Design of a versatile voltage based output stage for implantable neural stimulators

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***Abstract*—Neural stimulators have the potential of becoming very important devices for the treatment of a wide variety of diseases. One of the major problems with existing stimulators is the limited waveform adjustability. This precludes the use of sophisticated stimulation programs and thereby affects the efficacy of the therapy applied. For this reason a new type of stimulator is required.**

The physical principle underlying stimulation is based on elevating the tissue potential up to a particular level by injecting a particular amount of charge. Furthermore the injected charge needs to be canceled precisely in order to prevent tissue damage.

Most existing stimulators use a current based architecture in which the charge is controlled by enabling the stimulator for a particular amount of time. Voltage based stimulation however yields a much higher power efficiency.

A novel type of voltage based architecture using indirect current feedback of the tissue current is proposed. Using a current integrator with a very high dynamic range the injected charge can be controlled very precisely, while any arbitrary voltage waveform can be used for stimulation. Circuit simulations prove the feasibility of the approach and show a charge mismatch in the order of 0.1% paving the way to full charge balancing. Furthermore, they predict correct functionality over all process corners, including mismatch. The system only uses a single-ended supply and its quiescent power consumption is less than 15μW.

I. Introduction

Throughout the history of medicine, drugs have been one of the major ways for the treatment of diseases. However it is realized more and more that drugs alone cannot cure all diseases efficiently. Most drugs suffer from unwanted side effects and the spatial selectivity of drugs is usually low. Another form of treatment consists of electromagnetic stimulation of the body. As neural cells use electromagnetic signals to operate, these signals can be influenced by artificially generated electromagnetic signals in order to establish a desired effect.

Cardiac pacemakers are one of the most well known types of stimulators. During the last decades these heart muscle stimulators have gone through an extensive development: current pacemakers are fully implantable and include feedback (stimulation based on measured heart activity).

Neural stimulation is a form of electrical stimulation shown to be effective for a wide variety of diseases, including Parkinson's disease [1] and tinnitus [2]. Compared to pacemakers however neural stimulators are still very primitive devices. Their implantability is limited and as a consequence the device is implanted in the chest and subcutaneous wires lead to electrodes in the brain. These wires are a common source of malfunctions and complications. The size constraint which limits the implantability is mainly due to the battery in the device.

Furthermore the stimulation does not involve feedback. The stimulator is simply imposing some stimulation pattern on the tissue without knowing what the result is. It is better to use a more sophisticated approach using feedback, so that the stimulation can be adapted to the neural response.

Finally, neural stimulators suffer from another major problem: neural tissue tends to have a large adaptability. This means that due to the fixed stimulation pattern the tissue will gradually habituate, after which the symptoms of the disease return. Therefore it is important to have as much flexibility in waveform shape as possible. Additionally this might offer more effective stimulation compared to the block shaped pulses current stimulators are limited to.

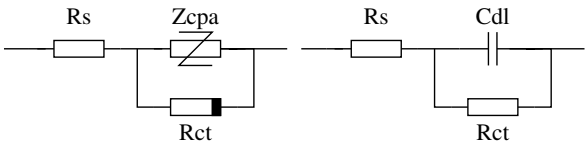
For all the reasons mentioned avove there is a need for a stimulator with the following properties:

- Small enough to be implanted in the skull. To reduce the size of the battery very low power consumption is required.
- Include feedback. Stimulation based on the response of the neural tissue to increase the effectiveness.
- Very flexible waveform adjustability. This will reduce tissue habituation.

In this paper the design of a stimulator meeting these requirements is discussed. In section II the physical principles underlying electrical stimulation of neural tissue are summarized. Based on these principles the system architecture is defined and a system level design is provided in section III. Finally in section IV the circuit level simulations of this design are presented.

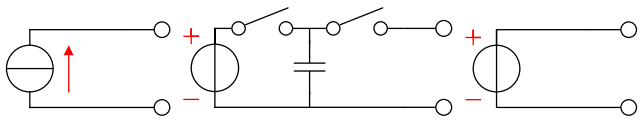
II. Electrical stimulation of neural cells

Neural cells (neurons) are surrounded by a cell membrane characterized by a particular potential difference between the inside and outside [3]. This potential difference is the result of a dynamic equilibrium of ion fluxes (mainly potassium and



(a) Complex model (b) Linear model

Fig. 1. Models for the Electrode-tissue interface



(a) Current steered (b) Charge steered (c) Voltage steered

Fig. 2. Three fundamental system architectures

sodium) through this membrane. When this potential difference is elevated above a particular threshold, a mechanism is started which will generate a sudden voltage change over the membrane: an action potential (AP). The AP will propagate through the membrane to other cells. In this way cells can send APs towards other cells, which can lead to some particular body functionality.

A. Artificially evoking or blocking APs

It is possible to implant electrodes close to the neurons and hereby the outer membrane potential can be changed. When this change is large enough an AP is either evoked or blocked. To understand how the outer membrane potential is influenced by the electrode, an electrical model of the electrode-tissue interface is required.

A common model used for this purpose is depicted in Figure 1a [4]. First of all it has a component called 'Constant Phase Impedance' (Z_{CPA}) which is described in the Fourier domain as:

$$Z_{CPA} = \frac{1}{(j\omega C_{dl})^{\beta}} \qquad 0 \leq \beta \leq 1 \tag{1}$$

Here C_{dl} is the 'double layer capacitance'. It represents the capacitive behavior of the interface. The other component is a highly non linear resistor R_{CT}. The current through this resistor is the result of charge transfer from the electrode into the tissue. The resistor has an double exponential voltage to current relation. R_s represents the impedance of the tissue itself.

A combination of an element described in the Fourier domain (for which no closed form in the time domain exists) and a highly non linear component makes this model very complex to analyse.

In order to work with this model it was linearized (Figure 1b). Z_{CPA} has a capacitive nature and is therefore transformed into capacitor C_{dl}. The nonlinear resistor is replaced by a linear resistor. The response of both models is subsequently compared in Matlab to verify whether the linear model is still accurate enough. It was found that the most important factors in the response of the tissue model (the maximum current and the frequency spectrum) do not change significantly by linearizing the model.

We therefore can conclude the electrode-tissue interface can be described using a capacitive model. This means that in order to elevate the tissue potential above a particular threshold, we need to inject a particular amount of *charge*. This implies that charge is the fundamental quantity associated with the electrical stimulation of the tissue.

B. Safety

It is important to stimulate the tissue without inflicting any permanent damage. The mechanisms associated with tissue damage are not yet completely understood, but it is known that among others two conditions must at least be satisfied to prevent damage.

First of all it is important not to inject any net charge into the tissue [5]. This means that after injecting a stimulation pulse consisting of a particular amount of charge, this charge needs to be withdrawn from the tissue.

The second safety constraint is that the tissue current is not allowed to exceed a particular maximum value. This means that the stimulator must incorporate some mechanism which ensures the tissue current is limited.

Another safety aspect that is important to consider is the large spread of parameters the system needs to deal with. Stimulation parameters can span several decades of magnitude (from several μA up to tens of mA). The same holds for the electrode-tissue model parameters which can vary significantly from patient to patient.

III. System architecture

When designing an output stage for a stimulator the first choice that needs to be made is with what electrical quantity the tissue is to be stimulated. Four fundamental quantities exist: voltage, current, charge and flux. The quantity flux is discarded, because it is hard to integrate flux based sources on chip for the low frequencies applied in neural stimulators. This leaves us with three fundamental architectures, depicted in Figure 2.

Almost all existing architectures use current steered stimulation (e.g. [6]). Usually the current source has a constant value and is switched on for a particular amount of time, yielding a block shape waveform. The reason for this approach is that in this way the charge is quite easily controlled: by switching the current source on for a particular period. Drawback of this approach is the need for accurately matched current sources, which are hard to implement. Furthermore there is limited waveform adjustability (as stimulation pulse can only be square shaped).

A charge steered approach is closest to the physical principles underlying stimulation. One way to make a charge source is using a capacitor. Considering the maximum stimulation parameters the capacitor needs to be in the order of $1\mu F$, which is hard to integrate on chip.

A voltage steered approach will yield a much more realistic implementation. As shown in [7] the voltage steered approach yields the highest power efficiency: 65%, 77% and 92% for

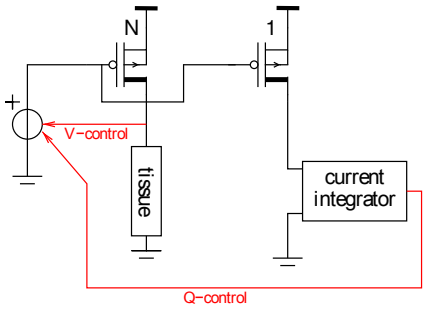


Fig. 3. Indirect current feedback with direct voltage feedback principle

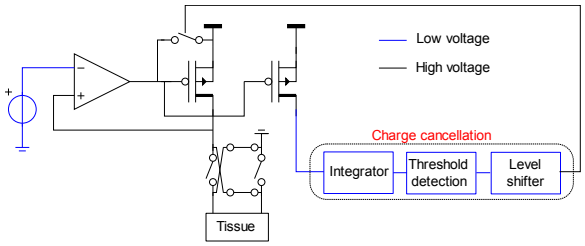


Fig. 4. Blockdiagram of the output stage employing double loop negative feedback

current, charge and voltage steered stimulation respectively. Since it is easy to make voltage sources, the area consumption can be expected to be low as well.

The drawback of the voltage steered approach is that there is no direct control over the charge injected anymore. Because of the large spread and time variance of tissue parameters the tissue current is not well defined. Therefore the injected charge is also not well defined. This means that some kind of charge control mechanism is required for a voltage steered approach.

It must be noted that the voltage source does not need to be very accurate. It only controls the waveform shape (and thereby the rate at which the charge is delivered to the tissue). This can once again save a lot of power and area in the implementation of this source.

A. Implementation of the charge control scheme

An easy way to keep track of the injected charge is to measure the tissue current and subsequently integrate it. This means there is a need for a current sensor.

In a *direct* current feedback network, the current sensor is placed in the tissue current path. This approach was chosen in [8], but suffers from two drawbacks:

- The sensor has floating terminals. When a passive sensor (e.g. a resistor) is used, this requires high common mode voltage readout electronics.
- The sensor needs to handle the relatively large stimulation current and voltage directly. This means the power consumption of the sensor will be higher as well. Due to the high voltage applied to the tissue the sensor requires high voltage components, which are area inefficient.

Therefore an *indirect* current feedback network is considered. In this approach the tissue current is sensed using a related quantity. One way to sense the tissue current indirectly is by copying an accurate fraction of the stimulation current. This copy can be easily created using a scaled transistor pair as depicted in Figure 3.

In this way the relatively large stimulation current can be directly fed into the tissue without losses. A small current and a low voltage can be used for the indirect feedback network, which reduces power consumption and area. Furthermore the feedback network can now be grounded, which yields an easier implementation.

Because of the indirect feedback topology chosen the tissue voltage is not directly controlled by the input anymore. A second feedback loop ensures the voltage across the tissue follows the input voltage.

B. Block diagram topology

Based on the topology depicted in Figure 3, a block scheme for the system can be designed. The result is depicted in Figure 4. The two feedback loops described in the previous section (indirect current feedback and direct voltage feedback), can be seen in this scheme. Furthermore the low voltage and high voltage parts are indicated using colors. As can be seen a large part of the system is using a low voltage supply yielding a low power consumption. A few blocks need some more attention to fully understand the operation of this system.

First of all it was chosen to implement a switch array to control the direction of the current into the tissue. In this way it is possible to create positive and negative currents through the tissue, while using only a single ended supply voltage.

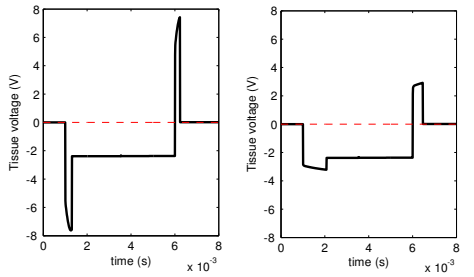
Furthermore it can be seen that the input voltage source is low voltage, while the tissue voltage can be high voltage. It was chosen to implement a gain of 10 by means of a 10 times attenuation in the voltage feedback network. In this way, the input voltage source can operate from a low voltage power supply, again resulting in lower power consumption.

IV. Simulation Results

All blocks described in the block diagram of Figure 4 were designed in the I3T80U AMIS 0.35μm high voltage (80V) technology at circuit level. Using the Spectre simulator in the Cadence environment the feasibility of this design was verified and the performance was analysed. It was chosen to have a $V_{DD,low} = 3V$ and $V_{DD,high} = 15V$. Especially $V_{DD,high}$ is more or less an arbitrary value, which can be adjusted without any modifications to the system.

A. Feasibility

To test the feasibility a sinusoidal waveform (1kHz, 2V amplitude and 7V offset) was injected into the tissue ($R_{tissue} = 10k\Omega$, $C_{tissue} = 75nF$). The charge threshold was set to 171nC. The resulting transient simulation result of the tissue voltage is depicted in Figure 5a. As can be seen, the tissue is first charged to about -2.3V during the first (negative) voltage pulse. Subsequently a positive pulse is injected to remove the charge at the tissue. As can be seen the resulting tissue voltage



(a) tonic sinusoid (b) tonic block, assymmetric

(c) burst, block (d) burst, triangle

Fig. 5. Illustration of the flexible waveform possibilities

is very close to zero, indicating the charge metering technique is working properly.

B. System performance

1) Power consumption: One of the design goals was to have a very low power consumption to increase the implantability of the device. Active power consumption is very dependent on the waveform used. Therefore it is hard to quantify the active power consumption or efficiency of the design. The quiescent power consumption is as low as 15μW. This is to the best knowledge of the authors among the lowest values for quiescent power consumption reported until now.

2) Safety: An important safety performance parameter is charge mismatch. In the waveform from Figure 5a the remaining charge was 1.5nC, corresponding to about 1%. About 50% of this mismatch is due to discharge of the tissue in the inter-pulse delay because of the finite off resistance of the switches in the switch array. This mismatch can therefore easily be removed when the inter-pulse delay is chosen to be shorter. Further another 40% of the charge inbalance is due to inaccuracies in the implementation of the integrator. When these inaccuracies are improved, the charge mismatch will become 0.1%. The remaining charge can be discharged from the tissue by short circuiting the tissue electrodes using the switch array if needed.

Another safety parameter is the ability to handle the large spread in stimulation and tissue parameters. The system is working for any combination in tissue parameters ranging from $1k\Omega < R_{tissue} < 100k\Omega$ and $10nF < C_{tissue} < 100\mu F$. Furthermore the system is also working over all process corners and process mismatches, preserving charge cancellation.

3) Versatility: Waveform adjustability: Because of the chosen architecture there are endless possibilities for waveform adjustments. In principle any waveform can be used: the charge metering mechanism will keep track of the charge injected in the tissue. It is therefore possible to use tonic stimulation, burst stimulation, asymmetric stimulation, subthreshold prepulses, excitatory and inhibitory stimulation, etc.

To illustrate this two waveforms for both tonic and burst stimulation are depicted in Figure 5. This figure illustrates the charge cancellation mechanism is working for a wide variety of shapes, since the final voltage is very close to zero.

V. Conclusion

A fundamentally new architecture for an output stage of a neural stimulator has been designed. The design goals were focussed on very low power consumption, while still having endless possibilities for waveform adjustment. The system was shown to be very robust, as it is working for a wide range of stimulation, tissue and process parameters.

Using this stimulator more effective neural stimulation patterns can be applied. It is expected that many kinds of diseases can be treated much more effectively in this way. The application is not only limited to neural stimulation, but it can be used in any kind of nerve stimulator such as peripheral nerve stimulators.

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