Design Considerations for Ground Referencing in Multi-Module Neural Implants

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Abstract—**Implantable neural interfaces have evolved in the past decades from stimulation-only devices to closed-loop recording and stimulation systems, allowing both for more targeted therapeutic techniques and more advanced prosthetic implants. Emerging applications require multi-module active implantable devices with intrabody power and data transmission. This distributed approach poses a new set of challenges related to inter-module connectivity, functional reliability and patient safety. This paper addresses the ground referencing challenge in active multi-implant systems, with a particular focus on neural recording devices. Three different grounding schemes (passive, drive, and sense) are presented and evaluated in terms of both recording reliability and patient safety. Considerations on the practical implementation of body potential referencing circuitry are finally discussed, with a detailed analysis of their impact on the recording performance.**

I. INTRODUCTION

In the past few decades, Implantable Medical Devices (IMDs) have shown remarkable success both as therapeutic devices in treating debilitating conditions, and as prosthetic devices, in restoring functions by bypassing dysfunctional organs/pathways. 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 [1]. 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 [2], [3]. 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 [4]. 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.

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. [5], [6]. 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 [7], [8].

Closed-loop applications are now requiring more interfacing channels (both for recording and stimulation) and at multiple locations. This is posing significant challenges for
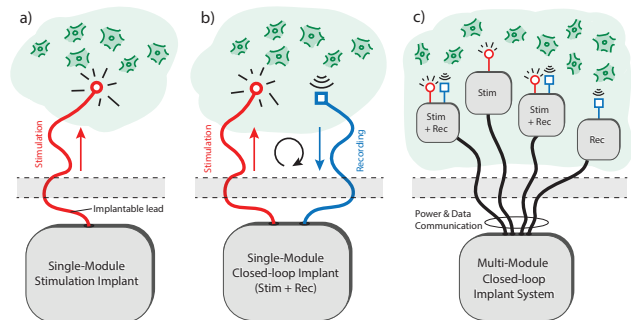

Fig. 1. Comparison between different IMD generations: (a) single-module pace-making stimulation implant, (b) single-module closed-loop stimulation and recording implant, and (3) multi-module closed-loop implant system.

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, e.g. [9], [10]. 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 [11].

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).

This paper identifies the ground referencing challenge in multi-module active implants and discusses three different grounding configurations for distributed neural implants, with particular focus on their impact to the recording performance and on device safety. The remainder of this paper is organised as follows: Section II describes the multi-module implant configuration, discussing key challenges when dealing with multiple power domains in an implant; Section III presents three different grounding schemes for dealing with this; Section IV analyses these (practical considerations, recording
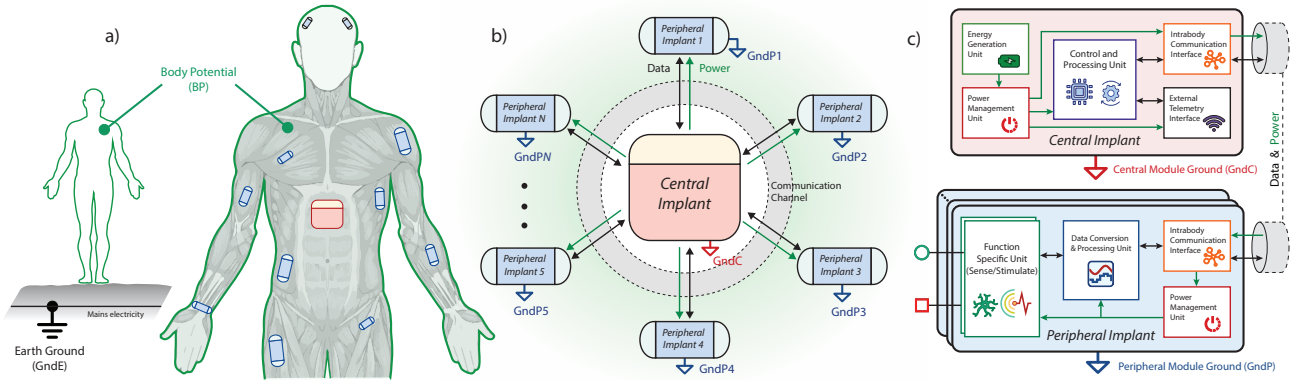
Fig. 2. Concept of the multi-module implantable system. (a) Conceptual body model with of a distributed implant system; (b) iter-module connectivity (one central implant, multiple peripheral implants) and; (c) block diagram of internal structure of both central and peripheral implants and their interfacing.

performance, safety); and Section V concludes the paper.

II. MULTI-MODULE ACTIVE NEURAL IMPLANTS

The system concept of the multi-implant system is illustrated in Fig. 2.b. 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).

A. Module Ground Referencing

In order to avoid a DC voltage bias across the conductors in the implantable lead [12], 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 (Vdr - Vss), hence the module ground rail (GndP = Vss), as shown in Fig. 3.a. [13]. This however, creates DC isolation between central and peripheral implants and therefore the electric potential of the module ground at the peripheral implant N (GndPn) 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 bedside hospital equipment, they are not directly connected to the AC electric power coming from the 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 a significant impact on: (1) safety of the patient (2) device reliability, and (3) recording quality.

B. 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 senses and amplifies the difference in potential between two electrode, while 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 rails 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 limits 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.
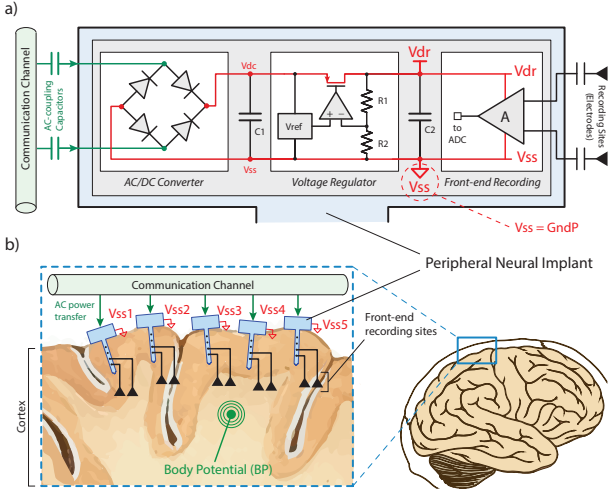
Fig. 3. Peripheral neural recording implant system: (a) power recovery and Vdr and Vss DC rail generation; (b) distributed multi-module brain implants.

III. 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. 3.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 V_{dr} or V_{ss}) 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. 4.a) with a resistive divider. The 'short' is obtained from this when considering either $R_1 = 0$, $R_2 = \infty$ ($BP = V_{dr}$), or vice-versa ($BP = V_{ss}$). 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. 4.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. 4.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
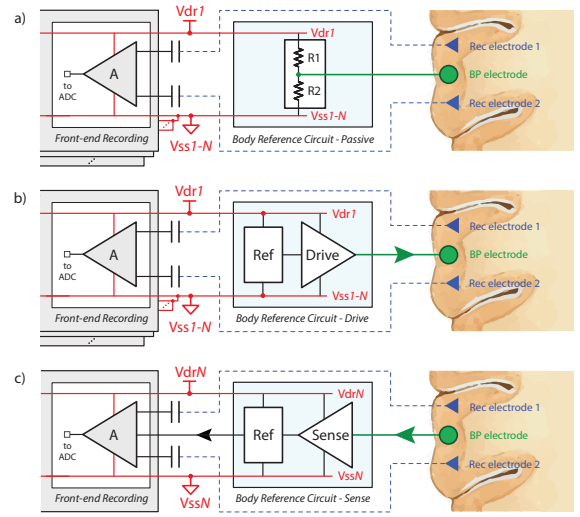

Fig. 4. Different grounding schemes: (a) passive; (b) drive; (c) sense.

($BP - GndP$) 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.

IV. DISCUSSION

A. Insights to Practical Implementation

So far in the paper, the interfacing between the BRCs 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. 5. 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 ($Vss < V_p < Vdr$). Such diodes ($D1$ and $D2$
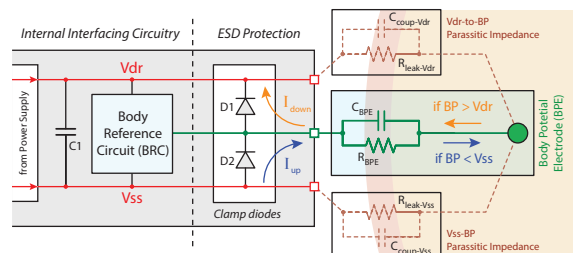

Fig. 5. Practical implementation interfacing of the body reference circuit (BRC) with the body potential (ESD, parasitic impedances, and BPE model).
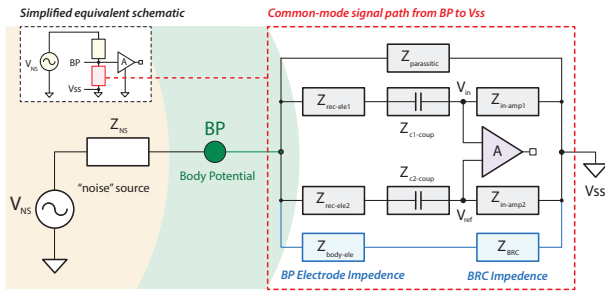
Fig. 6. Common-mode signal path in an experimental setup. Noise sources (V_{NS}) are observed at the recording amplifier inputs via stray noise coupling impedance (Z_{NS}). The purpose of the body reference circuit is to shunt this common-mode signal to recording amplifier ground such that it is attenuated 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.
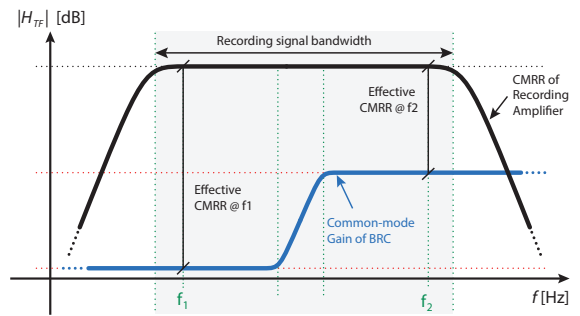

Fig. 7. Common-mode Rejection Ratio (CMRR) of Recording Amplifier and Common-mode gain of Body Reference Circuit (BRC) with the effective observable CMRR being the difference between these responses. The BRC configuration, i.e. impedance/drivability of BP relative to recording amplifier ground, determines the magnitude and frequency dependence of the common-mode signal observed at the amplifier inputs (Vin and Vref). This is critical for experimental design.

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.

B. Recording Performance

The three ground referencing schemes proposed in the previous section give a different contribution to the neural recording performance depending on the BRC implementation. As shown in Fig. 6, where a schematics of $BP\ to\ Vss$ path is illustrated, the BRC and BPE impedances intervene in the modification of such $BP\ to\ Vss$ total impedance. By maintaining the latter a low impedance path, thus reducing the BRC common-mode gain A_{BRC-CM}, the common-mode fluctuations seen at the recording amplifier inputs are minimized. However due to the frequency response of the BRC, 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. 7 by the effective $CMRR$ at frequency f_2. On the other hand, a zero impedance BRC (short circuit) would reduce to 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.

C. 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 $\leq 0.75\ \mu A/mm^2$ according to the directive in [14]. For this reason, current limiters are necessary when employing driving ground referencing topologies, alongside with implant encapsulation and implant-to-implant electrical isolation.

V. Conclusion

This paper has described the key design considerations in developing next generation multi-module neural implants. Such systems need to cross-reference isolated power supplies, in addition to facilitating a stable body potential with respect to the power rails of the individual modules. Three different grounding schemes have been identified based on *shorting*, *sensing*, and *driving* the body potential. The function of these schemes have been analysed by considering the coupling of common-mode interference to the recording inputs. It is revealed how ensuring a relatively low impedance path to the recording circuit ground has the effect of attenuating the common-mode observed at the electrode inputs, thus improving recording quality. Finally, the practical and safety considerations are assessed – in particular, due to the fact they exchange electrical current with the body, it is essential to guarantee operation within safe limits.

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