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Global digital controller for multi-channel micro-stimulator with 5-wire interface featuring on-the-fly power-supply modulation and tissue impedance monitoring

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A B S T R A C T

This paper reports a global digital controller for multi-channel high density micro-stimulator applications. By centralizing the essential functions including stimulation timing controls, tissue-electrode impedance (TEI) monitoring and power supply modulation in the global digital controller (GDC), digital blocks in the local electrode driver channels (LEDCs) can be greatly simplified, resulting in compact electrode drivers suitable for high density intracellular stimulations. Based on the proposed column-parallel row-scanning (CPRS) stimulator topology, the 5-wire global control ensures customized per channel stimulation with minimal interfacing overhead. A 4-column GDC stimulator prototype is fabricated in a 0.18 µm CMOS process. It features flexible stimulation with a power consumption of 18 µW/column while enabling global power-supply modulation and real-time TEI monitoring for improved stimulation efficiency and safety.

1. Introduction

Multichannel microstimulator is the main apparatus required for establishing a link between an artificial sensor and a high-density nerve array such as the retina and the cortex [1]. A stimulator triggers neurons by artificially modulating excitatory postsynaptic potential through sourcing and sinking charge packets into and from the axon hillock region of neurons. Recently, electrical stimulation of high-density electrode array emerges as an important engineering interest, engendered by technological advances appeared in nanofabrication of electrode array and deep-submicron high-density current driver circuits [2]. A micro-stimulator drives an external load impedance that can be approximated by a resistor (R_a) and a capacitor (C_{dl}), which are determined by the materials and geometries. The access resistance (R_a) arises from the hydrolyte at the tissue-electrode interface, while the double layer capacitance (C_{dl}) represents the charge-transfer contributed by ionic redox reaction in the Helmholtz layer adjacent to the metallic surface of the electrode [3]. The Faradaic resistance which lies in parallel with C_{dl} is large enough to keep the irreversible chemical reactions (which are physiologically harmful) inside the safety window.

With the ever-expanding applications of electrical stimulators, ranging from unraveling the blueprint of the neural network to the cure of various deceases originating from defects in the nervous system, various micro-stimulator systems have been proposed [4,5]. The recent development of high-density stimulation applications such as retinal prosthesis [6] and intracellular micro-electrode array (MEA) monitoring (~10,000 electrodes [2]) brings about increasing demands for multi-channel micro-stimulators (MCMS) systems. As the number of driving channels increases, the managing architecture of a MCMS has to be more delicately designed to efficiently supervise the timing, duration, polarity, and spatial patterns of electrical stimulations under power and area constraints. Various efforts have been made to achieve different aspects of GDC system optimization, including global-local stimulation data packet transmission protocol [7], programmable scanning pattern control [8], flexible waveform generation [9], compliance-voltage-aware power-supply control [10,11] and TEI monitoring [12].

Sivaprakasam et al. [7] sought for an efficient data transmission protocol by optimizing the packet error rate and the required bandwidth. However, their 60-channel stimulator is highly customized and the dedicated serial data bus and parallel 8-to-1 multiplexed-waveform-generator architecture can be a major bottleneck in high-density stimulator designs. In [9], a GDC which can generate programmable current waveforms with a time resolution of 4 µs was designed for a 1024-channel stimulator. Despite its capability of generating effective step-down stimulation waveforms [13] which can reduce the required

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compliance voltage by 10% − 15%, every LEDC requires a dedicated digital controller with a power and area overhead of ~10 μA at 1 MHz and 21%, respectively, which inevitably burdens the LEDC design and limits the stimulation density.

In high-density stimulation, it is important to minimize the resistive heat loss across the current drivers to regulate the tissue temperature increment to be under 4.5 °C [14]. In order to prevent overheating as a result of the voltage drop across the current drivers, many existing MCMSes [10,11,15] employed DC-DC converters to adaptively regulate the power supply and dynamically track the compliance voltage of the electrodes. Nevertheless, the scalability issue of the previously reported GDC remains unsolved as a result of the requirement for dedicated regulation circuits per stimulation channel [10] and a split supply configuration [13] which ultimately limits the stimulation resolution. To improve this situation, the GDC in [11] employs a time-multiplexed charge packet delivery scheme with multiple electrode drivers sharing a buck-boost DC-DC converter. However, the number of concurrently driven channels is still limited by the number of the time slots that can be allocated within a stimulation time frame. Apart from that, the requirement of a large number of interfacing wires (16 in [6], 26 in [9], and 31 in [16]) as a result of the intensive usage of multiplexing switches and sophisticated bus protocol also significantly increases the LEDC complexity. To further ensure the safety of MCMS, the GDC should ensure the tissue-electrode interface quality while preventing residual charge accumulation. This can be indirectly inferred by measuring the tissue-electrode impedance [12] by incorporating an impedance monitoring circuit in the stimulator system [17].

This paper presents a GDC architecture that is suitable for MCMS with a high-density MEA. The GDC can control a large variety of MEAs with varying geometries and channels by using a very lightweight LEDC controller which is comprised of only a set of shift registers. Particularly, the control signals are synchronized by the GDC in a row-wise manner through the proposed column-parallel row-scanning (CPRS) stimulator topology, relieving the LEDC from dedicated clock counters for complex timing control. In addition, on-the-fly dynamic power supply modulation and in-situ TEI monitoring for improved stimulation efficiency and safety can also be achieved. Each LEDC can be custom controlled by using only a 5-wire interface. The fabricated 4-column GDC prototype in 0.18 μm CMOS successfully demonstrates stimulation data distribution, flexible stimulation timing control, switching-mode power supply (SMPS) control and TEI safety monitoring, while consuming only 18 μW per column.

The rest of the paper is organized as follows. Section 2 discusses about the architectural considerations that underlies the proposed GDC. Section 3 explains the operating principles of the GDC. Experimental results are presented in Section 4. Section 5 concludes the paper.

2. Architectural considerations of the proposed GDC

In this section, we investigate the architectural considerations for MCMS driving a variable-size MEA, including the communication protocol, stimulation timing control method, power supply configuration, and impedance monitoring circuit. The GDC should distribute the data packets containing the stimulation waveform profiles (e.g. current amplitude) for every LEDC. As the attached MEA size increases, the required data flit size also increases. The **Stimulation Data Packet Transmission (Stim Data Pkt Tx)** block distributes the stimulation profiles (current amplitude) to each LEDCs. Fig. 1(a-c) illustrates three different data bus protocols: 1) parallel; 2) cluster; and 3) serial. The serial bus architecture is shown in Fig. 1(c). As all the LEDCs retrieve their data packets funneled through the serial bus, the packet dropout rate can be increased [7]. A solution is to utilize the fully parallel data bus architecture as shown in Fig. 1(a). The constant flit size packets are distributed to LEDCs simultaneously, leading to a reduced packet dropout rate but at the expense of the requirement for a local LEDC decoder. To achieve both a small LEDC size and a good data error tolerance, the cluster bus protocol where each cluster is accessed separately can be utilized, as shown in Fig. 1(b). In the CPRS topology, the LEDCs contained in a row are assigned with an identical address, thus the required bandwidth for address data transmission can be reduced by $1/N_{col}$, where N_{col} is the number of LEDCs in a row. Yet, the data error rate can be kept much lower than that of the serial data protocol as each cluster data packet can be independently distributed. In this work, we exploit the row-wise clustering MEA architecture and propose the CPRS topology (which is based on the cluster bus protocol) to achieve an efficient GDC implementation for high density MCMS.

To control the stimulation pattern on a high-density electrode array, multiple instances of timers should be operated coherently. The timing control protocol employed by the **Channel Controller** determines where the timers are located (e.g. centralized or distributed), and how the timers are shared and synchronized among LEDCs. Fig. 1(d–f) depicts the timer instances in three different timing control protocols: 1) local autonomous; 2) parallel; and 3) cluster. The GDC controls both the polarity and timing control of stimulation waveform of multiple LEDCs, which can be accomplished using the local distributed autonomous control method (Fig. 1(d)), the centralized parallel method (Fig. 1(e)), and the centralized cluster method (Fig. 1(f)). The stimulation timing sequence of the LEDCs in the MCMS, which is applicable to all the control methods to determine the scanning pattern. With the local distributed autonomous timer control [6,11], each LEDC contains a set of timing controllers and related building blocks to generate the local polarity and charge cancellation signals. Even though highly customized LEDC functions can be realized, the requirement for the control overhead in individual stimulation channels can limit the LEDC size. By centralizing the stimulation waveform generation in the GDC as shown in Fig. 1(e), the LEDC area overhead and the data packet size can be much reduced. This scheme is demonstrated in [9], where the GDC generates the initiation and termination signals for each LEDCs. Nevertheless, the multiple instances of concurrently running timing control circuits can increase the LEDC synchronization complexity and possibly lead to deadlock or livelock states. Also, independent sets of routing wires are required for every LEDC, increasing the probability of crosstalk and hence the packet dropout rate. In this work, the centralized cluster method which takes advantage of the row-wise clustering in the propose CPRS topology is utilized, as shown in Fig. 1(f). The waveform control timer in the GDC is shared among different stimulation rows through time multiplexing, which can balance the tradeoffs between the implementation complexity, LEDC size and stimulation frequency. Also, it can significantly reduce the number of global control wires while minimizing signal interference.

In MCMS, the power management circuit should improve the stimulation efficiency and reduce the heat energy loss so as to increase the number of simultaneously driven electrodes without violating the power and safety constraints (e.g. [10,11,15]). According to how energy is delivered between the power source and electrodes, there are three classes of power supply modulation methodologies for reducing the power loss across the current controlling transistors, namely: 1) fixed voltage (FV); 2) dynamic voltage scaling (DVS); and 3) direct voltage forming (DVF). In the FV supply based stimulator topology, current drivers pull energy from the fixed voltage regulator. The conduction loss across the current drivers can induce excessive heat, especially when the supply voltage is much higher than the electrode voltage. To reduce the power loss of the FV supply based stimulators, DVS supply based stimulators adaptively vary the supply voltage levels according to the compliance voltage required for successful biphasic stimulations (e.g. current amplitude, pulse duration, and electrode types). For DVF supply based stimulators [10,11], the power loss across the current controlling transistors can be completely remove by applying voltage waveforms directly derived from the instantaneous electrode voltage for driving a specific electrode. Yet, existing DVF supply based stimulators cannot be applied in high density MCMS designs due to the excessive overhead induced by dedicated