# POWERED HYBRID ASSISTIVE SYSTEM

**Popović[*]D.B., Schwirtlich[**]L., Radosavljević[**] S.**

[*] Faculty of Electrical Engineering, University of Belgrade

[**] Institute for rehabilitation "Dr Miroslav Zotović"
Belgrade, YUGOSLAVIA

## ABSTRACT

Use of patterned electrical stimulation results with the generation of functional motions. However, a number of handicapped persons does not benefit from the present techniques mostly due to insufficient control and muscle fatigue. These problems are extremely exposed in high-thoracic lesions and in low-cervical lesions, as well as, in mixed lesions resulting with denervation of some muscle groups or muscle atrophy. The external power, however, requires the use of an external skeleton which has a numerous disadvantages. This paper describes initial experimental work with a sequential hybrid assistive system. This system uses powered hip joint and controllable brake in knee joint in addition to four channel FES system. The general conclusion is that the use of powered hip joint improves the quality of the gait by all means. The powered hip joint adds the functional movement which resemble much more to the swing phase of the gait in comparison with the flexion reflex. The work presented here is a single patient study and it can not be generalized,but it suggests to some advantages of HAS in comparison with FES systems.

KEY WORDS: Sequential hybrid orthosis, powered external brace, FES, gait restoration

## INTRODUCTION

Considerable effort is now being directed to the FES of spinal cord injury patients in order to re-establish a safe walking pattern. Most of the research to- date has addressed the problems of electrode design and placement, stimulation patterns and microprocessor design. What needs to be recognized is that FES microprocessor system is replacing the central nervous system (CNS) and therefore must mimic the same strategies that the intact CNS uses. One of basic criterion that the CNS considers is the reliability and safety. It is impossible at this stage to mimic natural movements because the spinal injury results with numerous changes in neuromuscular functions.

One possible approach at this point is to integrate two assistive systems, the FES and external mechanical orthosis. Such an approach is known as hybrid assistive system (HAS). Hybrid assistive systems were suggested (Tomović et al, 1972) and several possible realizations of HAS were demonstrated. Most of suggested realization are combining relatively simple rigid mechanical structures for passive stabilization of lower limbs during stance phase (Andrews et al, 1989, Solomonov et al, 1989; Phillips, 1989). We have investigated the possibility to combine active brace, which will contribute to patterned stimulation in realization of some functional movements in addition to its supporting function (Popović et al, 1989).

Each trend in the design of HAS implies different applications as well as specific hardware and control problems. On the basis of accumulated experience, following features can serve as criteria for closer description of various HAS designs (Tomović et al, 1990): **1)** Partial Mechanical Support - PMS, **2)** Parallel operation of the biological and mechanical system - PHAS, **3)** Sequential operation of the biological and the mechanical system - SHAS.

The PMS refers to the use of braces to assist FES only at specific events within a walking cycle (Andrews et al 1989).

Parallel use assumes permanent exchange of power and control between the biological and mechanical system during a complete gait cycle. As a prerequisite for PHAS, reference joint trajectories of lower limbs must be stored in the controller. The main difficulty is, however, how to compensate the inadequacy of FES provoked muscle responses by external supply of power and control in order to match the desired gait performance. Is should be possible to arrive at such control algorithms but the sequential solution of HAS been preferred as being more appropriate to patient's needs. The reasons are given below.

The SHAS, in its pure form, consists of two contiguous phases:

a) gait segment driven by biological (muscles) actuators only,
b) gait segment driven by mechanical actuators only.

The phase a) is, clearly, under FES control with participation of internal sensory-motor control, if available; the phase b) is supported by the active brace. To avoid jerky transitions between the two segments and preserve the internal system energy for further gait activity, evidently, an overlapping interval must be provided to assure efficient and smooth transfer. Successful design of the transfer process in SHAS represents the key to its satisfactory operation.

## SEQUENTIAL POWERED HYBRID ASSISTIVE SYSTEM (SHAS)

The application of multichannel surface functional electrical stimulation suffers from numerous deficiencies: **1)** muscle fatigue , **2)** use of a withdrawal reflex as a replacement for swing phase of gait cycle, **3)** a delay of knee extension after switching-off of flexion reflex and the like. These elements are main limiting factors imposing slow ambulation with hand supports instead of regaining walking. The term "slow ambulation" is used for gait speeds which are less than v= 0.6 m/s. Duration and distance of ambulation are limited in these systems as well to only several minutes and few dozen meters. A multichannel implantable system still suffers from similar deficiencies, but a gait is pattern faster (up to 0.9 m/s) and distance rises up to several hundred meters. Our experience with the use of FES systems points out that in mid-thoracic lesion in addition to above mentioned deficiencies, a major problem arises from insufficient control of hip joints. This fact influenced a design of an external self-fitting powered orthosis to be used in combination with a four-channel FES system. Our main goal in this experiment was to test the usefulness of cybernetic actuator in hip joint to increases hip flexion, thus to shorten the swing phase and increase the gait speed.

**Patient selection.** We selected a patient who could walk and maintain standing with a four channel FES system, but who, in the same time, had inadequate hip control. Patient B.T., male, 43 years, fell down from 10 m height in June 1988. He suffered a spinal cord injury, he was treated surgically, and since than he has a complete motor paralysis below the level T6. He was admitted in Rehabilitation Institute "Dr Miroslav Zotović", Belgrade in June 1989 for gait restoration. Upper extremities are well developed, the grasp force is limited because of previous fractures of both wrists (during the accident in 1988). Paraspinal and back muscles are well preserved above the lesion and minimal activity of the upper part of abdominal muscles exists. Balance in sitting position, as well as in standing position is good. Spasticity is moderate and the patient does not use any antispastic drugs. The X-ray and CT scan shows no stress fractures or other deformities, neither major osteoporosis.

The SHAS integrates several components: **1)** multichannel electronic stimulator with surface electrodes, **2)** powered external orthosis, **3)** microcomputer controller, and **4)** sensors (Figure 1).

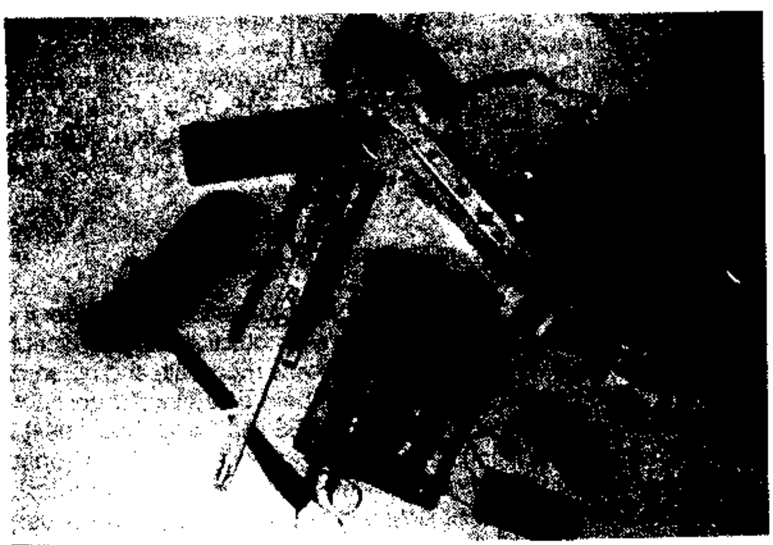

Figure 1. Hardware of an SHAS with powered hip joint

**1)The multichannel FES system.** The electronic stimulator with four isolated channels was applied. All three parameters in monophasic pulses can be controlled by the micro-computer: pulse width ($T = 50 - 600\,\mu s$), interpulse interval ($IPI = 10 - 100$ ms), and the amplitude ($I = 0 - 140$ ma). Current control provides constant current stimulation for impedances from 450 $\Omega$ to 1.8 k$\Omega$. The surface electrodes are conductive rubber with conductive gel or stainless steel mash with water soaked felt. The stimulator is battery operated and provides the stimulation with full power up to 6 hours. Rechargeable batteries are integrated.

**2)External powered mechanical orthosis.** Details on the Self-fitting modular orthosis (SFMO) are presented earlier (Popović, 1986). The knee joint is equipped with electrically controlled brake mechanism (Popović, 1986). The construction of the powered hip joint relies on the use of the motor unit which is located at the hip joint module. The actuator has two independent motor units, one providing flexion and extension and second controlling the stiffness from loose to locked state. Mechanical properties of such an device, called cybernetic actuator are explained elsewhere (Popović et al, 1990). The specific attention is payed to the attachment of the motor unit to the brace and positioning of the motor unit to allow easy mounting and to allow patient to perform other activities without interference. The weight of the motor unit is 369 g, the over all size is adequate to fit the self-contained system. The power (NiCd rechargeable batteries) is provided with 10 D cells located in a special belt.

**3)Microcomputer system.** The microcomputer system is based on self- contained micro-controller, with an 68HC11 microprocessor. Control algorithm is an skill based expert system.

**4)Sensors.** Sensory feed-back uses knee and hip angles and angular velocities, ground force reactions in shoe insoles and a pendulum potentiometer for the displacement of the thigh regarding the gravity line. The angles are recorded using a Hall effect transducer (Texas Instrument) and angular velocities are estimated (software) from angular changes. Ground reaction is recorded using Interlink polymer and conductive pattern printed on a mylar layer (Biomech Design, Edmonton, Alberta). The present system has strain gauge equipped under-elbow crutches. This information is used, only, for estimation of efficacy of the system. The force, in direction of under-elbow crutches, is measured.

## EXPERIMENTAL WORK WITH SHAS

The patient has been admitted to the rehabilitation institute a year after the onset of injury. At that point, he was not trained to use any ambulation device, except for his wheel-chair. His muscles were very weak, spasticity was moderate and no contracture were registered. He was included in physical therapy before he joined the project. Using slightly modified known procedure of four-channel FES (Kralj and Bajd, 1989) he was prepared to start standing and walking. The training protocol for muscle strengthening and muscle fatigue resistibility started with daily electrical therapy (45 minutes). At the beginning, as expected, the duration of sessions was shorter because of very weak muscles caused by delayed start of stimulation after the onset of the spinal cord injury. Two daily sessions were included in order to decrease the period of strengthening. Simultaneously B.T. was included in the program for standing and walking with callipers in parallel bars. Initially, we carefully checked the bone and joint status. Once, B.T. learned to perform swing-through gait in parallel bars we switched him to non-powered self-fitting modular orthosis. He was taught to done and doff orthosis on his own, to transfer with SFMO from the bed to the wheel-chair, to stand up from the wheel-chair in the parallel bars. The further step, required by our protocol, was to combine the non powered brace and FES. Finally, we included a powered hip joint, and electrically controlled locking device at the knee joint of the orthosis. Details on the motor unit, as well as, the protocol for the brace treatment and control method are explained elsewhere (Popović, 1986, Schwirtlich, 1984, Popović, 1987). The specific control method for this brace is explained in this book (Tomović et al, 1990).

This paper describes only objective criteria for determination of the efficiency of the powered hybrid brace. Objective criteria relies on the speed, uninterrupted duration of the use, distance of walk, energy consumption while standing and walking and arm forces. Kinematic analysis and some dynamic records were taken in order to estimate above mentioned quantities.

## RESULTS AND DISCUSSION

We will start with the data obtained in patient after the muscle strengthening program.

### Results with the four channel FES system.

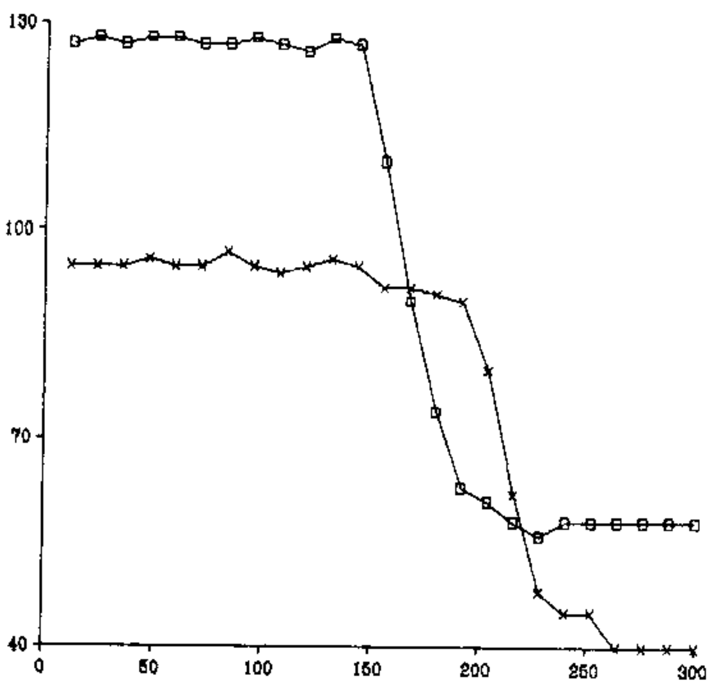


Figure 2. Maximal torque generated in the knee joint when quadriceps m. is stimulated with two large surface electrodes. The upper trace shows that contractile force starts to decrease from its maximal value ($M_{max}$=127 Nm) after 132 seconds, for measurements in flexed position (90°). The lower trace starts to decrease from its maximal value ($M_{max}$=95 Nm) after 204 seconds. The torque was reduced to 50% of its initial value after 192 seconds in flexed position, and after 264 seconds in full erected position. Stimulus parameters were fixed during stimulation (I=120 mA, IPI=50 ms, T=300$\mu$ s).

As suggested in many research results, the knee torques are essential for the efficacy of FES. Our test consists of two types of measurements, both for the muscle fatigue and force. In first one, quadriceps muscles were stimulated, supramaximal (tetanic) contraction for 6 seconds was induced and interrupted for resting periods (6 seconds). The specially designed chair for force recording with different knee angular positions, between $90^{0}$ and full extension was used. The record shows that the contractile force starts to decrease from its maximal value ($M_{max}$=127 Nm) after 132 seconds, when the leg is flexed. The torque starts to decrease from its maximal value ($M_{max}$=95 Nm) after 204 seconds. The torque was reduced to 50% of its initial value after 192 seconds in flexed position, and after 264 seconds in full erected position. (Fig.2).

The second test measures the duration of full extension of the knee joint, with a loaded thigh (resistive torque 12 Nm obtained with sand bags at the ankle joint). The left leg was slightly less fatigable, so the measured time is just about 4 minutes, and the right leg gave up after 3 minutes and 40 seconds (Fig. 3).

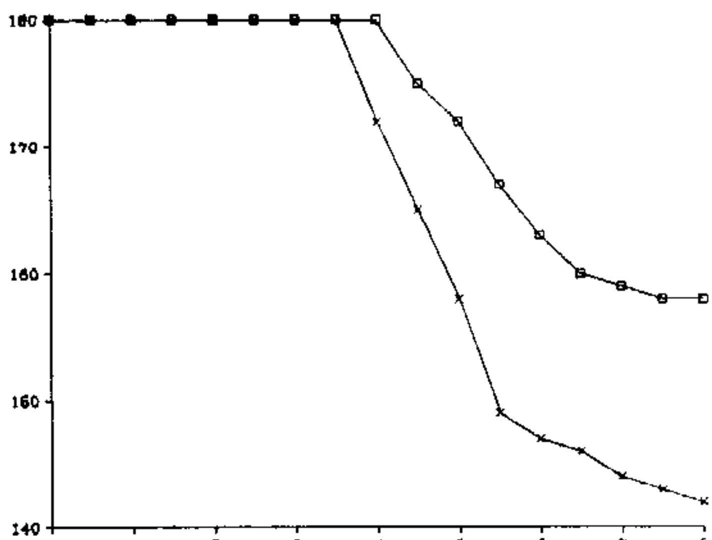


Figure 3. Knee angle in both legs of the paralyzed patient. The figure shows the result of uninterrupted stimulation of quadriceps m. with same parameters as in Fig 2. The shank of the leg was in horizontal position, and the leg was loaded with the sand bag (m = 4 kg) at the distance of 30 cm from the knee axes. As it is expected there is asymmetry between legs, and maximal duration is limited to several minutes (t ≈ 220 s).

The other important criterion is stimulated hip flexion and extension which is rather difficult to measure. The measurements were performed in standing position. In the same time it has to be pointed out that strong hip flexion, obtained through a withdrawal reflex, involves a painful and uncontrollable dorsi- flexion, external rotation and eversion of the foot and knee flexion which is completely inadequate for the swing phase of the gait. Thus, the stimulation strength has to be selected upon the desired knee and ankle movement. Maximal recorded hip flexion, for such a movement, was less than 35°. Our computer simulation points out that for the reasonable speed of gait at least 50° flexion angle is required. The average angular speed measured at the hip joint is less than 0.5 radians/second which is equally unacceptable for reasonable speed of gait (Fig. 4).

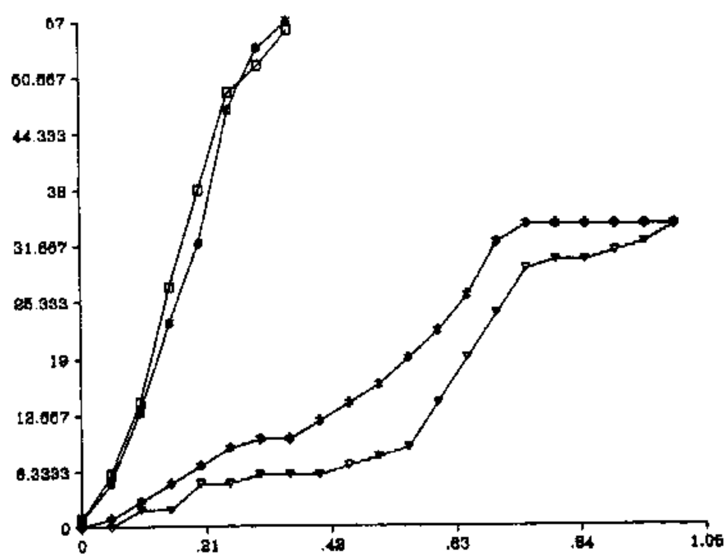


Figure 4. The angular change in hip joint while subject is standing in full erected position. Lower traces are for the FES generated movement in left and right leg, and upper traces for the SHAS generated movement in both legs. The use of SHAS, as expected, provides bigger hip flexion and corresponding angular velocity. The criterion for the setting of the stimulation strength was the knee angle and ankle movement during elicited movement.

The maximal recorded gait speed was 0.28 m/s, maximal distance was limited to 6 lengths of parallel bars (30 m), including turning around at ends of the parallel bars. Maximal standing period was less than 5 minutes. The patient was able to stand up from sitting position and to maintain upright posture in parallel bars more than 4 times, in series, every time standing for at least 3 minutes, while resting for five minutes between standing periods.

In this pilot study we have not measured the oxygen consumption and arm forces.

## Results with the use of SHAS.

Obviously there is no need to measure the muscle fatigue because of the presence of the brake mechanism in the knee joint and fully controllable hip joint. The brace limits the ankle movement and acts partially as ground reaction orthosis. Our main concern was how much we can increase the hip function with the powered external brace. According to selected motor unit we succeeded to increase the maximal hip flexion to $50^{o}$ and to increase the angular speed to 1.2 radians/second, which is still much less than in normal individuals. The angular velocity was limited because of the principles of self-contained portable battery operated system design and safety of patient skin, tissue and bones (Figure 4).

Maximal standing was prolonged to at least 30 minutes and it depended only on over all patient fitness. The maximal distance of gait was prolonged to hundred meters. The maximal measured speed in parallel bars was 0.7 m/s, while with the walker is still rather low, v = 0.47 m/s. Because of the short time we still have not completed the system, so the patient was not able to walk with the under-elbow crutches safely.

One of the most interesting results is presented in Figure 5. Angular changes in gait cycle with the four channel FES system and SHAS as well as time constants are pointing out the reasons for the use of the SHAS.

## PROSPECTIVE

This paper presents the first hybrid brace with a powered hip joint. Our main concerns were safety and reliability of the gait performance, increase of the duration of use and increase of the speed of gait.

The cosmetics was not our primary goal, however it is extremely important for an practical system. We concentrated on the use of surface stimulation, but flexibility of this system allows the use of any implantable stimulation system. The general conclusion is that the use of powered hip joint improves the quality of the gait by all means. The powered hip joint adds the functional movement which resemble much more to the swing phase of the gait in comparison with the flexion reflex. It has to be mentioned that the control algorithm combines the flexion reflex and hip flexion and quadriceps stimulation with the hip extension in order to improve the quality of the gait. The work presented here is a single patient study and it can not be generalized, but it suggests that the approach which has been suggested by our group several years ago, is a method which can improve and facilitate the broader use of FES systems in rehabilitation of spinal cord injured patients.

Our recent work considers some other combination of powered external brace and skill based expert system control for different incomplete spinal cord patients. The flexibility of the system allows easy combination of different types of braces and FES components, in addition to different sensory feed-back.

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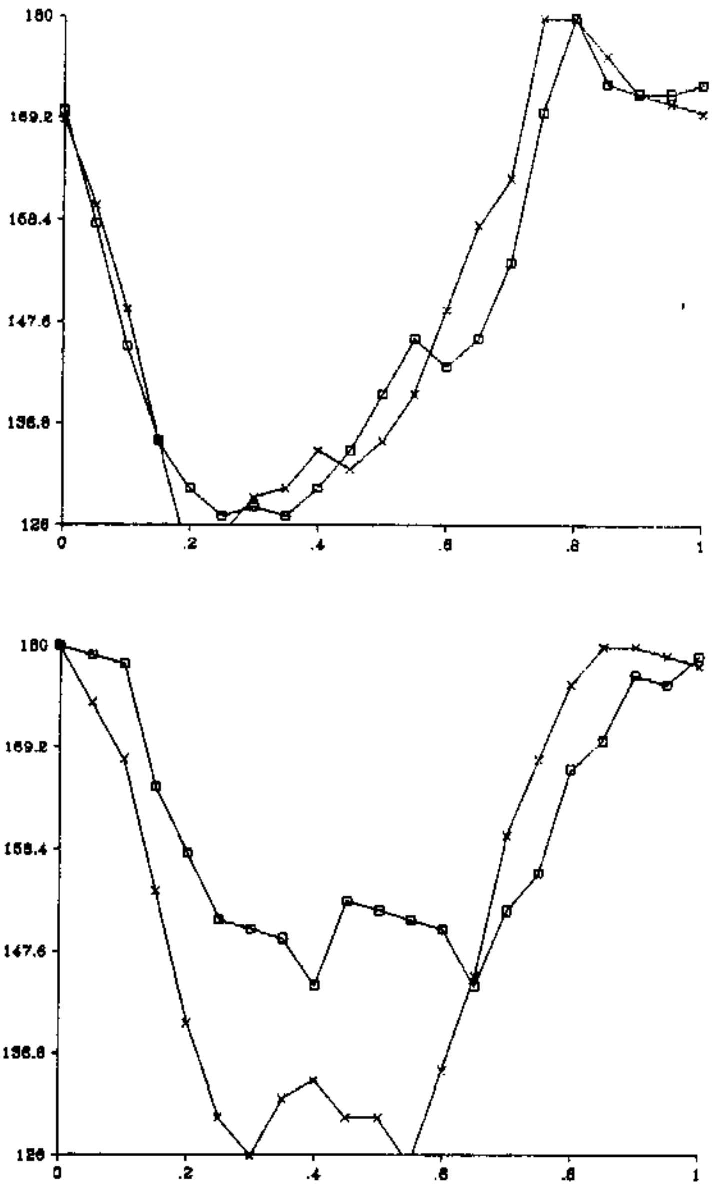


Figure 5. Knee and hip angles during an average gait cycle are presented. The duration of the gait cycle is normalized. The gait cycle with FES was T = 2.06 s, and it was much shorter with SHAS, T = 1.29 s. The average record is taken from 50 steps. The length of the step was 0.64 m for the FES, and 0.88 m for the SHAS. It can be seen that the knee movement is very similar, bur the difference occurs in the hip movement. Bigger flexion and faster movement is obtained in SHAS application, thus the faster gait pattern may be achieved. External brace was activated in periods marked with horizontal line

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