FUNCTIONAL ELECTRICAL STIMULATION—A NEW HOPE FOR PARAPLEGIC PATIENTS?

Alojz Kralj, Eng., D.Sc.
Assistant Professor
Faculty for Electrical Engineering
University of Ljubljana, Yugoslavia

Slobodan Grobelnik, M.D., D.Sc.
Head Physician
Rehabilitation Institute
Ljubljana, Yugoslavia

INTRODUCTION

Spinal-cord lesions result in a variety of symptoms in the area of motor activity and sensitivity, and consequently in deficiencies of numerous organs. Among them, one of the most crucial problems is the dysfunction of certain muscles or groups of muscles causing a restriction of motor activity. The inability of a subject to walk or stand has a bearing on a number of other very significant factors—space within which the subject lives is strongly limited. This has an effect on the acquisition of further life experiences; contacts with people within the subject's environment cannot be maintained as they used to be prior to the onset of disability. Moreover, the disabled subject would probably not be able to retain the job he had previous to his infliction, etc. Outside of the home, the majority of things are quite difficult or even impossible to reach due to architectural barriers, inadequate transportation, and other obstacles. The above-mentioned problems may inevitably result in psychological disturbances.

Man has always tried to overcome limitations in independent motor activity. Numerous pictures and other historical documents illustrate designs, development, and utilization of various aids such as crutches, braces, etc., aimed at helping the disabled overcome and/or limit these difficulties. Today two devices, designed some time ago, are utilized: the wheelchair and the orthopedic long-leg brace. Of the recently developed passive supporting devices, the Ortazur long-leg braces utilizing air cushions (15) and the swivel walker (16) should be mentioned.

As we know, long-leg braces cannot be prescribed or given to every paraplegic patient, particularly not to one who manifests a high dorsal spinal-cord lesion. Also, the appliance frequently has to be attached to a corset, so it inevitably becomes both cosmetically and functionally unacceptable. Utilization of braces by paraplegic patients with lower level lesions is much more successful. In such cases, sometimes only fixation of the knee and ankle joint is needed. These devices are much lighter and easier to fit. Experience and analyses have shown that even these aids are not so commonly used by the disabled. They do utilize them, even with relative success, when treated at rehabilitation centers where they are subjected to a day-to-day training program. But as soon as they get home, they do not make use of them any longer and prefer staying in wheelchairs. Research related to the success of our rehabilitation methods has been carried out. A special questionnaire was submitted to all the disabled rehabilitated at our Institute during the past 10 years. It has been established that long-leg braces have been utilized by only 20 percent of those people who had been making use of the device during their daily training at the Institute. The majority of these people rejected the device for its weight, the tremendous physical effort during walking, and inadequate design—it takes too much time to put on and take off the aid. Figure 1 confirms the above statements. As a result, we have decided to change our rehabilitation program. The disabled are told the way the device has to be utilized, they are informed about the possibilities of getting one if they wish, but we do not force them to use it if they do not want to. Instead, intensive training in wheelchair utilization is provided.

In our opinion the above decision is the road of least resistance, a kind of surrender resulting from our inability to give paraplegic patients a better aid than a wheelchair. For that reason we have been constantly searching for a more acceptable and successful walking aid for paraplegic patients. Use of functional electrical stimulation (FES) alone or combined with sophisticated mechanical bracing could be one of these possibilities. A suitably designed functional and technological approach might help these people partly regain the ability to stand, and such a brace would be easily applied which is not the case with conventional braces.

The idea of utilizing functional electrical stimulation to achieve control of motor action in patients is not new (1,2). In the past years many orthotic devices (3,4,5) based on FES have become commercially available, and we are expecting more growth in this area. A number of promising orthoses have already been developed (6,7,8, 9,10,23,24). A combination of the surface stimulation with the implantable stimulation technique in conjunction with devices based

FIGURE 1.—Paraplegic patient equipped with conventional long-leg braces.

on the multichannel FES will provide new ways for various rehabilitation appliances.

The present-day research and practical routine work in this realm is primarily concerned with treatment of hemiplegic patients. The first attempt to raise a paraplegic patient into a standing position by the use of functional electrical stimulation was carried out by Kantrowitz (11). The idea of applying a simple orthotic device using FES for locking the knee to enable the subject to achieve an erect posture was first reported in 1969 (12). In 1970 (13) the first two-channel implant was inserted for the stimulation of the M. quadriceps and hip extensors of a complete T5 paraplegic patient with a view to enabling the patient to stand in an erect posture and to ambulate. A couple of years later, a preliminary feasibility study of the use of FES in paraplegic patients was reported (14). We should like to describe our experiences with FES in paraplegic patients from the rehabilitation engineering standpoint.

FUNCTIONAL ELECTRICAL STIMULATION

Of various possibilities for external control in paraplegic patients, electrical power in the form of FES seems to be the best solution. The greatest problem of all active external assistive devices is power consumption. Only FES has possibilities to overcome this problem owing to the power generation in the electrically stimulated muscle. Figure 2 clearly illustrates the electrical energy triggering only the stored energy in the muscle. Power amplification may be observed between the electrical stimulus power and the power exerted on the load. The power gain factor may be 2,000 for surface stimulation, and up to 200,000 for implanted nerve stimulation (7,22). From the engineering point of view this is a very important advantage of FES. Orthoses using FES do not need force transfer attachments, levers, and fixations of the hardware to the body, because FES uses the natural exoskeletal system and levers. This is the second important advantage. We hope that such outstanding advantages of FES combined with a sophisticated bracing technique will enable better motor development in paraplegic patients.

From the engineering point of view it is important to be familiar with the "motor performance" of an electrically stimulated muscle, in relation to the mechanical properties versus electrical stimulation as the triggering input. The behavior of the "muscle motor" in general

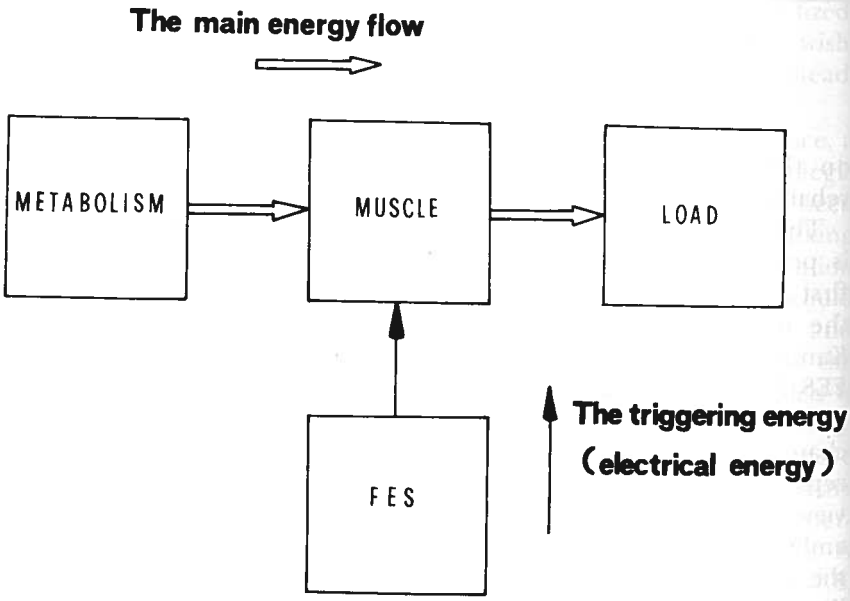


FIGURE 2.—Block diagram showing the main energy transfer sources by FES.

limits and determines the later properties of orthotic devices using FES of muscles. The triggering determines to some extent the circuitry which must take into account the mentioned "motor performances" too.

As a matter of fact, power triggering is the principal concept we must be familiar with if we want to understand and use it. Once this is done, we can compare a stimulated muscle "force generator" with the other known possibilities. As there have been several other facts, such as deficient knowledge and the absence of technology for afferent and spinal-cord stimulation, we have had to limit our work to efferent electrical stimulation only.

Another question has been the number of patients that can benefit from FES according to age, location of lesion, and lapse of time following damage of the spinal cord.

The third necessity was to know which paraplegic muscles respond to FES, to become familiar with the mechanical and dynamic properties of stimulated muscles, and to learn more about fatigue and methods for strengthening muscles.

METHODOLOGY

During our experiments, rectangular stimulation pulses were used with a pulse width between 0.3 and 0.7 ms. and repetition frequency of 50 Hz. It should be stressed that this is not the optimal stimulation. All the moments and forces illustrated should not be considered as the absolute maximal values if not stated otherwise.

Only surface stimulation was utilized. The tin electrodes used were coated with gauze and moistened with water. For stimulation of the peroneal nerve, the electrode sizes were 3 x 3 cm., for M. gastrocnemius 4 x 5 cm., and for large muscles such as M. biceps, femoris, or M. quadriceps sizes were 6 x 7 cm. An eight-channel stimulator, with a built-in programer, and two torque measuring braces were used throughout the experiments. One of them was used for the ankle joint and the second for the knee joint. In the experiments the isometric moment was measured as the response to FES of various paraplegic muscles.

For muscle force testing the Mayo Clinic method was used (19). Of course, the method is mainly performed for the testing of voluntarily controlled forces, but there are no limitations for the use of this method in the testing of forces produced by FES.

SELECTION OF PATIENTS

We first tried to find the percentage of paraplegic patients in

TABLE 1.—*Paraplegic Patients—Response of Muscles to FES*

No.	Sex	Age	Lesion		Fibularis & tibial. ant.		Gastroc-nemius		Biceps fem.		Quadriceps fem.		Gluteus max.	
			Site	Duration	L	R	L	R	L	R	L	R	L	R
1	2	3	4		5		6		7		8		9	
1	f.	26 y.	D 5–7	7 y.	−4	−4	−1	−2	−4	−4	−3	−3	3	2
2	f.	24 y.	L 1	4 y.	0	0	−2	−2	0	0	0	0	0	0
3	m.	33 y.	L 1	5 y.	0	0	−2	1	0	0	1	1	0	1
4	m.	36 y.	D 10	2 y.	0	0	0	0	0	1	0	0	1	1
5	m.	42 y.	D 9, L 3	31 y.	0	0	0	1	1	0	0	0	1	1
6	m.	29 y.	D 9	6 y.	3	5	1	1	3	3	−4	−4	−2	2
7	m.	21 y.	D 7, 8	6 y.	3	4	−4	−3	−3	2	−4	−4	−2	−2
8	m.	25 y.	D 5	4 y.	0	0	0	0	1	1	1	0	1	1
9	m.	39 y.	D 12	4 y.	−4	2	0	0	−3	−3	1	1	2	1
10	m.	41 y.	D 12	11 y.	0	0	0	0	0	0	0	0	0	0
11	m.	25 y.	L 1	4 m.	3	3	0	0	1	1	0	1	1	1
12	m.	22 y.	D 3–7	2 y.	−4	3	−2	−2	2	2	5	4	−2	2
13	f.	24 y.	D 11	9 y.	−2	−2	1	0	−4	−4	−3	2	1	−2
14	m.	40 y.	D 7–8	13 y.	−3	3	2	−3	1	1	2	−2	1	1
15	m.	30 y.	D 3	7 y.	−4	3	−2	−2	3	3	2	−4	−3	3
16	m.	23 y.	D 12	1 y.	0	0	0	0	0	0	0	0	0	0
17	m.	21 y.	D 3–4	1 y.	0	0	−2	−2	−4	−4	−3	−3	2	2
18	m.	57 y.	D 4	14 y.	−4	−3	1	1	−3	3	−3	1	−2	−2
19	m.	25 y.	C 7	1 y.	0	0					−3	1	Hypersens	

20	m.	43 y.	D 12	3 y.	3	−3	1	1	1	−2	−3	−2	1	1
21	m.	32 y.	D 9	5 y.	0	0	0	0	0	0	0	0	1	1
22	f.	31 y.	D 9–10	7 y.	−2	1	0	0	−2	2	1	1		
23	m.	36 y.	D 10–11	9 y.	0	0	0	0	0	0	0	0	0	0
24	m.	53 y.	D 7–8	15 y.	−3	2	−2	−2	−2	−2	−4	−4	−2	1
25	m.	31 y.	D 12	10 y.	−4	−4	0	0	−3	3	3	−4	−2	−2
26	m.	33 y.	D 12	4 y.	1	1	1	−2	2	−3	1	1	1	1
27	f.	27 y.	L 2	8 y.	0	0	0	0	0	0	4	4	0	0
28	f.	42 y.	L 1	1 y.	−2	0						Hypersens		
29	f.	26 y.	C 6–7	10 y.	0	−3	−3	−2	−3	−3	3	3	1	1
30	f.	44 y.	D	12 y.		3		0				−4		
31	m.	26 y.	C 5	2 y.	1	0	−2	−2	0	0	1	1	1	1
32	m.	37 y.	C 6	4 y.	4	1	4	−4	−3	3	−3	−3	2	−2
33	f.	21 y.	C 3–5	5 y.	0	3	1	0	−3	−3	1	1	1	1
34	f.	30 y.	C 6–7	2 y.	3	−4	1	1	−3	−3	1	1	1	1
35	m.	21 y.	C 6–7	3 y.	0	0	0	0	1	0	1	2	−2	−2
36	m.	17 y.	C 6	1 y.	0	0	−2	1	2	3	3	−4	−2	−2
37	m.	29 y.	C 5	12 y.	4	3	1	1	−3	−2	5	5	−3	2
38	m.	26 y.	C 5	2 y.	0	0	0	0	0	0	1	0	1	0
39	m.	34 y.	C 6–7	11 y.	0	0	1	1	3	3	2	−2	−2	−2
40	m.	38 y.	D 5	1 y.	0	0	0	0	1	1	−4	4	1	−2
41	m.	28 y.	D 4–7	3 y.	−3	4	0	0	3	3	−4	−3	−2	−2
42	m.	49 y	L 3	2 y.	4	−4	1	1	−3	−3	4	2	−2	−2
43	m.	24 y.	D 12	3 y.	−3	1	1	−2	−3	−3	−2	2	1	−2
44	m.	55 y.	D 11	3 y.	−2	2	0	2	1	1	1	1	1	1
45	m.	31 y.	L 1	5 y.	0	0	0	0	0	0	0	0	0	0
46	m.	35 y.	D 12	1 y.	2	−4	1	1	−3	−3	0	0	0	1
47	m.	34 y.	D 12	5 y.	0	0	1	0	0	0	0	−3	0	0
48	f.	32 y.	L 1	1 m.	0	0	1	0	0	1	0	0	0	2
49	m.	49 y.	D 3	2 m.	0	0	0	0	0	0	0	0	0	0
50	m.	39 y.	D 12+L 1	1 m.	0	0	0	0	0	0	0	0	0	0

TABLE 1.—*Paraplegic Patients—Response of Muscles to FES*

No.	Sex	Age	Lesion		Fibularis & tibial. ant.		Gastroc-nemius		Biceps fem.		Quadriceps fem.		Gluteus max.	
			Site	Duration	L	R	L	R	L	R	L	R	L	R
1	2	3	4		5		6		7		8		9	
51	f.	67 y.	D 6	2 m.	−2	1	0	0	−2	−2	1	1		
52	m.	25 y.	D 3	1 y.	0	−4	−4	−2	4	3	4	3	−2	−2
53	m.	34 y.	D	1 y.	1	1	0	1	−2	−2	1	1		
54	m.	59 y.	C 7—D 1	6 m.	3	2	0	0	−2	−2	1	1	−2	−2
55	m.	64 y.	C 6	2 m.		1	1	1	−2	−3	2	1	2	2

NOTE: Mayo Clinic muscle evaluation: 0 = Normal, −4 = Complete paralysis.

whom muscles were still able to produce movements through FES in regard to their age, location and duration of lesion and other possible conditions. The related testing should be the best answer to the question whether FES may be useful in rehabilitation of this group of the disabled or not.

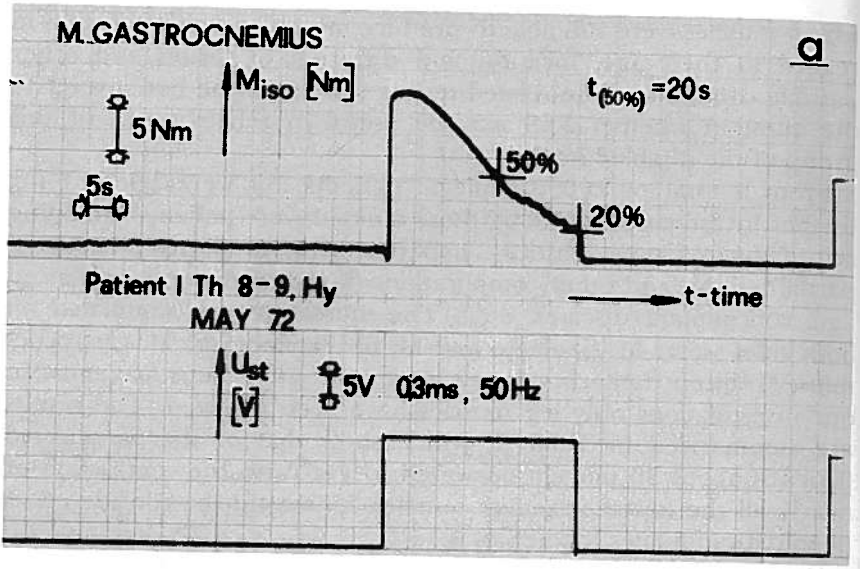
From a total of 100 paraplegic patients, 55 were selected for further investigations. Among those omitted were patients exhibiting skin damage, strong spasticity, hypersensitivity to electrical currents, partial lesions, and other complications (mental deficiency, advanced age, concomitant diseases, etc.). The muscle groups stimulated on both sides were: M. fibularis, and M. tibialis anterior, M. gastrocnemius, M. biceps femoris, M. quadriceps and M. gluteus maximus. In our investigations only leg muscles have been involved. The results are summarized in Table 1, which show that of the 55 selected patients, about 50 percent answered to FES with the contraction of nearly all the lower extremity muscles. These can be considered as potential candidates for FES.

During our experiments, spasticity was observed in some patients. As no exact method exists to measure spasticity, and the phenomenon in itself is still not clearly understood, the notes about spasticity are omitted. Generally speaking, however, patients with slight spasticity showed better response to FES than other patients.

It is obvious that no visible correlation exists between the location of the lesion and the properties of the muscles to be stimulated, and muscle force produced by FES. There is even no strong dependence of the muscle force generated by FES on the lapse of time after damage. This statement is clearly visible if we compare some patients mentioned in Table 1. Comparisons of patients 12, 42, and 52 with 4, 5, 38, 48, 49, and 50 show that the level of lesion does not strongly influence the response of the muscle to FES. The same could be said of the lesion duration. Compare, for instance, patients 1, 4, 12, 13, 14, 25, 42, and 52 with 4, 16, 23, 39, 40, 47, 48, 49, and 50. This comparison, compared with today's most used patient status correlated to the level of lesion, points out that the commonly used patient description is not appropriate and gives no measure regarding the use of FES.

During the muscle testing experiments, it could be clearly observed that the muscle force quickly decreased with time. Our attention was therefore directed toward measuring the muscle fatigue period associated with permanent stimulation.

After a general review of Table 1, we made some detailed analyses of two selected paraplegic patients: Patient I is a D 8–9 paraplegic patient, 24 years of age with slight spasms 4 years following damage. Patient II, (18 years old) has a D 10 lesion with very slight spasticity



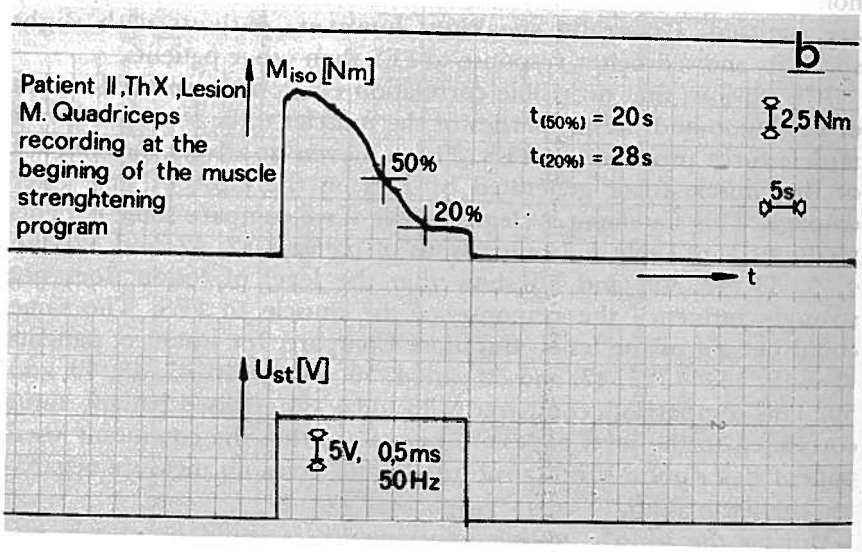


FIGURE 3.—a. Isometric moment plot versus time of electrically stimulated M. gastrocnemius. b. Isometric plot of M. quadriceps moment versus time exerted by FES.

2 years after lesion. In both of them permanent electrical stimulation and various stimulation experiments caused no change in the status of spasticity. During the test period, bladder function was carefully checked and no influence of FES on the existing bladder automation could be observed.

MOTOR PERFORMANCES OF ELECTRICALLY STIMULATED PARAPLEGIC MUSCLES

Fatigue was the first characteristic of electrically stimulated paraplegic muscle that we investigated. The knee joint of a patient was fixed into a torque-measuring brace. The isometric moment of the electrically stimulated quadriceps was recorded as a function of time. Owing to the brace construction we did not try to measure the absolute maximum moment which may be initiated with FES. The measured moments for the knee joint were in the range between 20 and 50 Nm, and those for the ankle joint from 5–20 Nm. A typical isometrical moment plot versus time ($M_{iso}(T)$) is given in Figure 3a for M. gastrocnemius and Figure 3b for M. quadriceps.

The moment decreases with time, and time for a 50 percent drop $t_{(50\%)}$ in most patients is in the range of 15–40 seconds. Fatiguing and $t_{(50\%)}$ and $t_{(20\%)}$ time is the function of the initial moment. For a high initial moment fatiguing is higher and slower when small initial moments are concerned. The initial force varies and can be 50 Nm for about 45 v. stimulation or only 5 Nm for 30 v. stimulation. The patients with spasms have good initial forces, and the $t_{(50\%)}$ is often higher in the order of 40–50 seconds.

The cyclical stimulation, e.g., 5 seconds stimulation and 5 seconds rest period without stimulation, increases the $t_{(50\%)}$ time. After a continuous stimulation trial, a rest period of 5 minutes suffices for the initial force recovering, and we obtain almost the same moment versus time plot. At the end of an experiment after the $t_{(20\%)}$, the change of stimulation frequency from 50 to 40, 60, and 70 Hz has no effect. Only increased stimulation pulse widths result in an increased moment. A change from 0.5 to 1 ms. following the $t_{(20\%)}$ time will result mostly in a 30–50 percent increase of the remaining force.

The dynamic properties of electrically stimulated muscle play an important role once we decide that we would use such a "motor" for performing functional tasks. The dynamic properties of the moment increase and decrease in response to a unit jump in electrical stimulation versus time are illustrated in Figure 4. This is a typical curve. It might be of interest to mention that in the same patient essential changes in the delay time t_{01}, rise time t_{11}, fall time t_{44}

among muscles were not established. M. quadriceps is recorded in Figure 5. The t_{01} time is in the range of 40–60 ms., the t_{11} time between 120–360 ms., the t_{04} time is typical 90 ms. ± 30 percent and the fall time between 90–180 ms. The quadriceps response for two unit jumps, one after another, is shown in Figure 6. The t_{02} time is in the range of 70 ms., t_{22} varies between 150–230 ms., t_{03} between 70–90 ms., and t_{33} between 100–120 ms. Figures 4 and 5 show that the t_{11} time is normally 2–3 times larger as compared to the t_{01}. T_{04} is 1.5 larger than t_{01}. The same can be said of the fall time. A preliminary statement would therefore be that an electrically stimulated paraplegic muscle from a dynamic point of view does not differ much from the normal one. The only noticeable difference appeared in the delay time t_{01}. In most cases the delay time in normally innervated muscles is in the range of 20–30 ms., as compared to paraplegic muscle value from 40–60 ms. For the rise time (t_{11}) and fall time great differences between the normal and paraplegic muscles have not been established.

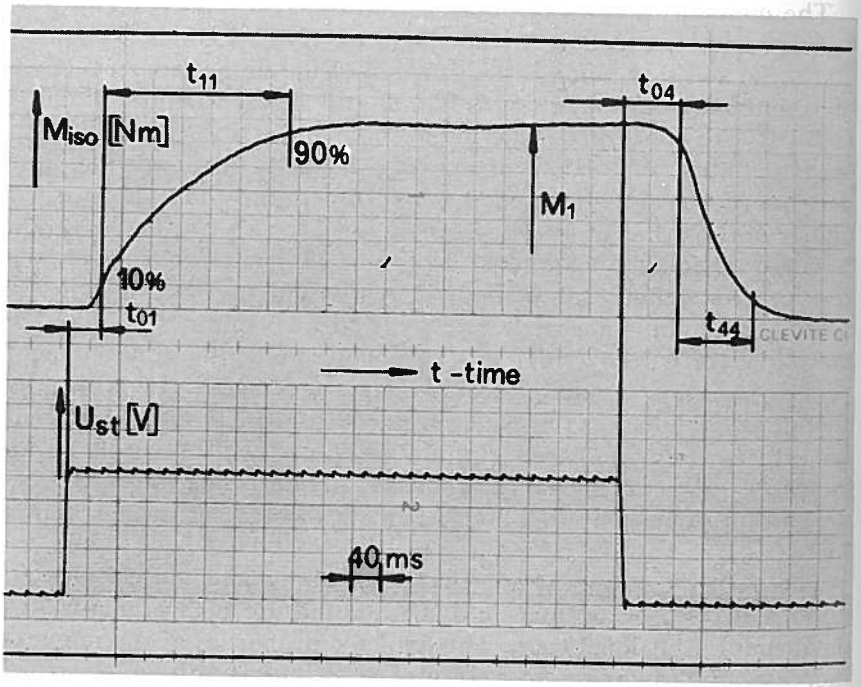


FIGURE 4.—Dynamic plot of the isometric torque time for an electrically stimulated muscle.

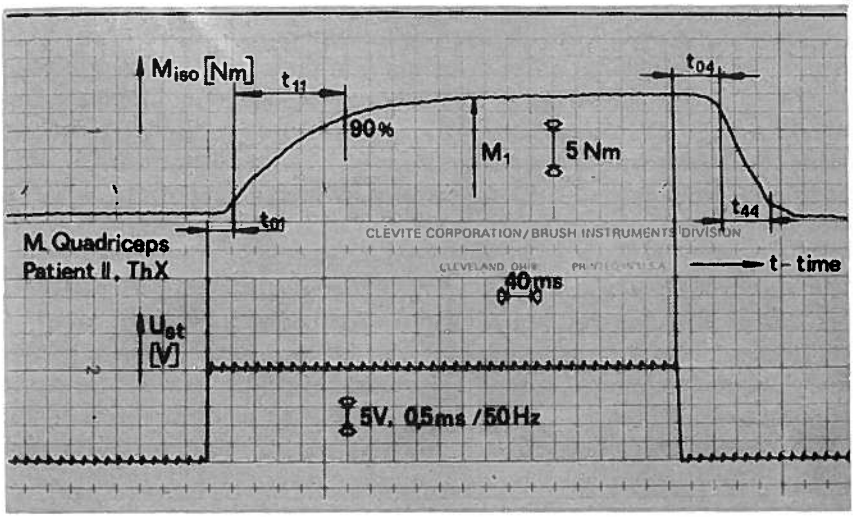


FIGURE 5.—Dynamic plot of isometric torque versus time for FES of M. quadriceps.

PARAPLEGIC MUSCLE STRENGTHENING PROGRAM

The initial forces with FES of paraplegic muscles diminish with time as fatigue occurs early. The question is, whether or not it is possible both to find a program for increasing the initial force and to slow down the fatiguing process. Willemon et al. (13) described a paraplegic muscle strengthening program based on cyclical FES. They reported a stimulation program with cycles consisting of 10

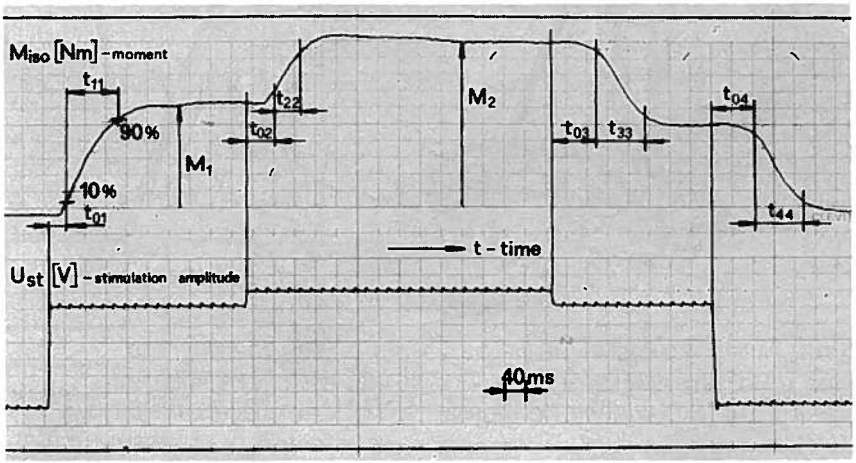
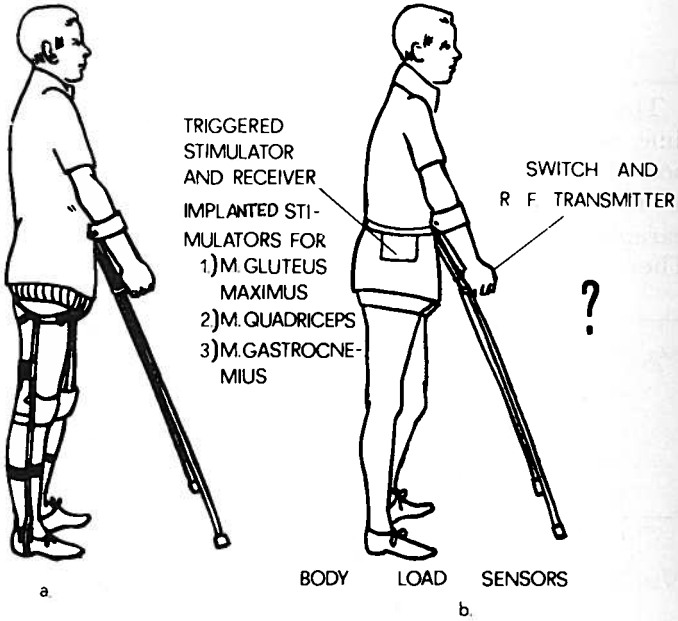


FIGURE 6.—Response record to two unit jumps in FES exerted over the M. quadriceps.

seconds stimulation and 20 seconds rest period, the whole process lasting 12 hours daily. After a 2-month stimulation the fatigue time increased from 15 seconds to "2 hours of sustained unremitting contraction." This muscle strength increase is very high and allows static muscle stimulation, e.g., for at least standing up and/or keeping the legs extended during the standing position of the patient.

Since 1968, we have been investigating the possibilities of developing an orthotic device based on FES for paraplegic patients. The basic idea of such a device is shown in Figure 7 (12).

Encouraged with the published results, we started our research with a slightly changed muscle strengthening program. As a matter of fact, in our first trials we got a response to electrical stimulation only after a 5–10 minute stimulation period. We believed that stimulation without muscle response could not enable increase of muscle strengthening. This was the reason why the stimulation time was not constant over the whole training period. During the



THE PATIENT STANDS BY HELP OF LONG LEG BRACES AND KEEPS THE BALANCE WITH CRUTCHES

THE LEGS ARE EXTENDED BECAUSE OF ELECTRICAL STIMULATED MUSCLES. THE BALANCE IS ACHIEVED BY HELP OF CRUTCHES

FIGURE 7.—Idea of a long-leg brace for the paraplegic patient using FES of muscles.

program, the stimulation time was increased and kept within the limits of muscle response to stimulation. The instrumentation setup in the stimulation program is given in Figure 8. At the beginning of the program each patient was stimulated three times for 10 minutes daily. The stimulation time was increased daily and after 2 months we stopped at three times 30 minutes daily. The torque in the knee exerted from M. quadriceps was measured regularly and so was the muscle bulk 15 cm. above the patella. The muscle recovery plot versus time is given in Figure 9. We see the initial muscle force and the joint torque increased almost five times and the muscle bulk to nearly 3 cm. Fatiguing of M. quadriceps for continuous stimulation recorded after 6 months of muscle strengthening is given in Figure 10. It can be seen that the $t_{(50\%)}$ time is 121 seconds. If we compare this with Figure 3b, we get fatiguing time which is six times larger. When comparing Figure 10 to Figure 11, we can see that a strengthened paraplegic muscle does not differ much from a normal muscle as far as fatiguing is concerned.

These results show that after a 3-month period of muscle strengthening by means of FES, increase in strength is negligible. The question is what the optimal muscle stimulation strengthening

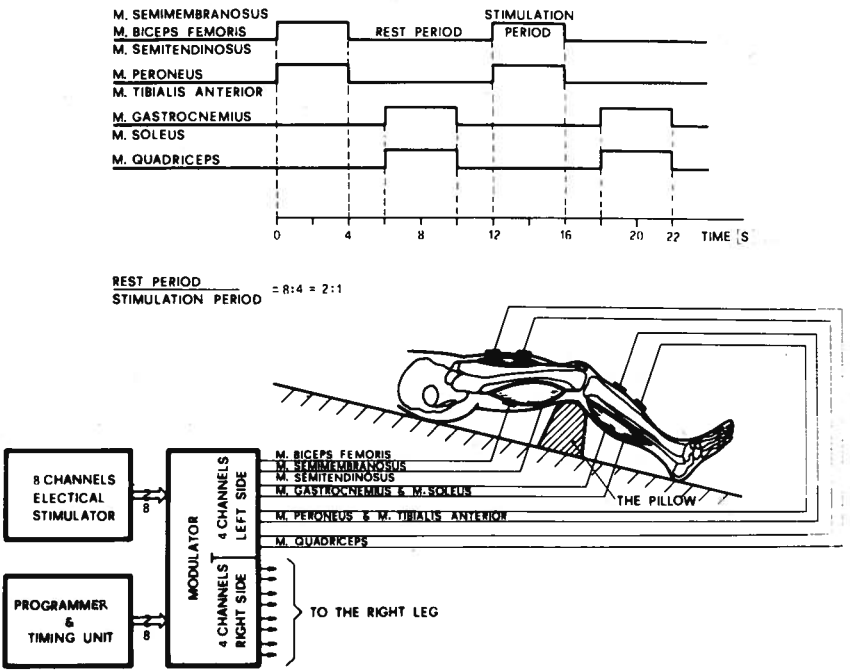


FIGURE 8.—Muscle training instrumentation setup.

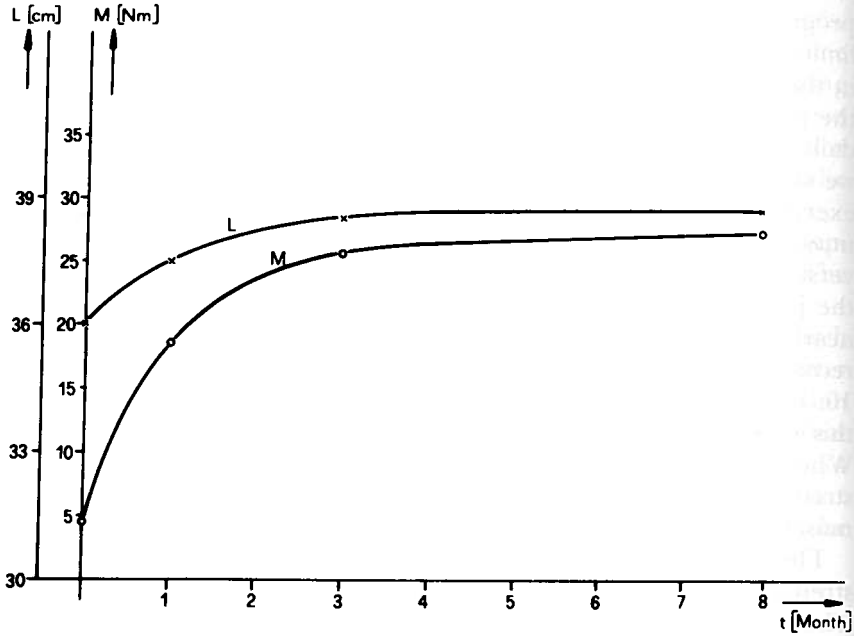


FIGURE 9.—Muscle recovery versus training time.

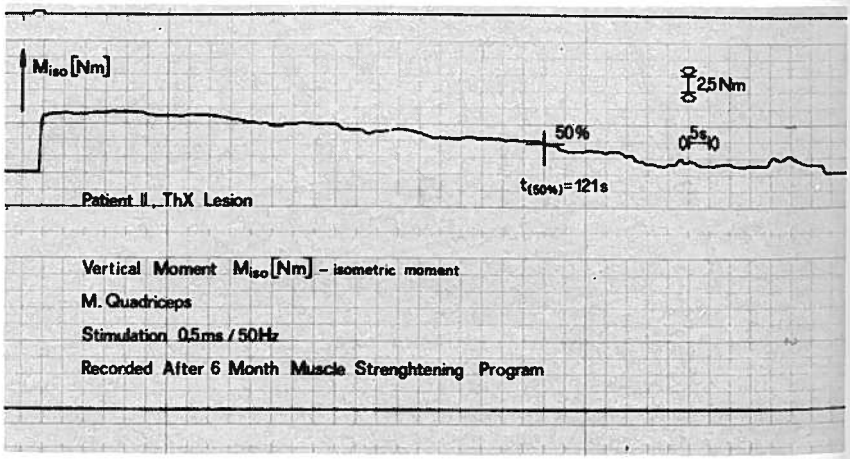


FIGURE 10.—The M. quadriceps fatigue versus time at the end of the training period.

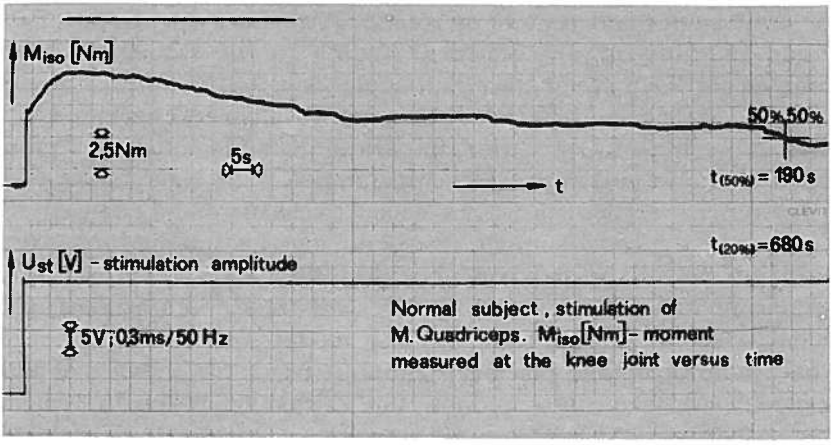


FIGURE 11.—Fatiguing time for a permanent normally innervated contraction of the M. quadriceps.

procedure should be like. Especially, the daily stimulation time should be determined exactly as well as the stimulation and rest periods of one stimulation cycle. Furthermore, the muscle loading during the program and the amount of load increments should be determined, since this plays an important role, as shown by Rose et al.(17). Various muscles have various optimal loads (Rose et al. (18), Krusen et al.(19)).

In regard to the above, this report could be considered as an experiment showing that under certain circumstances the paraplegic muscles could be strengthened with electrical stimulation. It would be interesting to mention that after a 7-month rest period without any stimulation the fatiguing curve showed the typical shape of the curve obtained before the program of strengthening with a typical quick fatiguing and decreased initial force. The muscle bulk decreased too. This shows that muscles that are not used are quickly subject to atrophy.

STANDING UP—PRELIMINARY EXPERIMENTS

In man the standing-up function is a simple task from a normal subject's point of view, but as far as the mechanisms involved are concerned, it is quite difficult to explain exactly how it happens. During this phase, both legs are supporting and lifting the body. This is achieved through an interaction of joint muscle balance, torque balance, and body balance. The body keeps and controls the center of gravity exactly midway between narrow limits according to

the joint angles and centers of rotation. An exact calculation of the whole standing-up procedure according to the stability criteria, muscle coordination, etc., via joint angles is very difficult, because all the needed data and knowledge are not available. Therefore, it is obvious we have to start with simplifications and models describing only the known and important happenings.

For our experiments it was necessary to calculate and compare the worst case of needed muscle forces during the standing-up procedure which will be performed with FES. As this comparison is promising it would be worth trying with the chosen patient. Our calculations have been based on the model shown on Figure 12, where natural muscle placing is simplified and the muscles of only one joint are taken into account. The calculation procedure can be understood observing the given equations 1) to 4) and Figure 12. For the beginning we must find the ground reaction force R (the body structure is considered as a rigid body) for the given external load force W of the structure (in our case the only external force is the body weight force). After this is accomplished, the internal structure and muscle forces can be determined (e.g., 2.), 3), and 4)). Once the internal and external forces are known it is easy to calculate the bone strains by help of Pauwels' (20) findings.

LIST OF EQUATIONS

$$W = R \qquad \text{.... 1)}$$

$$W \cdot h = F_{GM} \cdot r_{GM} = M_h \qquad \text{.... 2)}$$

$$W \cdot k = F_Q \cdot r_Q = M_k \qquad \text{.... 3)}$$

$$W \cdot a = F_{GS} \cdot r_{GS} = M_a \qquad \text{.... 4)}$$

W —Body weight (force)

r —Moment levers

F —Muscle forces

a,k·h —Moment levers to joints of the body weight force

R —Reaction force

A,K,H—Joints

GM —M.gluteus maximus

Q —M.quadriceps

GS —M.gastrocneumius and M.soleus

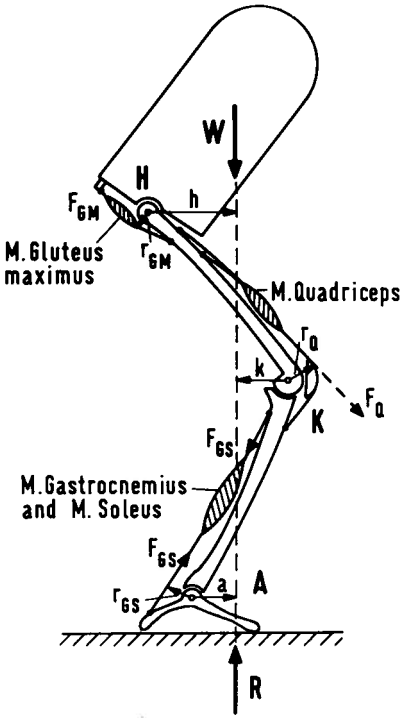


FIGURE 12.—Simplified model of the human structure at an intermediate position during rising from seated to standing position, with estimates of the main one-joint muscles.

For each moment during standing up, the body weight moment lever according to the hip, knee, and ankle joint could be determined with a few simplifications and with limited accuracy. The weight of the segments above the joints can be determined and calculated by help of Drillis et al. (21) data. Using these data it is easy to calculate the M_H, M_K, and M_A for each time interval and posture during standing up. It should be stressed that the optimal relation of joint angles versus time will result in minimal required torques and muscle forces. If we want to calculate muscle forces from the given joint torque the muscle torque levers are needed. The levers for the main leg muscles measured in the sagittal plane only are given in Figures 13, 14, and 15. The levers were measured at the Anatomical Institute of the Medical Faculty in Ljubljana by help of a special X-ray method developed for this purpose. Having in mind the shape of muscle lever curves versus joint angle, we can easily understand that with the maximal muscle force induced by FES in the region of small muscle levers, the joint torque would be small and insufficient. The muscle levers can be used also for estimating the muscle force achieved by FES and for comparing it with the force which could be obtained for the normally innervated

muscle of the same size. This is based on the rule that 1 sq. cm. of the muscle cross-section area is capable of producing a force near 100 N(Newton). We believe that an optimal posture can be obtained for each interval during standing up resulting in minimal consumption of muscle force during the task.

Only two leg muscles, the M. gastrocnemius and M. quadriceps have been stimulated thus far. The foot is placed flat on the ground. The balance support for the patient does not allow him to exert high vertical forces by help of hands if the catch up is put low enough (see Fig. 18). The M. quadriceps is stimulated in a pattern related to the activity in the normal standing-up pattern. The M. gastrocnemius muscle is correlated to this. In case the rise of the center of gravity of the segments about the knee joint does not pass too far from the knee joint rotation axis, the needed force of M. quadriceps

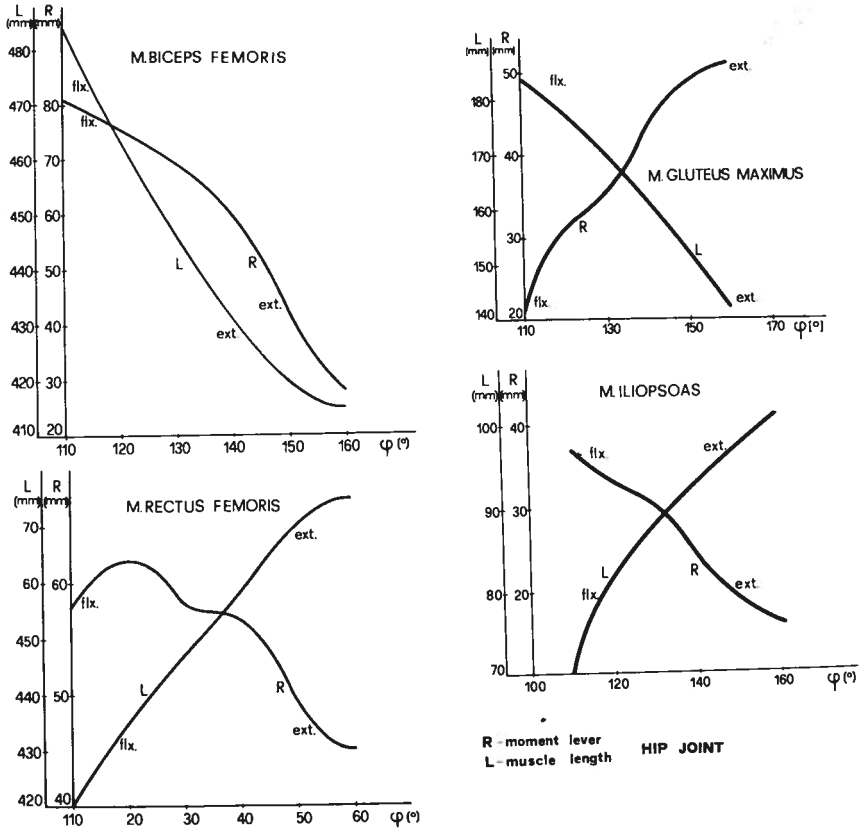


FIGURE 13.—Sagittal plane moment levers for the main leg muscles.

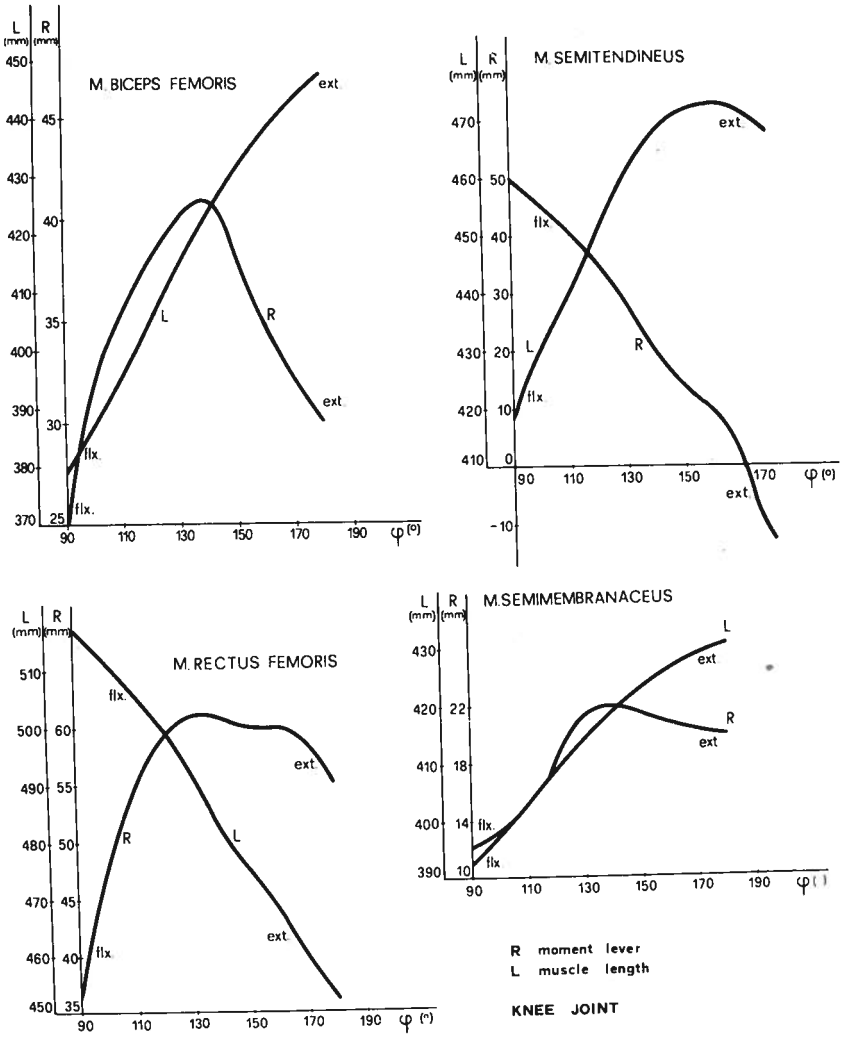


FIGURE 14.—Sagittal plane moment levers for the main leg muscles.

produced by electrical stimulation will enable the patient to stand up. The M. gastrocnemius helps the knee joint rotation axis to stabilize and rise from the ground. In case the center of gravity line through the proper program selection and electrical stimulation of both muscles is kept within the given limits in regard to the distances between the center of gravity line and joint rotation axis, the needed forces of both muscles may be relatively small.

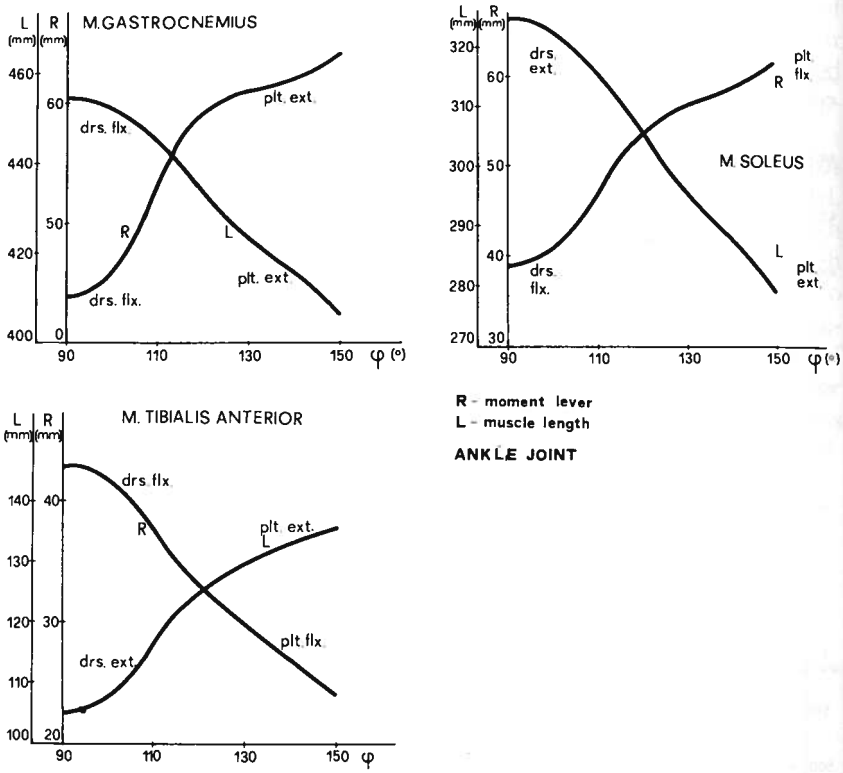


FIGURE 15.—Saggittal plane moment levers for the main leg muscles. (Term plantar extension equals plantar flexion.)

The basis for our stimulation program selection are the EMG records obtained during the standing-up function of a normal subject under conditions similar to those of our paraplegic patient's standing-up experiments. Figure 16 illustrates the joint angles and the EMG for the main muscles acting during the getting-up and sitting-down phase. On the bottom the foot switch record can be seen. The whole experimental setup could be understood from Figure 17. Till now we have made three standing-up experiments with FES. In these experiments our patient's hand support was not appropriate and the setting of the program was too clumsy. In spite of this the patient could get up by means of FES. The P.T. added very slight lateral, ventral, and dorsal corrections of the body position for equilibrium reasons. In case these corrections are not appropriate the forces produced in muscles are not sufficient for lifting up the patient. We believe that with slightly modified

hardware and a stimulation program, the patient would be lifted with FES only and without any external support except by means of the self-balance regulation.

Figure 18 shows the experimental setup arranged for the standing-up procedure. Figure 19 shows the patient during standing. The torque imbalance in the hip and ankle plantar flexion can be seen. Owing to the experimental setup it was difficult to prescribe the patient's posture and to put the body weight-line to a desired position.

Before concluding let us summarize the results and the goals of FES of paraplegic patients. Our experiments should be considered a basis for further research. Before definite conclusions about the future of FES for orthotic use and clinical application are to be made, new investigations should be performed. These should provide more knowledge about FES in general and the problem of fatigue of stimulated muscles which seems to be a great limitation

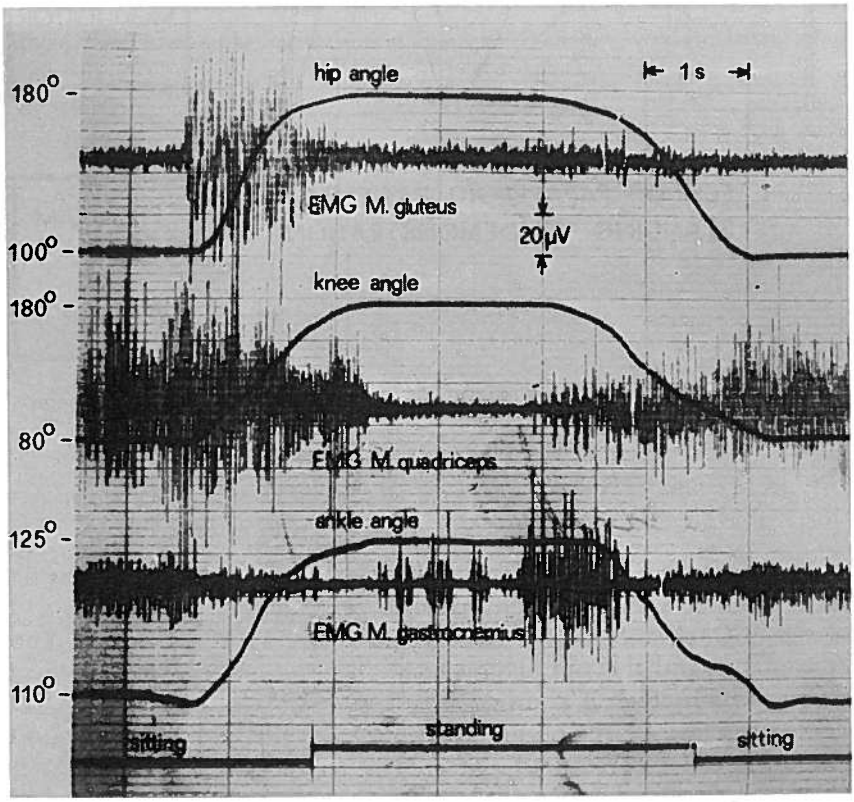


FIGURE 16.—Goniometer and EMG records of the main muscles during standing up in a normal individual.

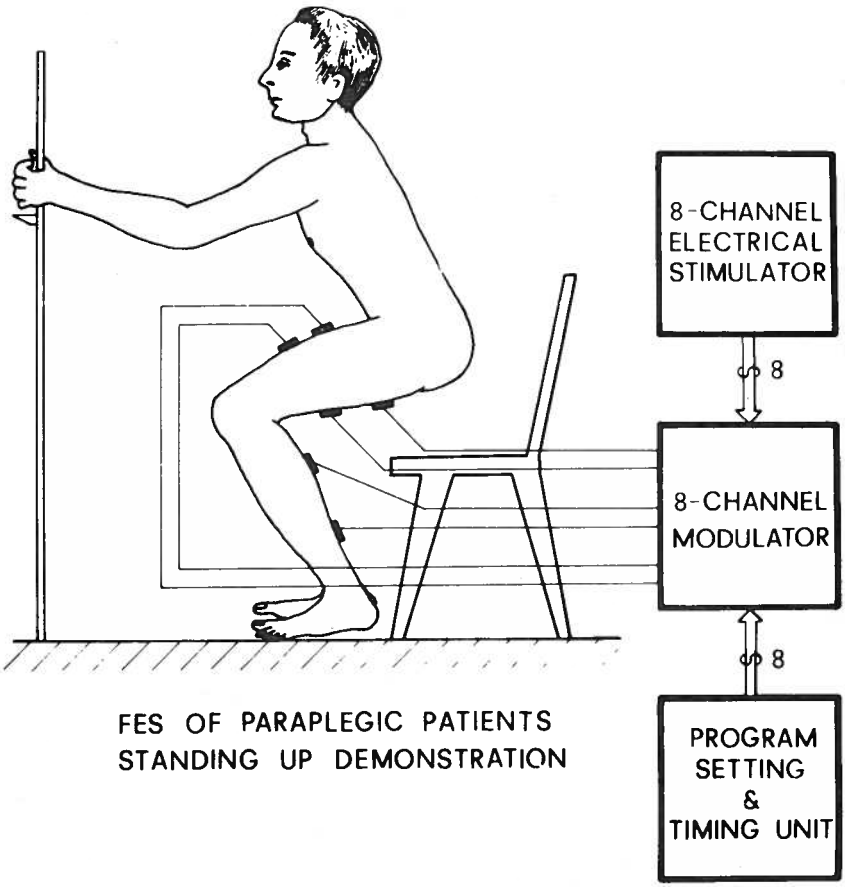
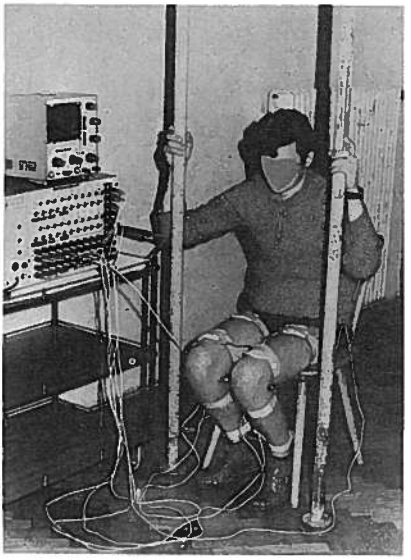


FIGURE 17.—Simplified illustration of the experimental standing up arrangement.

for FES. For this reason FES is difficult to use in weight-bearing tasks. Another very interesting point is the muscle coordination and balance maintenance across one or more joints. Of course the FES repeatability and the stimulation technique should be improved and the neurophysiological knowledge broadened. Effective methods for overcoming spasticity should be found. The whole feedback and sensory needs referring to the control problems still remain unsolved. In spite of the above, we believe that there is promising hope for better motor performance and mobility of paraplegic patients through further research in the area of FES.



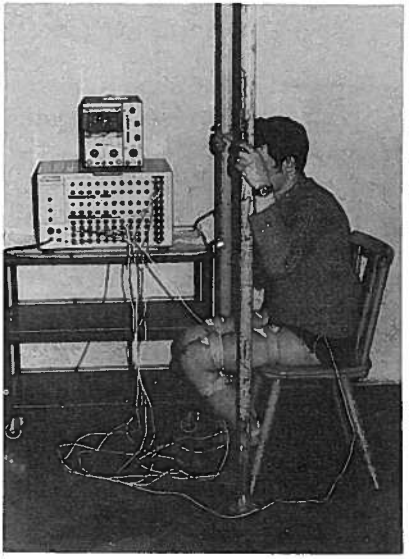


FIGURE 18.—The standing up experiment setup view from the ventral and lateral sides.

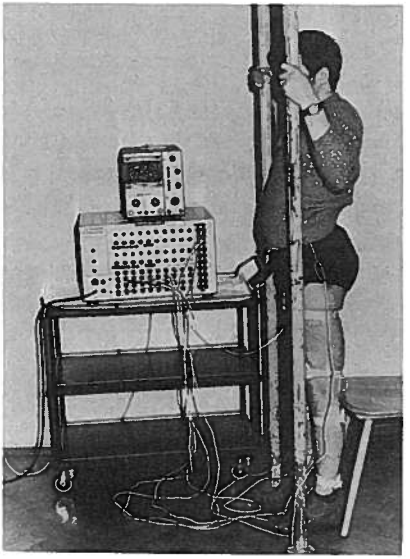


FIGURE 19.—Patient standing.

CONCLUSION

The performed experiments confirm that by electrical stimulation of selected lower-extremity muscles or muscle groups in paraplegic patients, we can obtain forces which are strong enough to enable the lifting of the patient's body from the sitting position. Our experiments could be considered as a basic step for the future introduction to the use of FES devices as a better and more functional substitution for the present-day crude and clumsy orthopedic appliances.

The experiments pointed out that permanent electrical stimulation did not have any influence on the patient's bladder function and that there was no special influence upon the patient's spasticity either.

The performed work carried out thus far proves that paraplegic patients can perform functional movements and tasks by means of FES. Considering the high power amplification of FES, this method seems today superior to all known methods, especially because an orthotic device using FES makes use of the natural bone support, and the lever system does not need any external weight-bearing or force transferring devices. FES can thus provide the best hope for the improvement of locomotion in paraplegic patients. We believe that paraplegics can benefit from FES from both therapeutic and rehabilitation points of view. Having in mind the above results, we are of the opinion that further research should be carried out.

ACKNOWLEDGMENTS

We should like to acknowledge Prof. L. Vodovnik, D.Sc., for his continuous encouragement, advice, and support during this work.

We should like to thank Prof. A. Sirca, D.Sc., from the Institute of Anatomy, Medical Faculty, Ljubljana (for his time-consuming work during the measurement of muscle torque levers and the special X-ray method developed for this purpose).

The authors also wish to express their gratitude to S. Rebersek, Eng., for assistance throughout the experiments and to M. Kovac, P.T., for her valuable work during the muscle strengthening program.

REFERENCES

1. Liberson, W.T., H.J. Holmquest, D. Scot, and M. Dow: Functional Electrotherapy: Stimulation of the Peroneal Nerve Synchronized with the Swing Phase of the Gait of Hemiplegic Patients. Proc. of the 3rd International Congress of Physical

Medicine, Washington, D.C. 1960, Chicago, Ill.: Westlake Press, pp. 705–770, 1962.
2. Long, C., II and V.D. Masciarelli: An Electrophysiologic Splint for the Hand. Arch. Phys. Med., pp. 499–503, Sept. 1963.
3. Technical Data of the Po-8 Electronic Orthopedic and Rehabilitation Walking Aid, Institute of the Socialistic Republic of Slovenia for Rehabilitation of the Disabled, Yugoslavia.
4. Medtronic Inc., Minneapolis, Minnesota, USA. Neuromuscular Assist (NMA) Device, Patient Information, 1972.
5. Philips, G.m.b.H.: Peroneal Muscle Stimulated Operator's Manual, 1970.
6. Jeglic, A., E. Vavken, and M. Benedik: Implantable Muscle or Nerve Stimulator as a Part of an Electronic Brace. Proc. of the Third Int. Symp. on External Control of Human Extremities, Dubrovnik 1969.
7. Kralj, A., A. Trnkoczy, and R. Acimović: Hemiplegic Gait Improvement by Means of a Three-channel Functional Electrical Stimulator. Elektrotehniski vestnik, Ljubljana, pp. a12–a15, 1971.
8. Jeglic, A.: Two-channel Implant Approach to an Orthotic Device. Proc. of the Fourth Intnl. Symposium on External Control of Human Extremities, August 1972, Dubrovnik, Yugoslavia (in press).
9. Vodovnik, L. and S. Rebersek. Myoelectric and Myomechanical Prehension Systems Using Functional Electrical Stimulation. Proc. of the Intnl. Symp.: The Control of Upper Extremity Prostheses and Orthoses, Oct. 1971, Göteborg, Sweden (in press).
10. Vodovnik, L., C. Long II, J.B. Reswick, A. Lippay, and D. Starbuck: Myo-electric Control of Paralyzed Muscles. IEEE Transactions on Bio-Medical Engineering Volume BME–12(3,4):161–172, July–Oct. 1965.
11. Kantrowitz, A.: Electronic Physiologic Aids. A Report of the Maimonides Hospital of Brooklyn, N.Y. 1963.
12. Kralj, A. and S. Grobelnik: Electronic Quadriceps Brace, Brief Description in the Progress Report No. 1, Grant, Social and Rehabilitation Service—YUGO 23–68, Apr. 1969.
13. Willemon, W.K., V. Mooney, D. McNeal, and J. Reswick: Surgical Implanted Peripheral Neuroelectric Stimulation. Rancho Los Amigos Hospital, Los Angeles, Internal Report, 1970.
14. Kralj, A. and S. Grobelnik: Functional Electrical Stimulation of Paraplegic Patients—Feasibility Study. Proc. Sixth Symp. on External Control of Human Extremities, Dubrovnik 1972 (in press).
15. Data Sheet, Orthese Pneumatique "ORTAZUR," Societe Aerazur –58, Boulevard Gallieni, 92 Issy- Les Moulineaux, France, 1971.
16. Rose, G.K. and J.T. Henshaw: Swivel Walker for Paraplegic, Medical and Technical Consideration. Bio-Medical Engineering, 7(9):420–425, Oct. 1972.
17. Rose, D.L., S.F. Radzimirski, and R.R. Beatty: Effect of Brief Maximal Exercise on the Strength of the Quadriceps Femoris. Arch. Phys. Med., 38:157–164, Mar. 1957.
18. Rose, D.L. and P.B. Page: Conscious Proprioception and Increase in Muscle Strength. Arch. Phys. Med. 50:6–10, Jan. 1969.
19. Krusen, F.H., F.J. Kottke, and P.M. Ellwood: Handbook of Physical Medicine and Rehabilitation. W.B. Saunders Co., Philadelphia, London, 1965.
20. Pauwels, F.: Gesammelte Abhandlungen zur funktionellen Anatomie des Bewegungsapparates, Springer Verlag, Berlin, Heidelberg, New York, 1965.
21. Drillis, R., R. Contini, and M. Bluestein: Body Segment Parameters. Artif. Limbs, 8(1):44–66, Spring 1964.
22. Kralj, A, A. Trnkoczy, P. Strojnik, and R. Acimovic: Three-channel Electrical Stimulation of Lower Extremities. Final Report of the Research Grant No. 19–P–

58391–F–01 from the Social and Rehabilitation Service, Department of Health, Education and Welfare, Washington, D.C. 20201, pp. 32–34, University of Ljubljana, Yugoslavia.

23. Kralj, A, A. Trnkoczy, and R. Acimovic: Improvement of Locomotion in Hemiplegic Patients with Multichannel Electrical Stimulation. Proc. of the Conference on Human Locomotor Engineering, University of Sussex, pp. 60–68, Sept. 1971.
24. Waters, R. and D. McNeal: Hip Extensor Weakness. Rehabilitation Engineering Center, Rancho Los Amigos Hospital, Downey, California, Annual Report of Progress, 40 pp., Dec. 1971–Nov. 1972.