

# Design of a Highly Efficient Circuit for Electrical Muscle Stimulation

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**Abstract**— A highly efficient multi-channel Electrical Muscle Stimulation (EMS) circuit is proposed. EMS involves the use of pulsed electrical current to stimulate muscle ‘motor points’, causing the corresponding muscle to contract. The EMS device generates an amplitude controlled pulsed electrical signal and passes it to the targeted ‘motor point’ via an electrode. For implanted devices, and with the increasing trend towards ‘body worn’ battery powered patient EMS devices, there is a need for reduced size and greater power efficiency.

## I. INTRODUCTION

For both implanted and external ‘body worn’ EMS devices, there is a need for reduced size and greater power efficiency. For implanted devices, although stimulation signal levels are very low, extended battery life is an obvious benefit. Unlike implanted devices however, external EMS devices need to produce much larger voltage and current levels. The need for increased voltage levels is due to large and variable skin-electrode impedances.

For external devices, EMS voltages and currents using surface stimulation electrodes placed over a motor point are of the order of 60-100 V with currents in the region of 40-60 mA, for pulse durations of the order 0.1-0.2 ms [1]. This paper gives a brief overview of typical EMS signals, the electrical circuit equivalent for the body load, and some typical circuits used to generate the stimulation signal (stimulus). An alternative, highly efficient circuit is then proposed.

## II. ELECTRODE-ELECTROLYTE MODEL

The body load impedance, including tissue, skin and electrodes, can be approximated by an equivalent network of a resistor  $R1$ , representing the tissue, serially connected to a parallel circuit of a capacitor  $C$  and a resistor  $R2$ , both representing the electrode-skin junction [2].

Absolute values for  $R1$ ,  $R2$  and  $C$  vary due to a number of reasons given by [3]:

- Current density
- Stimulus waveform shape

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Other factors affecting the model parameters are: Skin Type, Electrode composition, Electrode size and Body region.

Due to the highly variable nature of the body load, the approach adopted in the proposed circuit allows for a wide range of equivalent circuit parameters. However, the circuit as proposed can be optimised to meet the individual requirements of a specific application.

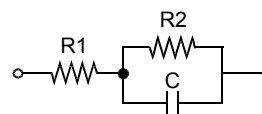


Fig. 1 Electrode-Electrolyte model

## III. ELECTRICAL MUSCLE STIMULATION

As shown in [3], for rectangular pulse waveforms the stimulus may be classified into two common types, constant voltage or constant current, as briefly discussed below.

### A. Constant Voltage

The voltage across the load is kept constant. In this situation, changes of the inter-electrode impedance may modify unpredictably the current and hence the effects of the stimulation. Therefore the stimulation current may be a non-linear function of stimulation voltage.

### B. Constant Current

The current through the load is kept constant. Since membrane depolarisation is related to current density, this output stage usually allows better response reliability and repeatability.

## IV. STIMULATION PATTERNS

The upper waveform of Fig. 2 details a typical biphasic EMS pulse train. The amplitude of the pulses is ramped up from zero to the desired intensity in rise time ( $T_r$ ), after a set period at this amplitude the intensity is reduced towards zero in fall time ( $T_f$ ). This pulse train consists of adjacent biphasic pulse doublets (i.e. pulse doublets A and B in Fig. 2), which consist of both a positive and negative phase.

The key EMS stimulation parameters are: Rise Time ( $T_r$ ), Fall Time ( $T_f$ ), Pulse Width ( $T_{pulse}$ ), and Frequency ( $F_{stim}$ ), with typical values given in Table I.

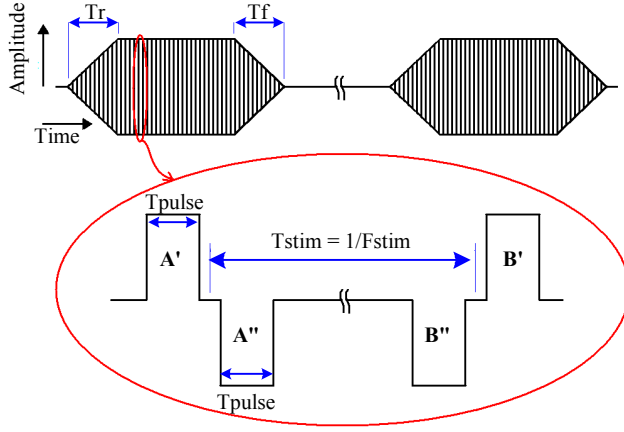


Fig. 2 Typical biphasic EMS stimulation envelope

In some applications it may be desirable to change the polarity of the leading pulse of each pulse doublet as shown in Fig. 2 (i.e. leading pulse of doublet A is positive current pulse A', while leading pulse of doublet B is negative current pulse B''). This is to eliminate any charge imbalance, which might be present between adjacent pulse doublets. Charge imbalance (build up of a net DC charge, which may cause skin irritation) may be caused by differences in amplitude and/or pulse widths of the positive and the negative pulse within the same doublet.

TABLE I. EMS STIMULATION PARAMETERS

No.	Simulation Parameter	Min	Max	Units
1	Frequency (Fstim)	4	140	Hz
2	Pulse Width (Tpulse)	50	2000	μs
3	Amplitude	5	100	mA
4	Rise Time(Tr), Fall Time(Tf)	100	∞	μs

## V. EXISTING EMS OUTPUT STAGES

In order to generate the required currents through the body load described in section II, it is necessary to generate voltages in the range 0-100V across the load.

The stimulator output circuit used in [4] (Fig. 3), makes use of two fly-back power converters (+150V and -150V)

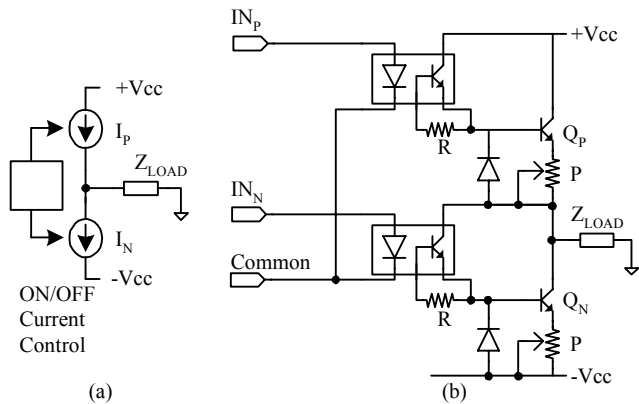


Fig. 3 EMS Dual Supply Output Stage proposed by [4].

and a push-pull constant current stage. Alternatively, [5] describes a circuit using a single +100V power supply and a constant current circuit (shown in Fig. 4). In this arrangement biphasic stimulation is catered for with the use of switching transistors in a 'H-bridge' configuration, enabling the voltage polarity across the load to be reversed.

Both [4, 5] use fixed voltage rail(s) and control the output current by dissipating power via constant current circuit stages. Although both solutions provide biphasic pulses, the method of controlling current in both cases is very power inefficient, as described in the following section.

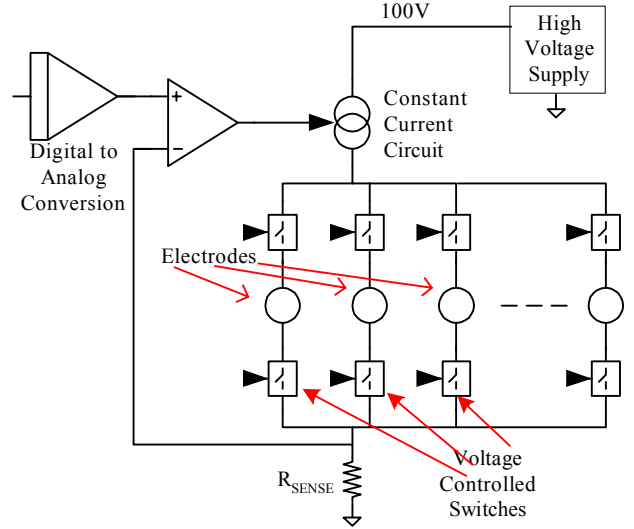


Fig. 4 EMS Single Supply Output Stage proposed [5].

## VI. POWER LOSSES IN OUTPUT STAGE

Fig. 5 shows the constant current pulse and the corresponding voltage waveform, for a monophasic output.

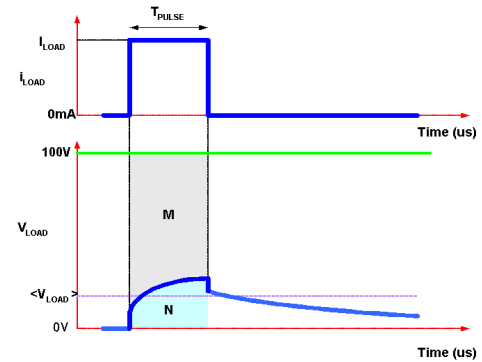


Fig. 5 Body load current and voltage waveform

The average voltage  $\langle V_{LOAD} \rangle$  over one single current pulse can be shown to be (1):

$$\langle V_{LOAD} \rangle = \frac{I_{LOAD}}{T_{PULSE}} \left\{ R_1 T_{PULSE} + R_2 \left[ T_{PULSE} + R_2 C e^{\frac{-t}{R_2 C}} \right] \right\} \quad (1)$$

Ignoring other power losses in the circuit, we may define the output efficiency  $\eta_{EMS\_STIM}$  as:

$$\eta_{EMS\_STIM} = \frac{P_{LOAD}}{P_{LOAD} + P_{CONST\_CURRENT}} = \frac{P_{LOAD}}{P_{TOTAL}} \quad (2)$$

Where  $P_{LOAD}$  (shaded area 'N', Fig. 5) is the power delivered to the load and  $P_{CONST\_CURRENT}$  (shaded area 'M') is the power lost in the constant current circuits of [4, 5].  $P_{CONST\_CURRENT}$  includes all losses in the EMS output stage but NOT in the switching power converter.

Table II, shows the average voltage and instantaneous power over a single pulse period. When delivering a range of constant current pulses (with a duration of  $T_{PULSE}=250\mu s$ ) from a fixed 100V supply into the following body load model:  $R1=100\Omega$ ,  $R2=1000\Omega$  and  $C=1\mu F$ .

TABLE II. EMS OUTPUT STAGE EFFICIENCY

	$I_{LOAD}$ mA	$\langle V_{LOAD} \rangle$ V	$P_{LOAD}$ W	$P_{TOTAL}$ W	$\eta_{EMS\_STIM}$ %
(a)	20	4.304	0.086	2	4.3
(b)	50	10.760	0.538	5	10.7
(c)	100	21.520	2.150	10	21.5

A significant improvement in efficiency could be achieved by generating only the voltage required for the load. To-date however, designs attempting to achieve this have either relied on an estimate (or measurement) of the load impedance or an iterative adjustment of the output such that the power loss is minimized [6]. Although power savings can clearly be made using these techniques, both fail to account for variations in output voltage requirements during the pulse delivery. And if the output voltage generated is too low or too high this is not detected until after the pulse has been delivered.

The design presented in the following sections enables the generation of only the required voltage during the entire stimulation pulse period. This is achieved by designing an adaptive power generation circuit that is capable of producing an output voltage, which dynamically adjusts to a typical body load and responds to variations in this impedance.

## VII. PROPOSED EMS OUTPUT STAGE

The proposed EMS output stage seeks to eliminate the previously described output stage power losses by generating only the required voltage, to deliver the desired load current. The major challenge in the proposed design is to design a power generation stage with fast transient response. This will dynamically provide the necessary voltage to maintain the load current over the complete load impedance/load current range.

The system required can be considered as a classic control system as shown in Fig. 6, where  $G_{PLANT}$  [7] represents the power converter and  $G_{CL}$  represents the compensation loop circuit including the load network.

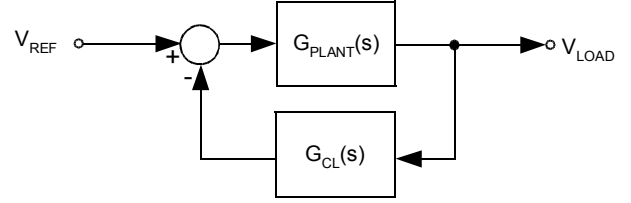


Fig. 6 Proposed closed loop feedback control

Closed loop feedback control is commonly used to sense an output voltage ( $V_{LOAD}$ ) and adjust for any differences between it and the reference signal ( $V_{REF}$ ). And although existing EMS devices may contain such a control circuit to control the high voltage output, the conventional approach is to treat this high voltage stage as separate from the constant current circuit. The proposed compensation circuit is shown in Fig. 7, where the body load ( $Z_{LOAD}$ ) is included as an integral part of the compensation arrangement. This alternative compensator aims to keep the voltage across  $R_{SENSE}$  constant, therefore keeping the load current constant.

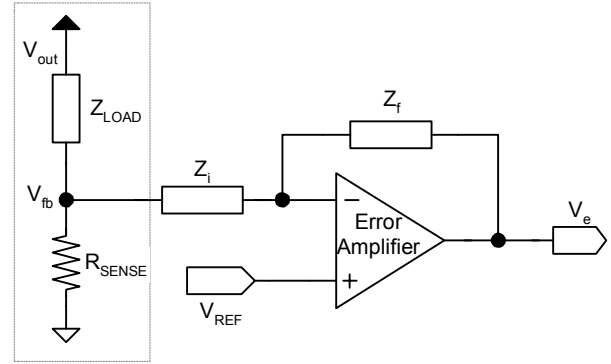


Fig. 7 Compensation circuit including load ( $G_{CL}$ )

The transfer function of this compensation circuit can be shown to be (3):

$$G_{CL}(s) = \frac{V_e}{V_{OUT}} = \frac{Z_f}{Z_i} \cdot K_H \cdot \left( \frac{\frac{s}{\omega_{zH}} + 1}{\frac{s}{\omega_{pH}} + 1} \right) \quad (3)$$

Where:  $\omega_{zH} = \frac{1}{R_2 C}$ ,  $\omega_{pH} \approx \frac{R_1 + R_2}{R_1 R_2 C}$

and  $K_H = \frac{R_{SENSE}}{(R_{SENSE} + K_{LOAD})} = \frac{R_{SENSE}}{(R_{SENSE} + R_1 + R_2)}$

The block diagram of the proposed system is shown in Fig. 8. With pulse widths ranging from  $50\mu s$  upward, transient time of the order of  $10\mu s$  or less are required. This dictates that a power converter with high switching frequency is required. For example, for a critically damped system (fastest response without any overshoot) at time  $t=1/f_c$ , the output voltage/current has reached 98.6% of its final value where  $f_c$  is the loop crossover frequency [7]. Also, in general  $f_{c\_MAX} = 10\% f_{SWITCH}$ . Therefore maximizing  $f_{SWITCH}$  is necessary for fast transient response time.

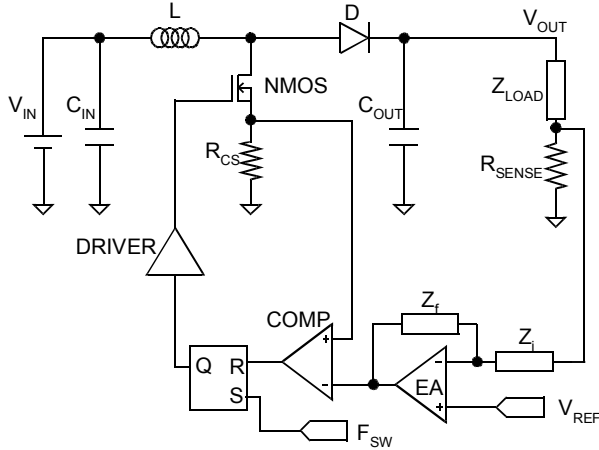


Fig. 8 Proposed EMS Circuit

The design of the compensation network ( $G_{comp}=Z_f/Z_i$ ) involves modifying the loop gain transfer function ( $T_{open}$ ), to ensure adequate phase margin. The challenge in the design of the control loop is to ensure stability but also a fast transient response, under all load conditions. Transfer functions of the proposed design are shown in Fig. 9.

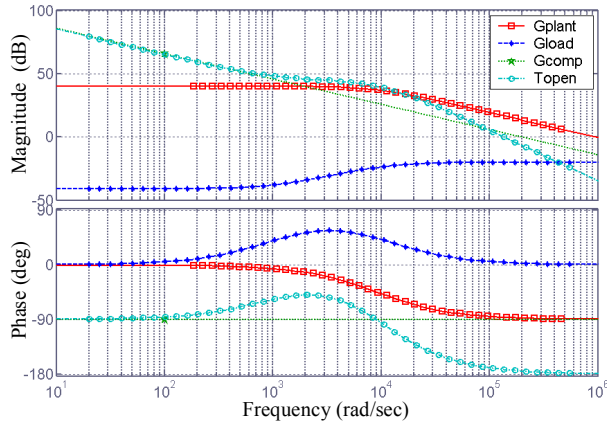


Fig. 9 Transfer functions

Table III details the efficiency in the output stage including losses for a series  $10\Omega$  current sense resistor and two  $5\Omega$  output switching transistors, when delivering a range of constant currents  $I_{LOAD}$  (pulse width  $T_{PULSE}=250\mu s$ ) into a body load model of:  $R_1=100\Omega$ ,  $R_2=1000\Omega$  and  $C=1\mu F$ .

TABLE III. PROPOSED EMS OUTPUT STAGE EFFICIENCY

	$I_{LOAD}$ mA	$\langle V_{LOAD} \rangle$ V	$P_{LOAD}$ W	$P_{TOTAL}$ W	$\eta_{EMS\_STIM}$ %
(a)	20	4.304	0.086	0.094	91.5
(b)	50	10.760	0.538	0.588	91.5
(c)	100	21.52	2.152	2.352	91.5

## VIII. SIMULATION RESULTS

Figure 10 shows the transient response of the proposed system for a pulsed load current ( $I_{LOAD}$ ) of 100mA.

Here, we have taken the small signal model for the power converter, linearized this model about the average voltage and design the compensation network to ensure both fast response and stability. However, the small signal model actually changes with load current and load voltage. With the constant current implementation, load voltage changes during the constant current pulse as seen in section VI; therefore a more accurate model, which dynamically changes as a function of load voltage, should be used.

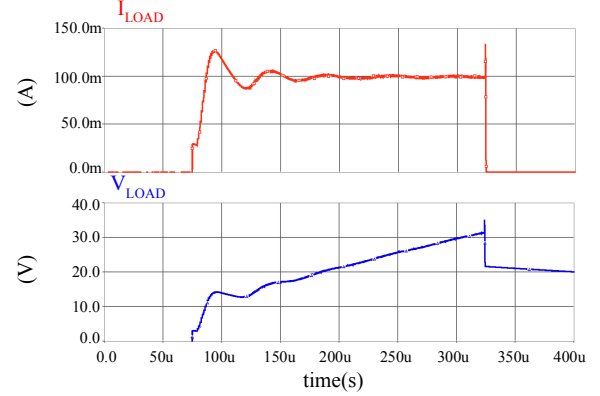


Fig. 10 Proposed EMS stage transient response

## IX. CONCLUSIONS

As shown from the results, a stable power converter with the required transient response to deliver even short  $\sim 50\mu s$  EMS pulses is achievable. This converter generates only the required voltage across the load and series components hence minimal excess power is wasted during EMS stimulus delivery, unlike solutions [4, 5]. The theoretical efficiencies of the proposed EMS output stage are shown in Table III and are significantly higher than those of existing solutions [4, 5]. Other methods of reducing the power lost by dynamically reducing the 'fixed' voltage rail have been described [6]. However, as well as added complexity, this has a relatively slow response time and still has considerable power losses associated with it. The proposed system represents a significant improvement in terms of energy efficiency compared to [4, 5].

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