



# Grounding of Implantable Medical Devices – A Tutorial

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*The material in this tutorial is based on Design Considerations for Ground Referencing in Multi-Module Neural Implant by Dorian Hacıy, Yan Liuy, Sara S. Ghoreishizadehy, Timothy G. Constandinouy and my own research. For more information, please write to [hosseinniri@hotmail.com](mailto:hosseinniri@hotmail.com).*

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**I**N The past six decades, implantable medical devices or systems have been advanced through developments in science and engineering, especially in microelectronics, biotechnology, and materials.

Jiang and Zhou have described that 8% to 10% of the population in America and 5% to 6% of people in industrialized countries have experienced an implantable medical device for rebuilding body functions, achieving a better quality of life, or expanding longevity. The vast majority of clinical grade IMDs are electrical stimulators. A well-known example is the cardiac pacemaker, which uses periodic electrical pulses as a stimulation means for heart contraction. More recently, neuromodulation devices such as cochlear implants for deaf people and deep brain stimulators (DBS) for Parkinson's disease and essential tremor (ET), have demonstrated significant impact to the quality of life of millions of patients. Besides these examples, implantable devices for treatment of Epilepsy have proven to be effective therapeutic solutions,

with vagus nerve stimulation (VNS) implants from Cyberonics playing a key role. Such neuroprosthetic IMDs are typically implanted in the upper chest and based on active electronics enclosed in a metallic or ceramic can, which houses also the battery. Implantable leads are used as means for reaching the area to stimulate and contact electrodes are responsible for delivering the electrical pulse to the target tissue.

## History of IMDs:

From Zoll's first report on electrical heart stimulation in 1952 to the first commercialized wireless blood pressure measurement system introduced by Cardiometrics in 2010, medical experts have exerted honorable efforts to improve the quality of patients' lives with various medical devices, such as the implantable cardiac defibrillator, cochlear implant, implanted bladder stimulator, and implantable wireless pressure sensor. Such implantable medical devices were developed to sense a physiological response in vivo or to actuate physiological organs. Recently, with the superminiaturization of electronic circuits and mechanical structures, many researchers have focused on the development of implantable real-time vital monitoring systems, which are continuously operated in subsecond periods. The implantable real-time vital monitoring devices may shift medical systems from

remedying incidents after they occur to self-managing incidents before they occur. For example, the EndoSure of Cardiometrics, which is inserted into the aorta to measure intrasac pressure during endovascular abdominal aortic aneurysm repair and during endovascular thoracic aortic aneurysm repair, can detect intraoperative leaks of the stent graft. After discharge from the hospital, patients can check the intrasac pressure daily at home, which was traditionally monitored by an angiographic method only at a hospital.

The first fully implantable pacemaker was encapsulated by araldite epoxy resin. In the polymer encapsulation, a low power consumption first-generation silicon transistor (OC460) integrated with electrical circuits was packaged with two sealed 60 mAh nickel-cadmium battery cells. The electrical circuit was connected with an inductive coil for recharging the batteries. The electrical-pulse-generating system was connected with electrodes via stainless steel wire covered with polyethylene. The total size of the first implantable pacemaker was 55 mm in diameter and 16 mm in thickness, similar to the dimensions of a shoe polish can from British Kiwi.

The epoxy resin encapsulation of the first-generation pacemaker was developed to protect the electronic circuits and batteries from body fluid penetration and tissue invasion to the system. Also, the encapsulation or housing worked as a barrier between the human body and hazardous batteries. However, the polymer-based epoxy resin swelled and dissolved inside the human body. Thus, the housing materials had to be replaced with ceramics and titanium to protect the electrical systems. Currently, the electric elements and a battery are packaged by a laser welding of titanium, which has a strong mechanical hardness, extreme resistance to corrosion, biocompatibility, and durability. Electrical connections between

the metal-housed electronic system and the lead wire of the pacemaker induced another sealing issue. Owing to the metallic housing, the electrical signal output part, which is connected to the lead, should be passivated by an insulator to prevent electrical connection to the metal housing. Therefore, polymer or ceramic-based electrical feedthrough was developed with perfect separation between the electronic system and the human body environment.

The first operation for implanting a pacemaker was done by the surgeon Ake Senning, who performed an open thoracotomy and suturing of 9 mm diameter electrodes on the left side of the myocardium. Afterward, the heart pace was electrically controlled for 3 hours. The next operation happened on following day and the device worked for 7 days. Currently, the pacemaker implant procedure takes around 1 to 2 hours with 4-5 mm diameter electrodes fixed on the right ventricle and the right atrium. One day after the operation, the patient is released from the hospital and can enjoy his or her regular life again with the help of the pacemaker. The small size and lightweight main body of the pacemaker allow the patient to avoid open-heart surgery, and the flexible platinum lead coils make an easy and short procedure possible.

This tutorial has been developed to help you understand what grounding in IMDs is, why it is important.

## What is an Implantable Medical Device (IMD)?

A medical device is defined as implantable if it is either partly or totally introduced, surgically or medically, into the human body and is intended to remain there after the procedure.

## Why grounding used in implantable medical devices?

1. Avoiding a DC voltage bias across the conductors in the implantable lead in Multi-Module Neural Implants.
2. Protection from electromagnetic Interference (Control EMI).

## Ground Referencing in Multi-Module Neural Implants:

The need for both stimulation and recording of the biosignal activity, often a requirement in modern therapeutic techniques, have caused a recent surge in the generation of closed-loop neuroprostheses. This relatively new category of implantable devices, in addition to stimulation, also introduces the need for front end sensing, signal conditioning, and real-time processing. Recently FDA-approved and now

clinically available examples of such devices include the closed-loop DBS Activa PC+S system from Medtronic and the responsive neurostimulation (RNS) system for refractory partial epilepsy implemented by NeuroPace. Closed-loop applications are now requiring more interfacing channels (both for recording and stimulation) and at multiple locations. This is posing significant challenges for the connectivity, for example, the number of implantable the connectivity, for example, the number of implantable leads, and conductors in

each lead, the number of electrical connections into a hermetic package, the challenge of trying to observe microvolt-level biopotential signals through length leads.

Additionally, surgical and safety considerations place additional requirements on performing certain functions in certain locations. for example, a chest-mounted IMD can occupy more volume (i.e. larger battery) and dissipate more power (i.e. more capacity for on-board processing) than one mounted in the head. To address this, a new generation of implantable medical devices is now emerging. Instead of the centralised approach that uses a single active implantable device, the system is now being partitioned and distributed across multiple active implantable devices each with specific functions, and located at different sites .This approach addresses some of the limitations of single-module implants, however it poses a new set of challenges, mainly related to inter-module connectivity, functional reliability and patient safety. The implantable leads are now additionally used to facilitate transfer power between modules and for bidirectional communication.

Moreover, having distinct active modules implies the need for electrical isolation from each other to avoid direct current paths between them. This means each implantable module has its own power domain, and therefore different reference potentials between the electronic circuits (module grounds).

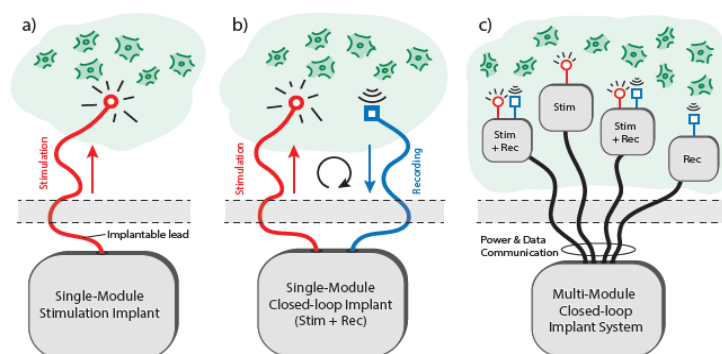


Fig. 1. Comparison between different IMD generations: (a) single-module pace-making stimulation implant, (b) single-module closed-loop stimulation and recording implant, and (3) multi-module closed-loop implant system.

## Multi-Module System Concept in Medical implant:

The typical configuration comprises of one **central implant**, or central unit, and **multiple smaller peripheral implants**, or peripheral units, each located in close proximity to the target interface sites. The former includes the energy source (battery), power management circuitry, a control and processing unit, and communication interfaces to transmit wireless data externally and to communicate with the peripheral units through an intrabody network. Although there are several research groups investigating different methods of implementing wireless intrabody networks, all clinically applicable devices currently require a wired connection between the individual modules.

We will thus focus only on a wireline connection. This is based on multi conductor implantable leads serving as communication channel (bidirectional) and power delivery medium (from central to peripheral). At the 'receiving' end of this network, the peripheral units must also contain an intra-body communication interface, to interact with the network and to retrieve power. This will power up: (1) the front-end function-specific units, which can be both recording and stimulation; (2) the data conversion circuitry (analog-to-digital and digital-to-analog); and (3) the digital processing unit (Fig. 2.c).

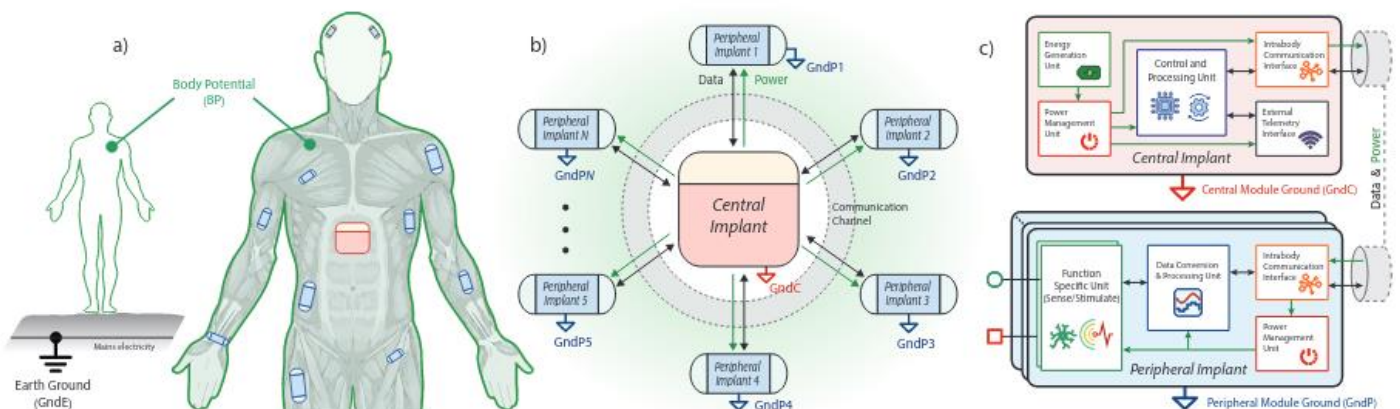


Fig. 2. Concept of the multi-module implantable system. (a) Conceptual body model with of a distributed implant system; (b) inter-module connectivity (one central implant, multiple peripheral implants) and; (c) block diagram of internal structure of both central and peripheral implants and their interfacing.

## What is Ground?

'Ground' has become such a common term in electrical and electronic circuits that even in electronic devices which have no real contact with the physical earth, the ground plane on a PCB or a circuit (with reference to which all voltages on a PCB are measured) is called the 'ground' of the circuit. This is a convenient way to tie all points with this circuit ground symbol together. voltage is a differential quantity. To measure the voltage of a single point, a reference point must be selected to measure against. This common reference point is called "ground" and **considered to have zero voltage**. For safe voltage levels or insulated enclosure, this does not have to be Earth ground. There can be more than one circuit ground, due to finite conductance in wires.

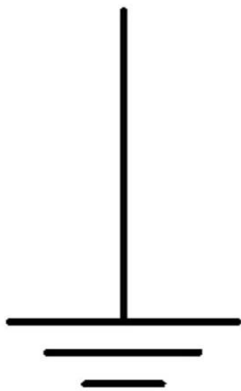


Fig. 3. Common ground symbol.

## Why it is necessary to common all the ground in circuit (groups of circuit need to interface with each other)?

It is necessary when there are groups of circuit need to interface with each other. This is obvious because ground is the reference. If the grounds are not connected together, ground actually loses its meaning.

Example for better understanding:

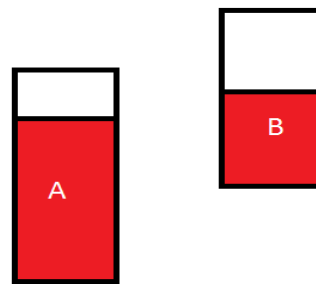


Fig. 4.

Consider this case, you have a liquid in jar 'A' and jar 'B'. Level of liquid in Jar B looks more isn't it? But the level of jar A is higher. So same thing happens with the potential difference in the circuit. If the reference is not common, the results may be undesirable. So now what to do to avoid this?

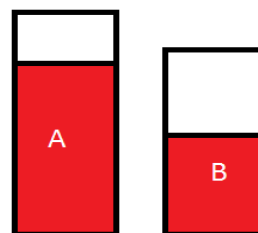


Fig. 5.

You bring both the jars at same level (You common the ground). Now can you say that the liquid level in Jar A is more than level in Jar B? You can compare it easily, in fact any other jar



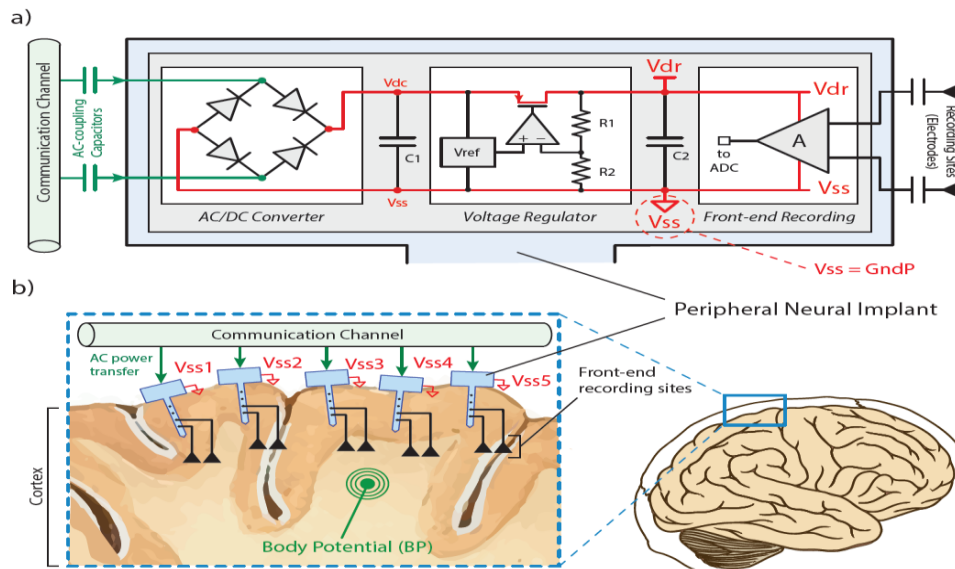


Fig. 6. Peripheral neural recording implant system: (a) power recovery and  $V_{dr}$  and  $V_{ss}$  DC rail generation; (b) distributed multi-module brain

kept at same level, you can compare the levels with these 2 jars. the exact same logic applies with the level of potentials in the circuit.

### Module Ground Referencing:

In order to avoid a DC voltage bias across the conductors in the implantable lead, and thus to maintain the charge balance over time, power transmission schemes based on AC coupling are typically required in wired implant systems. This precludes unwanted current paths from being formed between different modules if damage of the insulation occurs.

Accordingly, AC/DC power domain conversion is implemented at the peripheral side, generating a suitable DC supply voltage for the front-end circuitry ( $V_{dr} - V_{ss}$ ), hence the module ground rail ( $GndP = V_{ss}$ ), as shown in Fig. 6.a. This however, creates DC isolation between central and peripheral implants and therefore the electric potential of the module ground at the peripheral implant N ( $GndP_n$ ) is not referred to the module ground at the central implant ( $GndC$ ).

By definition, implantable neural systems are placed inside the human body, hence surrounded by biological tissue. Unlike beside hospital equipment, they are not directly connected to the AC electrical power coming

from mains (grid power), they instead receive the electrical energy from a battery enclosed within the implant or inductively from external Power sources. Unless the body touches the mains, the generated potential rails are to be considered floating and not referenced to the earth ground. The human body, however has its own electric potential, **Body Potential (BP)**, as shown in Fig. 2.a. In electrical equilibrium (no current flow within the body), such potential is considered the same over the whole body. The **BP** has to be taken into consideration when dealing with ground referencing in the neural implant system, as the potential displacement of  $GndC$  and  $GndP$  from BP may have significant impact on: (1) safety of the patient (2) device reliability, and (3) recording quality.

### Grounding Impact on Recording Quality, Device Reliability and Patient Safety:

Implantable devices that sense biopotential signals, for example neural recording systems, connect the sensing electrodes that are in direct contact with the tissue, i.e. thus with BP, to high impedance amplifier inputs. The observed biopotential thus consists of this large baseline (BP) in addition to a relatively weak AC signal (desired activity to be recorded). A differential front-end amplifier is typically used to record

such signals, for example, in intracortical recording: this sense and amplifies the difference in potential between two electrode, suppressing the common potential (common-mode rejection, CMR). Moreover, to deal with the unknown DC offset between the BP and recording circuit (power rails Vdr and Vss/GndP), biopotential amplifier inputs are typically AC coupled through capacitors. This leaves the BP having a relatively high impedance to the recording circuit power rail that effectively makes the tissue an 'antenna' for picking up common-mode noise. It is thus essential to maintain a relatively low impedance connection between the recording circuit power rails and BP to attenuate the observed common-mode noise (at the amplifier inputs). Furthermore, in terms of safety, a significant difference in electric potential between GndC/GndP and BP (voltage) may lead to corrosion/failure of encapsulation that would result in current flow into/from the tissue exceeding the safety limit defined by the international directives on active implantable devices.

For these reasons, an appropriate ground referencing scheme is required in order to ensure good recording quality, but also device reliability and ultimately patient safety.

Body referencing circuit >> BRC

### Ground Referencing Schemes:

Given that the central implant is enclosed within a case serving as DC electrical screen from the body, with no direct contact between internal circuitry and body tissue, there is no need for GndC to be referenced to BP. On the other hand, the peripheral implants have a direct path to the tissue through the recording electrodes (Fig. 6.b), hence the need for proper referencing of the DC rails to the body potential.

The **most common** and **simple** ground referencing configuration is the passive connection of a peripheral device power rail (either Vdr or Vss) to the body potential. In acute experiments, the peripheral module ground GndP is often shorted directly to the BP

through a ground electrode, so to reference the tissue to the lowest potential of the module. The

**passive scheme** is represented in (Fig. 7.a) with a resistive divider. The 'short' is obtained from this when considering either  $R1 = 0$ ,  $R2 = 1$  ( $BP = Vdr$ ), or vice-versa ( $BP = Vss$ ). Depending on the initial value of the body potential, compared to the shorted rail, there will be current flow from/to the tissue in order to equilibrate the potential between the two nodes. In the scenario of multiple peripheral implants close to each other, it is usually preferred to short the BP to all the peripheral grounds (GndP1, ...GndPN). The **second category** of body referencing circuits (BRC) is composed of active components. This can be further divided into two topologies: (1) driving BRC; and (2) sensing BRC. The driving reference circuit is shown in Fig. 7.b consisting of a reference voltage generation circuit (e.g. a resistor divider) and an output buffer with a low output impedance to drive the BP. In both passive and drive schemes, there is current flow between implant and tissue. This ensures the voltage between BP and GndP to be fixed, thus reducing the common-mode signal at the input of the recording amplifier, at the expense of current exchange between peripheral module and body. Unlike for the passive scheme, the BRC with driver/buffer ensures that the current is always injected into or out of the tissue providing a level of stability to the source (peripheral implant) from inherent fluctuations of the potential around BP. Fig. 7.c presents a **third scheme**, based on a **sensing BRC**: this time, a front-end sensing circuit is used as body reference circuit to measure the BP instant value and references the peripheral rails level accordingly. The sensed voltage ( $BP - GndP$ ) can thus be sent to the recording amplifier, which uses it to adjust the input biasing point of the latter. By doing so, the body potential fluctuation itself, seen as common mode from the recording input electrodes, is employed in fact as part of the common-mode rejection of the amplifier.

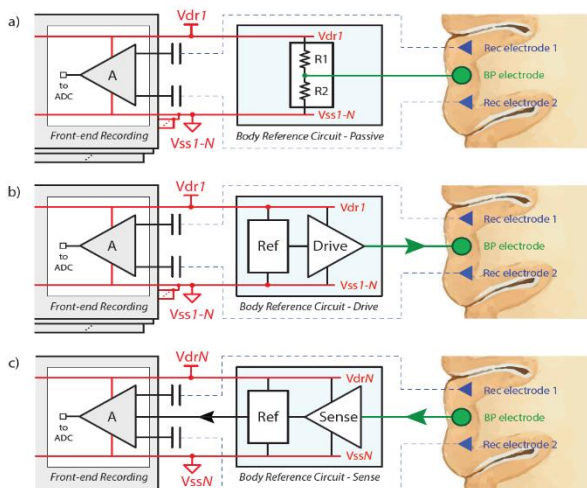


Fig. 7. Different grounding schemes: (a) passive; (b) drive; (c) sense.

## ESD (Electrostatic Discharge):

Electrostatic discharge (ESD) is the sudden flow of electricity between two electrically charged objects caused by contact, an electrical short, or dielectric breakdown. A buildup of static electricity can be caused by tribocharging or by electrostatic induction. The ESD occurs when differently-charged objects are brought close together or when the dielectric between them breaks down, often creating a visible spark.



Fig. 8. ESD

## ESD protection:

Protecting against the effects of ESD, **Electrostatic Discharge** is essential when dealing with electronic technology - from the home hobbyist to full scale manufacture ESD protection is very important.

## ESD protection requirement:

With electronics devices used in today's electronics equipment being susceptible to electrostatic discharge, it is necessary to employ ESD design guidelines that ensure that devices used will be protected against its effects. The ESD design guidelines and the protection used is of particular importance where any connections are on the periphery of the equipment and may be accessed via the user.

When accessing external ports, users will not take any precautions against ESD, if they even understand about it. therefore it is necessary to provide full protection for any external ports that may exist.

Electronics devices manufactured today are often required to survive a discharge of 8kV contact discharge (i.e. where the 8 kV is discharged directly onto the pin via a metallic contact) or a 15 kV air discharge (where the 15 kV point is close to the pin and discharges across an air gap). While this is the aim, not all devices will survive this and in many cases the discharge may be greater than this. It is therefore wise to add additional protection.

## ESD design methods:

The key to the ESD design guidelines for protecting the devices on any external Input / Output (I / O) lines, is to prevent the voltage rising above a level that will damage the interface device. This may be achieved using a circuit that clamps the maximum voltages to just outside the maximum operating extremes. typically this may be just above the rail voltage and just below the zero volt line.

A typical circuit that can be used for clamping voltages employs reverse biased diodes from the input line to the voltage rail and to ground. This ESD protection circuit must ensure that the



voltage excursions on the input line are limited. The diodes must also have a low level of residual current, and the capacitance must be low to ensure that the frequency response / data rate and other input parameters are not impaired.

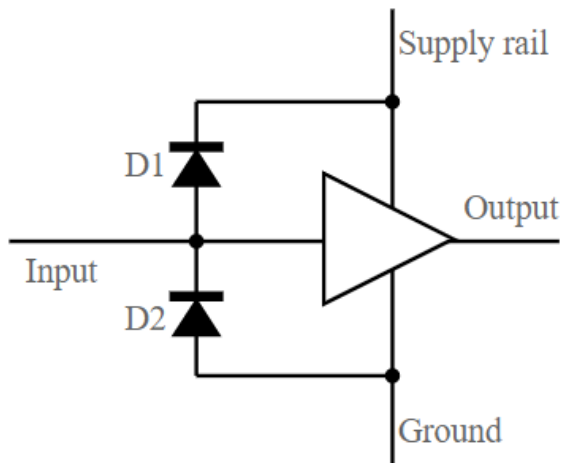


Fig. 9.- Diodes used on the circuit input for protection

The operation of the circuit is very simple in that the diodes, D1 and D2 are reverse biased under normal operating conditions. However if a pulse occurs that raises the input voltage above the rail voltage the top diode, D1, will conduct. Similarly if the voltage falls below the ground voltage, the other diode, D2, will conduct. Using ordinary signal diodes, the maximum voltage excursions that might expected on the input line in the first analysis may be +0.5V above the rail and -0.5V below ground. However this is not always the case as seen below.

The typical response curve for an electrostatic discharge is defined by IEC61000-4-5 and it simulates a typical electrostatic discharge curve. The waveform has a rise time of about 1 ns and the current level peaks at 30A. To suppress these voltages, very effective clamping circuits are required and ESD design guidelines need to specify acceptable components and performance limits.

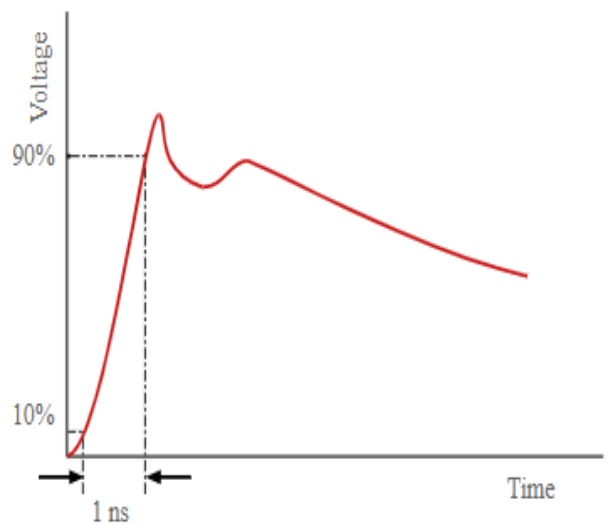


Fig. 10. - IEC 61000-4-2 pulse waveform used for ESD simulation

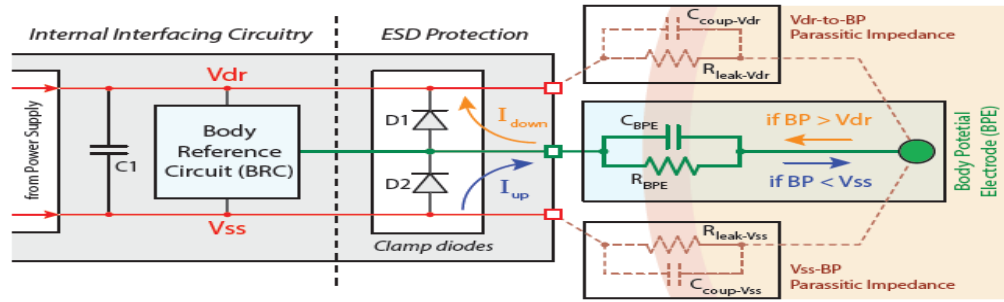


Fig. 11. Practical implementation interfacing of the body reference circuit (BRC) with the body potential (ESD, parasitic impedances, and BPE model).

### Practical implementation of RBCs interfacing with the body potential:

In the previous sections the interfacing between the RBCs and the body has been considered as ideal, with negligible parasitic elements and considering an ideal body potential electrode (BPE). When practically implementing the circuits in a specific technology and connecting them to the biological tissue, new challenges arise due to: (1) electrode interface (metal-tissue) non-idealities and; (2) circuit implementation constraints.

The electrical model of the electrode needs to be taken into account when evaluating the impact of the ground referencing in the different schemes, both for recording quality and safety limitations of current injection. A simplified model is presented in Fig. 11. Also, capacitive coupling and current leakage from the power rails, represented by  $R_{leak}$  and  $C_{coup}$ , affect the implant-tissue interfacing. This depends mainly on encapsulation and circuit technology.

The latter introduces another interfacing aspect: typically, in CMOS technology, integrated circuits require electrostatic discharge (ESD) protection at the interfacing pads, in order to prevent device malfunction or breakdown. This is usually implemented using reverse-biased pn-junction diodes, which ensure the potential at the pad doesn't exceed the upper and lower limits of the power rails ( $V_{ss} < V_p < V_{dr}$ ). Such diodes (D1 and D2 in Fig. 5) add

to the parasitic impedances of the implant-tissue interface, nevertheless, they prove to be advantageous in clamping the body-potential within the implant rail-to-rail range by providing a 'weak' leakage path.

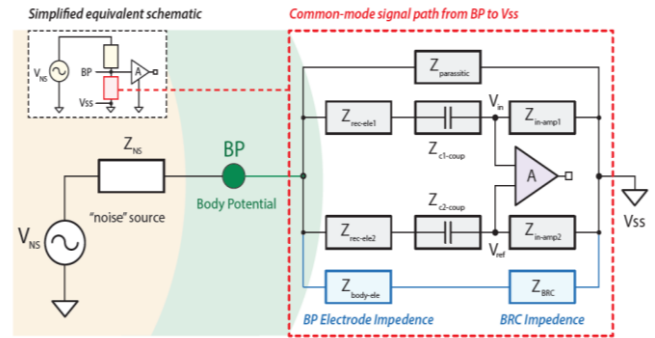


Fig. 12. Common-mode signal path in an experimental setup. Noise sources ( $V_{NS}$ ) are observed at the recording amplifier inputs via stray noise coupling impedance ( $Z_{NS}$ ). The purpose of the body reference circuit is to shunt this common-mode signal to recording amplifier ground such that it is attenuated at the recording amplifier input. There is essentially a potential divider between  $Z_{NS}$  and ( $Z_{BODY-ELE} + Z_{BRC}$ ). It is therefore essential to maintain a low impedance BP drive relative to the recording circuit ground.

### Recording Performance:

The three ground referencing schemes proposed in the previous section give a different contribution to the neural recording performance depending on the RBC implementation. As shown in Fig. 12, where a schematic of BP to  $V_{ss}$  path is illustrated, the BRC

and **BPE** impedances intervene in the modification of such BP to  $V_{ss}$  total impedance. By maintaining the latter a low impedance path, thus reducing the **BRC** common-mode gain  $A_{BRC-CM}$ , the common-mode fluctuation seen at the recording amplifier inputs are minimized. However due to the frequency response of the **RBC**,  $A_{BRC-CM}$  may increase with frequency and possibly reach to its maximum theoretical value of 0 dB (open circuit), i.e. the common-mode noise add directly on the recording inputs ( $V_{in}$  and  $V_{ref}$ ) through the recording path. This is represented in the example plot in Fig. 13 by the effective CMRR at frequency  $f_2$ . On the other hand, a zero impedance BRC (short circuit) would reduce 0 the  $V_{NS}$  potential division on BP, thus cancelling out the common-mode signal. This scenario is theoretically possible, however, due to parasitic and electrode impedances, it remains a practical implementation challenge.

### Safety Considerations:

Besides altering the neural recording performance, the chosen grounding scheme and the technology features are key aspects for body safety requirements, since they define the electrical behaviour at the implant-body interface (BP - GndP potential difference and current flow). Except for its intended function, active IMDs are required to be electrically neutral when in contact with the body and the maximum current density at the surface of any electrode is set to  $<0.75\mu A/mm^2$  according to the directive in. For this reason, current limiters are necessary when employing driving ground referencing topologies, alongside with implant encapsulation and implant-to-implant electrical isolation.

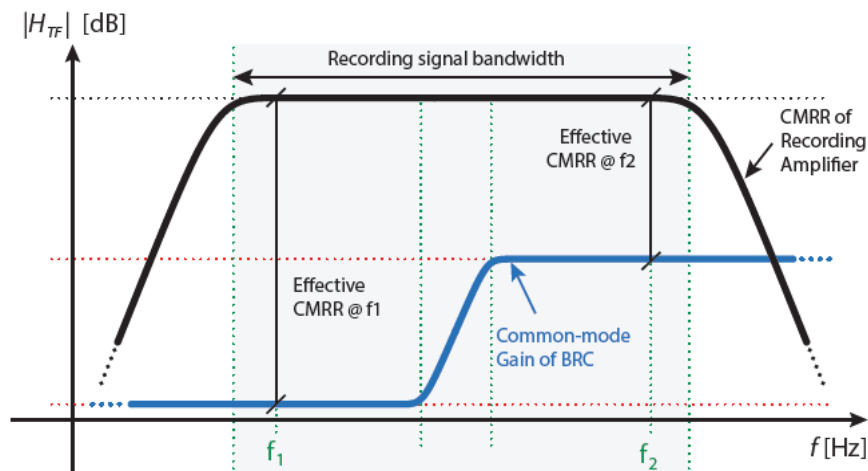


Fig. 13. Common-mode Rejection Ratio (CMRR) of Recording Amplifier and Common-mode gain of Body Reference Circuit (BRC) with the effective observable CMRR being the difference between these responses. The BRC configuration, i.e. impedance/drivability of BP relative to recording amplifier ground, determines the magnitude and frequency dependence of the common mode signal observed at the amplifier inputs ( $V_{in}$  and  $V_{ref}$ ). This is critical for experimental design.

**EMI Definition:** Electromagnetic interference (EMI), also called **radio-frequency interference (RFI)** when in the radio frequency spectrum, is a disturbance generated by an external source that affects an electrical circuit by electromagnetic induction, electrostatic coupling, or conduction. The disturbance may degrade the performance of the circuit or even stop it from functioning. In the case of a data path, these effects can range from an increase in error rate to a total loss of the data. Both man-made and natural sources generate changing electrical currents and voltages that can cause EMI: ignition systems, cellular network of mobile phones, lightning, solar flares, and auroras (northern/southern lights). EMI frequently affects AM radios. It can also affect mobile phones, FM radios, and televisions, as well as observations for radio astronomy and atmospheric science

### **What is EMI and Where Does It Come From?**

EMI occurs when a device's function is affected by the electromagnetic fields generated by a nearby device. These fields can be traced to a wide variety of electrical and magnetic sources, as well as some non-electrical sources. Since EMI can be radiated, it can travel through the air and originate from many possible sources, including everyday consumer devices, such as cell phones, microwaves, radio frequency identification (RFID) equipment, anti-theft devices, and metal detectors. People also can be exposed to EMI in work environments and through industrial sources, such as arc welding, electric and combustion motors, radio towers, power supplies, and electron microscopes. Environmental factors, such as lightning and solar flares, also can generate EMI. EMI generated by medical procedures is another concern. Dental equipment, transcutaneous electrical nerve stimulation (TENS), neurostimulation, magnetic resonance imaging (MRI), and defibrillators all can create

EMI; the latter two, in particular, can cause intense EMI that is very difficult to guard against.

External defibrillation involves the administration of a 200+ volt shock to a patient's chest in an attempt to halt fibrillations and restart the heart with a normal rhythm. The shock is delivered in short pulses, which can travel through the patient's tissues and accidentally wreak havoc on any implanted devices. In the case of ICDs, this high-voltage spike actually stems from the device itself. By the same token, MRI equipment creates a magnetic field of up to three teslas (3T), equal to the strength of 600 common refrigerator magnets. Magnetic fields are dangerous to any kind of electronics, and can wipe out the digital memory of devices exposed to them. Thankfully, implantable devices can be designed to survive — and even continue operating normally — during intense voltage spikes and magnetic pulses.

### **How Can a Circuit Be Protected From EMI?**

A conductive case, known as an EMI shield, can prevent radiated EMI from reaching a device. However, since medical devices are not closed systems, EMI shields must have openings that allow the shielded device to transmit signals or deliver treatment. Unfortunately, the wires used to transmit data and affect treatment can act as antennas, beaming harmful noise signals directly into devices, mixing with the intended signals, and resulting in potentially disastrous consequences.



Figure 14 — EMI shields, like this shield EMI shield for a power supply, are designed to prevent electromagnetic radiation (EMR) from escaping and acting as EMI in neighboring devices.

Implantable devices cannot be completely shielded from their environment, as many of them must interact with the body by sensing and distributing electrical impulses. Others even need the capability to interact with external devices. In some such cases, external adjustments to implantable devices are made using an external magnetic field or an electrical signal, so treatment can be adapted to the patient's needs. For example, some neurostimulators can be recharged wirelessly through the skin, and the frequency and strength of the pulses used to treat chronic pain can be controlled by patients using a handheld device.

Since implantable medical devices need to be both well-protected from EMI and able to send and receive signals from their environment, device engineers must employ EMI filters capable of separating the signals from the noise.

### How Can We Filter EMI?

Filtering can either be active or passive. The simplest form of filtering, suitable for most of the high-frequency noise present in the environment, is passive filtering, generally achieved through the use of a capacitor.

A capacitor can filter electromagnetic noise by absorbing and smoothing out an incoming signal. High-frequency changes in voltage

quickly charge and discharge capacitors and, in the process, cause the high-frequency noise to interfere with itself and cancel out the disruptive signal. Absorbing this energy to ground can neutralize or block certain frequencies from passing through a circuit.

When a capacitor is embedded in an EMI shield, feedthrough filters typically are the EMI filter of choice. A single feedthrough filter is usually round and shaped like a donut so that a lead, which carries the signal, can pass straight through the capacitor, while the capacitor's exterior is attached to the EMI shield. Filtered feedthroughs have a low equivalent series resistance (ESR) — one measure of the energy that the filter dissipates — and can be hermetically sealed and designed for high or low voltage, depending on the application.

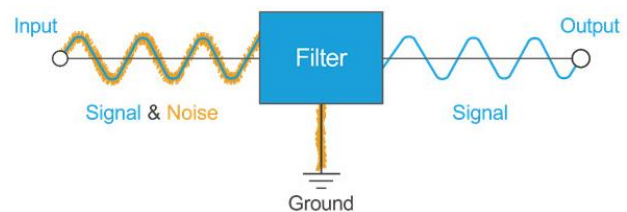


Figure 15— This diagram illustrates the basic concept behind passive EMI filtering: A low-frequency signal is passed through while high-frequency noise is removed.

Active filters comprise multiple active and passive circuit components, such as capacitors and operational amplifiers. Active filters, while suitable for many applications, require a power source to operate and can be limited in their upper frequency, so they are not commonly chosen for filtering in implantable medical devices.

Filtering also can be performed programmatically by initiating mathematical operations on the received signal to separate the signal from the noise. However, this can be computationally expensive and require larger circuitry and batteries to support the processing power. In addition, each signal line requires more computational processing, whereas



multiple signal lines can be filtered through a single capacitor array. Often, a physical filter and computational filtering will be combined to optimize effectiveness and efficiency.

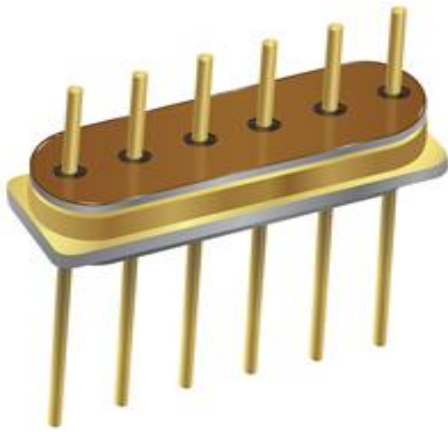


Fig. 16 — Six signal lines filtered with a capacitor array. Each line has its own separate capacitor built into the array so that each line can be filtered individually. The capacitor grounds are connected to the metal feedthrough surrounding the capacitor

#### Additional Concerns for Filtering in Implantable Devices:

Since many implantable devices are life-sustaining, they can be difficult to insert or remove, making the cost of device failure extremely high; and, if the device does fail, it is not easy to replace. This alarming concern brings to focus the issue of battery life. A dead or weakened battery can require removal of the current device and implantation of a new one, which can result in a lengthy, costly process in terms of both the number of doctor visits and overall patient health.

Selection of a feedthrough filter has an impact on battery life that can be twofold. To start, a very tiny amount of current always is flowing between the plates of a charged capacitor. Since one capacitor pole handles signal and the other is connected to a ground, leakage current can drain the battery over time. A strong dielectric of appropriate thickness has the ability to resist this flow of current as much as possible, significantly lessening battery drain.

In addition, filters in implantable medical devices should be designed to minimize loss of the intended signal. A filter's insertion loss measures how much a signal will be reduced or lost at each frequency. A good filter will have high insertion loss for noise frequencies and low insertion loss for the signal frequencies. Due to unavoidable effects, such as the internal resistance and inductance of the filter, some energy from the intended signal always will be lost, meaning device batteries need to work harder to achieve the same effect. However, careful design can minimize this energy loss. Less power dissipated means that less energy is demanded from the battery, extending battery life and improving efficiency.

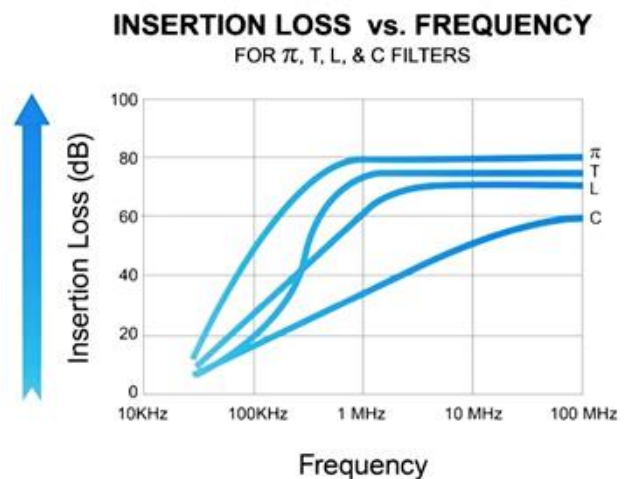


Fig. 17 — Example of insertion loss vs. frequency for several types of feedthrough filters. These filters are designed to have high insertion loss for high frequency signals, thereby blocking these signals (also known as radio frequency [RF] waves).

Through a combination of effective EMI shielding and filtering, implantable medical devices can be designed to withstand most daily exposure to EMI. If the protection is not implemented properly, the device will remain vulnerable to malfunction, damage, or a reduced lifespan. Careful EMI shield and filter selections and design-ins can optimize functionality and maximize the overall lifespan of the device.

## References

[1- ESD Design Guidelines: for electronic circuits](#)

[2- Development of Implantable Medical Devices: From an Engineering Perspective](#)

[3- Protecting Implantable Medical Devices From Electromagnetic Interference](#)

[4- Design Considerations for Ground Referencing in Multi-Module Neural Implants](#)

[5- Why it is necessary to common all the ground in circuit?](#)

[6- Using Grounding to Control EMI](#)

[7- Electromagnetic interference](#)