ADVANCED VERSION OF THE LJUBLJANA FUNCTIONAL ELECTRONIC PERONEAL BRACE WITH WALKING RATE-CONTROLLED TETANIZATION

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Summary

The original Ljubljana Functional Electronic Peroneal Brace was designed so that parameters of electrical pulses except the amplitude of the stimulus are fixed before the application of the device. The constancy of duration of the tetanizing train did not satisfy all functional requirements when the patient wanted to change his walking speed. Therefore, an improvement of the Ljubljana FEPB was introduced to provide continuous automatic adjustment of duration of the tetanizing train in accordance with the patient's walking speed. The tetanizing time lasting 85 to 90 per cent of the swing phase has proved most suitable.

In designing the advanced version of the Ljubljana FEPB the following requirements were considered: the basic part of the device is a circuit measuring the time of the swing phase and determining the tetanizing train duration. The electronic solution of these requirements including low energy concumption, small number of elements and miniaturization is described and the related problems are discussed. This type of the brace was tested on five patients and the results are under evaluation.

Three years experience with the application of the Ljubljana Functional Electronic Peroneal Brace to the patients with hemiparesis or hemiplegia due to cerebral vascular accident or head injury being a successful substitute for the classical mechanical splints, with the aim of improving their gait shows us: that this device beside being a successful substitute for the classical mechanical splints, which prevent drop foot, also produces functionl movement (i. e. dorsal flexion and eversion) of foot in some stages of walk and promotes the recovery of function of paretic peroneal muscles [1, 2, 3]. The device has been designed in such a way, that parameters of electrical pulses applied to the peroneal nerve trunk should be fixed before the application. Amplitude of the stimulus is the one parameter which can be modified in course of use by the patient himself. These properties make the Ljubljana Functional Electronic Peroneal Brace an appropriate device to help most patients in the

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stage of rehabilitation when any considerable influence on the walking speed lies beyond their possibilities. Yet, in advanced stages of rehabilitation when the patient can easily change the walking speed by himself, the constancy of most parameters of the stimulus, especially duration of tetanizing train, does not satisfy all functional requirements. The duration of tetanizing train fixed beforehand turns out to be too long in case the patient's gait becomes faster, respectively too short in case his gait becomes slower (Fig. 1).

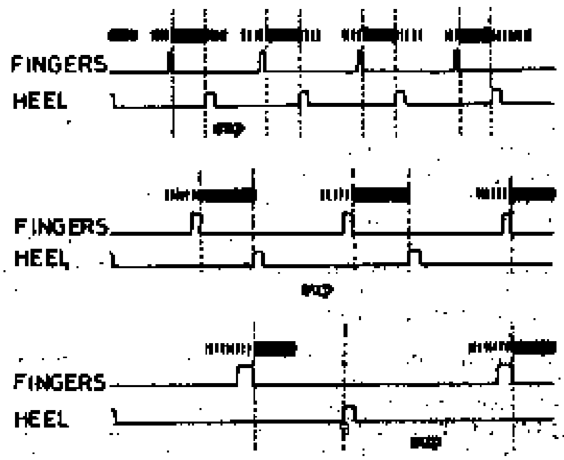

Fig. 1.

Thus, the functional electrical stimulation of the paretic peroneal muscles does not agree with the mechanics of walking which can result in an unsteady gait and even other undesirable consequences. Also, the transmission of information along the afferent pathways to the spinal cord which improves the motor function of the paretic muscles is not synchronized with single stages of walk any longer. Such a stimulation is temporally no more adequately programmed. This can seriously affect the organization of the spinal reflex motor activity involved in walking, the organization which has just been improved to a certain extent.

Considering these facts, we decided to improve the Ljubljana FEPB to such a degree that the tetanizing train duration would be determined by the patient's walking speed and, consequently, could be accomodated at every step. The mean tetanizing time would be slightly shorter compared to the swing stage which suits well all requirements of FES of the peroneal muscles.

The following requirements for the electronic circuit are included in the design of the brace:

The fundamental element of the brace is the circuit measuring the time of the swing phase. According to this measurement the electronic circuit shall properly adjust the time of the stimulation

From the electronic point of view, requirements are simple, but the design itself is not as simple as that if we add basic requirements for all orthotic and prosthetic circuits, i. e. small energy consumption, a small number of elements, and miniaturization.

When raising the foot, a shoe-heel switch is triggered and measurement of the swing phase time begins. Each time the heel touches the ground the swing phase is completed, as well as the time measurement. The time interval which, due to the gait changes from 0.3 to 2 sec., is converted by the time voltage converter into a time proportional voltage, which is stored in the memory circuit. The time measuring and conversion circuits are shown by the block diagram (Fig. 2).

BLOCK DIAGRAM

Fig. 2.

By the triggering of the switch in the shoe-heel we get an increasing linear voltage — the time base — opening the modulator over the Schmitt trigger and the stimulation begins. The voltage of the time base corresponds to the swing phase time of each step. In a simplified circuit the time base could be replaced by a simple charging of the RC constant. Given voltage is then stored in a suitable RC constant representing the memory. The same time base is applied for comparison with the memory voltage which corresponds to the length of the previous step. As the voltage of the time base and memory is equal it gives rise to the steep transition of voltage on the differential amplifier serving as the comparator. The pulse generated in such a way activates the Schmitt trigger which shuts the modulator and interrupts the stimulation. In this way the length of the previous step determines the stimulation time during the present step. Between the modulator and the Schmitt trigger, the RC circuit is built in to decide upon the rising and falling of the stimulation train. If the present step is shorter than the previous the switch in the shoe-heel blocks the Schmitt trigger interrupting the stimulation. In this case the situation does not correspond to the requirements according to which the time of the stimulation (t_s) should be shorter than that of the swing phase (t_r).

In all other cases that may occur taking into account the dynamics of the gait, requirements concerning $t_a < t_r$ are more or less fulfilled. We are to point out that at the very first unchanged step all prescribed parameters for the stimulation are normal and functionally correct. As for the circuit of stimulation, we have chosen two versions: the universal one, using the converter and the output stage with the supply voltage Ucc=100 V, and the circuit we have taken from the standard peroneal brace [4, 5]. This circuit makes use of the modified blocking oscillator serving as an output stage. Final energy consumption depends upon the design of the output stage. The smallest consumption is obtained when the transmission over the output stage feeds the current only during the width of the stimulation impulse. For this reason it is more convenient to use the design where the output stage is energized by the suitable voltage, with regard to the desired amplitude of the stimulation.

The memory must have the contact circuit or transmitting information from the measuring to the memory part. There are three possibilities: the application of photo-coupler, a relay or a diode-switching circuit. The existing photo-couplers have essential weak points preventing the application in our case in spite of the ideal principle. One of the main weak points is the great energy consumption, i. e. the great supply voltage for the light source. Our design will make use of the relay because of its simplicity, though this design is not finally optimised concerning the energy consumption. The final brace will make use of the diode circuit which is economical but more complicated.

In general our circuit gives wide possibilities, and enables simple changing and adjusting of all the important parameters important for the experiment. The relay is adjusted so as to provide interruption at 8 V, which means that the battery must be recharged in order to prevent damage.

The electronic circuit of the brace is shown in Figure 3.

Transistors and single parts of the circuit carry out the following functions: the circuit with transistors T_1 and T_2 is the Miller sawtooth generator, the transistors T_3 and T_4, or T_8 and T_9 represent the phase inverter and impedance transformation. The transistor T_5 and the condenser C_8 form the memory and the impedance transformation. The transistors T_6 and T_7 from the differential amplifier comparing the memory voltage with that of the time base. The Schmitt trigger consists of the transistors T_{10} and T_{11}. A stable multivibrator determining the frequency of stimulation consists of the transistors T_{12} and T_{13}, and the transistor T_{14}, which together with the time constant C_{10} $(P_{11}+R_{28})$ determines the width of the output stimulation impulse. The transistor T_{15} serves for the modulator. The increasing and decreasing of the stimulation train is set by the value of the condenser C_7. The transistor T_{16} is the impedance transformer, the T_{18} transistor is the amplifying and phase inverting stage, while the T_{19} is the output stage. The transistor T_{17} converts the battery voltage from 9 to 100 V. The amplitude of electrical stimulation is adjusted by the potentiometer P_{12}.

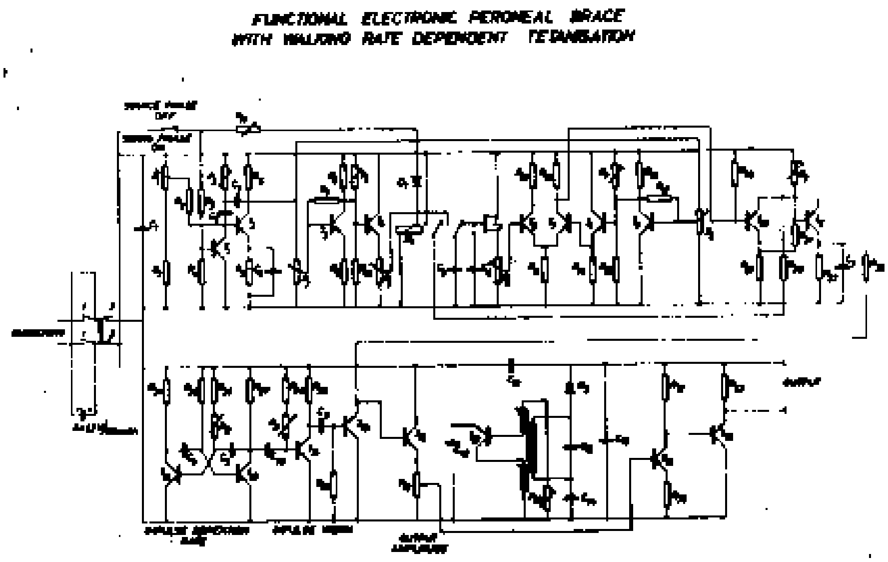

Fig. 3.

The more important characteristics are:

— output voltage can be set between 0 and 80 V;
— stimulation frequency range from 15 to 150 Hz;
— width of the rectangular stimulation impulse can be set from 15—1500 µS, and the tetanizing train duration with regard to the swing phase duration is 0.3 to 2 sec;
— adjustment of relation t_s/t_r is possible within the limits of 0.5 and 0.9.

Dependency of the stimulation time upon the duration of the swing phase is shown in Figure 4, $k=0.7$ ($t_s=k \cdot t_r$).

The time of the increasing stimulation train is 5 per cent of t_s, while the decreasing time is 15 per cent of t_s. The time of increasing or decreasing can be chosen by C_7, and for greater changes by adding a resistor in the collector lead between R_{22} and C_7.

The steady state on current of the circuit with the stimulation off is 12 mA. At the normal gait ($t_r=1$ sec) the energy consumption at the maximum output voltage on 2 kohms load is 23 mA. (the stimulation frequency was 50 Hz and the impulse width 0.3 msec). We presume that the final consumption using the diode switching circuit would be smaller, and therefore more preferable for the final design.

This type of brace was tested on 5 patients and the results are in course of evaluation. So, any definitive opinion about it would be premature. However, stimulus duration t_s of 10 to 15 percent shorter than the swing duration t_r turned out to be the most adequate. The necessary mean tetanizing train duration will be fixed more precisely by means of kinesiologic studies. Such a stimulation could possibly raise the question of new electronic solutions of the problem.

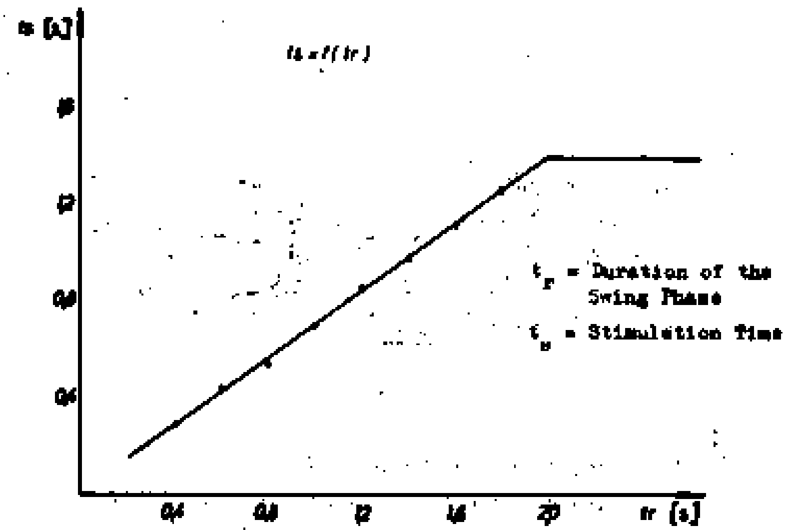

Fig. 4.

It would be interesting to find out whether k does or does not depend upon the duration of the swing phase, and what effect it has on swing-phase duration.

Miniaturization is one of the main problems while designing a prototype. The circuit in Fig. 3. will be somewhat less complicated without some emitter followers. The energy consumption will remain within the limits of 25 mA, which means that the capacity of batteries during the 10 hour walk will be 250 mA hours. This is more than is needed becuse only a few patients will make use of the maximum amplitude of the stimulus. While designing the standard peroneal brace it was found that economically and technically feasible miniaturization would be within the limits of the battery volume. The volume of the complete peroneal brace with the automatically adjusted stimulation time would be appr. 150 cm^3, using NiCd standard batteries.

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