ECE 1352- Analog Circuit-I

Term Paper

A Survey of Implanted Devices for Neuroprosthesis

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1. Introduction

In mammals, normal function depends on the continual electrical activity of nerves and muscles. The sources of this activity are the brain and spinal cord, sensory receptors within the tissues, local neural networks in the peripheral nervous system, and rhythmic or pacemaker circuits in cells of the heart and gut. The number of individual electrical impulses generated every second is very large, but against this background of general activity there are many circumstances in which it is desirable clinically, or useful from a research perspective, to modify the instantaneous electrical field density at a specific location. Such a perturbation is typically intended to generate action potentials in a given population of neurons, to activate a muscle directly, or to initiate a propagated depolarization of the myocardium [1]. Adequate selectivity of stimulation can normally be achieved only through the use of localized electrical fields, and therefore by means of small electrodes. For current to flow within the tissues these electrodes must be linked by wires through the skin to an external generator, or by electrical power. In both clinical and research applications there are considerable disadvantages to having wires that cross the skin, although in circumstances where it has proved unavoidable it has been managed successfully. The devices discussed in this paper are examples of the other two approaches, in which at least part of the stimulating system is entirely implanted.



Figure 1: Schematic representation of implantable stimulator.

Functional electrical stimulation (FES) and evolving neuroprostheses are dramatically changing therapeutic strategies for many neuromuscular conditions, especially for individuals with spinal cord injury. This includes FES for restoration of movement and posture in the spinal cord injured individual, phrenic nerve stimulation for respiration, spinal root stimulation for restoration of bladder and bowel function, cardiac assistance with skeletal muscle and dynamic graciloplasty for sphincter function [2]. While efforts during the initial research phase relied on surface or percutaneous stimulation, implementation of chronic clinically viable, acceptable and reliable systems depended on the introduction of fully implanted stimulation systems, as has been done for cardiac pacemakers. Neural prostheses are devices that utilize electrical stimulation to activate the damaged or disabled nervous system to restore function. Neural prostheses employ the technique of functional electrical (or neuromuscular) stimulation (FES or FNS) to generate action potentials selectively and thus to produce contraction of muscles. Stimulation is delivered in a coordinated manner, so that functional movements are achieved. This enables the individual to regain some control over their paralyzed muscles. Recently, neuroprostheses have reached wider clinical circulation, with the approval by the FDA of the first generation neural prosthetic systems for hand control and for bladder/bowel management. Standing systems are nearing clinical trials. This has been possible due to the implementation of implantable stimulator technology.

Current clinical applications use a third-generation implantable technology [3], which enables powering and control of limited number of muscles without having adequate data acquisition system for feedback control. Advanced FNS applications, precise control of muscle with higher resolution, require an increased number of channels of stimulation and simultaneous back-telemetry of multiple sensor and/or bio-potential signals from within the body or device itself. These telemetered signals are used to provide command control and/or feedback signals for an external control unit (ECU). The movement functions are under supervision from the ECU operated by the user. A schematic representation of the implantable stimulator is shown in figure 1. The requirement of more number of channels stem out from the fact that larger number of neurons have to be stimulated simultaneously through electrodes without cross-talking between them. In this paper, the basic configuration of single and multi-channel implantable devices will be discussed along with their present status and bottleneck of development, future improvement and prospect.

2. Implantable Device for Neuromuscular Stimulation

Many thousands of electronic devices have been implanted into human patients, of which the first, and still the most widespread, example is the cardiac pacemaker. The key features that have to be considered in the design of an implantable neuromuscular stimulator are *encapsulation*-the device and its leads and electrodes must be well tolerated by the tissues, *size*- a common rule of thumb that an implanted device should not exceed 2% of the body weight but depends on the place and area for which they are intended to be used, *lifetime*- no integral power supply and usually powered by electromagnetic coupling between an external and an internal coil, *programmability*-either use microprocessor and on-board memory compromising the size and complexity or retain much processing and control in the external device through appropriate protocol and communication signals to a relatively passive implanted system, *availability*- should support mass production ensuring proper clinical needs, *safety*- proper software to have user friendly control to minimize the risk associated with the maintenance [4].

A. Single-Channel Stimulator

On the evolution of implantable devices, the third generation of its kind extensively used microelectronics techniques and miniature hybrid surface-mount components to create a small single-channel unit that can be implanted in numerous quantities near the individual targets [5]. Since these devices use traditional glass sealing and metal-glass feed-through techniques, it is

difficult to expand them to a multi-channel system. Figure 2 shows the overall block diagram and data transmission protocol for a 5-b addressable single-channel microstimulator. The microstimulator receives power and data through an inductively coupled link and has overall dimensions of $2 \times 2 \times 10 \text{ mm}^3$. It consists of: 1) a silicon substrate supporting a stimulating electrode at each end and providing multiple feedthroughs, 2) a receiver circuit chip, 3) a hybrid capacitor used for charge storage of the stimulation pulse, 4) a hybrid receiver coil for power and data reception, and 5) a custom-made glass capsule which is electrostatically bonded to the substrate to protect the receiver circuitry and hybrid elements from body fluids [6].



Figure 2: System block diagram and data transmission protocol for a 5-b addressable single-channel microstimulator.

The device receives power and data through RF telemetry which is a transformer-like coupled pair of coils. This is achieved by amplitude modulation of a 2-MHz carrier. A 2-MHz carrier allows adequate miniaturization while permitting enough power to be transmitted to the receiver without much RF absorption and tissue heating. In order to supply enough power to the receiver coil, a high-efficiency transmitter/amplifier was used. Class-E power amplifiers show efficiencies in the 80–100% range and are suitable for this application where the coupling between the transmitter and receiver is usually very weak [7]. The receiverstimulator circuit design depends primarily on the data/power transmission format chosen for the microstimulator. The flow of data and power from the time that the transmitter is turned on is illustrated in figure 2. After power-on, the carrier amplitude is momentarily turned low to reset all of the blocks on the receiver chip and is then amplitude modulated to transmit a 5-b address to as many as 32 microstimulators located within the volume of the transmitter coil. After a specific microstimulator is selected, the carrier signal is turned high and then back to low again. It is maintained at this low level for a period of up to 200 μ s. The selected microstimulator will then deliver a constant current pulse of 10 mA into loads of 800 Ω through the stimulating electrode pair for the period of time that the carrier is low. Finally, the carrier is turned back high again, which will indicate the end of the stimulation period to the selected microstimulator.



Figure 3: Single-channel microstimulator system electronics.

The receiver circuitry for the microstimulator contains five main circuit blocks: two voltage regulators, an envelope detector for data demodulation, a clock recovery circuit, a control circuit, and an output pulse delivery circuit. The receiver coil picks up the amplitude-modulated RF carrier from the external transmitter and performs the following functions: 1) generates two regulated voltage supplies, one 4.5 V for the control circuitry and one 9 V for the output stage; 2) charges a 1- F tantalum chip capacitor to the 9-V supply; 3) regenerates the carrier clock; 4) decodes the modulated carrier to recover the control data for use by the control circuitry; 5) compares the demodulated address code with the microstimulator internal identification code (which is programmed into the control circuitry by laser-trimming metal lines in the circuit); 6) sends a pulse (up to 200 μ s in duration) from the control circuit to the output circuit, signaling the stimulation period if the correct microstimulator has been selected; and 7) discharges the storage capacitor through the stimulating electrode and body tissue. At the end of the stimulation period, the storage capacitor is recharged to the 9-V supply. The detailed receiver circuitry is shown in Figure 3. The voltage regulators consist of a full-wave bridge rectifier, storage capacitor C₂ and two shunt regulators implemented with zener diodes. The clock is regenerated from the RF carrier by taking the 12-20 V peak amplitude, 2-MHz sinusoidal carrier input and generating a 4.5 V square wave output. The function of data detection circuitry is shown in Figure 4. It consists of mainly three parts: a lowpass filter (LPF), a highpass filter (HPL), and a Schmitt trigger for envelope detection and noise suppression. The requirements for the output current delivery circuitry are to provide a cathodic first constant current pulse with fast rise and fall times for the entire pulse duration, and to charge-balance the storage chip capacitor during the time between two consecutive stimulating pulses.



Figure 4: Block diagram illustrating the operation and waveforms in the AM envelope detector.

The output circuit contains charge-balance circuitry, stimulus current regulator circuitry, and startup circuitry. The chargebalance circuitry provides a current of 100 µA through the M₁₃-M₁₄-M₁₅ current mirror with M₁₅ providing a reference current of 50 μ A and being twice the size of M₁₃. The stimulus current regulator circuitry generates a stable and supply independent current pulse. Transistors Q1 and Q2, and resistor R1 are used to create this stable stimulus pulse. Transistor M17 is used to switch this reference current on and off, while $M_{11}-M_{12}$ is used to provide a constant load. The drive transistor is Q_5 and has an area of 200 times that of Q1 in order to provide the constant current drive of 10 mA through the electrodes. Transistor M18 is used to achieve faster switching times and aid in charge removal from the base of the drive transistor. Transistors M19, Q9- Q11, and resistor R_6 are used to maintain a constant load current for the $M_{11}-M_{12}$ current mirror when no pulse is delivered to external electrodes. The startup circuitry consists of transistor M_{16} and diodes $D_1 - D_3$, which will prevent the circuit from remaining in a stable state in which zero current flows in the circuit, even when the power supply voltage is nonzero. The maximum stimulation frequency is a function of the amount of injected charge per stimulation period, the magnitude of the charge-balance current, and the time required to send the address data stream. This is to ensure that the electrodes are charge balanced at the end of each stimulation period. The control circuitry receives the unsynchronized envelope detector output and clock from the input circuitry. It generates a synchronized output and detects the logic ones and zeroes used to represent the address of the microstimulator. A '0' is differentiated from a '1' by the duration of a high pulse. Each microstimulator has five address bits which are programmable. Following synchronization, the received address bits are sequentially shifted into a 5-b register which is controlled by a 3-b state counter. The contents of this address shift register are compared with the programmed address of the stimulator to enable the stimulus control. The duration of the next low pulse received from the transmitter will determine the duration of the stimulus pulse. Upon completion of the stimulus pulse, the circuitry resets all of the counters and awaits the receipt of another address.

The overall system and individual circuit blocks developed for the implantable microstimulator is claimed to be applicable to a large class of devices which are remotely powered and controlled through RF telemetry. In some applications where the power demand is not as rigorous as in FNS, e.g., retinal and peripheral nerve stimulation, the hybrid coil and capacitor can be substituted by electroplated on-chip coil and thin-film capacitor. This will further reduce the size of the device and makes it more suitable for implantation in restricted anatomical regions. The microstimulator can also be expanded to a multichannel system by placing more electrodes on a longer substrate or using a flexible silicon ribbon cable with multiple electrodes.

B. Multi-Channel Stimulator with telemetry

Advanced FNS applications will require an increased number of channels of stimulation and simultaneous back-telemetry of multiple sensor and/or biopotential signals from within the body or device itself. These telemetered signals are used to provide command control and/or feedback signals for an external control unit (ECU). The movement functions are under supervision from the ECU, which is operated by the user. Through the system described below, various configurations of command control, feedback control, and stimulation will be achievable with implantable components, reducing the complexity and maintenance to the user while enhancing the functions available. The implantable hardware will be used to generate and telemeter both command control and feedback control information and to supply stimulus signals for a neuroprosthetic system employing multichannel activation of paralyzed muscles. This system could be used in FNS applications which seek to provide hand grasp and release in cervical level SCI, stroke, and head injury or standing and walking in thoracic level SCI, stroke, and head injury.



Figure 5: Block diagram of implantable stimulator telemeter system showing all possible functional elements.

Figure 5 shows a functional block representation of the Implantable Stimulator and Telemeter (IST) system [8]. The IST is a modular system, composed of functional blocks providing a wide range of stimulus and telemetry functions and a large number of stimulus outputs and telemetry channels. By selecting, combining, and fabricating only those functional blocks necessary for a target clinical application, an implantable circuit can be fabricated having only the features and capabilities needed for that patient function. This minimizes physical size, power consumption, cost, and fabrication effort for the device. Similarly, by combining a modular system of hermetic encapsulation, flexible lead wires, stimulating electrodes, recording electrodes, and sensors, an implantable system of minimum size can be suitably fabricated for long term surgical implantation in a specific clinical application. The IST system can be configured with the following functions.

- Up to 32 independent channels of stimulation (or sensory feedback), with independent control of stimulus pulse interval, pulse duration, pulse amplitude, interphase delay, and recharge phase duration (for biphasic stimulus waveform).
- Up to eight independent telemetry channels for sensors, with independent control of sampling rate and pulse powering parameters of the sensor (power amplitude and duration).
- Up to eight independent telemetry channels for processed (rectified and integrated) myoelectric signals (MES), with independent control of sampling rate and provisions for stimulus artifact blanking and processing control.
- Up to eight independent telemetry channels for unprocessed MES channels, with independent control of sampling rate.
- Up to eight independent telemetry channels for system functions, providing control or sampling of internal system parameters, such as internal voltage levels.

Due to overall timing constraints, implant circuit size, implant capsule size, number of lead wires, circuit and sensor power consumption, and external control and processing requirements, it is not practical to realize a single device having the maximal capabilities outlined above. However, the intent of the IST system is to provide the means of realizing an optimal implantable device having all the necessary circuitry and packaging to meet the anticipated clinical applications, without requiring design or engineering effort beyond that of fabricating the device itself. In addition, the IST system allows the use and coordinated control of up to four uniquely addressable implant devices under supervision of a single external controller for applications that may require larger functional capability or greater physical distribution within the body.

The functional elements required to realize the IST system include the following: 1) an RF receiver for recovering power and functional commands transmitted from an ECU using the method of load-shift keying (LSK) using circuit configuration modulator (CCM) [9]; 2) control logic circuitry to interpret the recovered signals, execute the command function, and to supervise functional circuit blocks; 3) multichannel stimulation circuitry for generating the stimulus pulses that are sent to the stimulating electrodes; 4) multichannel signal conditioning circuitry which provides amplification, filtering and processing for the signals to be acquired (MES and sensor signals); 5) data acquisition circuitry for sampling and digitizing these signals; 6) modulation circuitry for telemetering the acquired data through the RF link; 7) power regulation and switching circuitry for

selectively powering the included functional blocks of the circuitry, as needed, to minimize power consumption of the device; 8) system control circuitry to allow interrogation or configuration of the operation of the device.



Figure 6: IST command control structure and timing.

The basis of the IST system is its command control structure and is the means by which stimulus, back telemetry, and system control are accomplished (Figure 6). Two basic types of commands are provided; *stimulus control* commands and *data-acquisition control* commands. The communication structure utilizes a digitally coded pulse burst followed by several pulse durations. The coded pulse burst is used to indicate the specific type of operation that will be performed and enable a specific channel for that type of operation. It is decoded during the coding window. The number and timing of the pulses that follow

depend on the type of command that was issued (stimulus or data acquisition) and are used by the external controller to synchronize the phases of the function being executed. They are decoded during the duration window. The coded pulse burst contains 14 bits and encodes the implant identification code, command function; channel addressing, and any stimulus or sensor powering amplitude levels needed. It contains start and stop bits used to frame the pulse burst for proper decoding, and a parity bit for error detection of the data. In the case of a stimulation command, one of up to 32 stimulus channels can be addressed with one of eight stimulus amplitudes selected. For data acquisition, the command is further encoded to indicate the specific type of data acquisition; system, sensors, processed MES or unprocessed MES. For each type of data acquisition one of up to eight channels can be addressed with selection of one of eight powering amplitudes when applicable. The coded pulse burst is followed by one or more variable duration pulses, each pulse separated by an active low pulse which synchronizes the operation with the external controller. The handshake window is common to all commands and provides a short period for the external controller to receive the handshake signal back from the implant. This indicates that the command was received and decoded into a valid format, and that it is waiting to proceed with the rest of the command. During this time the implant remains in a ready state waiting for the next falling edge of the command. If the handshake signal is not received, then the command was either not sent correctly, not received correctly, or had the wrong device identification code. The external controller then has the options of continuing with the command, aborting it, or possibly trying the command a second time. For a stimulus command, up to three phases of variable pulse timing are provided. The stimulus duration specifies the width of the stimulus pulse. The interphase delay specifies the duration of the delay between the two phases of a biphasic stimulus pulse (anodic and cathodic). The recharge duration specifies the duration of the cathodic phase of the pulse, and allows the partial recovery of the injected stimulus charge if desired. For a data acquisition command, up to five phases of variable pulse timing are provided. The sensor power duration specifies the duration of the powering pulse of the sensors and/or any signal processing circuitry associated with the data acquisition channel addressed. During this phase, the sensor and associated circuitry are powered and allowed to settle and then, at the end of the pulse, all signals are sampled and held. This is followed by up to four *data pulses*. The data pulses initiate the sequential digitization of each signal element and its subsequent telemetry. For any command, at any time during the duration window, an end pulse can be used to terminate a command. Therefore, in a stimulus command, it is not necessary to control the delay or recharge phases directly if it is not desired. For the processed MES signal acquisition channels, MES signals are continually input to signal processing circuitry, and are then typically rectified and integrated. Since commands are input to the IST system over a single RF link, it follows that all commands must time-share the communications link.

To implement the capabilities of the command structure, extensive digital control circuitry is needed. To maintain adequate control processing with minimal circuit size and minimal power consumption, the control logic was implemented as an application-specific integrated circuit (ASIC). This ASIC combines all command decoding and control functions into a very small circuit area with a very low power consumption. It also provides the current regulators that are used by the stimulus output stages and the sensor powering circuitry. Figure 7 shows the functional block diagram of the ASIC control logic. Incoming

commands are decoded and functions are executed using a 1-MHz master system clock. The control command is decoded by using two time windows generated by the ASIC logic. The binary pulse burst is decoded within the coding window and enables the appropriate system functions and addressing. The duration window decodes the duration pulses and outputs these pulse durations and timing signals to the enabled channels and control logic in proper sequence. During the command decoding, any errors encountered are latched into the system status register and can be read (telemetered) by the external control system with a subsequent system command.



Figure 7: Functional block diagram of IST ASIC.

Figure 8 shows the simplified schematic diagram of implant circuit and Figure 9 shows the line drawing of the implant packaging. An error can be detected by the controller during the handshake window by a lack of handshaking in response to a command. Some small problems with the current device remain and additional work is required to correct these and to develop additional modules for other clinical applications. Some noise pick-up by the sensor circuitry from the RF field used to power and control the implant is experienced. This is manifest primarily as a small dc offset on the digitized sensor data, the magnitude of the offset being dependent on the relative positions of the pair of RF coils and the power density of the RF field. Ongoing effort is to alternate command control schemes and improved analog circuit design and component layout to eliminate the problem. Work is ongoing in the design, development, and interfacing of myoelectric recording as well as multiple implantable joint angle transducers (IJAT) devices.



Figure 8: Simplified circuit schematic diagram of implant circuit.



Figure 9: Line drawing of implant packaging.

C. Different Multi-Channel Stimulator

Among some recent example of multi-channel stimulator one is developed at the Technical University of Denmark [10] without any reverse telemetry. It only facilitates 4 channels with 5 MHz inductive link. The chip is able to generate chargebalanced current pulses with a controllable length up to 256 μ s and amplitude up to 2 mA for stimulation of nerve fibers. The quiescent current consumption of the chip is approx. 650 μ A at the supply voltages of 6-12 V, and its size is 3.9 x 3.5 mm². It uses pretty simpler command word and 8-bit words are chosen to represent the pulse amplitude and duration, and to select the output channel for stimulation by which the same transmission protocol can be employed for stimulators with up to 256 output channels. It employs cyclic redundancy check (CRC) for error detection. A simplified block diagram of the chip is shown in Figure 10. It uses Phase-locked-loop for the clock recovery from the input data. A novel method for charge-balancing is incorporated where the duration of the discharge pulse is calculated from the charge in the stimulation pulse.



Figure 10: Chip block diagram

A 100-channel neurostimulation circuit comprising a complementary metal oxide semiconductor (CMOS), ASIC has been designed, constructed and tested [11]. The ASIC forms claimed to be a significant milestone and an integral component of a 100electrode neurostimulation system being developed by the authors. Communication of both data and power across tissue via radio-frequency (RF) telemetry such that externally programmable, constant current, charge balanced, biphasic stimuli may be delivered to neural tissue at 100 unique sites. An intrinsic reverse telemetry feature of the ASIC has been designed such that information pertaining to the device function, reconstruction of the stimulation voltage waveform, and the measurement of impedance may be obtained through noninvasive means. To compensate for the scarcity of data pertaining to the stimulation thresholds necessary in evoking a physiological response, the ASIC has been designed with scaleable current output. The ASIC has been designed primarily as a treatment of degenerative disorders of the retina whereby the 100 channels are to be utilized in the delivery of a pattern of stimuli of varying intensity and or duty cycle to the surviving neural tissue of the retina. However, it is expected that other fields of neurostimulation such as cochlear prosthetics and functional electronic stimulation may benefit from the employment of the system.



Figure 11: Block diagram of system operation

Figure 11 explains the basic block diagram of the system. External image processor takes in an image from the environment. Image is pixelized according to a programmable compression protocol. External Encoder/Transmitter translates the pixelized image into a series of encoded RF sequences according to a programmable stimulation protocol. RF sequences are broadcast through the ocular tissue via the Data and Power Link. Implanted Receiver/Decoder/Stimulator receives the RF sequences, decodes the data and delivers stimulus from an active electrode (total 50 electrodes), through the neural tissue and returning to an indifferent electrode (total 50 electrodes). Telemetered stimulus data (the end-of-phase voltage across the current source) is detected externally by way of monitoring the transmitting antenna. The detailed ASIC operation is shown in Figure 12. RF energy (received by antenna circuit) is rectified and stored in Power Supply where it is divided and regulated to provide VDD, VLOGIC, and VREF for analog stimulation, digital logic, and reverse telemetry, respectively. Simultaneously, the RF signal passes through packet and sequence detectors then into signal divider for cycle counting and division by eight. These signals are passed to the relevant sub circuits via the data bus. Burst router determines which packet burst of a given sequence is being received, and directs the clock signal to the appropriate module of the circuit accordingly. Initially, synchronization burst detector is activated. An invalid synchronize burst evokes a reset signal from the error/completion detector (via error bus), thus placing the circuit in a state wherein it awaits the next valid synchronize burst. Otherwise, the Burst Router activates the Row/Column Counters & Polarity circuit such that the Row/Column Buses are configured according to the quantity of Clock cycles received. Clock-1 circuit filters the clock signal, subtracting one from the clock signals during each burst. Signals on the Row/Column buses determine which of the Row/Column switches within active and indifferent electrode switching circuits are connected to the output of the phase routing circuit. As a future improvement this device requires improved digital-to-analog converter (DAC) system to support multiplexing and parallel stimulation of the electrodes.



Figure 12: Detailed ASIC operation

Another kind of implantable device for FES is being developed in University of Southern California named BION (bionic neuron) [12]. Each BION microstimulator consists of a tiny, cylindrical glass capsule whose internal components are connected to electrodes sealed hermetically into its ends (Figure 13). BIONs receive power and digital data from an amplitude-modulated RF field via inductive coupling described before. One such transmitting coil can power and control up to 256 individually digitally addressable implants. BION implants generate digitally controlled, current-regulated pulses. They have been shown to be located permanently and stably in the target muscles, resulting in consistent thresholds and muscle responses over extended periods of time. If additional channels of stimulation are required to treat the patient's condition, they are easily added. BIONs

are small enough to implant by injection in an outpatient procedure that can be performed by any physician. They can be placed in small, deep or hard-to-reach muscles that have been impossible to stimulate selectively from the skin surface. Eliminating the expense and morbidity of surgery reduces total clinical costs, an important factor for the ultimate commercialization and reimbursement potential of this treatment. None of the external or internal components is intrinsically expensive to manufacture, so economies of scale will eventually make BIONs highly cost-effective for rehabilitation of a wide range of neuromuscular dysfunctions as reported [12]. Each operation of each implant is initiated by a command consisting of 3 data bytes and various formatting and parity bits, requiring a total of 288 µs for transmission.





The electronic circuitry in each implant consists of a self-resonant receiving coil, a custom integrated circuit (IC) chip, a Schottky diode, and two electrodes (Figure 14). The IC derives direct-current (DC) power by rectifying and filtering the carrier energy picked up by the coil. The carrier itself provides a synchronous clock and its amplitude modulations encode a serial bit stream, which is decoded by a state machine in the IC. The first data byte specifies an address, which is compared with the address specified by a hardwired read-only memory (ROM) in each IC. If they match, the subsequent data bytes are decoded to specify the desired operation. Stimulation operations require a pulse width and a pulse amplitude specification. The power delivered during a maximal stimulation pulse (30 mA×17 V=0.51 W) is about 100 times higher than the power that can be delivered by the inductive link. This problem can be overcome because the stimulation typically required to activate a muscle consists of relatively brief pulses ('0.2 ms) at low frequencies. During the interpulse period (typically \propto 50 ms), energy is stored on an electrolytic capacitor consisting of the tantalum electrode and the body fluids. This results in about 5 μ F capacitance with less than 1 μ A DC leakage at the +17 V DC compliance voltage of the BION. The counter-electrode is activated iridium, which forms a metal–electrolyte interface that resists polarization under any achievable sequence of charging and discharging of the

capacitor electrode. When the carrier is on but the implant is idling, the capacitor electrode can be charged at one of four selectable rates (0, 10, 100 and 500 μ A) until it becomes fully polarized to the +17 V DC compliance voltage. Sensing functions require a back-telemetry link that operates during pauses in the external carrier, during which the external coil acts as a receiving antenna. The self-resonant coil in the implant acts as the tank circuit for an oscillator that is amplitude-modulated to transmit digitized data obtained from a previously commanded sensing operation. Sensing and back-telemetry functions under development for the BION2 system are required to provide command and feedback signals for FES applications. The specifications and how their signals will be integrated into a portable control system for a particular clinical requirement have not been decided yet.

Implants

External Components



Figure 12: BION system architecture

3. Conclusion

Replacement of damaged or lost nerves by artificial implants with recording and/or stimulating electrodes has been a goal of many efforts since several decades. In this paper recent improvement in the development of single and multi-channel implantable devices is discussed. The successful implementation and application of a neuroprosthesis is a common matter of different fields of research, such as neurobiology, medicine, computer science, microelectronics, microtechnology, surface science, electrophysiology and electrochemistry. The high complexity of both structure and function of the nervous system is, however, a real obstacle on the way to apply simple neuroprostheses, whereas really functioning, easy-to-handle and long-term stable

neuroprostheses are still not available yet. Nevertheless, implants are applied successfully in a few particular cases, which mean that a partial restoration of the natural function can be achieved over a longer time period. These special cases are the cochlea implant and implants for bladder stimulation. Other applications, as pain management or stimulation of the limbs, are still not in the state as broad applicability and reliability could be claimed. In the field of replacement of sensory functions besides hearing by cochlear implants, visual prostheses are the target of ongoing efforts. Since the visual system is the most complicated sensory one, the natural impediments are particularly large. A satisfactory prosthetic replacement of visual function and microelectronic improvements which will allow parallel processing of detectable photic stimuli. FES applications differ greatly in the nature of their command and control problems, and individual patients differ in the details of their sensorimotor pathology and their musculoskeletal mechanics. A parallel effort is underway to develop a general computer modeling environment in which it will be feasible to create accurate models of specific musculoskeletal systems and FES interfaces and control systems in order to understand these requirements [12].

Regarding the success of ready-to-implant device and data processing, in the past, many approaches failed because of encapsulation, connection and stability towards leaking and corrosion. Realization of an appropriate data transmission and processing is still a big challenge, particularly for sensory neuroprosthesis where multi-channel stimulation is required. On-chip electronics, multiplexing and parallel processing are key features to manage the huge amount of data necessary for a useful stimulation. There is an intrinsic difficulty in developing a stimulation system for physiological excitation of tissue wherein the appropriate current threshold and modulation parameters are yet to be thoroughly defined. Recent improvement in packaging, encapsulation, size of chip and power consumption indicating a good sign of successful implementation for these kinds of devices. Wide dynamic nature of neuromuscular junctions at different parts of the body imposes a challenging task to analog microelectronic designer.

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