# Wireless Microstimulators

Abstract

Wireless microstimulation

## Synonyms

Wireless microstimulation

## Definition

Electrical currents for neural stimulation have conventionally been delivered via metal wires to the electrode that is in contact with the tissue. The wire connections attached to these rigid electrodes, floating in a very soft medium like neural tissue, not only damage the surrounding cells from tethering forces but also limit the longevity of the implant due to wire breakage in chronic implants. Wireless transfer of stimulus energy as well as the pulse parameters to a floating electrode or an array of electrodes has gained interest in recent years as a method to eliminate the associated problems with tethering wires. Considering the properties of the neural tissue, different types of energy transfer mechanisms have been proposed for energizing the implant wirelessly: electromagnetic radio-frequency (RF), optical, and acoustic waves. The implanted electrode(s) may or may not have active electronics for storing the pulse parameters. Active devices also require an energy storage mechanism for powering the circuit. On the other hand, passive devices that can instantaneously convert the incident energy into the electric stimulus do not need to store energy or require programming, often maximizing the stimulus energy and implantation depth.
The enthusiasm for replacing wired stimulators with wireless versions is primarily motivated from the complications that arise from tethering forces associated with the micromotions of the electrode assembly, the risk of infection from percutaneous connections, and the failures from wire breakage. Additionally, with passive versions of the wireless devices, the electrode shank needed for housing the electrode contacts and the connecting wires is no longer needed, thus displacing less tissue. This article discusses the critical parameters associated with microstimulators that use wireless links. The wireless microstimulation concept is illustrated in Fig. 1.

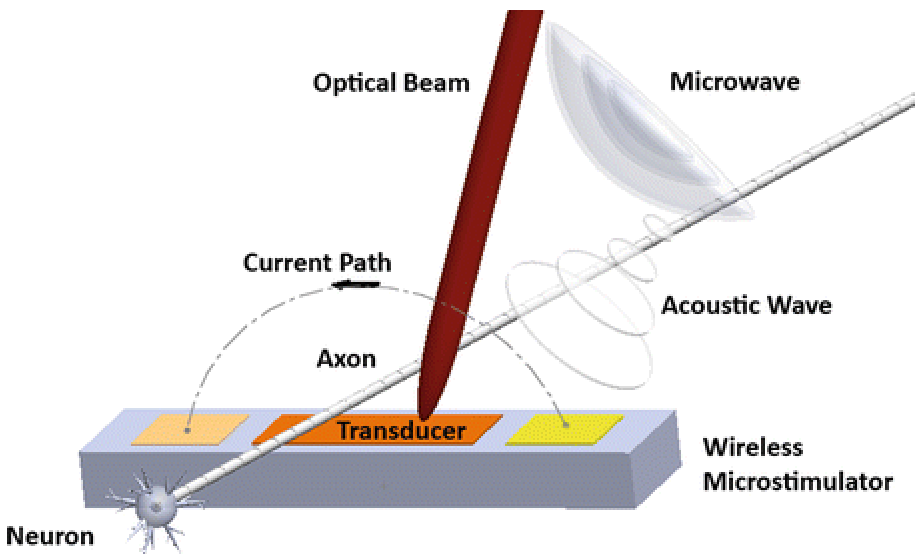


Fig. 1
Illustration of a wireless microstimulator. Energy can be transmitted to the stimulator using RF, optical, or acoustic waves. Current generated by the microstimulator travels from the anode to the cathode of the device, inducing action potentials in nearby neurons

## Detailed Description

## Active and Passive Microstimulators

The stimulation pulse parameters (amplitude, duration, frequency) need to be adjustable in most neuroprosthetic applications to achieve the best results. Thus, when discussing wireless microstimulators, it is useful to classify the devices as either active or passive, depending on whether the device is actively controlling the stimulation signals or if the device passively converts transmitter energy directly into electrical stimulation. In the passive case, the pulse parameters must be controlled by the external unit in the form of energy packages. Passive wireless devices can have the benefit of maximizing the device area for energy conversion. Active devices often contain complex electronics that control the pulse parameters and stimulus rates and must dedicate an area to this function. On the other hand, active wireless devices can operate autonomously with minimal intervention and only at times when the pulse parameters need to be modified or the battery needs to be recharged.

## Energy Transfer

### Battery Storage

The stimulus energy may be provided from a battery incorporated into the implant. The latest version of the BION™ is a good example of an implantable wireless stimulator with a battery (Loeb et al. 2001; Schulman 2008; Kane et al. 2011). Battery systems are attractive because they do not depend on a wireless link for uninterrupted operation, and they are not bound by energy transfer limitations inherent to wireless energy transfer through tissue. Drawbacks include their large volume, required for the battery, and battery capacity. Batteries need to be replaced or recharged (through a wireless link), limiting their use in chronic implants. The BION™ stimulator reportedly requires charging every few days.

### RF Electromagnetics

The most common wireless paradigm for microstimulation is via RF links due to the low absorption of these electromagnetic waves in the tissue. The size of the loop antenna, critically the receiver, can be the limiting factor in the case of RF-powered devices (Rabaey et al. 2011). Other important parameters are whether to use near-field inductive coupling (for increased power coupling efficiency, relevant at lower frequencies) or far-field coupling (larger distances at higher frequencies, but with lower coupling efficiencies) and the RF center and bandwidth (Poon et al. 2010). Heetderks in the 1980s analyzed this system and defined the characteristics of magnetically coupled power transfer systems (Heetderks 1988). Magnetic induction devices work best when the external exciter coil and the internal power receiving coil are in their near fields. This means that very small loop antennas associated with injectable-sized neurostimulators must be located relatively close to the body surface. Otherwise the nearly cubic rate of decay of the magnetic intensity in the loop far field requires substantial power generation and battery drain on the body surface exciter to overcome such losses.

It is worth noting that radio waves will be shortened when traveling through the tissue, allowing smaller antennas as compared to those used in air. This can become important at microwave frequencies above 300 MHz, reducing the antenna size of the microstimulator.

Direct conversion of microwave energy to generate neurostimulation electrical pulses in potentially injectable devices have been reported (Towe et al. 2012). These devices are basically small dipoles of about 1 cm length and 800 um or less diameter that are capacitively coupled by way of electric field coupling to a body surface antenna (Towe et al. 2012). These devices differ from more traditional methods of powering neuroimplants by low frequency magnetic induction coil systems that depend primarily on the magnetic field component of the EM wave to induce power into open-loop induction coils implanted in tissue.

#### Advantages

Powering implants at microwave frequencies through electric field coupling has been treated at length by Poon et al. who showed that efficient power transfer to moderately deep implanted devices can be accomplished at GHz frequencies (Poon et al. 2010). Emission of significant RF power from a body surface transmitter and radiating antenna is limited by most countries to specific frequency bands. In the USA unlicensed operation is permitted at 915 MHz and 2.45 GHz defined as the industrial, scientific, and medical (ISM) bands. These frequencies are consistent with efficiency as reported by Poon.

#### Disadvantages

Microwave penetration into the human body varies substantially depending on the tissue type and frequency. Tissue heating is the main concern at elevated microwave pulse powers needed to provide sufficient energy for conversion to neurostimulation pulses by an injectable size implant 5-7 cm deep. Such application might be, for example, for pain relief in the spinal canal. The specific absorption rate (SAR) is the primary indicator of microwave dosimetry in biological objects. The regulatory limit in the USA, for example, is 1.6 W/kg for 1 g of tissue (Poon et al. 2010). Since neurostimulation currents are pulsed and have low duty cycles (typically on the order of 1 %), the peak pulse power of microwave radiation can be relatively large, for example, in the 5-20 W and higher while maintaining an SAR-limited average power of less than a watt applied to the skin surface by an antenna over the implant. By using relatively high peak pulse powers in the tens of watts range, the microwave exciter antenna can be spaced away from the body by fractions of a meter (Towe et al. 2012) if the stimulation current demands of the implant are low, e.g., less than a milliampere with 250 μs pulse widths at 10 Hz.

#### Current State of the Art

In reported works, implanted microwave dipoles are generally untuned to the microwave frequency in order to achieve small size, and so they depend entirely on capacitive electric field coupling through the tissue. Reported versions have the dipole coupled to a low-threshold Schottky diode that rectifies the high-frequency microwave and turns it into a high-frequency pulsating DC that is functionally applied directly to the excitable nerves and tissues.

### Optical

Microstimulators using light in the near-infrared (NIR) spectrum (wavelengths on the order of 700-900 nm) can be used for wireless power transfer. NIR light has relatively low absorption and low scattering in the neural tissue, as compared to visible light, and it can be converted into electric currents with high efficiencies using semiconductors. Advantages of the optical method include the ability to focus light before the beam enters the tissue, which maximizes energy transfer, and that optical radiation frequency bands are not regulated by governments, unlike RF waves. Floating light-activated microelectrical stimulators (FLAMES) have been shown to produce functional forelimb forces via stimulation of the rat spinal cord gray matter (Abdo et al. 2011b). An optical stimulator with a fiber optic pigtail has also been demonstrated to generate muscle activation by stimulation of a frog sciatic nerve (Song et al. 2007). Fiber optic connections prevent classifying this approach as a wireless method; nonetheless a pigtail fiber connection has a potential to increase the number of optical channels easily without adding a new fiber for each additional stimulation channel. Active stimulators with optical links are still under development and have yet to be demonstrated in vivo (Freedman et al. 2011).

#### Advantages

Optical power transfer for passive neurostimulation is a relatively less complex method of energy transfer due to the lack of carrier-wave generating systems needed for radio and acoustic techniques. Laser diode infrared emitters and photodetectors are readily available. When powering neurostimulators placed within a few millimeters, they can be efficient in battery power consumption compared to other methods. The elimination of a rigid substrate and the electrode shank is a shared benefit by all wireless methods with a single stimulation channel. The FLAMES are reportedly in the submillimeter range and therefore suitable for CNS applications with minimal tissue replacement. Optical methods also allow addressing of multiple single-channel stimulators via wavelength selectivity.

#### Disadvantages

Light scattering in the tissue makes it difficult to activate a large number of disjoint stimulators. Passive stimulators of reported designs intrinsically generate monophasic pulses and so must either use classic capacitive coupling to perform electrode charge balancing or engage the use of an integrated circuit for more precise control which then entails additional complexity and requirements for its own powering. All reported passive wireless stimulators, including RF, optical, acoustic, and microwave excited, have an intrinsic orientation dependency in their coupling to the body surface exciters. For example, photodetectors must be oriented planar relative to the optical driving source. This requires that the physician would have to be aware of this with device implantation.

#### Current State of the Art

A small number of optically powered microstimulators are currently being researched. Wavelength-selective optical devices have reported efficiencies around 11 % (Abdo et al. 2011b) and 43 % (Song et al. 2007) for non-wavelength-selective devices.

#### Stimulation Depth

Although NIR light penetration into the neural tissue is greater compared to visible light, the light attenuation is still relatively strong. In the rat brain, the intensity was reduced to 1.85 % of the subdural intensity at 1,000 μm and to 0.15 % at 2,000 μm (Abdo et al. 2013). However, the prototype devices were able to stimulate the spinal cord gray matter and produce forces above 0.8 N (Abdo et al. 2011b) at an implantation depth of 2.35 mm. Scattering of light photons is the dominant effect that limits the penetration depth in the neural tissue, although the g factor (diversion of photon from straight path after scattering) strongly favors the forward direction (g=~0.9). The scattering coefficient, which indicates the number of scattering events on average per unit length of the photon path, is larger than a 100 per cm in the rat brain (Abdo et al. 2013). Thus, the primary limitation on stimulus power is the temperature increase near the surface where the photons enter the tissue. Direct measurements of temperature with a micro thermocouple sensor suggests that the power level used in the rat spinal cord (Abdo et al. 2011a) study would only cause a small fraction of a degree elevation in temperature.

### Acoustic

Ultrasonic techniques can efficiently transfer energy to a microstimulator through mechanical vibrations at high frequencies. The work by Towe et al. showed that millimeter-order sized chips of PZT were capable of intercepting sufficient ultrasound power to accomplish direct conversion of the energy to do neurostimulation through 7 cm or more of tissue while maintaining a form factor that was consistent with direct tissue injection of the neurostimulator (Towe et al. 2009). These systems depend on a body surface ultrasound exciter to generate the needed pulse characteristics modulated in the ultrasound carrier wave to evoke the desired neurostimulation waveform from the piezoelectric converter system in tissue.

#### Advantages

The advantage of acoustic powering of implants is that relatively high-energy densities can be carried by ultrasound waves compared to electromagnetic and optical techniques, and this improves power transfer and allows significant functional depths. This is because they are waves in matter rather than coupled through the electric or magnetic permeability of free space. High frequencies of ultrasound in the MHz region are associated with high accelerations but minuscule displacement of the tissue with the end result being significant momentum carried by the waves compared to a similar amount of power in generating, for example, an RF magnetic field. Another advantage is that unlike electromagnetic field frequencies below 400 MHz used for power and communication with conventional neurostimulators, ultrasound energy in the medical frequency region can be focused into tight beams by small acoustic lenses and hence has high-energy transfer efficiency in targeting a neuroimplant at a known location. This focality of ultrasound can thus result in reduced battery power consumption in some cases compared to lower-frequency magnetic induction techniques, particularly where the neural implant lies deep in the tissue to small devices where magnetic coupling techniques require relatively large coils.

#### Disadvantages

Ultrasound energy is strongly absorbed by the bone at MHz frequencies, and so ultrasound losses along the path length through the bone are severe amounting to 3-5 dB/cm (Wells 1977). The losses result from both absorption and scattering of the sound and can cause a disproportionate amount of the beam energy to be deposited in the bone leading to concerns about temperature rise. Ultrasound dosimetry is a limitation if trying to reach targets through the skull in the deep brain or possibly underneath the vertebrae to reach the spinal cord. Employing highly focal transducers to further reduce battery drain or reach deep implants is accompanied by the need to keep the skin surface or shallowly implanted transducer aimed with precision at the implant, a task which may not be practical in all patient applications where the patient wears the ultrasound exciter on his body. A potential method to overcome this problem was reported to be based on electronic feedback by way of monitoring remote skin potentials produced by an implanted neurostimulator and the use of an ultrasound scanning beam (Gulick and Towe 2012).

#### Current State of the Art

Conversion efficiency of ultrasound energy by piezoelectric elements integrated as part of a neurostimulator system can be on the order of 10 % or more (Ozeri and Shmilovitz 2010). Ultrasound attenuation due to viscoelastic path losses in the tissue amounts to 1-3 dB/cm at 1 MHz, depending strongly on the tissue along the beam path and the ultrasound frequency. Ultrasound path losses increase nonlinearly as the frequency increases (Wells 1977), and so optimal frequencies for power transfer are not the same as diagnostic ultrasound imaging but rather lower and typically in the range of 500 kHz to 1 MHz depending on the path length to the powered implant.

#### Stimulation Depth

The primary biological effects of concern using ultrasound as an energy transfer medium to an implant are those of tissue heating. FDA limits medical diagnostic ultrasound average intensities to 720 mW/cm$^2$ and peak pulse intensities to 190 W/cm$^2$ (Center for Devices and Radiological Health 2008). The safe peak pulse intensity specification means that very large 40-50 dB path losses of ultrasound can be tolerated since typically less than a milliwatt is required for neurostimulation.

## Concerns and Limitations of Wireless Microstimulators

### Tissue Heating

Essentially all reported energy transfer methods are limited in their ability to power deeply implanted device by their energy dissipation along the tissue path length. This dissipated energy can cause tissue heating. Tissue, especially water-rich tissues, such as the brain, muscles, or skin, can absorb and convert significant amounts of RF, optical, acoustic, or microwave energy into heat. It is imperative that the temperature at any focal point in the tissue is kept at minimal. A temperature elevation limit of 0.5 °C may be taken as a conservative value until more definitive data becomes available, based on the finding that 1 °C produces observable changes (Kiyatkin 2004).

### Material Safety

Like all biomedical implants, they must be hermetically sealed and not release any harmful chemicals into the tissue. Encapsulation and sealing sufficient to last a person's lifetime can add significantly to the overall size of these devices. Whether active or passive, implantable stimulators are typically made with semiconductors that may contain materials, such as copper oxides or arsenic oxides, and can be toxic. Solutions include encapsulation of the microstimulator with dielectrics, such as ceramics, glass, and/or organic polymers, such as parylene-C or polyimide.

### Channel Selection

Effective microstimulation for some applications such as brain-machine interfaces or spinal cord stimulators requires addressability of multiple different stimulation sites or channels. Optical energy transfer methods can permit some spatial directivity and by an ability to encode, over a range, in infrared wavelengths. Digital encoding for optical device selection is technically possible (Freedman et al. 2011). However, simplicity and some efficiency are lost with the requirement to divert received power to digital logic. Such digital approaches that incorporate microwave or ultrasound powering methods have not been reported so far. Multichannel ultrasound is enabled in principle by the ability to scan a focal beam over a tissue region and serially power multiple devices.

Bipolar Versus Monopolar

Wireless stimulators contain both cathodic and anodic contacts on the device, and thus the inter-contact distance is limited by the device size. Multichannel, active stimulators may have a few millimeters between the cathode and anode, and they may be considered to behave as monopolar stimulators especially if the reference electrode is significantly larger than the stimulating contacts. However, with single-channel passive devices, the primary objective is to make the device as small as possible, perhaps a few hundred microns, to minimize the tissue replacement. Thus, the cathodic-anodic contact distance is similarly very short, and the device should be considered as a bipolar stimulator.

Device Size

An ideal microstimulator would displace zero volume in the tissue. Realistically, a stimulator will use the minimum volume needed. In bipolar stimulation, the stimulation strength decreases steeply with distance from the electrodes (Sahin and Pikov 2011). However, simulations suggest that the structures in a volume that is in the same order as the size of the device should be feasible to stimulate while keeping the current in the microampere range (Abdo and Sahin 2011).

Device Geometry and Contact Placement

Large aspect ratios of device geometry should be preferred to maximize the anode-cathode separation and thus the stimulation effect for a given stimulus current. In a 3D device fabrication process, the contacts can be placed at each end of a longitudinal structure. This may rather be difficult to achieve using standard micro-fabrication processes that are typically applied on wafer-like structures. If the contacts have to be placed on the top surface of a rectangular prism with dimensions in proportions of 2:2:5, the stimulus strength reduces by 8.5 % compared to the case where the contacts are placed on each end (2:2 surfaces, (Abdo and Sahin 2011)). A high aspect ratio is also advantageous for insertion of the stimulator into the targeted neural tissue. Electronic device fabrication utilizes only a very shallow region of the semiconductor on the surface, allowing the total device thickness to be reduced down to 50 μm or less. Mechanical strength of the semiconductor substrate becomes the practical limiting factor that determines the maximum aspect ratio of the structure that renders the device too fragile under bending. In the case of acoustic devices, similar issues apply to piezoelectric ceramic materials like the PZT, in addition to the fact that the piezoelectric device geometry also determines its resonance frequency at which the energy harvesting efficiency is at a maximum. With all three energy transfer paradigms considered here, the device orientation inside the tissue with respect to the external energy source is critical in order to maximize the energy transfer.

Contact Material

The interface between the stimulator and tissue is a critical parameter because the available energy in a wireless microstimulator is always limited. Thus, materials with high charge-injection capacity are desired for the contacts of a wireless microstimulator. Examples include porous platinum or platinum-iridium alloys, sputtered or electroplated iridium oxide, titanium nitride, and conductive polymers like PEDOT (Cogan 2008). The area and the roughness of the electrode contacts also have a significant effect on the charge-injection capacity and typically have a nonlinear relationship between geometric area and the charge-injection capacity. For example, a sputtered iridium oxide microelectrode with a 200 nm surface coating has a charge-injection capacity of 5.3 mC $cm^{-2}$ when the area is approximately 2,000 $\mu m^2$ but decreases to 1.7 mC $cm^{-2}$ with a relationship that decays exponentially with increasing surface area.

Voltage Compliance

Simulations suggest that the output voltage of a microstimulator is an important parameter to optimize the energy transfer from the implant to the surrounding neural tissue (Abdo and Sahin 2011). There is a trade-off between the contact size and the required device output voltage. The electrode-electrolyte interface voltages at the contacts are in series to the voltage that develops across the tissue. The device output voltage (compliance) has to be large enough to accommodate the electrode impedance, which is inversely proportional with the contact size for a given material (i.e., charge-injection capacity), and the device current. Therefore, one can choose large contacts, allowing the device surface area to increase, in return for a smaller output compliance requirement. Thus larger contact sizes permit a smaller output voltage and so allow a larger stimulus current assuming the total stimulus energy is constant. Conversely, if the device electrode surface area has to be kept below a certain level, this dictates an upper bound for the stimulus current as well. In conclusion, the design of a microstimulator is an optimization problem that intertwines geometrical and electrical parameters of the device.

## Representative Examples

### RF (Inductive Coupling)

The BION™ neural stimulator (Loeb et al. 2001; Schulman 2008) (Kane et al. 2011) is an example of an active floating microstimulator that uses magnetic inductive RF coupling (2 MHz) to transfer energy and also incorporates pulse synthesis and decoding of channel data and for neural stimulation. It has an internal capacitor which integrates energy supplied continuously over time that is then metered out according to digital instructions. Externally, the exciter is characterized by a battery pack and relatively large exciter applicator compared to optical or ultrasound methods. The exciter is worn as a coil placed over the implant and connected to a remote battery pack. The diameter of the implant device is 2 mm and the length ranges from 10 to 16.7 mm. Current generations of the device have 256 individual channels that can be independently addressed and a custom ASIC as a pulse former. Stimulation parameters allow currents to be generated in the hundreds of microamperes, up to 40 mA, with voltage compliance values between 8.5 and 17 V. Using iridium and tantalum or platinum-iridium electrodes, the BION™ can deliver up to 50 pulses per second. A distinguishing feature of the BION™ stimulator is its hermetic sealing of either glass or ceramics. Accelerated moisture testing of recent devices shows excellent device reliability for use in multiyear implants.

### Optical (NIR Coupling)

The FLAMES (Abdo et al. 2011b) neural stimulator is a passive floating microstimulator that uses diffused near-infrared light to transfer energy for neural stimulation. The device size is approximately 500 μm × 200 μm and about 100 μm thick and is shown in Fig. 2. The devices can be addressed by using different wavelengths of infrared light. Stimulation parameters allow currents to be generated on the orders of tens to hundreds of microamperes with voltage compliance values below 2 V. Contacts have been made out of gold coated with PEDOT for good charge-injection capacity under zero bias conditions. Other contact materials and long-term encapsulation with parylene-C are currently being investigated.

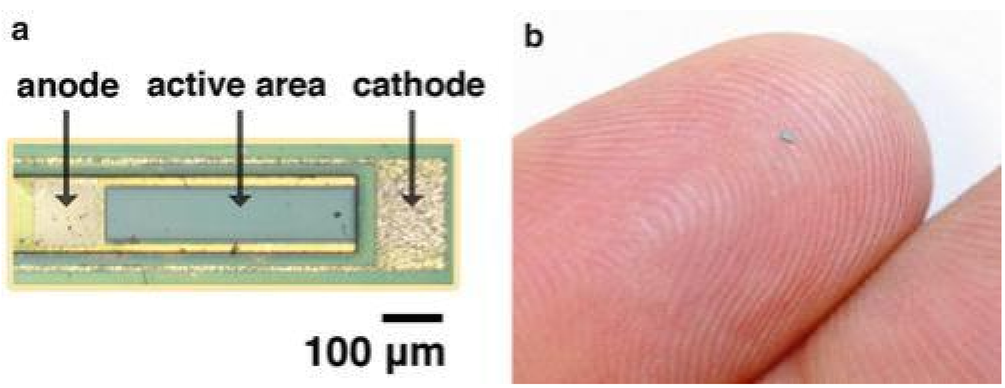


Fig. 2
(a) Micrograph of an optically powered microstimulator with the anode, cathode, and active area of the device shown. (b) Photo of the optical-powered microstimulator device

### Acoustical (Ultrasonic Coupling)

Ultrasonic-powered stimulators have been demonstrated (Towe et al. 2009). In this system, a 1 MHz ultrasonic emitter with a diameter of 26 mm at a distance of 12 cm was used. A piezoelectric receiver, made of PZT-5A, with a 1.13 mm diameter and 1 mm thickness, was used. In a Sprague-Dawley rat, currents on the order of a milliampere were generated with average powers between 10 mW/cm$^2$ and 147 mW/cm$^2$. Figure 3 shows a photograph of this type of device.

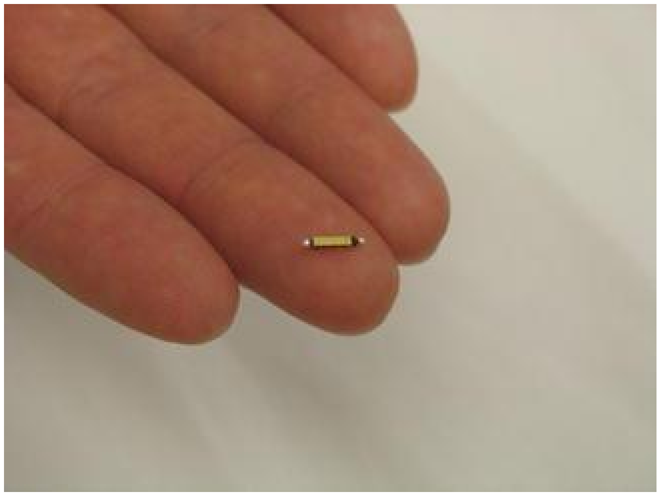

Fig. 3
Photograph of an ultrasound-powered microstimulator made from the piezoelectric plastic polyvinylidene fluoride (PVDF)

Microwave Coupling

Coupling through primarily the EM wave electric field component at microwave frequencies using a thin dipole antenna doubling as an electrode system was demonstrated in the stimulation of a rat sciatic nerve (Towe et al. 2012). An example of this type of device is shown in Fig. 4. This device has a size on the order of 800 μm in diameter and 1.5 cm length, and it was constructed using a low bias Schottky diode connected to an internal 100 μm diameter platinum wire dipole antenna. This device operates untuned in the ISM band at 915 MHz (US) or 2.45 GHz (international). It develops milliampere-order currents at 7 cm tissue depths when pulsed at 10 W peak.

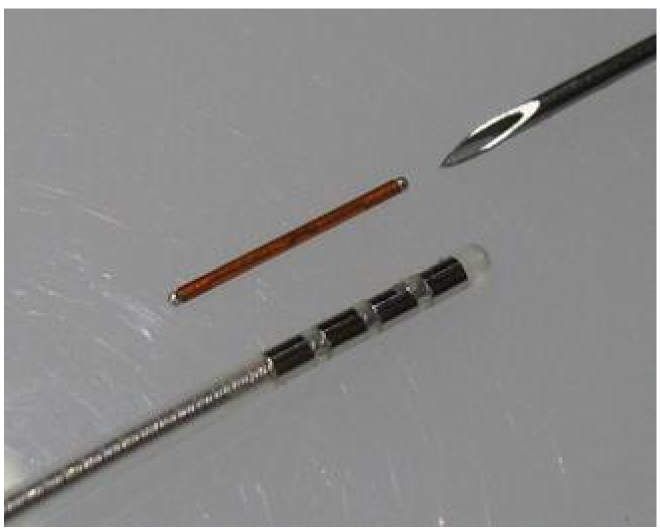

Fig. 4
Photograph of a 915 MHz microwave-powered passive neurostimulator (center) along with a 16-gauge syringe needle and a multielectrode catheter used for spinal neurostimulation (Medtronic)

## Cross-References/Related Terms

Brain Imaging and Optogenetics
Brain Machine Interface and Neuroimaging
Brain Machine Interface: Overview

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