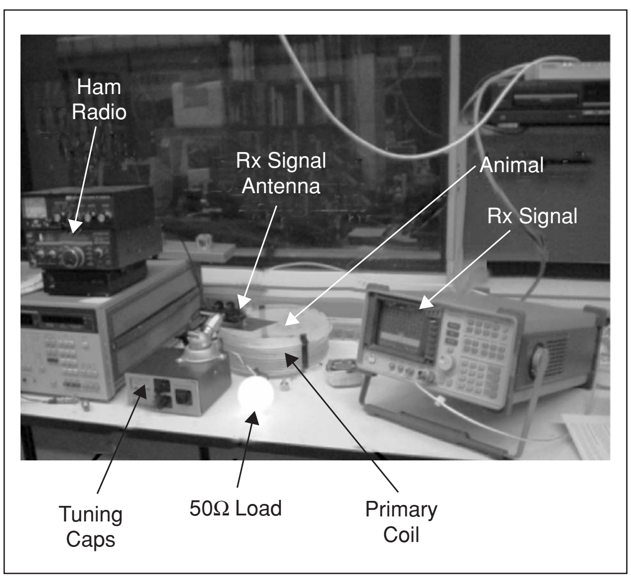


Fig. 8. A photograph of the complete inductive powering and RF telemetry setup photograph.

Once the magnetic field has been picked up and the voltage present at the inductor terminals has been maximized by tuning the parallel capacitance C_{p2}, this ac voltage needs to be converted into the dc voltages necessary to drive the implant. The simplest way to accomplish this is by using single diodes to half-wave rectify the ac signal, thereby providing either a purely positive or negative signal to the charge capacitors as shown in Figure 5. More complicated designs offer improved efficiency at the expense of complexity.

In a simple half-wave rectifier, diodes are combined with charge capacitors sufficiently large to store charge over the period of time that the load presented by the implant attempts to drain it off. The product of the integrated circuit (IC) load impedance and the charge storage capacitor is the RC time constant of the transceiver, which must be greater than the period of the ac drive signal. Therefore, the use of a higher powering frequency will allow for the smaller charge-storage capacitors with smaller footprints and will yield a smaller overall package size. Although rectifiers come in many flavors, two parameters are of particular importance. The first and most important parameter is the forward voltage drop. For an implant requiring a 2-V power supply, a rectifier with a 1-V forward voltage drop would consume 33% of the total power by itself! Because of this problem, and the fact that the voltage drop increases as a function of current in metal-oxide-semiconductor (MOS) devices, most rectifiers are not implemented in MOS technology. Fortunately, surface-mount bipolar-type devices can be found with forward voltage drops approaching 0.1 V. The second important rectifier parameter is the reverse leakage current, which represents the power that flows back through the diode and is lost during the half-cycle of the ac voltage in which the device is reverse-biased. Good rectifiers will have negligible reverse leakage currents; unfortunately, devices with particularly small forward voltage drops have larger leakage currents. In adding only minimal complexity to the rectifier circuitry, one can use full-wave rectifiers built from bipolar diodes in a bridge topology [8] to double the efficiency of the ac to dc conversion. Of substantially greater complexity are fully integrated CMOS rectifiers [10]. These can be integrated with the recording or stimulating circuits on the implant IC, allowing for a much smaller overall device size without sacrificing power conversion efficiency.

In applications where the implanted device is mobile with respect to the drive coil, the power picked up in the device will vary as a function of the tilt of the secondary coil with respect to the primary coil. Maximum coupling is achieved when the two are coaxial, and zero power is transferred when the axes are orthogonal. Thus, the current required in the primary coil to couple a given amount of power to the secondary coil varies as a function of the tilt of the animal with its implanted receiving coil. Figure 6 shows the percentage of the nominal driving current required to achieve the necessary pickup power as the implant is tilted away from a coaxial alignment. An important observation to make, from Figure 6, is that for small variations in tilt angle, there is a negligible effect in coupling.

Using all of the design information presented in this section, an inductive powering system was constructed for a miniature, implantable, neural recording device. The details of this design are given below.

Implementation and Results

In our specific case, an implantable neural recording device that required 5 mW of power was developed. The device records neural signals from electrodes implanted in the brain. These signals are amplified, FM modulated onto a 3-GHz carrier wave, and transmitted to an external receiver for processing. The details of the neural recording implant are described in the literature [11]. To power this implant, we need an inductive powering system capable of delivering 5 mW to the implanted device in a freely moving rat in a large enclosed area. The implanted device sits like a hat on top of the test subject's head, as shown in Figure 7. This allows for short wires and, therefore, less noise pickup between the electrodes implanted in the brain and the IC. The drive coil, like the pickup coil, is built with the highest number of turns possible while still keeping the SRF above the drive frequency of 2 MHz. The coil is made of 22-gauge single-strand copper wire that has a dc impedance of $50\,\Omega/\mathrm{m}$. The inductance of the coil is measured using an impedance analyzer at the actual drive frequency, and both the number of turns and the series capacitance are determined at this frequency. Due to the stationary nature of the enclosure, the coil can be driven with an inexpensive Ham radio transmitter. Ours has a variable transmit power of up to 100 W at 2 MHz. The coil consists of nine turns around a 30-cm diameter enclosure. The measured inductance at 2 MHz is $18\,\mu$H; therefore, the parallel and series capacitances needed to tune the coil impedance to 50 Ω are 264 pF and 87 pF, respectively. Using off-the-shelf surface-mount capacitor values of 270 pF and 86 pF should yield a $50\,\Omega$ impedance match at a frequency of 1.989 MHz. In practice, the impedance of the coil changes, depending on the exact environment in which it is placed. Therefore, the value measured when the coil is connected to the impedance analyzer is different than the actual value when the coil is moved to the test bench, and that value changes when an animal is placed inside the coil. Similarly, the placement of large conductors near the setup during operation can have the effect of detuning the matching circuit, reducing the flux generated in the coil, and turning off the inductively powered devices. A simple and effective solution is to place standard 100-W incandescent lightbulbs in series with the primary coil. These bulbs have impedances close to $50\,\Omega$ at 2 MHz and provide visual feedback on how matched the coil is. We then use the derived matching capacitor values as a starting point and move the coil to its permanent location on the test bench. We then place the animal in the coil and power up the transmitter. By tweaking the tunable capacitors to maximize the brightness of the bulb, we maximize the generated magnetic flux as well. Once this is done, further tuning is usually not necessary unless a major conductor is placed or removed from the immediate vicinity of the primary coil. Even then, with the bulb brightness as a guide, retuning is quick and easy. For the tuning, we use capacitors rated to 2 kV to handle the high voltages that will be present within the impedance matching circuit at high power levels. A photograph of our running setup is shown in Figure 8.

In order to produce both a positive and a negative voltage rail on the receiving side, two coils are used to pick up the drive signal in an identical configuration. Each consists of a ferrite-core surface-mount device (SMD) inductor that has a high quality factor Q and small footprint. Since miniature customizable center-tapped coils are not available, two coils with different inductances must be used to generate voltage rails of different magnitudes. In our case, one coil has an inductance of $47\,\mu$H and the other $22\,\mu$H. These coils are matched with 130-pF and 270-pF capacitors, respectively, to resonate at the drive frequency and maximize the energy transfer. The signal induced on the pickup coils is fed into discrete Schotky rectifying diode bridges, one configured to positively rectify the coupled sine wave and the other to negatively rectify it. Each rectified signal is then fed to 1-μF capacitors for charge storage. The diodes in the bridge have a forward voltage drop of 0.1 V. Smaller capacitors, as small as 10 nF, could theoretically be used at 2 MHz for charge storage, but in practical applications, nothing is lost with larger capacitors of similar footprint, which provide added stability to the system by storing a greater supply of charge. The two dc voltages stored, one positive and the other negative, drive the implant. The pickup coils, rectifying diodes, and charge capacitors are all integrated with an IC onto a microstrip board (Figure 9), which is then encased in biologically compatible epoxy. The size of the implantable device is less than $1 \times 0.5 \times 0.5 = 0.25$ cm^3.

The IC is contained within the gold subpackage. The board onto which all the components are soldered is made in-house with low-loss microstrip material. The neural recording device was successfully powered and stayed on everywhere within the enclosure without noticeable effect to its performance from tilts of less than 40°; the drive coil was powered with 20 W from the Ham radio transmitter.

Conclusion

The study of complex neuroscientific phenomena such as fear, epilepsy, and aggressive behavior is currently being limited by the physical and psychological effect of the test environment itself. In a macroscopic correlate of the uncertainty principle, the process by which researchers observe voluntary or subjective animal actions decreases the certainty with which they can establish any conclusive link between cause and effect by interfering with the animal's natural behavior on a conscious level. What becomes desirable in such studies is a means of observing the electrophysiological activity in the animal model without interfering with its environment. This way, the subject being studied does not know that it is being studied. An essential step along the path to making this technology practical and available lies in the successful development of an inductive power link allowing the researcher to untether animals for long-term chronic recording.

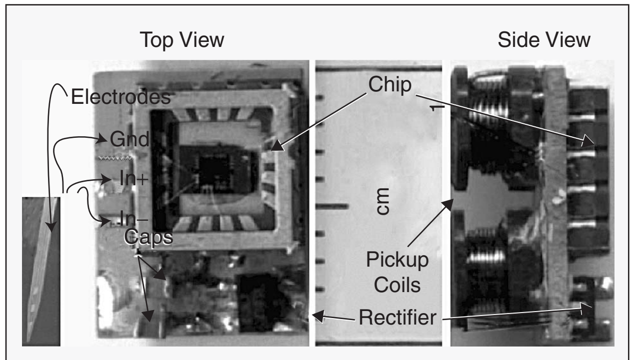
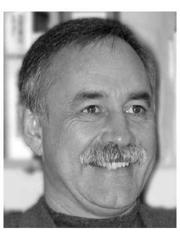


Fig. 9. Packaged transceiver photograph.

We have described in detail the theory behind the design of such inductive powering systems and the practical trade-offs involved in their design and construction. A complete inductive powering system consists of two major parts—a drive coil and a pickup coil, and their associated impedance matching networks. The drive coil is designed to maximize the magnetic field within the desired enclosure at the drive frequency. Similarly, the pickup coil is designed to maximize the amount of magnetic flux density converted to power for the implant, while minimizing it's own dimensions. Finally, to illustrate the application of the theories and design process described in the text, one specific inductive powering system tailored to the needs of a biological implant has been described in detail.

Pedro P. Irazoqui received his B.Sc. and M.Sc. degrees in electrical engineering from the University of New Hampshire (UNH), Durham in 1997 and 1999, respectively, and the Ph.D. in neuroengineering from the University of California at Los Angeles (UCLA) in 2003 for work on the design, manufacture, and packaging of implantable integrated circuits (ICs) for wireless neural recording. Together with three partners, he helped found and was vice president of IC development at Triangle Biosystems Inc., Research Triangle Park, North Carolina. There, he headed up development of analog ASICs for wireless and tethered headstages, including RF oscillators, amplifiers, filters, and neural preconditioning circuits for implantable wireless neural recording and stimulating systems.

Currently he is an assistant professor in the Weldon School of Biomedical Engineering at Purdue University; his lab is pursuing research into ASIC design for biomedical applications.



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His research interests include novel ferromagnetic MEMS, nanomagnetomechanical systems, chemical sensors, and wireless sensor networks and also include neuroengineering devices and systems. He is dedicated to developing and improving graduate-level training programs in the multidisciplinary engineering fields of MEMS/nanotechnology and neuroengineering.

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