Wireless smart implants dedicated to multichannel monitoring and microstimulation

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Abstract

This invited paper covers several techniques and methods employed to build high reliability circuits and systems dedicated to implement advanced implantable and wirelessly controlled smart medical devices such as sensors and microstimulateurs. A global view of typical micro-devices such us neural signals monitoring device, cortical multichannel stimulator, as well as peripheral nerve interfaces to recuperate bladder functions, is given. In addition, case studies related to peripheral and cortical neural systems are reported. In all these devices, special attention is paid to low-power management of involved systems and to the design of corresponding circuits of such typical implantable multi-disciplinary microsystems.

1. Introduction

The advent of microelectronics allowed the emergence of implantable microdevices (stimulators and sensors) and the interface of such microdevices to complex biological systems. However, the needed tools, for elaborated diagnostics and subsequent flexible processing to achieve neuromuscular functions control, still missing. Microelectrodes interconnected to sensors have been applied to the exploration of the brain and its physiological functions [1-4]. These measurements help researchers to understand the behavior of the primary cortex and may allow physiological studies of the brain in order to address specific dysfunctions at the cortical as well as the deep brain levels and then to control prosthetic devices of paralyzed people. Simultaneous recording of large group of cells located in specific areas of the cortex is becoming a necessity. Brain dysfunctions such as autism, epilepsy, and depression/schizophrenia are among the numerous applications of the proposed monitoring tools [5-8]. Required devices must work at ultra low-power level and be powered wirelessly and communicate via a RF link.

On the other hand, electrical neurostimulation has been carried out to recuperate several peripheral organs functions. More particularly research on bladder control for urine voiding and to prevent incontinence [9] is elaborated. The currently employed techniques do not allow to reduce the detrusor–sphincter dissynergia which induces high detrusor pressure that can eventually lead to incontinence. Also, these techniques do not avoid the detrusor hyperreflexia which results from the hyperactivity of the autonomic nervous system.

The maximum efficiency of transcutaneously provided power to implants is below 20% [10]. It is even less when more power is required and depends on a wide range of input conditions such as implant depth, transmitter placement, etc. Available systems are not able to handle the needed bidirectional high-speed wireless data rate transmission and the high number of measurement and stimulation channels. More particularly, they can not be used for recording or stimulation several primary cortical sites. Available devices lack efficient sites positioning, noise rejection, and embedded signal processing module. Also, Several circuit techniques were introduced to address the various issues of these devices, such as low noise and low power, offset cancellation, DC level blocking and filtering, etc [11, 12].

We are designing and implementing autonomous and reliable multichannel monitoring and stimulating microsystems suitable for permanent implantation. The multifunction devices are powered and communicate data wirelessly. Overview of the proposed systems is given in the following sections. Next to the system description, we focus on few main building blocks such as the analog front-end, the telemetry RF data link, etc, followed by results and conclusions.

2. Description of system architectures

Embedded integrated circuits with microprobes using flip chip technique are used to meet the required flexibility in cortical interfaces. Power consumption and small integration area are major concerns when designing such implantable circuits and systems. The following sections include system architectures of three main undertaken projects by our PolySTIM team dealing with implantable devices.

2.1. The cortical monitoring devices

The global cortical monitoring system is composed of two main parts: An external controller and an implant. The front-end of the implant includes analog conditioning circuitry for each channel. The massively parallel structures necessitate to employ analog multiplexing of channels, and digitizing the extracted data for further wireless data transmission through an inductively coupled link. The transmitted data are then fed to a PC for software data processing and storage. Figure 1 shows a block diagram of an implantable multichannel monitoring device [12]. Up to 1000 channels will be integrated on one implantable device. Several small complexity arrays will be routed to one internal multiplexer before the analog to digital conversion and subsequent transmission.

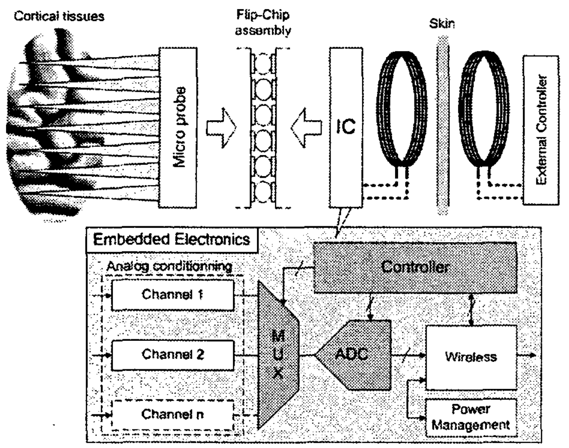

Fig. 1. Block diagram of a cortical monitoring device.

Low amplitude neural signals (1-10μV) are very sensitive to noise. Thus, special attention is paid to reduce the input noise level of the front-end preamplifier and critical stage of the conditioning chain. Also, the neural signal must be sufficiently amplified within a bandwidth of 20 kHz. Dominant noise sources in CMOS amplifiers are thermal and $1/f$. The thermal noise is directly related to the DC bias current of the transistors. Thus increasing the supplied current will generally lower the thermal noise floor. $1/f$ noise decreases almost linearly until it reaches the thermal noise floor at the corner frequency of the amplifier. This $1/f$ noise level can be reduced when width and length of the amplifier's input transistors are increased. This approach usually makes the preamplifier to dominate the entire integration area.

The proposed front-end mitigates this trade-off with chopper modulation technique [10, 12] that allow to decrease considerably the preamplifier's transistors. Each channel of the front-end monitoring system includes a low-power chopper amplifier which is composed of an input modulator, a low-noise preamplifier, a 2^{nd} order band-pass filter, a demodulator, a low-pass filter, and a gain amplifier.

A fully differential topology, combined with transistors operating in strong inversion, is used to achieve to achieve low input noise level. The transistors from other stages operate in weak inversion to reduce the power consumption. The filters are based on Gm-C structures and a multi-input transconductor to reduce circuit complexity [12]. Next to channel multiplexing stage, an analog wavelet core is involved for pattern detection and data selection to transmit to the external controller.

2.2. The visual cortical stimulation

Similarly to the monitoring devices, the visual implantable monitor and stimulator is based on two main parts (external controller and implant). A thin and modular approach is adopted, as opposed to fully monolithic designs [13]. The device is composed of several small size stimulation modules lying flat on the cortex, connected to penetrating microelectrode arrays, and an interface module, placed away from the stimulation sites and interfacing wirelessely with the external controller. All components forming the implant are to be laid out on a common thin flexible substrate, eliminating the need for connectors. Fig. 3 shows a simplified schematic of the device to be implemented from the modules described in the next sections [14, 15].

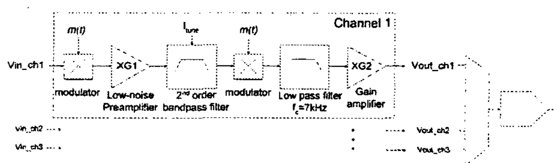

Fig. 2. Block diagram of a monitoring channel.

The stimulation modules generate constant current biphasic pulses, using monopolar or bipolar configurations. In the case where monopolar stimulation is used, the return current reference electrode can be distant from the stimulation sites, connected to the interface module, or distributed among inactive stimulation microelectrodes.

2.2.1. Typical stimulation module. A typical stimulation module has four parallel channels, each one drives four stimulation micro-electrodes (sites). All parameters of stimulation pulses are programmable. Pulses frequency, as well as train and inter-train durations, are determined by the rate at which individual pulses are being triggered by the external controller. Electrode shorting can be performed between pulses to ensure charge balance of biphasic pulses.

In order to minimize power consumption, the controller uses a low voltage supply and a configurable communication protocol, minimizing data transfers and its operating clock frequency. To achieve this, every stimulation parameter is sent to the stimulator with a minimal resolution defined in an initial configuration sequence, or is omitted in case a constant parameter can be used. Stimulation on parallel channels can be either sequential or synchronized. In the latter case, parameters are stored to the appropriate channel controller when received and all stimulations are activated by a subsequent trigger instruction. In addition, bi-directional current sources are made of complementary thermometer code digital to analog converters [14], and have four operating ranges (from 17 to 140 μA).

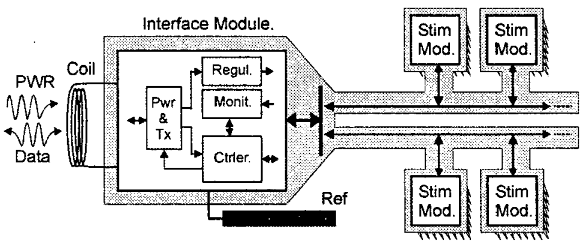

Fig. 3. Block diagram of the cortical stimulator.

Finally, to prevent data transmission errors, every configuration parameter is parity encoded prior to be stored, and valid parity of every parameter is continuously verified. Upon data corruption detection, stimulation is inhibited, and registers have to be reloaded to continue stimulation normally. Regarding the interface module, it decodes and transfers stimulation instructions to the appropriate stimulation module and sends telemetry data to the external user. The interface module controller manages half-duplex communication cycles, transfers appropriate commands to one or many stimulation modules in a specific or broadcast manner, manages monitoring operations and results, sets the reference voltage, and manages stimulation module activity to minimize power consumption.

2.2.2. Stimulation types. Several stimulation types can be performed. A constant input voltage follower is used to keep the electrode sinking and sourcing the total stimulation current at a constant voltage [15]. To increase the allowable stimulation current, the return electrode voltage can be dynamically set to the positive or negative supplies by the controller. Monitoring stimulation voltage and current should be performed to ensure that stimulation stays within safe limits and for troubleshooting. The voltage of one monitoring buffer is enabled and sampled by an analog-to-digital converter.

For current monitoring, only a single stimulation site in the system can be active, configured for monopolar stimulation, and other electrodes have to remain in a high impedance state.

2.3. The bladder control

The proposed bladder implant performs two complementary types of stimulation: selective for bladder voiding and permanent for detrusor hyperflexia curing [16]. The system includes an external controller that powers up the implant and allows the selection of parameters by the user and the generation of stimuli (Fig. 4). The selective stimuli are generated continuously while the inductive link remains between the implant and the controller. To choose the type of stimulation as to determine the length, frequency and amplitude of the stimuli, a friendly user interface is managed by a Finite State Machine and followed by a power amplifier for the energy transfer by RF inductive link.

On the other hand, an implant having selective and permanent stimulation modes is used. The device is composed of data and power recovery blocks, an embedded battery, selective and continuous stimuli generators and a selector. The recovery block receives the data sent by the external controller. The battery is dedicated to the PIC only because this operation needs to be performed without the external controller. The selective stimulator, based on a programmable controller (FPGA), gets its energy from the external controller via the power recovery block. It is the one in charge of the decoding of the data as so of the verification of the validity of those data. Furthermore, it has to generate selective stimuli or transfer the control to the PIC for permanent stimulation. The PIC generates low amplitude stimuli and supervises the power switching between the RF received voltage and the battery one. The selector configures the connection of one of the signal generators to the bipolar electrode.

3. Power and data issues

The transcutaneously received energy to power up the implant must be rectified and regulated. This regulation can be achieved using a low drop-out (LDO) linear regulator. Regulators using N-type pass devices show fewer stability problems, better regulation and lower Ohmic output impedance characteristics than their P-type counterparts. However, regular N-type pass devices require their gate voltage to be significantly higher than their source output, making the LDO impossible if the feedback amplifier is powered by the input voltage. This would result in reduced power efficiency and stimulating current compliance, limited by the voltage swing of the stimulator output stage [17]. This problem is eliminated using a native transistor. These transistors have their threshold voltage much lower than regular transistors, at the expense of showing larger discrepancies between samples. On the other hand, a regular transistor can be used for the low voltage output since the high regulated voltage provides sufficient voltage swing at the output of the feedback amplifier.

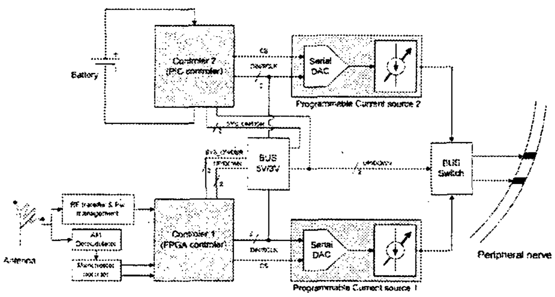

Fig. 4. Block diagram of the bladder implant.

The reference voltage, required by regulators, is set by a bandgap circuit, for which the most important characteristic is its line rejection. As a matter of fact, temperature does not vary substantially, but the rectified voltage shows strong ripple because a small tank capacitor must be used at the input for size considerations. However, both the regulator and the reference are then interdependent. So a start-up circuit is used to provide the bandgap reference with the rectified voltage on power up, and then switch to the regulated voltage when the latter becomes sufficient.

On the other hand, data communication is ensured by a half-duplex data link using On Off Keying (OOK) and Load Shift Keying (LSK) techniques for downlink and uplink respectively [15, 18]. In this data link, a digital demodulator is used for extracting clock and data signals from an OOK modulated 13.56 MHz carrier. Envelope detection is performed by directly digitizing the input carrier using Schmitt inverters with different threshold levels. A high-level envelope results in both inverters toggling at the carrier frequency, while a low level results in constant high outputs. However, because of the gradual decay/rise of the carrier envelope, caused by the channels limited bandwidth and the stored energy in the receiver resonant circuit continuing oscillations when the rectifier cuts off, the inverters will cease their toggling activity for different periods. The output of the narrow threshold inverter is used for sampling the state of the wide threshold inverter. Also, to minimize the risk of glitches on envelope level transitions, three consecutive samples are used as inputs to a combinatorial logic majority detector. The energy is dissipated in this circuit by the Schmitt inverters. It is maximum when the input signal ceases and stabilizes between logic input levels. The inverters are optimized for presenting the largest possible output resistance while allowing their output to toggle at the carrier frequency.

The digitized carrier and envelope signals are fed to a digital clock and data extractor. The carrier is used for generating a data clock at a constant frequency. Data modulation uses a constant high period for ones and a short low pulse at the end of the data period for zeros. Envelope low levels reset the carrier counter, which restarts running when the envelope goes back to high. The LSK modulation technique is used for uploaded telemetry. Changing the rectifier from full-wave bridge to half-wave configurations, hence its equivalent input impedance, resulting in detectable load variations.

The data link is managed by a communication protocol, in which each cycle is composed of an initializing header, any number of data instructions, followed by a reply from the implant including monitoring results, and flags that are set in case communication errors were detected in the preceding downlink data stream, or if volatile configuration parameters are corrupted. A stream of ones for 400ms resets the system. On power-up, an initial handshake allows the user to ensure that coil alignment and transmitted power are adequate for error free data transfer. This is followed by a configuration sequence, in which data rates are determined, as well as links packet lengths.

4. Results

All analog and mixed-signal circuits of presented systems have been fabricated on a standard 0.18 µm CMOS process. Although the presented components are designed for a reduced size prototype, and complete integration as well as final assembly remains to be realized, the parameters measured allow us to make reasonable assumptions on the performance of complete devices. Layout of 3.5x3.5 mm^2 representing a module of the cortical stimulator is given in Fig. 5.

Regarding the data transmission rate, a possible implementation of a complete visual stimulator comprising 1024 stimulation sites (32 modules of 32 electrodes each) distributed on 6 channels. Considering that 5 bits are required for addressing the sites, and using the configurable communication protocol of the modules for defining instructions with 20 bits including data encoding for error detection and correction, stimulation patterns can be sent in 13.3 µs at a data rate of 1.5 Mbps.

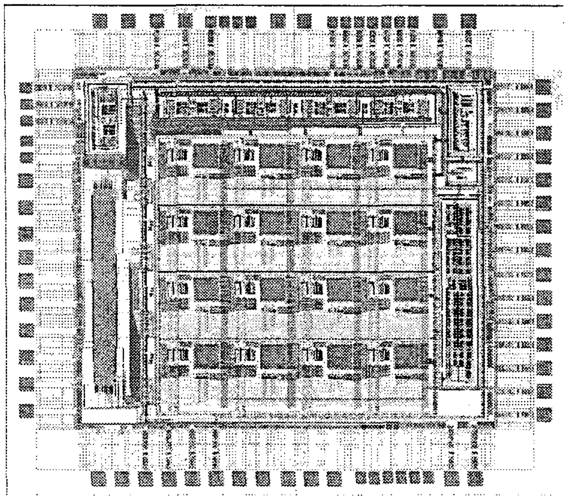

Fig. 5. Layout of a stimulation module.

The power consumption of such a stimulator, including the dissipated power by the LDO is below 50 mW. Assembly of the modules and the micro-electrodes on a flexible miniature substrate are to be completed before the device can be implanted for in-vivo testing. For some presented devices, non implantable prototypes have been implemented using commercial components to validate the critical parts and to validate operation chronically in animals such as rats, cats and dogs. For example the bladder implant has been validated chronically in dogs based on elaborated protocol. These experiments allowed to validate both stimulation techniques (Permanent and selective), and the long term (up to 8 months) operation of the devices. The average voided urine volume was improved more than five times comparing with conventional stimulation of the detrusor. These experiments allowed to reconsider steps and to adjust few parameters of the full custom designs.

5. Conclusion

The design and realization of prototypes of the main components of presented systems (monitoring devices, visual cortical stimulator, and bladder controller) have been described. The devices offer full latitude on parameters and configurations. Low power is achieved by combining low and high supply voltages in most designs.

In the cortical stimulator, attention was paid to propose a low-power architecture with dynamic voltage low impedance return current electrode, as well as by the usage of programmable parameters, reducing data and clock rates. Safety features include volatile memory checking, and stimulation voltage monitoring. Measurements on prototype show that the most stringent performance requirements for a stimulator with 1000 sites are met.

For the bladder, the proposed stimulator allowed to confirm the benefit of combining both selective and permanent stimulations that have great promises for bladder rehabilitation with no drawback. The fabrication of the monitoring system parts is undertaken and a device assembly is our next step to validate it as well as the cortical stimulator in vivo in Monkey soon.

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