Topical Review

Electrical connectors for neural implants: design, state of the art and future challenges of an underestimated component

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Abstract

Technological advances in electrically active implantable devices have increased the complexity of hardware design. In particular, the increasing number of stimulation and recording channels requires innovative approaches for connectors that interface electrodes with the implant circuitry.

Objective. This work aims to provide a common theoretical ground for implantable connector development with a focus on neural applications.

Approach. Aspects and experiences from several disciplines are compiled from an engineering perspective to discuss the state of the art of connector solutions. Whenever available, we also present general design guidelines.

Main results. Degradation mechanisms, material stability and design rules in terms of biocompatibility and biostability are introduced. Considering contact physics, we address the design and characterization of the contact zone and review contaminants, wear and contact degradation. For high-channel counts and body-like environments, insulation can be even more crucial than the electrical connection itself. Therefore, we also introduce the requirements for electrical insulation to prevent signal loss and distortion and discuss its impact on the practical implementation.

Significance. A final review is dedicated to the state of the art connector concepts, their mechanical setup, electrical performance and the interface to other implant components. We conclude with an outlook for possible approaches for the future generations of implants.

Keywords: biomedical microsystem, brain–machine interface (BMI), multichannel connector, electronic contact, implantable hardware, neural prosthesis

(Some figures may appear in colour only in the online journal)
1. Introduction

Since the first fully implantable cardiac pacemaker in 1958 [1], electrically active implantable devices (AIMDs) have become increasingly predominant in the diagnosis and treatment of many previously intractable diseases [2]. Nowadays, their application is beyond cardiology, with neuronal implants targeting the central and the peripheral nervous system. They help restoring body functions such as hearing (cochlear implants, CIs), vision (retinal implants), and motor functions (prostheses, functional electrical stimulation (FES)) [3–5]. They are also designed to alleviate symptoms of a variety of diseases such as hypertension [6], chronic pain, epilepsy, Parkinson’s disease, and psychiatric disorders [7].

Generally, the design of AIMDs includes three main components:

- Electrodes interfacing the target tissue.
- Implantable pulse generator (IPG) with controlling electronics and energy supply systems, including batteries and telemetric setups. Its size and accessibility often require implantation at a location distant to the stimulation electrode.
- Cables electrically connecting the electrodes with the stimulation unit. Their length ranges from about 4.5 cm in cochlear implants [8] to about 45 cm in deep brain stimulation (DBS) [9].

In the very first implantable devices, cardiac stimulators, these three components were permanently joined together, forming a one-piece set up [10–12]. The life span of those systems was rather short as engineers and researchers just started to investigate how to achieve long-term stability of implantable hardware [1]. Their complexity was kept at minimum to avoid introducing further break points.

To increase the implant lifetime and to minimize the impact of the surgical intervention, the development of detachable systems was an important step for implants comprising distributed modules. The ability to connect and disconnect the different components from each other eased the installation, replacement and troubleshooting of the implants [13]. Each component is implanted at the desired site separately and joined together afterwards. System failures and battery replacements can be remedied locally, without removing the complete system. Especially for very sensitive targeted tissues such the brain, this is a crucial feature with electrode assessment being a critical process [14]. Additionally, the possibility to separate implant parts might prevent the spreading of infections along the cables in severe cases [15]. During the surgery, low-profile cable ends can be tunneled underneath the skin where otherwise incisions would have been necessary along the cable length [16]. This minimizes the risk for wound infections and tissue scars. Further, mechanical stresses during surgery are produced only over one implant section without passing on to the other parts. For these reasons, connectors have been always an integral part of complex, distributed AIMDs.

With the objective of more efficient recording and stimulation, the level of complexity in neural implants rises. The development of new therapies demands a greater number of electrodes and connection nodes [17]. At the same time, the organization of implant systems in modules provides flexibility and shorter development times [18]. Further, some novel approaches refrain from a single concentrated electronics unit but distribute the circuitry over several modules. They reduce noise induction by signal processing close to the electrodes [19, 20] and provide configurability, functional redundancy and customization of implant architecture [20, 21]. All these trends particularly challenge the reversible electrical interconnections in the AIMD. A plurality of electrical channels has to be interconnected without compromising the lifetime and electrical properties and while keeping the geometries to a minimum.

Still, the variety of current connector designs remains limited, and only a few well-proved concepts have been adopted for new target applications. Their design provides a robust connection of a few channels (figure 1), but becomes obsolete for a multitude of conductors [17] since it requires an excessive amount of space. Like this, the connectors become a bottleneck for reaching the next level of complexity in neural implants. Only very concentrated implant designs as CIs and retinal prostheses do not require connectors with respect to implantation procedures. Therefore, the need of reconnections between the electronics and the electrode arrays does not necessarily limit the numbers of channels in these devices.

In general, as implant hardware development mostly happens in industrial environments where confidentiality and intellectual property prevents the sharing of knowledge, only few publications exist currently on this topic. General connector standards have been established for cardiac applications (the so-called IS-1 (figure 1), IS-4, DF-1 and DF-4), which in turn encouraged the discussion on different technical solutions [11, 22, 23]. Yet, in the field of AIMDs applied to lower patient numbers, no standards exist. Without this need for compatibility between different manufacturers, less technical specifications on connector systems are shared.

The next generation of implantable connectors will require joint efforts. Engineering disciplines such as electrical, material and mechanical engineering are required to face
the challenge of a functional, long-term reliable setup with minimal geometries. Biological testing accompanies the transfer from prototype design to application in the patient and is a mandatory tool for risk reduction. Medical and clinical expertise is essential for a design compatible with setups and practices in the surgical theatre (section 2.1). Further, the clinical experience in implant-based therapies of well-proven concepts and weak points guides the development. This work aims to provide a theoretical ground for implantable connector development with a focus on neural applications. Still, we contribute a large part of this review to contact physics, since it has not been considered for publications in the delicate field of AIMDs yet, but represents a fundamental discipline for every work on electrical contacts.

Aspects and experiences from several disciplines are compiled from an engineering perspective. The state of the art of connector solutions are then introduced and discussed. Finally, an outlook is provided for possible approaches for the future generations of implants.

2. Requirements for clinical implantable connectors

Functionally, a connector ‘provides a separable connection between two elements of an electronic system without unacceptable signal distortion or power loss’ [24]. The electrical performance of a connector is given essentially by the performance of its weakest parts. Thus, all its components including supply tracks, bulk and interconnections to the cables, require a careful design. Regardless of the actual connector design, universal components can be identified (figure 2). What is required for each of those components? The following sections discuss those questions and provide a theoretical background. Whenever available, general design guidelines are presented. An overview of requirements that apply is given in table 1.

2.1. Clinical requirements

The development and design of implant components like connectors always has to take into account surgical procedures and clinical needs. Those specific requirements are guided by the principle to keep the impact on the patient at a minimum.

To minimize tissue injury, the size of the implanted component should be as small as possible. Beyond reducing the impact at the actual implantation site, small dimensions enable minimally invasive surgical procedures like tunneling of cables. This decreases the risk for wound infections and tissue scars. Further, surgical techniques and the use of surgical tools demand a robust design that can withstand clamping and other mechanical stress. As a ‘back-end’ of low profile electrodes and cables, a connector should be compatible with stylets, i.e. allow extracting the stylet from the lead body after positioning of the electrodes and leads. To reduce contamination, the connector’s contact area should be easy to clean from tissue and blood residues e.g. being wipeable before the final assembly during surgery.

Especially for components, whose functionality depends on actions undertaken by the surgeon, ‘Human Factors Engineering’ (or ‘Usability’) is crucial. For example, every multi-channel connector requires a design that allows only one (the intended) way of closing an electrical connection (foolproof). Risk related to use error has to be assessed and minimized. For this purpose, normative frameworks can be found in the European standard IEC 62366-1:2015 [25] and the US Code of Federal Regulations 21 CFR part 820 [26] (accompanied by a guidance document [27]) provided by the FDA.

Further, operative duration is a critical factor [28] requiring the setup of the device to be time-efficient. Consequently, the complexity of mechanisms (e.g. locking of the connection) needs to be reduced.
Table 1. Target specifications for implantable connectors: qualitative requirements, clinical needs and specifications given by standards and manufacturers during the certification process (as summary of safety and effectiveness data (SSED) and premarket approval application (PMA)).

| Connector properties | Requirements and clinical needs | Specifications |
|----------------------|---------------------------------|---------------|
|                      |                                 | By standard (ISO5841-3 (IS-1), ISO 14708-1) | As applied in SSEDs/PMAs |
| **Design**           |                                 |               |
| Channel count        | Defined by application          |               |
| Size/geometries      | No sharp corners or edges       |               |
|                      | No rough surfaces unless required for function |               |
|                      |                                 |               |
| **Mechanical**       |                                 |               |
| Insertion Force      | <Force needed to buckle the cable\(^a\) | \(\leq 14\) N (for gauge pin: \(\leq 9\) N) (2 channels) | <6.67 N (4 channels) \([36]\) |
|                      |                                 |               |
|                      | No buckling \([37]\)             |               |
|                      | \(\leq 1\) N (4 channels) \([38]\) |               |
|                      | \(\leq 9.0\) N (8 or 12 channels) \([39]\) |               |
|                      | \(\leq 8.9\) N (8 channels) \([40]\) |               |
| Withdrawal force (intended disconnection) | <Force where cable is damaged\(^a\) | 14 N (2 channels) | 4.45 N (4 channels) \([36]\) |
|                      |                                 |               |
|                      | No damage \([37]\)              |               |
|                      | \(\leq 2\) N (4 channels) \([41]\) |               |
|                      | \(\leq 9.0\) N (8 or 12 channels) \([39]\) |               |
| Retention force (unintended disconnection) | >Pull forces occurring within body and during surgery\(^a\) | Retention force needs to be specified. | 4 N \(\pm 0.5\) N, for 60 s, straight separating pulls (4 channels) \([41]\) |
|                      |                                 |               |
|                      | \(\geq 10\) N (wet, 8 or 12 channels) \([39]\) |               |
|                      | \(\geq 15\) N (8 channels) \([39]\) |               |
| Fixation setup and forces | Forces need to be specified, in accordance with IEC 62366-1:2015 (EU) \([25]\), 21 CFR part 820 (U.S.) \([26, 27]\) or corresponding national standards, \(\pm0.025-0.028\) Nm \([36]\) | Two M2 set screws at a torque of 0.15 Nm (+/-0.01 Nm) | Tightening of setscrews to 0.025-0.028 Nm \([36]\) |
|                      | Active fixation by surgeon (user need) | Securing mechanism forces shall not deform connector to the extent that insertion and withdrawal forces are excessive. |               |
| **Electrical**       |                                 |               |
| Contact resistance   | \(<1\)% of the electrode impedance | --- | --- |
| Insulation impedance between conducting parts | \(<1\)% of current leakage | --- | --- |
|                      |                                 | \(\leq 50\) kΩ in saline solution \((t = 0 \text{ and } t = 10\) d, test signal: 50 Hz \(< f < 120\) Hz, 100 mV \(< U < 250\) mV RMS) | \(\geq 50\) kΩ (after 10 d in saline solution) \([39]\) |
|                      |                                 | Max. DC leakage current density: 0.75 \(\mu\)A mm\(^{-2}\) |               |
|                      |                                 | Dielectric strength: \(2 \times \) peak voltage (saline solution) |               |
| **Thermal**          |                                 |               |
| Warming              | Tissue should not be harmed | \(<2\) K on outer implant surfaces | --- |
| RF heating in MRI    | Internal parts: no degradation | Heating will origin from cables (depending on their length), effect only secondarily \([42]\) | --- |
### Material

- **Biocompatible and biostable**
  - Tissue, organs or systems should not be harmed
  - No degradation

- **Sterilizable and resistant to cleaning procedures**
  - Withstand physical and chemical sterilization and cleaning techniques (e.g. wiping, flushing)

- **Displacement forces and torque in MRI**
  - Avoid ferromagnetic materials [42]

- **Testing in accordance to ISO 10993, ‘biological evaluation of medical devices’**
- **Validated sterilization process (sterility assurance level \( \text{SAL} < 10^{-6} \))**

### Connector properties

| Requirements and clinical needs | Specifications |
|---------------------------------|----------------|
| Depending on application:       | By standard (ISO5841-3 (IS-1), ISO 14708-1) |
| >max. number of expected revisions | As applied in SSEDs/PMAs |
| Depending on application:       |                         |
| >Implant lifetime               |                         |

- **Number of mating cycles**
  - 5 [36, 37]
- **Long-term stability**
  - 10 [39]

- **Quality of connection shall not degrade during use**
- **Re-connection shall be possible without a degradation in the performance**

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*All force limits should be impinged with a safety margin.*

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**Table 1.** (Continued)
2.2. Requirements on material and geometries

The challenge of developing long-term reliable electronic systems embedded in a harsh environment as the human body strongly restricts the selection of applicable technologies, materials and designs. Contradictory, most of the efforts in the design optimization of electrical connections do not consider such challenging environment. Here, we introduce the important general and task-specific materials and geometrical characteristics for implant components.

The fabrication material is defined by its biostability and biocompatibility. Neither it should degrade inside the body nor should harm the tissues, organs or systems [29]. Therefore, it should not be prone to leaching of hazardous substances or abrasion of particles. Materials should withstand degradation mechanisms, corrosion and embrittlement, and maintain their functionality (electrical and mechanical) over the entire implant lifetime. Test setups and procedures to evaluate a material biocompatibility are described within the international standard ISO 10993 ‘Biological evaluation of medical devices’.

Corrosion may occur via several mechanisms, affecting metallic components. To prevent galvanic corrosion, combining dissimilar metals should be avoided, including alloys and coatings. If this is not possible, the differences in nobility should be minimized. Further, the surface exposed to the electrolyte by the nobler metal should be as small as possible [30].

Metals commonly used as electrical conductors in AIMDs (i.e., electrode contacts, wires, connectors and feedthrough terminals) are platinum and platinum alloys (mainly Pt/Ir 90/10 or 80/20 for mechanical reasons), tungsten, gold and gold alloys, Elgiloy™, Phynox™, stainless steel (L316LVM) and MP35N™. The metals of choice for structural elements (i.e. bulk and housing) are titanium, MP35N™ and stainless steel.

A variety of polymers has been used for different implant applications. None or only a slightly deterioration and inactivity can be observed with high-density polyethylene (HDPE), polypropylene (PP), polytetrafluoroethylene (PTFE) and polyetheretherketone (PEEK) [29, 31]. Polymers that can be structured very precisely (therefore employed for delicate structures within neural interfaces) include polydimethylsiloxane (PDMS), polyimide (PI) and parylene C (poly(dichlorop-xylene)) [32].

Non-metallic and inorganic materials, ceramics like Alumina (Al₂O₃), and Zirconia (ZrO₂), have been successfully employed as substrate, bulk and encapsulation materials. Additionally, hermetic encapsulation as well as hermetic feedthroughs can be fabricated using biocompatible and biostable glasses (e.g. soda lime glass and borosilicate glass) [29, 33].

Beside the chemical and physico-chemical aspects of biocompatibility, the structural factors should also be considered as the implant’s mechanical properties and geometries should match the target tissue mechanics. Sharp edges that cut into the tissue and hard bodies within soft tissue promote the formation of scar tissue, skin erosion, inflammation and the formation of thick tissue capsules [33, 34]. The dimensions (especially the profile) should be kept as small as possible to minimize their impact onto the body [13]. In deep brain stimulation (DBS) systems, the connector profile has been identified to influence on infections and erosion problems [35].

The need for miniaturization calls for fabrication and assembling technologies that enable high precision without affecting biocompatibility (e.g. by toxic process residues or wear). Suitable structuring technologies include but are not limited to laser cutting, lithography and etching, screen-printing, micro milling, injection molding and spin coating. Typical assembly methods for conductors include welding, bonding, brazing and soldering. Insulating materials are applied by gluing, encapsulation or re-melting.

All the involved materials and joining techniques require compatibility with cleaning procedures and at least one sterilization method. Undercuts and crevices provide a trap for germs and infectious residues and should be avoided.

2.3. Electrical requirements

2.3.1. Contact interface. The core element of every connector is the region where the electrical signal is transferred from one connector side to the other. This area is commonly referred in the literature as ‘contact zone’ or ‘contact interface’ [24, 43]. The physics behind the electrical contact is a broad research field, with many processes at the contact interface still remaining unclear [43]. Many factors have to be taken into account to describe an electrical interface, including the electrical load, mechanical and environmental conditions, material properties and geometries [44]. This section does not aim to cover in detail the contact theory, but to point out the most important aspects with a focus on biomedical applications. For a in-depth discussion of this topic the reader is referred to the work of Slade [45], Braunovic [43] and Vinaricky [46].

2.3.1.1. Apparent contact area— and real conditions. On a microscale, the contact surface is always rough (figure 3). When such rough contact surfaces come into touch, the highestasperities will make contact first. After penetrating through insulating surface passivation layers, opposing ‘peaks’ establish a metal-to-metal contact. Due to high local pressures, those contact spots will be further compressed and asperities of less height will make contact. The process stops when the mechanical contact of asperities is sufficient to withstand the applied contact load.

This indicates that the real area of mechanical contact A_m is much smaller than the apparent (‘nominal’) contact area A_n, with this difference reaching almost two orders of magnitude [43]. Further, the nominal contact size does not even contribute to the area of real mechanical contact [47]. Given the same contact load F, a large apparent contact area with a multitude of contact spots ends up with the same amount of real contact area as a small apparent contact area with less, but more compressed contact spots [48, 49].

The material and topography define a critical contact pressure value above which the deformation of a contact spot
changes from elastic to plastic. These deformations occur throughout the entire contact interface [46], which makes it difficult to rely on simplified formulas based on purely one deformation mode. Further, only a small fraction of mechanically contacting areas contributes directly to the electrical connection.

Depending on the presence of contaminating films and their properties, different load-bearing areas can be classified in:

I. relatively thick, mechanically stable films with high resistance values and films with semiconducting behavior give rise to an additional film resistance $R_{\text{film}}$ (see figures 3(b) and (d) (blue frame)) [46, 50].

II. films that are sufficiently thin (2–3 nm [46]) to allow the tunneling of electrons create quasi-metallic areas where no further losses due to the resistivity of film material occur (see figures 3(b) and (d) (light blue frame)). A typical example are chemisorbed layers of oxygen atoms that are formed on any metal surfaces in air [50].

III. areas of metallic conduction without contaminating films. Here, electrical current can pass through the contact interface without perceptible transition resistance (see figure 3(d) (dark blue frame)) [46]. Those separate small contact spots are referred to as $\alpha$-spots. The proportion of such areas $A_\alpha$ within the load-bearing contact area $A_m$ is considered to be much smaller than 1% (see figure 3(b)) [43]. Thus, the volume of electrically conductive material is largely reduced to the interface areas (b) and (c). This constriction of current flow effects the constriction resistance $R_{\text{constr}}$ [46, 49].

The total contact resistance $R_{\text{cont}}$ is made of the transition resistance $R_{\text{film}}$ introduced by (a) and the constriction resistance $R_{\text{constr}}$ contributed by areas (b) and (c). In cases where $R_{\text{film}}$ is comparable to $R_{\text{constr}}$, both resistance parts can just be summed up:

$$R_{\text{cont}} = R_{\text{constr}} + R_{\text{film}}$$

where as a first approximation, $R_{\text{constr}}$ and $R_{\text{film}}$ are independent from each other. This changes when $R_{\text{film}}$ becomes considerable larger than $R_{\text{constr}}$. In these cases, most of the conduction happens within metallic and quasimetallic areas and equation (1) is not valid anymore [46].

More crucial than the sheer increase in contact resistance are its fluctuations, which can be another result of surface insulating films. Unpredictable behavior of the electrical junction can damage the entire system circuitry [46]. The following sections will concentrate on the different effects contributing to the contact resistance.

2.3.1.2. Constriction resistance. For a single, circular $\alpha$-spot with a constriction radius $a$ considerable smaller than the conductor radius, the resistance introduced by the current flow constriction is given by

$$R_\alpha = \frac{\rho}{2a}$$

where $\rho$ is the electrical resistivity [50].
The overall constriction resistance of an electrical joint represents a parallel connection of a multitude of $\alpha$-spots. For contact spots lying close to each other, the potentials of the single current lines overlap [46]. This interaction of current increases the resistance compared to a purely parallel connection. This is also the reason why one cannot extrapolate from a measured contact resistance to the real area of electrical contact [51]. Therefore, the constriction resistance is based on the number of current pathways connected in parallel and the size (radius) and distribution (interaction of the pathways) of the contact spots. Again, those parameters are defined by a plurality of properties, including the shape of the contact partners, the contact force, temperature, crystal lattices, material properties and topographical parameters such as waviness and roughness [46].

Many theories have evolved to define number, size and distribution of the microcontacts. On the one hand, model based theories used mathematical functions to describe asperity geometry and distribution. On the other hand, profilometry based theories took real topographies that have been determined experimentally. Both approaches usually used boundary cases of either fully plastic or fully elastic deformation depending on the choice of contact material (mechanical strength). Further assumptions and simplifications are necessary to describe the interaction of contact spots. Thus, constriction resistance has to be considered a statistical parameter. Under given mechanical conditions, only an approximate value can be provided [46].

We will only focus on practical connector design, with a general understanding of the interplay of the various properties and minimizing contact resistance. In practice, details on the microtopographical conditions are usually missing. Therefore, empirical formulas are employed to describe the constriction resistance by means of more accessible parameters [46, 49]. A common feature of such formulas (supported by theoretical work by Greenwood and Williamson and Harada and Mano [48, 52]) is the proportionality between constriction resistance and contact force [46]:

$$R_{\text{constr}} \sim \rho F_{\alpha}^{n}.$$  (3)

The values of the exponent $n$ ($n \in \mathbb{Q} \geq 0$) vary with the presence of contaminating films, the material and the mode of deformation. Theoretical considerations, which have been verified experimentally, found values of $n$ equals to 1/3 for pure elastic deformation and 1/2 for pure plastic deformation [46, 50]. Another study suggests values of $n$ for clean surfaces between 0.9 and 1 whereas for surfaces covered with film values less than 0.9 [46].

With increasing contact load, the constriction resistance decreases since more contact spots are established, which create further parallel current pathways. More importantly, this happens in a way that the greatest proportion of the decrease occurs in a small contact force range, remaining practically constant with high contact forces [46]. Reliable and stable connections require high contact forces.

Further, the correlation between constriction resistance and contact force depends on the contact material properties such as the resistivity $\rho$ and mechanical properties (hardness $H$). For the assumption of fully plastic deformation, the correlation given in (3) can be approximated as

$$R_{\text{constr}} = \sqrt{\frac{\rho^2 H}{4F}}$$  (4)

with $\eta$ being an empirical coefficient of order unity. This relatively simple expression does not take into account the complex physical effects at the contact interface. Still, it is a well-accepted approach to provide a rough estimation of the constriction resistance [49].

Regarding surface topography, theoretical and experimental results suggest higher roughness values to increase the constriction resistance [46, 52, 53]. One possible explanation are less $\alpha$-spots to be developed with rougher contacts. Compared to smoother surfaces, the peak asperities have to be further deformed before asperities of less height get in contact with each other. Since the increased deformation is accompanied by work hardening, higher pressure levels per contact spot are reached. Thus, less $\alpha$-spots are needed to counteract the contact force [53]. However, the influence of contact surface roughness on the formation and stability of contaminating films lacks a general correlation. Thus, characterization is required for each application respectively.

### 2.3.1.3. Contamination on contacts

Different forms of contaminations between the contact surfaces disrupt electrical conduction. Particles originating from abrasion or the body environment (i.e. clogged blood, pieces of tissue etc) can be trapped during mating. Cleaning immediately before closing the connections (rinsing or wiping) is a common prevention measure [54, 55]. However, these measures do not prevent impurity layers. During normal operating conditions, a totally film-free metal surface cannot be obtained. Usually, reactions between the contact surface and its environment create several atomic layers of contaminants separating the contact surfaces and changing the distribution of contact spots. The interactions can be classified into physical adsorption (physisorption), chemical adsorption (chemisorption) and chemical reactions.

#### 2.3.1.3.1. Adsorption processes

Adsorption is the binding of molecules to a surface forming a thin impurity layer. Several binding mechanisms may be involved in this process. Physisorption by means of van der Waals forces establishes only weak bonds ($\approx 0.05 \text{ eV}$) [50]. In contrast, chemisorption includes covalent, metallic or ionic bonds of significantly higher strength ($1-8 \text{ eV}$) [46, 50]. One prominent adsorption process is the formation of passivation layers on metal surfaces. This effect acts as protection of the bulk material against detrimental environments [46].

Further, surgical environments suggest the adsorption of thin layers of water on the contact surfaces. This has been shown to increase the contact resistance in the low contact force range ($\leq 10^{-3}$ N). However, those layers can be penetrated easily. They will introduce an offset between the mechanical and electrical contact point but are not expected to have severe effects on the contact performance [46, 56].
2.3.1.3.2. Chemical reactions. Unlike adsorption processes, chemical reactions induce the formation of thicker layers often growing over time. Usually, they are made up of metal oxides and sulfides showing semiconducting behavior [46]. Corrosion products are one example of those layers. In addition, organic layers can be formed when organic vapors condensate and degrade or polymerize afterwards. Precursors for this process originate from organic solvents and detergents or silicone sealants, greases and oils [46, 57].

2.3.1.3.3. Effects of contamination. The additional resistance $R_{\text{film}}$ by contaminating films (see equation (1)) can be calculated using the approach for sheet resistances with a circular area of contact (radius $a$):

$$R_{\text{film}} = \frac{\sigma}{\pi a}$$

with $\sigma$ being the resistance per area of the film (also referred to as skin resistance) [43, 46]. For values of $\sigma$ in the range to $10^{-8} \ \Omega \text{m}^2$, no contact disturbance is expected. This is the case for passivation layers on gold, silver, platinum and some of their alloys (i.e. PtIr 90/10, PtIr 80/20, AgPd 70/30, PtNi 92/8) [46, 58]. $R_{\text{film}}$ can be neglected for thin films since most of the electrical conduction happens via metallic contact at points of film rupture. In turn, the biostability of chromium alloys (i.e. MP35N™) and titanium relies on thicker, more stable passivation layers [59]. Thus, those materials are not suited for electrical contact interfaces. Failure due to contamination may be reflected on a permanent increase in the contact resistance, with higher power dissipation and heat generation [60].

Further, foreign layers diminish adhesive forces between the metal surfaces. Noble, ductile or soft metals with clean surfaces are often prone to ‘cold-welding’, a major cause of wear. For electrical contacts in implant operating environments, this effect can be neglected [43, 46, 57].

2.3.1.3.4. Motion induced contamination and contact degradation. All the processes described so far occur during static contact. Additionally, sliding motions of the contact surfaces induce contact degradation. Even in locked connections, external vibrations trigger micro-motions with amplitudes ranging from a few micrometers up to 100 $\mu$m. Possible sources in implants include heartbeats, breathing and movements of limbs or the intestine. Two kinds of degradation mechanisms with different contaminant products can occur. For base metal contacts with tendency to oxide formation, ‘fretting corrosion’ leads to insulating oxide debris. Still, even metals with little tendency to film formation show a similar effect. ‘Frictional polymerization’, the formation of amorphous organic coatings, occurs especially with metals of the platinum group (i.e. Pt, Pd, Ru, Rh and many of their alloys) [46, 50, 57, 61]. However, films formed by fretting are supposed to be tougher than those formed by frictional polymerization. Both degradation mechanisms are influenced by frequency and amplitude of the micromotions. With higher amplitudes and decreasing frequencies, the volume of debris is increased and contact degradation accelerated [61]. In turn, a high contact force diminishes the degradation effect since it facilitates the penetration of the debris or the polymer layer [61].

Still, external forces have to exceed the friction between the mating contact surfaces for fretting to occur. Mechanical decoupling of the contact zone from environmental influences lowers the impact of external forces and can prevent fretting (i.e. by strain-relief at the interface cable-connector or by compliant bulk designs with ‘floating’ contact pairs [61]). Further, higher contact forces increase frictional resistance and counteract fretting motions. Regarding the surface roughness, no direct correlation to contact resistance evolution under fretting conditions has been investigated up to date. However, its effect on underlying mechanisms (adhesive wear) suggests increasing fretting corrosion rates with increased surface roughness [57].

On the macroscopic scale, sliding motions might occur during opening and closing of the connection. They can support the removal of contaminants and disrupt insulating surface films. However, they come mostly along with large engagement forces, which induce wear, one of the most prominent failure mechanisms for connectors undergoing multiple mating cycles. Wear involves several mechanisms, all resulting in material transfer, debris, increasing contact resistance and loss of electrical contact. Higher contact material hardness reduces the real area of contact and thereby the probability of material removal and susceptibility to wear (see equation (3)). While fluid lubricants can help to stabilize the contact resistance [61], they have to meet all requirements regarding compatibility to the biological environment and to all other materials in charge (e.g. silicone rubber). The most important factors influencing wear are count and frequency of mating cycles [57]. In most fields of connector application, resistance to a high count of mating cycles is a prominent factor. However, for fully implantable systems the number of mating cycles that a connector has to withstand over its lifetime is rather low. Revisions are a rare event and many disconnecting actions include the exchange of one connector part. In fact, testing specifications for connectors in several neural implants require connection of only 5–10 cycle counts [36, 37, 39].

2.3.1.4. Electrical phenomena of current conduction. Beside the conditions on the contact surfaces, electrical current conduction is accompanied by physical phenomena, which generally affect connector performance. The following section evaluates two common phenomena in electrical connectors and sets them in reference to AIMD applications.

2.3.1.4.1. Skin effect. The basic expressions for the constriction resistance were derived under the assumption of direct current (DC) whereas alternating current (AC) conditions are the usual case for implant applications. The major difference between AC and DC constriction resistance stems from the skin effect, which limits the penetration depth $\delta$ of the electromagnetic field in a conductor for higher frequencies:

$$\delta = \sqrt{\frac{\rho}{\pi f \mu_0}}$$

(6)
with the magnetic permeability of free space \( \mu_0 \) (\( 4\pi \times 10^{-7} \text{ H m}^{-1} \)) and \( f \) the AC excitation frequency [62]. Thus, the AC constrictive resistance will deviate from the DC value for penetration depths smaller than the diameter of the conductor [43].

Maximum frequencies for both, biosignal recording and electrical stimulation do not exceed 10 kHz for most neural implants [63] (CIs taking a special role with stimulating with up to about 50 kHz [64]). However, these are still low enough to achieve skin-depths (penetration limits) in the range of several millimeters for typical conductor materials. Hence, the conductance through contact spots with typical diameters in the sub-micrometer range should not be affected.

2.3.1.2. Contact heating. The flow of electrical current induces heating of the conductor. In electrical junctions, the maximum temperature occurs at the contact spots due to the constriction of current lines [46]. In fact, material softening or melting is a crucial degradation factor in many connector applications. However, softening due to Joule heating requires current amplitudes larger than 100 mA [46, 57, 65]. With maximum currents in the range of few tens of milliamperes, AIMD applications lie far below this range [65]. Nevertheless, heating of implanted electrical junctions needs to be minimized for reasons of patient safety. Normative requirements limit thermal heating on the outer implant surfaces to 39 °C [34]. Enlarging and maintaining the area of real contact will reduce heat generation [43].

2.3.2. Insulation/gaskets. While a stable electrical connection is an essential attribute of a connector, for high channel counts and body-like environments, insulation can be even more crucial than the electrical connection itself. The prevention of signal loss and distortion requires sufficient electrical insulation from neighboring channels and the environment.

Guidelines regarding insulation requirements can be derived from international standards, which allow a maximum DC (leakage) current density of 0.75 \(\mu\text{A mm}^{-2} \) on any conductive surface in AIMDs [34]. Further, a proof of dielectric strength for at least twice the applied peak voltage after previous immersion in saline solution is also required [34]. Again, the ‘access resistance’ of the electrodes at the front end can be used as indicator. Not more than 1% of current should be lost by shunting between individual conductors. For electrodes with impedance values in the lower kOhm-range, this provides ambitious insulation impedance requirements (e.g. an insulation impedance above 1 MΩ for an electrode with an impedance of 5 kΩ) [66]). However, for high impedance-electrode arrays, such requirement seems to be not feasible. Still, normative requirements for implantable pacemaker connectors apply considerable lower values. For the IS-1 connector, insulation impedances of 50 kΩ with frequencies up to 120 Hz are required [67]. The same value (at 100 Hz) was applied in connector test specifications for SCS devices [39].

Throughout hardware design of AIMDs, encapsulation is an essential step to hinder unintended interactions between the implant and its environment, including corrosion, short-circuiting or tissue intoxication. The need for an implant-body barrier is commonly addressed either by hermetic capsules (made of ceramics, metals or glasses) [13] or by conformal (non-hermetic) polymer coatings [68]. These permanently applied barriers allow even non-biocompatible materials in the package. Yet, they are not suitable for setups, which require access to conducting parts after fabrication such as dis- and re-connectable electrical connections. Conditions during surgery require the connector system to be compatible with the entering of body fluids in the electrical contact zone. Without a permanent barrier, solely biocompatible and biostable materials can be used. Further, conducting parts have to be sealed against each other in a non-permanent manner.

Gaskets fulfill several general functions like separation of media, electrical insulation and compensation of geometrical tolerances [69]. They may even reduce the risk of fretting corrosion by absorbing micromotions [70]. Electrical insulation is achieved by preventing electrolytic leakage via fluid bridges between channel pairs and towards the environment. Additionally, lip seals in ‘wiping’ connectors act as excluders that keep contaminants away from the contact zone [71].

Different methods can be applied for sealing purposes. Compression and attrition (combination of dragging action and compression) act purely mechanical while lip expansion (edge swelling) includes liquid uptake by the gasket material [72]. Still, for AIMD applications, the insulation has to be initially effective and swelling can only serve to improve a mechanical seal. Sealing by fusion of the gasket material (i.e. uncured silicone rubber, melted polymers) may provide effective sealing but disqualifies for connector applications as it is irreversible. In turn, compression seals rely on the blockage of leakage pathways by deformation or by plastic flow of the gasket material when it is forced against a counter surface.

In analogy to the electrical contact theory (see 2.3), the micro-scale roughness of the seal surfaces causes areas of real contact to be substantially smaller than suggested by the apparent geometries. However, this case does not require a maximum transfer (of current) over the contact gap but rather a minimum transfer (of fluid medium) throughout the gap. Thus, the lateral expansion of real contact areas becomes very important. To establish a functional seal, the areas of contact have to block all pathways by forming a continuous cluster. While this is the theoretical case for loads above a percolation threshold where the ratio of real contact area to nominal contact area reaches 50% (for random roughness structures) [73, 74], in practice, leakage occurs even for sufficient contact ratios since shape deviations and imperfections play a large role [73]. Further, the continuity of contact area and the percolation threshold are scale-dependent as no seal is perfect. Macroscopically effective seals still show leakage pathways on a micro-scale [75].

Elastomeric materials should be favored for gaskets in implantable connectors. Showing a low modulus of elasticity, elastomers undergo large deflection while requiring low stress. Additionally, they show comparably low creep, which is important for maintaining sealing stress for long time periods. With a Poisson’s ratio of almost 0.5, elastomers are virtually incompressible. In fact, they transmit pressure evenly through the seal structure. The inherent resilience ensures...
that the seal will return to its original geometry when the distortion force is removed. Further, elastomers can be applied as molded-in-place seals which are retained on the bulk face. This process reduces the number of seal components, prevents additional leakage pathways and enables complex seal shapes with robust alignment.

An important aspect for a good choice of gasket material is its resistance to the applied fluids. Silicone rubber (Polydimethylsiloxane, PDMS) became the material of choice. Combining the advantages of elastomers with well-proven biocompatibility and essential semipermeable characteristics, it has been used as sealing material in all successful connector systems up to now. It is highly flexible and conforms to minor imperfections in a ‘flat’ surface. Water vapor can penetrate through its molecular network, but the pass of ions is prevented. In a saline environment, an osmotic gradient is established, which results in well-defined swelling characteristics with stable electrical insulation properties.

PDMS can be reversibly sealed to glass, hard plastic, silicon, flat metal and native PDMS surfaces. The sealing properties have been well characterized in low-pressure microfluidic applications. Simple van der Waals contact already provides watertight seals that even withstand small pressures applied in the fluidic channels (up to ~340 hPa). The hydrophobicity of PDMS surfaces might further contribute to the sealing quality. Poor wetting promotes the trapping of micro- or nano-bubbles in cavities of the gasket surface (on a microscale rough). Gas filled cavities, in turn, were identified to improve sealing by blocking leakage pathways.

Yet, PDMS shows relatively low tear resistance and is therefore susceptible to wear. Sliding under dry conditions should be avoided as silicone rubber surfaces show high and erratic friction. Mechanical loading generally accelerates the deterioration of polymers. Further, in a connector setup, the force for compressing the gaskets impedes the handling as it adds upon the force required to close the connection. Therefore, only the minimal force required to ensure proper sealing should be applied. Still, gaskets represent the connector part that is most susceptible to ageing and wear processes. Despite many approaches for long-term lifetime prediction of elastomeric seals by accelerated ageing, there is no consensus on an effective and reliable method yet. Consequently, the rule of thumb of installing gaskets in the connector part that is most likely to be exchanged (for AIMDS using batteries usually the stimulator site), evolved as good engineering practice.

2.3.3. Noise and interference. Especially for the case of recording where amplitudes of the transmitted signals are low, interferences and noise have to be considered for each component of the implant system. In the following, we will shortly discuss these phenomena regarding an implant connector interface.
2.3.3.1 Noise. Noise is a purely stochastic phenomenon that is inherent in any electronic system but can also be picked up from external sources.

Several kinds of noise can be differentiated. Johnson-Nyquist noise or thermal noise is due to the thermal agitation of electrons and present in any electrical component, including contacts. It is proportional to the square root of the temperature, the resistance and the frequency bandwidth (and therefore ‘white noise’). While thermal noise limits the ultimate threshold sensitivity of amplifiers, for contact applications as connectors it is insignificant compared to other types of noise [50].

More specific for electrical contacts is contact noise, the fluctuation of contact resistance [62]. Several mechanisms affect contact noise characteristics, including diffusion on the contact surface, oxidation of surface films and variations in the dimensions of the conducting spots [62].

One effective measure to reduce contact noise is increasing the amount of contact points. For n equal contact points, the noise level is expected to be lowered by the factor $1/\sqrt{n}$ (compared to the case where $n = 1$) [50, 62]. Further, for the case of mechanically generated load variations, noise can be reduced by applying higher contact pressure [50].

2.3.3.2 Interference. In contrast to the statistical nature of noise, interferences (including the phenomena of artifacts and crosstalk) refer to the pick-up of signals or electrical events other than the ones of interest [81]. They can originate from both, sources within the recording system and external sources.

Galvanic interferences occur due to common coupling impedances between an interfering circuit and the interfered circuit. Further, alternating electrical fields cause capacitive interferences. In this case, voltage changes in the interfering circuit generate alternating electrical fields which themselves generate displacement currents in the interfered circuit. On the other hand, inductive interferences occur due to an alternating magnetic field of an interfering circuit [82].

In systems that include both, stimulation and recording, stimulation artifacts can affect the recording of signals, for example in CIs that use electrically evoked compound action potentials (ECAPs) to assess implant functionality [83]. However, most current implants are used in time-multiplex modes. This means that recording blocks and stimulation blocks are never passed simultaneously over cables and connectors. In this case, direct interference between the lines does not have to be considered.

Movement artifacts are a known phenomenon regarding displacement of electrodes. Still, they can be an issue of the electrodes’ subsequent electrical connections as well. In this context, Edell and Dweiri et al have observed the effect of mechanical interaction of insulation materials between bundled leads. This can cause spikes up to 40 $\mu$V, which can be interpreted as neural activity [84, 85]. Movement has also been shown to increase the noise level in cables, presumably due to leakage between the cores [86]. Thus, relative movement of leads, especially those comprising insulation materials with dissimilar charge affinity should be avoided.

Regarding the periphery of an active implant, the main point to reduce interferences should be the cables. For concentrated elements as connectors, effective measures are good conducting properties within and high insulation impedances between conductors. In fact, general design considerations for contacts are in good accordance with steps to reduce noise and interferences.

There are no standards for characterization of noise and interferences regarding implant components as connectors or cables. However, the experience shared by some publications can help to assess these phenomena [81, 85, 86].

3. State of the art of implantable connectors

In this section, actual connector concepts shall be discussed to evaluate how the requirements stated above can be achieved. Figure 4 provides an overview of those concepts while table 2 summarizes their advantages, disadvantages and fields of application.

3.1 Contact design, arrangement and channel count

The overall connector appearance is defined by the design of the electrical contacts, and for more than one channel by their arrangement and the channel count.

The traditional concept of a connector comprises pins as ‘male’ contacts and receptacles as ‘female’ contacts, where a pair of pin and receptacle is matched for each channel. This classical approach has also been employed in implantable connectors. The so-called ‘Craggs-connector’ (figure 4(d)) is one of the few well-described connector systems that can be found in the scientific literature [87]. It provides up to four channels (in a modified version up to 12 [88] (figure 4(j))) and has been successfully implemented for sacral and lumbar anterior root stimulation, bladder stimulators, drop foot devices [89, 90] and grasp prostheses [77, 91–94].

Less described examples of pin connectors were used in early DBS systems [7] (figure 4(i)), ventricular assist devices [95] and phrenic pacemakers [96–99] (figures 4(a) and (g)).

As the channel density increases, the overall geometry of the connector needs to be reduced and pins become more fragile. Non-implanted miniature connectors were designed to address this issue by embedding each pin in a protecting recess [100]. They have been adopted and customized for acute animal trials [101] or first prototypes [19, 102], but up to now they have not been proven eligible for long-term implantation.

A slight different approach by Letechipia and colleagues used pins on each side of the interface that are connected by a center spring as receptacle (figure 4(b)) [16]. This single-channel connector system performed with reliable functionality in many prosthetic devices [15, 103–111].

Current commercially available connectors for neuro-modulators, spinal cord stimulators and cardiac applications refrain from this traditional design. They still comprise an overall ‘male’ and ‘female’ connector part, but contain multiple contacts. Axially arranged ring contacts on the male side are contacted by terminal spring contacts on the female side.
Table 2. Connector concepts, their advantages, disadvantages and target applications.

| Connector concept | Male (Picture reference) | Female (Picture reference) | Advantages | Disadvantages | Applications | Channel count/connector unit | Mode of operation | Designation/Company + References |
|-------------------|---------------------------|----------------------------|------------|---------------|--------------|---------------------------|------------------|----------------------------------|
| **Perpendicular** |                           |                            |            |               | FES (lower limbs, drop foot), bladder stimulators, grasp prostheses | 1-4              | ![ ] | Cragg's connector [66, 77, 87, 88, 90-94] |
| **Pin**           |                           | Receptacle figures 4(a), (d), (f), (g) and (j) | Design easy to implement - Simple cable connection by crimping - Arrangement (related to cable axis) | UPC super large geometries - Insertion force - channel count - Only one revision possible (permanent sealing) - Difficult cleaning - Fragile pins | Ventricular assist devices, phrenic pacemakers, early DBS systems | 1-4              | ![ ] |  | [95-98] |
| **Spring receptacle** figures 4(b) and (k) |                           |                            | Active clamping prevents disconnection - Multiple reconnections - Robust design | UPC super large geometries - Insertion force - channel count - Difficult cleaning | [Acute animal trials, prototypes] Intended applications: Neurostimulation, Prostheses, Biotelemetry | up to 16 | ![ ] ( ) intended | Omnetics micro and nano circular connectors [19, 100-102] |
| **Circular**      |                           |                            |            |               | Prostheses | 1              | ![ ] ![ ] | [15, 16, 16, 103, 105-108, 108, 110, 110, 111, 146, 147] |
| **Ring contact**  | Wire basket figure 8(b)   | Low insertion forces        | High contact forces - Fixation and contact establishment in one action - No self-locking - Complex design - Insertion force - channel count - Sliding of contact surfaces during insertion - Upscaling entails large geometries - Difficult cleaning | Few contact points (line contact) - Limited miniaturization | Cardiac devices, DBS extensions | 2-4              | ![ ] ![ ] ![ ] | ![22, 148] |
| **Spring**        | Cantilever figure 8(a)    | High contact forces         | Extraction force - Insertion force - Low insertion force | Few contact points - High insertion forces - Linear force curve characteristics - Limited miniaturization | Cardiac devices (pacemakers, defibrillators) | 2-4              | ![ ] ![ ] ![ ] | ![11] |
| **Spring pin**    | figure 8(c)               | Low profile - Robust        | Extraction force - Insertion force - Low insertion force |  | DBS, SCS, Cardiac devices (pacemakers, defibrillators) | 4-12             | ![ ] ![ ] ![ ] | ![Bal Seal Inc., 17, 131, 133, 140] |
|                   |                           |                            | Insertion decoupled from contact force buildup | |  | 4              | ![ ] ![ ] ![ ] | ![Neurospace, 38, 134] |

a Modified design
Details on different spring contact designs will be discussed later in this review. In connectors with multiple channels, the arrangement of the electrical contacts becomes an important design aspect as they define the total volume of the connector. The channels can be generally arranged in line or perpendicular to the cable. In-line concepts provide low profiles, but the connector length increases with higher channel counts. A perpendicular arrangement in turn keeps the connector relatively short, but results in bulky geometries in high channel counts. Although both arrangements are applicable for any contact designs, up to now pin and receptacle contacts have only been applied in perpendicular channel arrangements, while contact pads and rings are used in the in-line concepts (figure 4).

The channel count of AIMDs has substantially increased in the last decades. Future implant generations with increased recording and stimulation selectivity will comprise even more electrode sites. Especially for applications with electrode arrays, small and high-density connectors are required [112, 113]. Direct upscaling of available connector solutions is not possible due to space and mechanical limitations (see 3.2 and figure 4(j)).

To some extent, the lack of appropriate connectors can be bridged by distributing the channels over several low channel count connectors. This splitting allows falling back on proven connector solutions. However, it does not solve the problem of space limitation, introduces more cables and increases the complexity of the surgery. The single channel in-line connector by Letechipia has been used in several neural prostheses providing good reliability, but requiring a large number of cables (figure 4(k)) [15, 103–106, 108, 110, 111]. A neural prosthesis developed by Guiraud and colleagues adopted the cardiac IS-1 connector system (figure 1) [114]. The space needed for the header makes this type of connector impractical for high-density electrode arrays (figure 4(l)).
Twelve is the maximum of channels per connector that have been implemented with multi-channel in-line-connectors (Spinal Cord Stimulation (SCS) applications and eight channels for neuromodulators [39, 54]). Higher channel counts require several connector units to be stacked (figure 4(i)). In some cases, the resulting bulky headers contribute almost 50% to the overall IPG volume. However, this technique still allows a great extent of upscaling.

The highest channel count served up to now is 32, whereby all channels originate from the same electrode array, are divided into four leads and connected to the IPG via four eight-channel in-line connectors [55] (figure 4(m) for the connector and figure 5(a) for the electrode array).

Implants with higher channel counts are limited to permanent electrical connections. During early development stages of high channel implants (animal models and first clinical trials), transcutaneous setups can be used where the connector part is either fully (transcutaneous cables) or at least partly (transcutaneous connector) kept outside the body (see figure 6) [115–124]. However, transcutaneous setups increase infection risks and this strategy can only serve as a short-term solution. Especially brain–computer-interfaces (BCIs) require the development of fully implantable systems [113, 125] and comprise much more individual channels than can be served by state of the art connector concepts (see figures 5(b) and (c) for the electrodes). Therefore, the first fully implanted high-channel BCI systems were hard-wired [126–128]. Beside other drawbacks (see section 1), this substantially increases the required level of miniaturization, since only concentrated system designs are

Figure 7. Setscrews are applied for mechanical fixation and electrical contact establishment: (a) DBS extension connector where setscrews ensure both, electrical contact and mechanical fixation; reproduced with permission of Medtronic, Inc. (b) Cardiac pacemaker header with one setscrew (mechanical fixation and electrical connection) and one spring contact; reproduced with permission of Medtronic, Inc. (c) + (e) spring contacts ensure electrical contact, the setscrew only serves mechanical fixation; with (c) SCS-system, image courtesy of NuVetra™ and (e) DBS extension, image courtesy of Boston Scientific Corporation (d) design without any screws where a plastic part is pressed in for fixation; From the Museum of Health Care at Kingston. Used with permission.

Figure 8. Different kinds of in-line lead connector terminals: (a) setscrew contacts provide only a single line contact and are elaborate for multiple contacts. Spring contacts establish a connection automatically. Several designs are possible: (a) leaf springs are simple but take up much space and come along with high forces. (b) The wire basket design has been established for the IS-1 standard, but is no longer manufactured (image courtesy of Hypertac Hypertronics Corp.). (c) As the current state of the art, coiled spring contacts provide definable (low) force characteristics and multiple contact points. Reprinted Courtesy of Bal Seal Engineering, Inc. Copyright © Bal SEAL Engineering, Inc. 2018. Bal Conn electrical contacts and the Bal Spring canted coil spring (SYGNUS) are trademarks of Bal Seal Engineering, Inc. All rights reserved.
Figure 9. The NeuroPace RNS Neurostimulator connector concept (NeuroPace Inc., Mountain View, CA). First, the low profile leads are inserted without considerable engagement forces ((a) modified from [38]). Second, electrical contact is established by applying a cap that presses the leads onto the contact sites ((b) modified from [38]). Reprinted with permission of NeuroPace Inc.

compatible with implantation procedures. Thus, the insufficient channel counts makes implantable connectors a bottleneck for further development of AIMDs [129, 130].

3.2. Mechanical concept

As discussed, the contact force has a large influence on the electrical contact and insulation performance. Thus, a uniform force distribution over the entire contact zone and well-defined and long-term stable force characteristics are required. To provide good handling and prevent cable damage, the insertion (or connecting) force and the force for intended disconnection should be as low as possible. On the other hand, the pullout (or disconnecting) force for a closed connector should be high to prevent unintended disconnection. Various mechanical setups such as setscrews, clamps, springs, compression sleeves or sutures could be used to apply those forces. Yet, the setscrews have prevailed so far. They allow for a precise control of the applied force even among different users by dedicated torque wrenches.

Three kind of necessary mechanical actions can be distinguished in an implantable connector in general. Depending on the mechanical concept, one or several mechanisms are required to take those measures.

First, one may differentiate the overall locking or securing of the connection. This measure can but does not have to be directly related to electrical functionality. It fixes the separable connector parts, ensures their proper alignment and counteracts any kind of disconnecting forces (pulling, tearing). The second required mechanical measure is by pressing the contact surfaces together. By providing the required contact force, it ensures a constant electrical contact and prevents fretting corrosion due to micro movements of the contact partners [57]. The last measure to be described is by compressing the gaskets for watertight seals between conductors and towards the environment.

In pin connectors, receptacles are elastically deformed when the connection is closed. The deformation action raises the required (counter-) contact force. Special designs of receptacles (see figure 4(b)) might provide even ‘active clamping’ of the pins [16]. Receptacles in the shape of wired coils stretch when pull-out forces occur. In turn, their diameter decreases, and the contact pressure between receptacle and pin increases. Pull-out forces for this design were not characterized quantitatively but have been proven to lie above the forces required for lead failure [16]. Still, additional measures to withstand disconnecting forces and to ensure proper sealing are usually needed. Gluing with silicone adhesive, often in combination with silicone sleeves, provides a simple solution, but also inhibits multiple connection cycles (see 2.3.2) [7, 87].

The multichannel in-line concept as found in IPG headers employs setscrews for fixation. In the first IPG generations, the electrical and the mechanical connection were established the same way, requiring one setscrew per channel (figure 7(a)). In fact, for round connector designs, only a line contact is achieved between the contact partners (figure 8(a)). Further, for multiple channels, individual installation and testing of contacts is not feasible. Functionality should not rely on the physician to provide a secure connection. Every screw introduces a point of potential failure by losing, cross threading, damaging its head, deterioration by body fluids or nicking the lead. Setscrew problems have been identified to play a significant role in defibrillator returns [11].

Alternatively, recent in-line connectors employ spring terminal contacts in the receptacle. They establish electrical contact ‘automatically’ [22, 131]. For mechanical locking, the number of screws could be reduced to only one per connector unit (figures 7(b), (c) and (e)). Even functional designs without any screws have been developed, but did not prevail (figure 7(e)) [22]. The first set screw alternative, a ribbon of leaf springs using stainless steel, titanium and MP35N, was established by Medtronic and is still prevalent in IPG headers. However, leaf springs require certain dimensions limiting miniaturization and increase of channel counts. They provide relatively few contact points, which increases the risk for higher contact resistance (figure 8(a)). Further, leaf springs show linear force curve characteristics, which can result in variations of insertion and extraction forces [13].

Other approaches used spring wires as contact on the female side. The ImplanTAC™ design by Hypertac Hypertronics Corporation (now Smiths Interconnect Inc.) connectors employed ‘wire baskets’, hyperboloid-shaped baskets of individual platinum-iridium spring wires (figure 8(b)). This design provides small insertion and extraction forces and a low contact resistance [131]. Still, the requirements for the basket dimensions limit the opportunities for miniaturization. Despite its accordance to the IS-1 and IS-4 cardiac connector standard [132], this design has been withdrawn from the market.

The contact design by Bal Seal Inc. employs canted-coil springs made from platinum-iridium (figure 8(c)). These provide many contact points, a low contact resistance and good compensation of surface irregularities. In contrast to beam contacts, the canted coil springs show non-linear force curve
characteristics. Thus, the design of the spring defines insertion and extraction forces. Depending on the coil orientation, uni- and bi-directional versions are available. A high resistance to fatigue results in reliable constant force characteristics. For the smallest lead diameters (1.3 mm), insertion forces below 0.8 N and extraction forces below 1.4 N apply (per one connector unit comprising seal, contact spring and housing) (see table 1) [133]. This concept has been established in more than a million implants worldwide [130].

For all those concepts including ‘active clamping’, the contact surfaces slide along each other during closing of the connection. On the one hand, this ‘wiping’ might remove contaminants and disrupt insulating surface films. On the other hand, engagement forces and the tendency for wear are prone to increase [57]. Higher channel counts require higher insertion and extraction forces. Those need to be compensated by the lead bulk. Low-profile connectors do not allow for large bulk structures, so the stability of the lead defines the maximum acceptable insertion force (where no bending occurs) and the maximum count of channels (see table 1) [17, 130].

Therefore, to connect low-profile, high-channel electrode leads, alternative mechanical solutions are required. It is preferable to decouple contact forces from the insertion process and avoid sliding of the contact surfaces. One successful example is the NeuroPace RNS system (figure 9). Instead of a shaft with ring-shaped spring contacts, its female connector part comprises a groove with embedded spring contact pins. For insertion of the cables, they are placed in the groove longitudinally. Compression for electrical and mechanical connection is applied by a covering cap [134]. Further connector design concepts for low-profile leads exist, but have not yet been implemented in an implant system successfully [135–138].

In-line contact arrangement allows for low profiles, but it results in connectors of excessive length. In turn, extending the profile enables channel organization in arrays with reduced length and more stable bulk geometries. Several designs comprising an array-arrangement of contact sites have been proposed [139–145]. However, they still have to leave the development stage and lack a solution for cable management during surgery (tunneling only for low-profile structures possible).

### 3.3. Electrical performance

#### 3.3.1. Conduction performance

Reports considering the electrical performance of the implantable connector part separately are a rare find in literature. Further, they are subject to varying measurement conditions, which impedes comparison of the different concepts. When evaluating a connector’s electrical performance, the electrical properties of the electrodes to which it connects require also special considerations. These depend strongly on the type of electrodes. Most electrodes show impedance values in the range from 1 to 100 kΩ [115, 149–153]. Lower values of 400–500 Ω apply for some grid and foil electrodes [66, 153] while for microfabricated high-density electrode arrays impedance values up to 1 MΩ are reported [154]. Regarding the resistance introduced by the connection, an upper limit of 1% of the electrode impedance can be set. Thus, for low impedance electrodes, the contact resistance should remain below 4 Ω.

For the Craggs connector, resistance values range between 5 and 15 mΩ in a four-wire dry measurement setup [87]. Letichipia and colleagues state a resistance of their connector assembly below 1 Ω [16]. However, it remains unclear what parts of the lead periphery where included in their measurements and under which conditions. Ball Seal specifies their connector system to show dry static resistances between 40 and 600 mΩ [130].

A four-wire setup for resistance measurement is preferable as it excludes the connectors’ periphery. Here, the probe points define the resistance to be characterized (contact resistance only or connector resistance including interconnections). To evaluate the contact (not the insulation) during development and fabrication, dry resistance measurements are adequate. Still, the final connector setup should be closed and tested in saline solution as specified in cardiac connector standards (see table 1) [67].

#### 3.3.2. Insulation performance

To meet the insulation requirements, early connector concepts renounced multiple connection cycles. In the original Craggs connector setup, the mating surfaces and the protecting fairing have to be glued with silicone adhesive during surgery [87]. This not only provides sufficient insulation but also prevents unintended opening by distributing the stress eventually spread from the cables. However, there is none opening and reconnection possible. Therefore, Rushton developed another sealing method, which allows at least for one revision. Both, initial and secondary sealing, provided insulation impedances of more than 1 MΩ after 106 d of saline soaking [66].

Similarly, the single channel in-line connector by Letichipia and colleagues comprises an insulating silicone rubber cuff. It is closed and secured by sutures; rubber rings around the lead wire prevent the cuff from sliding down the lead. This setup has been proven to show leakage currents below 0.01 µA for a 30 V bias in saline solution [147]. Disconnecting is possible, but as the reassembly involves gluing of a repair sleeve, the design is limited to only one revision also [146].

In contrast, the well-established multichannel in-line connectors employ silicone O-rings. They are stacked alternately with the spring receptacles in the connector cavity. This design can provide resistance values of more than 250 kΩ [17, 133]. However, each additional O-ring element increases the insertion force and therefore the mechanical stress onto the inserted lead (see 3.2 and table 1 for values). New generations of connectors circumvent this by separating compression and insertion actions as follow described [17, 134].

#### 3.3.3. Interface connector to periphery (cables or IPG)

Besides introducing a non-permanent (‘reclosable’) electrical connection, the connector system itself comprises several (permanent) electrical interfaces with the cables or the IPG. Equivalent to a series connection, the overall performance of the connector system is dominated by its weakest
part (conduction wise and insulation wise). The overall resistance introduced by the connector system is a result of adding up the resistance of the actual electrical transition zone $R_{\text{contact}}$ (see 2.3), the bulk resistance of the conductors and the resistance of the interconnections to the connector periphery. Similarly, the insulation depends not only on gaskets at the non-permanent interface, but also on bulk material properties and sealing of permanent interconnects.

Depending on the substrate design and material, different interconnection technologies can be applied. Still, from a large selection of interconnection technologies, only few are applicable for implants. Common methods used to connect electrodes are laser-welding, gap welding and soldering [155]. Further, Microflex is used for bonding flexible and rigid substrates in a biocompatible and stable manner [155, 156]. However, as this technology originates from chip size integration, it is not applicable to connect with cable wires. Soldering has been proven to provide sufficient reliability in implants [157] and can provide bond strengths higher than welding [155]. However, it bears the risk of leaching of toxic substances and corrosion due to dissimilar (especially non-noble) metals (see 2.2). Further, the formation of intermetallic phases and the diffusion of solder material have been observed [113].

Welding provides the opportunity to establish a bond just from substrate materials. This again reduces the corrosion tendency. From the welding technologies, parallel gap welding proved better reproducibility and higher bond strengths than laser-welding for the distal electrode interconnection [155]. Still, for the electrode endings proximal to the connector unit (in-line connector setup), laser welding is the method of choice to connect wire strands to the contact rings [158, 159].

Beside these metallurgical methods, crimping has been employed for implant connector assemblies, especially with Cooper cables (see Craggs connector [87]) and in combination with Cooner wires [87, 160]. Early implant generations even used crimping during surgery, before non-permanent connections were established [161, 162].

However, none of the interconnection technologies is supposed to withstand higher pulling forces acting on the cables. Most interconnections are casted in insulating material (e.g. silicone rubber) to maintain insulation between the channels in moist environments, mechanically stabilize the connection and to distribute stress. Still, dedicated strain relief measures are necessary. They prevent stress at the interconnection site by ‘mechanically shortcutting’ more distal cable parts and the connector housing. Pulling forces are diverted from the cable to the housing bulk without passing the interconnection part. Possible strain relief structures include basket weave setups [60], springs [16] or fairings [87].

4. Conclusion

Connectors play a key role in neural implants. The requirements on channel count, biocompatibility and long-term reliability in a very delicate environment make connector development a great challenge. Thus, only few functional design concepts can be found. Up to date, in-line connectors comprising low-profile contact rings on the electrode site are the design of choice.

Implantable electrical connections comprise many factors, which influence their performance and involve a variety of technologies and fields of knowledge. Some of them can be controlled by pure engineering design, but many are dictated by the delicate environment of the human body. Some factors even show no distinct impact but evoke different, sometimes contrary effects.

From the previous discussions, the following conclusions can be found:

- High contact force increases the number of contact spots, helps the penetration of insulating films and hinders fretting motion. Therefore, it is the main factor for low and stable resistance values.
- A large electrically conductive area is not achieved by designing large nominally contact surfaces.
- Surface contaminant films will be always present due to environmental conditions. They are influenced by the material and surface characteristics.
- Gaskets and a reasonable geometrical design can reduce the ingress of substances that promote corrosion and other detrimental reactions.
- Long-term stable electrical insulation requires a sufficient level of gasket compression and proper material choice. However, it does not need watertight materials.

Connector solutions for future implant generations with increased channel counts require compromises:

- For spring contacts, more channels require higher insertion forces. Bulk structures need to provide sufficient stability. On the other hand, higher channel counts require a higher grade of miniaturization. This results in designs that are more fragile.
- Non-spring contact designs have to find a way for ‘automatic’ contact establishment providing a well-defined contact pressure.
- In-line channel arrangement is required for lead tunneling but results in unacceptable connector length.

Some groups have suggested distributed active systems that were connected with wires [163–166]. The requirements of connectors with respect to contact and insulation resistance remain the same, while power supply and data transmission should be investigated in detail. Since this topic goes far beyond the questions of connectors and their reliability but addresses design questions of future implants, it will not be covered in this review.

High-channel AIMDs with increased selectivity will lead to new fields of applications and considerably improve electroceuticals therapy. However, taking the step to fully implantable systems requires an innovative and feasible approach for the connector part. This needs a sound experience in implant manufacturing, close collaboration with neural surgeons and the flexibility to choose between different technologies and concepts to overcome the drawbacks of current systems.
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